University of Strathclyde

Bioengineering Unit

Comparison of Biomechanical and Proprioceptive Function following Resurfacing Arthroplasty and Standard Total Hip Replacement

by

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Table of Contents

		PAGE
ABSTRACT		**xxiii**
LIST OF ABBREVIATIONS		**xxv**
1	**INTRODUCTION**	**1**
1.1	Background and Purpose of the Study	1
1.2	Content of the Thesis	3
2	**LITERATURE REVIEW**	**5**
2.1	Hip Joint Structure and Function	6
	2.1.1 Anatomical considerations	***6***
	2.1.2 Biomechanical considerations	***8***
	2.1.2.1 Influence of anatomy on hip joint kinematics	*8*
	2.1.2.2 Influence of anatomy on hip joint kinetics	*11*
2.2	Osteoarthritis and Degenerative Joint Disease	15
	2.2.1 OA disease process	***16***
	2.2.2 Diagnosis, clinical and surgical implications of OA	***18***
2.3	Total Hip Replacement	20
	2.3.1 History and development of THR	***21***
	2.3.2 Conventional arthroplasty	***23***
	2.3.3 Wear and total hip replacement	***25***
	2.3.3.1 Biomaterials and wear	*26*
	2.3.3.2 The effects of friction and lubrication on wear	*27*
	2.3.3.3 Geometry of the bearings	*30*
	2.3.4 CoC and MoM for the young and active patient	***32***
	2.3.4.1 Ceramic-on-ceramic THR	*32*
	2.3.4.2 Metal-on-metal THR	*34*
	2.3.5 Resurfacing arthroplasty	***34***

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Table of Contents

		PAGE
ABSTRACT		**xxiii**
LIST OF ABBREVIATIONS		**xxv**
1	**INTRODUCTION**	**1**
1.1	Background and Purpose of the Study	1
1.2	Content of the Thesis	3
2	**LITERATURE REVIEW**	**5**
2.1	Hip Joint Structure and Function	6
	2.1.1 Anatomical considerations	***6***
	2.1.2 Biomechanical considerations	***8***
	2.1.2.1 Influence of anatomy on hip joint kinematics	*8*
	2.1.2.2 Influence of anatomy on hip joint kinetics	*11*
2.2	Osteoarthritis and Degenerative Joint Disease	15
	2.2.1 OA disease process	***16***
	2.2.2 Diagnosis, clinical and surgical implications of OA	***18***
2.3	Total Hip Replacement	20
	2.3.1 History and development of THR	***21***
	2.3.2 Conventional arthroplasty	***23***
	2.3.3 Wear and total hip replacement	***25***
	2.3.3.1 Biomaterials and wear	*26*
	2.3.3.2 The effects of friction and lubrication on wear	*27*
	2.3.3.3 Geometry of the bearings	*30*
	2.3.4 CoC and MoM for the young and active patient	***32***
	2.3.4.1 Ceramic-on-ceramic THR	*32*
	2.3.4.2 Metal-on-metal THR	*34*
	2.3.5 Resurfacing arthroplasty	***34***

PAGE

2.3.5.1 History and evolution of resurfacing arthroplasty 35

2.3.5.2 Renaissance-2nd generation resurfacing arthroplasty 37

2.3.5.3 Resurfacing arthroplasty-Contending with the standard 40

2.4 Anatomical reconstruction and surgical procedure 40

2.4.1 Surgical approach ***40***

2.4.2 Surgical procedure using posterior approach ***41***

2.4.3 Differences in surgical procedure for THR and hip resurfacing ***41***

2.4.4 Differences in anatomical reconstruction for THR and hip resurfacing ***42***

2.4.4.1 Application of anatomical reconstruction to hip joint kinematics 43

2.4.4.2 Application of anatomical reconstruction to hip joint kinetics 44

2.4.5 Selection of arthroplasty type ***45***

2.5 Motion Analysis 47

2.6 Gait Analysis 50

2.6.1 Features of normal gait ***50***

2.6.1.1 Kinematics of normal gait 51

2.6.1.2 Kinetics of normal gait 52

2.6.2 Gait and osteoarthritis ***54***

2.6.3 Gait following THR ***55***

2.6.3.1 Kinematic assessment following hip arthroplasty 56

2.6.3.2 Kinetic assessment following hip arthroplasty 57

2.6.4 Factors influencing abductor muscle function ***58***

2.6.5 Study of outcomes following hip resurfacing arthroplasty ***59***

2.6.5.1 Resurfacing arthroplasty-Clinical indicators and subjective data 60

2.6.5.2 Resurfacing arthroplasty-Objective comparisons with standard THR 61

2.7 Analysis of Stair Negotiation 64

PAGE

2.7.1 ***Significance of stair negotiation*** ***64***

2.7.2 ***Differences with level walking*** ***64***

2.7.2.1 Temporospatial parameters *65*

2.7.2.2 Sagittal plane kinematics and kinetics *66*

2.7.2.3 Frontal and transverse plane kinematics and kinetics *68*

2.7.3 ***Critique of stair climbing simulation*** ***70***

2.8 Proprioception 72

2.8.1 Proprioceptive function for the control of movement ***72***

2.8.2 Proprioceptive function of a joint ***74***

2.8.2.1 *Joint position sense and motion detection sense* *74*

2.8.2.2 Proprioception studies of lower limb joints *76*

2.8.2.3 Proprioception studies of the hip joint *78*

2.8.3 Measuring proprioception ***82***

2.9 Study Aims and Hypotheses 84

3 DESIGN OF MOTION DETECTION SENSE PROTOCOL **86**

3.1 Design Aims 86

3.1.1 Purpose ***86***

3.1.2 Previous designs ***86***

3.1.3 Design requirements ***90***

3.1.3.1 Neurological and physiological considerations *90*

3.1.3.2 Mechanical requirements *92*

3.1.3.3 Positional aspects-Freedom and restraints *93*

3.1.3.4 Comfort and safety *95*

3.1.3.5 User requirements *95*

3.2 Design Features 95

3.2.1 Basic structure ***95***

3.2.2 Suspension of the limb ***97***

3.2.3 Limiting motion to the hip joint ***99***

3.2.4 Movement of the limb ***100***

			PAGE
	3.2.5	***Control of limb movement***	*101*
	3.2.6	***Measuring the amplitude of movement of the limb***	*103*
3.3	Protocol Design		105
	3.3.1	***Protocol overview***	*105*
	3.3.2	***Recording angular motion***	*106*
	3.3.3	***Recording angular velocity***	*107*
3.4	Pilot Testing		108
	3.4.1	***Validation of the MDS protocol***	*108*
	3.4.2	***Selection of the angular velocity of MDS stimulus***	*108*
	3.4.3	***Reliability of the MDS protocol***	*109*
4	**METHODOLOGY**		**111**
4.1	Study Design		111
4.2	Subject Recruitment		111
4.3	Instrumentation		115
	4.3.1	***Subjective instruments***	*115*
	4.3.2	***Kinematic analysis***	*116*
	4.3.3	***Kinetic analysis***	*121*
	4.3.4	***Tools for functional analysis***	*121*
	4.3.5	***Proprioception rig***	*123*
4.4	Procedure		125
	4.4.1	***Preparation and baseline measures***	*125*
	4.4.2	***Kinematic analysis***	*125*
	4.4.2.1 Creation of the global co-ordinate system		*125*
	4.4.2.2 Marker set		*127*
	4.4.2.3 Waist segment and pelvic co-ordinate system		*129*
	4.4.2.4 Thigh segment and femoral co-ordinate system		*132*
	4.4.2.5 Shank segment and tibial co-ordinate system		*135*
	4.4.2.6 Foot segment and foot co-ordinate system		*138*
	4.4.2.7 Overview and application of marker set		*139*
	4.4.3	***Functional activities***	*140*

		PAGE
	4.4.4 Subjective assessment	*141*
	4.4.5 Proprioception	*141*
4.5	Pilot Studies	144
	4.5.1 Proprioception rig validity and reliability	*144*
	4.5.2 Motion analysis reliability tests	*144*
5	**DATA ANALYSIS METHODS**	**145**
5.1	Outline of Data Management	145
5.2	Disclosure and Management of Confidentiality Issues	145
5.3	Management of Subjective Data	146
5.4	Management of Motion Analysis Data	146
	5.4.1 Processing of markers and motion trajectories	*146*
	5.4.2 BodyBuilder code	*149*
	5.4.3 Management of data output from BodyBuilder code	*150*
	5.4.4 Additional data outputs	*152*
5.5	Management of Proprioception Data	153
	5.5.1 Raw motion detection sense data	*153*
	5.5.2 Management of motion detection sense data	*153*
5.6	Statistical Methods	156
	5.6.1 Subjective data statistics	*156*
	5.6.2 Statistical methods for motion analysis data	*156*
	5.6.3 Statistical methods for proprioception data	*157*
6	**DESCRIPTIVE STATISTICS AND SUBJECTIVE DATA**	**158**
6.1	Formation and Refinement of the Sample Group	158
	6.1.1 Recruitment	*158*
	6.1.2 Initial sample group selected for the study	*159*
	6.1.3 Final sample group	*161*
6.2	Descriptive Characteristics of the Sample Group	162
	6.2.1 Descriptive data of the sample group	*162*

PAGE

6.2.2 Aetiology and social characteristics of the sample group 164

6.3 Results of the Subjective Data Collection 166

6.3.1 Oxford Hip Score 166

6.3.2 Satisfaction Score 169

6.3.3 Visual analogue scores 170

7 MOTION ANALYSIS RESULTS **172**

7.1 Subject Characteristics 172

7.1.1 Subject exclusions 172

7.1.2 Descriptive assessment of motion 173

7.2 Results from Level Walking 175

7.2.1 Comparison of hip angles between limbs for level walking 175

7.2.1.1 Individual angle profiles during level walking 175

7.2.1.2 Mean peak hip angles during level walking 177

7.2.2 Comparison of hip moments between limbs for level walking 180

7.2.2.1 Individual moment profiles during level walking 180

7.2.2.2 Mean peak hip moments during level walking 182

7.2.3 Comparison of hip angles between arthroplasty groups for level walking 186

7.2.3.1 Average group angle profiles for level walking 186

7.2.3.2 Mean peak hip angles for level walking 193

7.2.3.3 Average group moment profiles for level walking 194

7.2.3.4 Mean peak hip moments for level walking 199

7.2.4 Summary of within- and between-group differences for level walking 200

7.3 Results from Stair Ascent 201

7.3.1 Comparison of hip angles between limbs for stair ascent 201

7.3.1.1 Individual angle profiles during stair ascent 201

7.3.1.2 Mean peak hip angles during stair ascent 202

PAGE

7.3.2 Comparison of hip moments between limbs for stair ascent 205

7.3.2.1 Individual moment profiles during stair ascent 205

7.3.2.2 Mean peak hip moments during stair ascent 206

7.3.3 Comparisons between arthroplasty groups for stair ascent 208

7.3.3.1 Average group angle profiles for stair ascent 208

7.3.3.2 Mean peak hip angles for stair ascent 213

7.3.3.3 Average group moment profiles for stair ascent 213

7.3.3.4 Mean peak hip moments for stair ascent 215

7.3.4 Summary of within- and between-group differences for stair ascent 218

7.4 Results from Stair Descent 219

7.4.1 Comparison of hip angles between limbs for stair descent 219

7.4.1.1 Individual angle profiles during stair descent 219

7.4.1.2 Mean peak hip angles during stair descent 220

7.4.2 Comparison of hip moments between limbs for stair descent 223

7.4.2.1 Individual moments profiles during stair descent 223

7.4.2.2 Mean peak hip moments during stair descent 224

7.4.3 Comparisons between arthroplasty groups for stair descent 227

7.4.3.1 Average group angle profiles for stair descent 227

7.4.3.2 Mean peak hip angles for stair descent 231

7.4.3.3 Average group moment profiles for stair descent 232

7.4.3.4 Mean peak hip moments during stair descent 237

7.4.4 Summary of within- and between-group differences for stair descent 238

7.5 Temporospatial Characteristics of the Motion Tasks 239

7.5.1 Velocity characteristics of level walking 239

		PAGE
	7.5.1.1 Gait velocity of the sample group	*239*
	7.5.1.2 Relationship between gait velocity and moment variables	*240*
	7.5.2 Stance and cycle duration characteristics of the motion tasks	***241***
7.6	Comparative Summary of Functional Tasks	244
8	**PROPRIOCEPTION RESULTS**	**246**
8.1	Subject Characteristics	246
8.2	Early MDS Results - 3 month Data	247
	8.2.1 Comparison of MDS between the operated limb and the control limb	***247***
	8.2.2 Comparison of MDS between the standard THR and resurfacing group	***250***
	8.2.3 Variability of the 3 month post-operative data	***252***
8.3	Late MDS Results - 12 month Data	255
	8.3.1 Comparison of MDS between the operated limb and the control limb	***255***
	8.3.2 Comparison of MDS between the standard THR and resurfacing group	***257***
	8.3.3 Variability of the 12 month post-operative data	***260***
8.4	Comparison of Data at 3 and 12 months post-operation	261
8.5	Summary of MDS findings	263
9	**DISCUSSION**	**264**
9.1	Motion Analysis Findings	265
	9.1.1 Hip angle data: Differences due to hip arthroplasty	***265***
	9.1.1.1 Angular restriction of the operated limb	*266*
	9.1.1.2 Inverse angle patterns between-limbs	*269*
	9.1.2 Hip angle data: Differences due to arthroplasty type	***274***

		PAGE
	9.1.2.1 Differences in sagittal plane angles due to hip arthroplasty type	*274*
	9.1.2.2 Differences in coronal plane angles due to hip arthroplasty type	*277*
	9.1.2.3 Differences in transverse plane angles due to hip arthroplasty type	*278*
	9.1.3 Hip moment data	***279***
	9.1.3.1 Differences due to hip arthroplasty	*280*
	9.1.3.2 Differences due to arthroplasty type	*282*
	9.1.4 Temporal aspects of the motion tasks	***286***
	9.1.5 Comparison of ambulation modes	***287***
9.2	Proprioception Findings	288
	9.9.1 Influence of hip arthroplasty on MDS of the joint	***289***
	9.9.2 Differences in MDS due to arthroplasty type	***292***
9.3	Relating Findings to Subjective & Descriptive Data	294
	9.3.1 Use of subjective outcome measures to predict function	***294***
	9.3.2 Variability in activity levels of subject group	***295***
	9.3.3 Influence of psychological factors	***296***
9.4	Merits and Limitations of the Motion Analysis Study	296
	9.4.1 Merits of the motion analysis protocol	***296***
	9.4.2 Limitations of the motion analysis protocol	***296***
9.5	Merits and Limitations of the Proprioception Study	297
	9.5.1 Merits of the MDS protocol	***297***
	9.5.2 Limitations of the MDS protocol	***298***
	9.5.3 Effectiveness of MDS as an outcome measure	***299***
9.6	Implications of the Study	300
10.0	**CONCLUSION AND FURTHER STUDY**	**304**
10.1	Conclusions of the Study	304
10.2	Recommendations for Further Study	306

	PAGE
REFERENCES	**308**
APPENDIX 1	**330**
Influence of Component Design and Placement	
APPENDIX 2	**331**
Modification of Femoral Offset	
APPENDIX 3	**332**
Definition of Severity of Pain	
APPENDIX 4	**333**
Oxford Hip Score	
APPENDIX 5	**334**
Satisfaction Questionnaire	
APPENDIX 6	**335**
BodyBuilder Code	
APPENDIX 7	**352**
Standardised Instructions for MDS protocol	
APPENDIX 8	**355**
UCLA Activity Score	
APPENDIX 9	**356**
Results Figures	

LIST OF FIGURES

	PAGE
Figure 2.1: Hip joint anatomy	6
Figure 2.2: Hip joint capsule	7
Figure 2.3: Extracapsular ligaments of the hip joint	7
Figure 2.4: Intracapsular ligaments of the hip joint	7
Figure 2.5: Muscles stabilising the hip joint (posterior view)	8
Figure 2.6: Femoral anteversion and acetabular inclination (right side)	10
Figure 2.7: Anatomical and technical axis of the femur	12

PAGE

Figure 2.8: Anatomical features of the proximal femur	13
Figure 2.9: Forces acting on the hip joint during single leg stand	14
Figure 2.10: The normal hip and the osteoarthritic hip	15
Figure 2.11: Radiograph of OA hip (Right)	19
Figure 2.12: Smith-Petersen Mould Arthroplasty	22
Figure 2.13: Charnley "low friction arthroplasty" prosthesis (a) at 22 years (b)	23
Figure 2.14: Stribeck curve describing modes of lubrication	31
Figure 2.15: Ceramic-on-Ceramic Trident™ THR prosthesis	33
Figure 2.16: Resurfacing arthroplasty	34
Figure 2.17: Timeline of the evolution of resurfacing arthroplasty	38
Figure 2.18: Metal-on-metal Durom™ resurfacing prosthesis	39
Figure 2.19: Femoral head preparation for resurfacing arthroplasty	42
Figure 2.20: Normative kinematic data for the hip during level gait	52
Figure 2.21: Normative kinetic data for the hip during level gait	53
Figure 2.22: Comparison of hip flexion angles during walking, stair ascent and descent	67
Figure 2.23: Comparison of external hip flexor moments during walking, stair ascent and descent	68
Figure 2.24: Reciprocal stair gait pattern on a four step stair case	71
Figure 3.1: MDS methodology by Karanjia and Ferguson (1983)	87
Figure 3.2: "Mechanical Stimulator" test rig by Grigg et al (1973)	88
Figure 3.3: Basic structure of the MDS rig, opened for subject access	96
Figure 3.4: Limb suspension mechanism-(a) Beam, (b) Proximal attachment, (c) Distal attachment	97
Figure 3.5: Limb suspension attachments-(a) connector attachments, (b) cradles	98
Figure 3.6: Pelvis fixation in (a) supine and (b) side lying positions	99
Figure 3.7: PROM brace (supine position, 0° knee flexion)	100
Figure 3.8: Pulley System	101
Figure 3.9: Viscous damper	102

PAGE

Figure 3.10: Electrogoniometer placement for measuring (a) hip joint and (b) beam movement	104
Figure 3.11: Circuit set up	107
Figure 3.12: Graph of mean threshold angles as a function of velocity of motion	109
Figure 4.1: Inclusion and exclusion criteria for the recruitment of subjects	113
Figure 4.2: Visual analogue scales of subjective pain and stiffness	115
Figure 4.3: Vicon camera and adjustable tripod stand	116
Figure 4.4: Camera position within the lab space	117
Figure 4.5: Patch-panel with force plate panels (a), Vicon data station (b) and PC (c)	118
Figure 4.6: Calibration Instruments	119
Figure 4.7: Individual 14mm markers (a) and marker clusters for thigh (b) and shank (c) segments	120
Figure 4.8: Pointer for anatomical landmark identification	120
Figure 4.9: Adjacent Kistler force plates	121
Figure 4.10: Force plate steps	122
Figure 4.11: Instrumented staircase	122
Figure 4.12: Proprioception rig	123
Figure 4.13: Overview of MDS instruments for use with rig	124
Figure 4.14: Axis definition for the global co-ordinate system	126
Figure 4.15: L-frame position for static calibration (global axis system)	126
Figure 4.16: Dynamic calibration of the capture volume	127
Figure 4.17: Waist segment marker cluster (anterior and posterior aspects)	130
Figure 4.18: Pointer trial calibration of anatomical landmarks of the pelvis (right ASIS and right PSIS)	130
Figure 4.19: Pelvis anatomical frame and co-ordinate system	131
Figure 4.20: Axis definition for the pelvis co-ordinate system	131
Figure 4.21: Axis definition for the hip joint co-ordinate system	132
Figure 4.22: Thigh segment marker clusters (anterior and lateral aspects)	132

PAGE

Figure 4.23: Pointer trial calibration of anatomical landmarks of the femur (right lateral and right medial epicondyle)	133
Figure 4.24: Axis definition for the femoral co-ordinate system	133
Figure 4.25: Axis definition for the knee joint co-ordinate system	134
Figure 4.26: Joint co-ordinate system for the knee joint	135
Figure 4.27: Shank segment marker clusters and malleoli markers	136
Figure 4.28: Axis definition for the tibial co-ordinate system	136
Figure 4.29: Axis definition for the ankle joint co-ordinate system	137
Figure 4.30: Joint co-ordinate system for the ankle joint	138
Figure 4.31: Foot markers (lateral, anterior and posterior aspects)	138
Figure 4.32: Axis definition for the foot co-ordinate system	139
Figure 4.33: Overview of kinematic modelling method; (a) anatomical reference frames and (b) joint co-ordinate systems	139
Figure 4.34: Subject positioning and set up for abduction MSD trials	143
Figure 4.35: Subject positioning and set up for flexion MSD trials	143
Figure 5.1: Example of subjective pain and stiffness VAS results	146
Figure 5.2: Erratic marker trajectory	148
Figure 5.3: List of data variables output from BodyBuilder code	149
Figure 5.4: Example of raw ASCII data output from BodyBuilder code	150
Figure 5.5: Flowchart of Matlab processes in application to motion analysis data	151
Figure 5.6: Example of real time 3-channel raw MDS output from LabView	153
Figure 5.7: Example of 3-channel MDS data	154
Figure 5.8: Goniometer output and identification of onset of movement	154
Figure 6.1: Subject distribution according to arthroplasty group and assessment phase	162
Figure 6.2: Gender distribution according to arthroplasty group at 3 and 12 months	163
Figure 6.3: Employment distribution according to arthroplasty group	165
Figure 6.4: Results of total OHS according to arthroplasty group	166

	PAGE
Figure 6.5: Comparison of pre-op individual item OHS results between groups	168
Figure 6.6: Comparison of 3 month individual item OHS results between groups	168
Figure 6.7: Comparison of 12 month individual item OHS results between groups	168
Figure 7.1: Comparison of stair turning direction between arthroplasty groups	174
Figure 7.2: Mean 3-D angle profiles typical of operated and control limbs during level walking	176
Figure 7.3a: Sagittal plane view of stick figures during level walking	178A
Figure 7.3: Comparison of 3-D hip angles for standard THR group-operated and control limbs during level walking at (a) 3 months and (b) 12 months	178
Figure 7.4a: Coronal plane view of stick figures during level walking	179A
Figure 7.4: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during level walking at (a) 3 months and (b) 12 months	179
Figure 7.5a: Transverse plane view of stick figures during level walking	180A
Figure 7.5: Mean 3-D moment profiles typical of operated and control limbs during level walking	181
Figure 7.6: Comparison of 3-D hip moments for standard THR group-operated and control limbs during level walking at (a) 3 months and (b) 12 months	183
Figure 7.7: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during level walking at (a) 3 months and (b) 12 months	184
Figure 7.8: Time-Limb variable interaction for mean peak abductor moment during level gait	185
Figure 7.9: Mean hip adduction angles at (a) 3 and (b) 12 months	188
Figure 7.10: Mean hip internal rotation angles at (a) 3 and (b) 12 month	189

PAGE

Figure 7.11: Mean hip flexion angles at (a) 3 and (b) 12 months	190
Figure 7.12: Change in sagittal plane angles with Time and THR type	191
Figure 7.13: Comparison of 3-D hip angles between standard THR and hip resurfacing during level walking at (a) 3 months and (b) 12 months	192
Figure 7.14: Mean hip adductor moments at (a) 3 and (b) 12 months	195
Figure 7.15: Mean hip internal rotator moments at (a) 3 and (b) 12 months	196
Figure 7.16: Mean hip flexor moments at (a) 3 and (b) 12 months	197
Figure 7.17: Comparison of 3-D hip moments between standard THR and hip resurfacing during level walking at (a) 3 months and (b) 12 months	198
Figure 7.18a: Sagittal plane view of stick figures during stair ascent	203A
Figure 7.18: Comparison of 3-D hip angles for standard THR group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months	203
Figure 7.19a: Coronal plane view of stick figures during stair ascent	204A
Figure 7.19: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months	204
Figure 7.20a: Transverse plane view of stick figures during stair ascent	205A
Figure 7.20: Mean internal rotator moment profiles typical of operated and control limbs during stair ascent	205
Figure 7.21: Comparison of 3-D hip moments for hip resurfacing group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months	207
Figure 7.22: Mean hip adduction angles at (a) 3 and (b) 12 months	209
Figure 7.23: Mean hip internal rotation angles at (a) 3 and (b) 12 months	210
Figure 7.24: Mean hip flexion angles at (a) 3 and (b) 12 months	211
Figure 7.25: Comparison of 3-D hip angles between standard THR and hip resurfacing during stair ascent at (a) 3 months and (b) 12 months	212

PAGE

Figure 7.26: Mean hip adductor moments at (a) 3 and (b) 12 months 214

Figure 7.27: Mean hip internal rotator moments at 12 months 215

Figure 7.28: Comparison of 3-D hip moments between standard THR and hip resurfacing during stair ascent at (a) 3 months and (b) 12 months 217

Figure 7.29: Mean 3-D angle profiles typical of operated and control limbs during stair descent 219

Figure 7.30a: Sagittal plane view of stick figures during stair descent 221A

Figure 7.30: Comparison of 3-D hip angles for standard THR group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months 221

Figure 7.31a: Coronal plane view of stick figures during stair descent 222A

Figure 7.31: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months 222

Figure 7.32a: Transverse plane view of stick figures during stair descent 223A

Figure 7.32: Mean 3-D moment profiles typical of operated and control limbs during stair descent 224

Figure 7.33: Comparison of 3-D hip moments for standard THR group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months 225

Figure 7.34: Comparison of 3-D hip moments for hip resurfacing group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months 226

Figure 7.35: Mean hip adduction angles for stair descent at 12 months 228

Figure 7.36: Mean hip flexion angles at (a) 3 and (b) 12 months 229

Figure 7.37: Comparison of 3-D hip angles between standard THR and hip resurfacing during stair descent at (a) 3 months and (b) 12 months 230

Figure 7.38: Mean hip adductor moments during stair descent at (a) 3 and (b) 12 months 234

Figure 7.39: Mean hip internal rotator moments during stair descent at (a) 3 and (b) 12 months 235

PAGE

Figure 7.40: Mean hip flexor moments during stair descent at (a) 3 and (b) 12 months 236
Figure 7.41: Comparison of 3-D hip moments between standard THR and hip resurfacing during stair descent at 12 months 237
Figure 7.42: Gait velocity during level walking 240
Figure 7.43: Cycle duration for stair negotiation 242
Figure 7.44: Comparison of stance phase duration between groups during stair descent 244
Figure 8.1: Comparison of mean threshold detection angles for operated and control limbs of subjects with standard THR 248
Figure 8.2: Comparison of mean threshold detection angles for operated and control limbs of subjects with MoM resurfacing 249
Figure 8.3: Comparison of mean threshold detection angles of the operated limb between arthroplasty groups 250
Figure 8.4: Comparison of mean threshold detection angles of the control limb between arthroplasty groups 252
Figure 8.5: Comparison of between-subject variability of the operated and control limb for subjects with (a) CoC standard THR and (b) MoM resurfacing 253
Figure 8.6: Comparison of mean threshold detection angles for operated and control limbs of subjects with CoC standard THR 255
Figure 8.7: Comparison of mean threshold detection angles for operated and control limbs of subjects with MoM resurfacing 257
Figure 8.8: Comparison of mean threshold detection angles for subjects with CoC standard THR and MoM resurfacing 258
Figure 8.9: Comparison of mean threshold detection angles of the control limb for subjects with CoC standard THR and MoM resurfacing 260
Figure 8.10: Comparison of mean threshold detection angles between 3 and 12 month for hips with CoC standard THR 262
Figure 8.11: Comparison of mean threshold detection angles between 3 and 12 month for hips with MoM resurfacing arthroplasty 262

Figure 9.1: Femoral offset comparison between resurfacing (a) and standard THR (b) 283

LIST OF TABLES PAGE

Table 2.1: Full active range of motion of the healthy hip joint 9
Table 2.2: Comparison of hip joint centre estimation methods 49
Table 2.3: Comparison of resurfacing and standard arthroplasty during gait 63
Table 2.4: Literature review of staircase dimensions and characteristics 70
Table 2.5: Literature review of velocity of motion used in lower limb joint MDS studies 75
Table 3.1a: Design specifications for MDS apparatus 91A
Table 3.1: Motion detection terminology 105
Table 4.1: Marker set used for kinematic analysis 128
Table 6.1: Reason for exclusion from study 159
Table 6.2: Standard THR subject exclusions from 3 month data analysis 161
Table 6.3: Reasons for reduced follow-up at 12 months 162
Table 6.4: Age data for sample group 163
Table 6.5: OHS results summary 167
Table 6.6: Satisfaction score summary at 3 month assessment phase 169
Table 6.7: Satisfaction score summary at 12 month assessment phase 169
Table 6.8: Visual analogue pain scores 170
Table 6.9: Visual analogue stiffness scores 170
Table 7.1: Frequency of preferential stair turning patterns 174
Table 7.2: Mean peak hip angles of the operated limb during level walking 193
Table 7.3: Mean peak hip moments of the operated limb during level walking 200
Table 7.4: Mean peak hip angles of the operated limb during stair ascent 213
Table 7.5: Mean peak hip moments of the operated limb during stair ascent 216

PAGE

Table 7.6: Mean peak hip angles of the operated limb during stair descent 231

Table 7.7: Mean peak hip moments of the operated limb during stair descent 237

Table 7.8: Gait velocity during level walking at 3 and 12 months 239

Table 7.9: Stance phase duration summary for all ambulation modes 243

Table 7.10: Percentage stance phase for all modes of ambulation 243

Table 8.1: Data summary for the operated limb at 3 months 251

Table 8.2: Data summary for the control limb at 3 months 251

Table 8.3: Within-subject variability of MDS for the operated limb at 3 months 254

Table 8.4: Within-subject variability of MDS for the control limb at 3 months 254

Table 8.5: 12 month data summary for the operated limb 259

Table 8.6: 12 month data summary for the control limb 259

Table 8.7: Within-subject variability of MDS for the operated limb at 12 months 261

Table 8.8: Within-subject variability of MDS for the control limb at 12 months 261

Abstract

Resurfacing arthroplasty is gaining popularity as an alternative method of hip joint reconstruction for younger individuals. Hip resurfacing has several perceived benefits over the conventional total hip arthroplasty (THA), including a greater range of motion offered by the larger diameter bearings and improved abductor muscle function due to femoral neck preservation. Literature reports that hip resurfacing often provides a 'natural' feel and allows greater functional return, which may be related to improved proprioception. However, existing studies have failed to fully substantiate the functional merits of resurfacing arthroplasty over those provided by the traditional stemmed THA.

Outcomes were evaluated for 31 patients with primary unilateral resurfacing or standard stemmed THA at 3 and 12 months following surgery. Kinematic and kinetic data were collected while subjects completed level walking, stair ascent and stair descent activities. 3-dimensional hip moments and angles were compared between the arthroplasty groups. Threshold motion detection sense of the hip was tested to quantify hip joint proprioception, by administering passive abduction and flexion motion stimuli using a validated test rig and comparing threshold detection angles.

Hip moments showed no statistically significant difference due to arthroplasty type. Slightly greater peak hip angles were achieved by those with standard THA. The resurfacing group showed greater walking velocities. Threshold detection angles were statistically similar.

Resurfacing and standard THA demonstrated equivalent proprioceptive and functional outcomes. Motion detection sense did not differ due to arthroplasty type. Preserving the femoral offset did not appear to benefit abductor function and the greater diameter resurfacing bearings did not result in greater functional range of motion. The increased offset and greater head-neck ratio provided by the prosthetic

neck may benefit abductor function and increase motion to an extent which meets the benefits of hip resurfacing. Given that there are greater risks and difficulties associated with hip resurfacing, standard THA may be viewed as the more desirable alternative for young active patients.

List of Abbreviations

AJC	Ankle joint centre
Al_2O_3	Aluminium oxide or Alumina
ALVAL	Aseptic lymphocytic vasculitis-associated lesions
ASIS	Anterior superior iliac spine
BHR	Birmingham Hip Replacement
CoC	Ceramic-on-ceramic
CoCr	Cobalt-chromium
CoCrMo	Cobalt-chromium-molybdenum
COM	Centre of mass
DDH	Developmental dysplasia of the hip
F_x, F_y, F_z	Force $_x$, Force $_y$, Force $_z$
FP	Force plate
GT	Greater trochanter
H	Height
HJC	Hip joint centre
JCS	Joint co-ordinate system
KJC	Knee joint centre
L	Length
M-PE	Metal-on-polyethylene
MDS	Motion detection sense
MoM	Metal-on-metal
MSE	Mean squared error
MT	Metatarsal
M_x, M_y, M_z	Moment $_x$, Moment $_y$, Moment $_z$
NHS	National Health Service
OA	Osteoarthritis
OHS	Oxford Hip Score
PC	Personal computer
PE	Polyethylene
PELF	Origin of pelvis co-ordinate system

PROM	Passive range of motion
PSIS	Posterior superior iliac spine
RA	Resurfacing arthroplasty
ROM	Range of movement
SACR	Sacrum
SD	Standard deviation
THA	Total hip arthroplasty
THR	Total hip replacement
UHMWPE	Ultra-high molecular weight polyethylene
VAS	Visual analogue score
VI	Virtual instrument
W	Width
WB	Weight-bearing
ZrO_2	Zirconium dioxide or Zirconia

1 Introduction

1.1 Background and Purpose of the Study

Osteoarthritis of the hip joint is extremely prevalent and is responsible for a vast number of hip arthroplasties conducted in order to relieve pain, restore function and improve the health related quality of life of those suffering from hip joint disease. Total hip arthroplasty (THA) is one of the most successful joint replacement procedures and the conventional arthroplasty, with a metal-on-polyethylene articulation is well established and provides excellent clinical results as demonstrated by elderly patient groups. However, the use of conventional THA in young and active patients is associated with high failure rates due to osteolysis and aseptic loosening of the components in response to polyethylene wear debris, leading to eventual early revision of the implant.

There are several arthroplasty options available for young and active individuals. These feature low wear bearing combinations, such as metal-on-metal and ceramic-on-ceramic, incorporated into the standard stemmed THA design. However, the growing popularity of 2nd generation metal-on-metal hip resurfacing has provided a potentially superior method of hip arthroplasty, which has possible merits over the standard THA.

Resurfacing arthroplasty is bone conserving and maintains the anatomic neck of femur, so may allow greater ease of revision surgery, thus suiting it to the younger patient group who might outlive the conventional arthroplasty. Preservation of the femoral neck is also thought to result in superior reconstruction of the hip, to better reflect the anatomy of the pre-diseased joint, compared to the use of a stemmed prosthesis. Moreover, hip resurfacing uses larger diameter bearings which result in greater range of hip motion in vitro. Therefore, the more anatomic reconstruction method combined with a large diameter bearing is thought to benefit in vivo biomechanics of the hip by optimising kinetic and kinematic function of the joint.

Early results of hip resurfacing have been promising, although the outcome measures used have largely been subjective and relate to clinical indicators as opposed to objective measures of post-operative function. Subjective reports from patients document a 'natural feel' following hip resurfacing, which is not reported following standard THA, and these patients often return to greater levels of function. These reports suggest that resurfacing may provide greater proprioceptive feedback and stability, and may optimise kinematic and kinetic function of the joint as suggested by the greater post-operative activity levels associated with this patient group.

Only one published study has assessed objective outcomes relating to function following resurfacing arthroplasty. Mont et al (2007) compared kinetic and kinematic hip outcome measures between subjects with resurfacing and standard THA during gait analysis, and concluded that hip resurfacing provided superior function which better reflected normative data. There were however, potential errors in the interpretation of the data, and the results were largely inconclusive.

The need for greater information regarding the differences between resurfacing and standard THA have been identified (Loughead et al 2005). Currently there is limited information which orthopaedic surgeons can use to adequately inform their patients for guiding the selection of their type of hip arthroplasty; a ceramic-on-ceramic standard THA or a metal-on-metal hip resurfacing arthroplasty. Research is required in order to substantiate the claims in favour of hip resurfacing. Such research may allow provision of evidence based informed consent for patients and greater confidence for the surgeon in his or her selection of hip arthroplasty prosthesis.

The current study aims to investigate the differences in functional performance of individuals having had hip resurfacing arthroplasty compared to those with conventional THA, by assessing biomechanical aspects of their level gait and stair climbing activities and by characterising joint proprioception. By quantifying such differences between the subjects with resurfacing or standard THA, this research intends to present recommendations for the use of the two main arthroplasty options currently available for the young active population.

1.2 Content of the Thesis

The following chapters document the content of work completed in order to substantiate the current research question, test the hypotheses of the study, manage and present the data obtained and relate these results back to the research question.

Chapter 2 presents a literature review highlighting the structural and potential functional differences between resurfacing and standard hip replacement. The functional anatomy of the hip joint will be documented and the relevant kinematic, kinetic and proprioceptive function will be discussed with respect to the natural and reconstructed joint. An overview of osteoarthritis and hip arthroplasty will be conducted and placed into context of the current young study population. Aspects of hip arthroplasty prostheses will be considered in order to form an understanding of, and justification for the use of the current stemmed hip replacement and hip resurfacing, including the components, bearing materials and procedures used. Functional ambulation and proprioception will be reviewed and the methods of objectively recording outcome measures relating to human motion and proprioception will be assessed. Following a balanced assessment of this information, the aims and hypotheses of the study will be formulated.

Critique of the methods of hip joint proprioception measurement, as reviewed in chapter 2, will form the basis for the design of a new motion detection sense protocol. Chapter 3 will present the planning of the protocol and the design features, construction and implemented use of a custom built proprioception rig in order to objectively measure detection of hip motion, by way of characterising hip joint proprioception.

Chapter 4 will present the methods used for the analysis of the functional ambulatory tasks and implementation of the motion detection sense protocol. Subject recruitment and means of descriptive and subjective data collection will also be provided. The data obtained from the motion analysis and proprioception methodologies will be managed in a standardised means, as reported in chapter 5, in order to allow the direct comparison of outcomes relating to the arthroplasty groups. Chapter 5 will also outline the statistical methods used for data analysis.

The results chapters which follow will outline the descriptive data relating to the sample group and the subjective results (chapter 6), followed by the motion analysis results (chapter 7) and finally the proprioception results (chapter 8). For all outcome measures, the differences due to hip arthroplasty (comparison between limbs) and post-operative time (comparison between test intervals) variables will be assessed as well as the main variable of the differences due to arthroplasty type.

The findings will then be appraised in chapter 9 and conclusions made regarding the difference in motion and proprioception outcomes due to the type of hip arthroplasty. The importance of the findings will be highlighted and the implications of the results for the patient, clinical and industrial population discussed. Recommendations for further study will be stated in chapter 10. The most appropriate type of hip arthroplasty for the young and active population will be judged in response to the findings of the study and recommendations for its use promoted.

2 Literature Review

The content of the current chapter aims to outline the relevant background to the study and the published literature substantiating the research question. As the study aims to determine the differences in function of patients with hip resurfacing arthroplasty compared to patients with a standard THR, the fundamental differences between the types of arthroplasty will be discussed, and an argument as to why there is an expectation of functional differences between the patient groups will be presented.

Differences in the structural geometry of the prosthetic implants and differences in the surgical procedure for implantation will be considered for resurfacing arthroplasty compared to standard stemmed total hip arthroplasty (THA). Consideration will be given to anatomical and biomechanical features of the natural hip joint and the possible effects of hip joint reconstruction on anatomy and biomechanics will be discussed.

A synopsis of the evolution of hip joint arthroplasty will be given to establish an understanding and justification of the use of the current types of arthroplasty in this study. This section will review developments in the structure and material properties of various THR designs in order to enhance performance and longevity of implants.

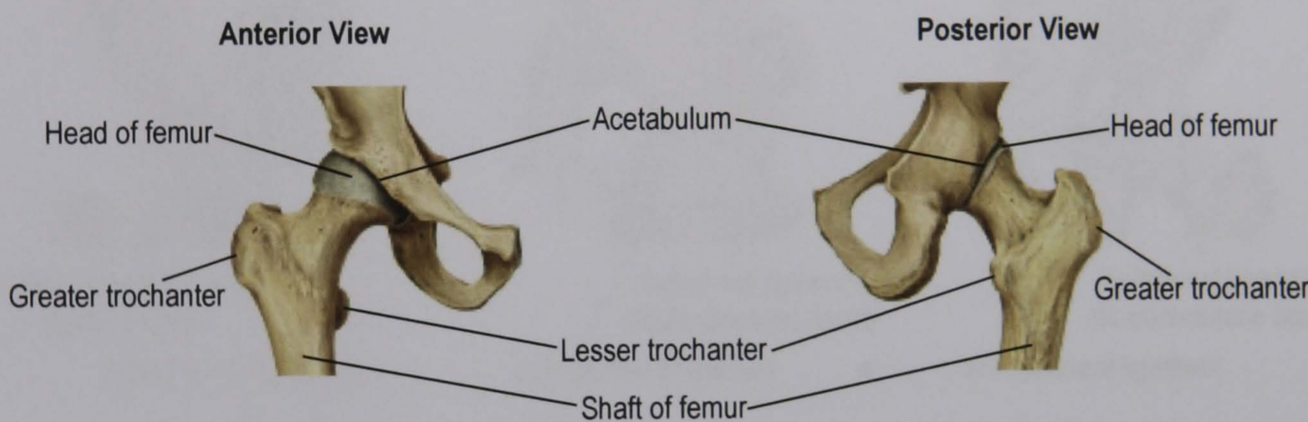
Aspects of function and ambulatory assessment will be discussed with reference to the current subject group. Previous studies concerning motion analysis of the THR population will be reviewed and the various parameters for measurement and possible ambulatory tasks with which to measure functional performance will be evaluated. Following discussion of biomechanical function, the relatively uncharted subject area of proprioceptive function of the hip joint will be investigated. The importance of hip joint proprioception, methods of measuring proprioception and the possible effects of hip arthroplasty on proprioception will be reviewed.

2.1 Hip Joint Structure and Function

The anatomical structure and mechanical function of the hip joint were considered for the 'normal' healthy joint. Alterations in anatomy and function in response to joint pathology implicated by osteoarthritis and due to hip joint reconstruction following arthroplasty were also investigated and will be alluded to in the current section and elaborated on in sections to follow.

2.1.1 Anatomical considerations

The hip joint is the articulation between the pelvis and the femur, characterised by the joint surface apposition of the acetabulum and the head of femur (figure 2.1) creating a synovial ball and socket, or ovoid joint. The structure of the hip joint provides a unique balance of stability and diverse mobility (Heybeli and Mumcu 1999).



Pictures adapted from www.thiemeteachingassistant.com

Figure 2.1: Hip joint anatomy

The stability of the hip joint is achieved by the depth and congruence of the articular surfaces, further deepened and stabilised by the acetabular labrum. The fibrocartilagenous labrum increases acetabular coverage to just over half of the surface area of the femoral head. Stability is further enhanced by the presence of the strong joint capsule (figure 2.2), the capsular ligaments and the surrounding joint musculature. The ligaments surrounding the joint include the iliofemoral, ishiofemoral and the pubofemoral ligaments (figure 2.3), so called due to their bony attachments. The intracapsular ligaments include the ligamentum teres and the transverse ligaments (figure 2.4), although their functional role remains limited and bears little influence on joint stability.

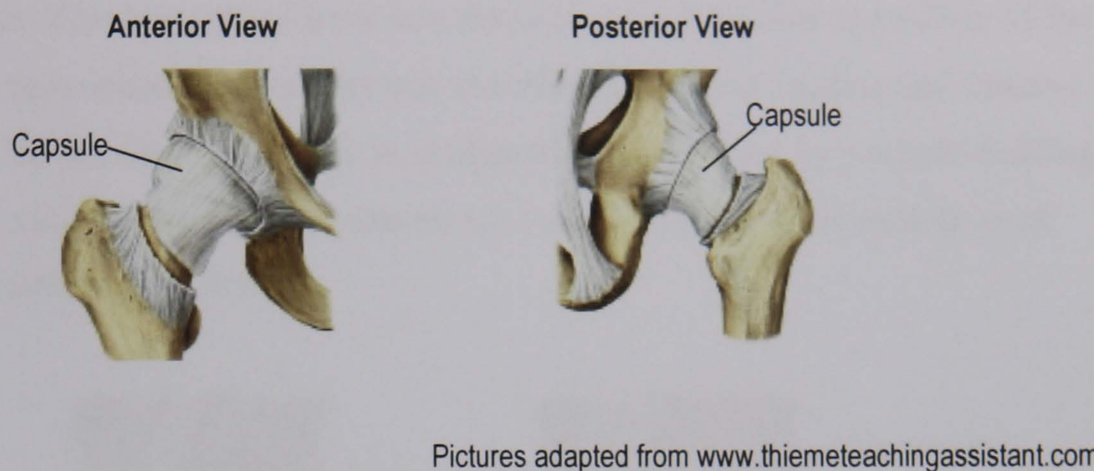


Figure 2.2: Hip joint capsule

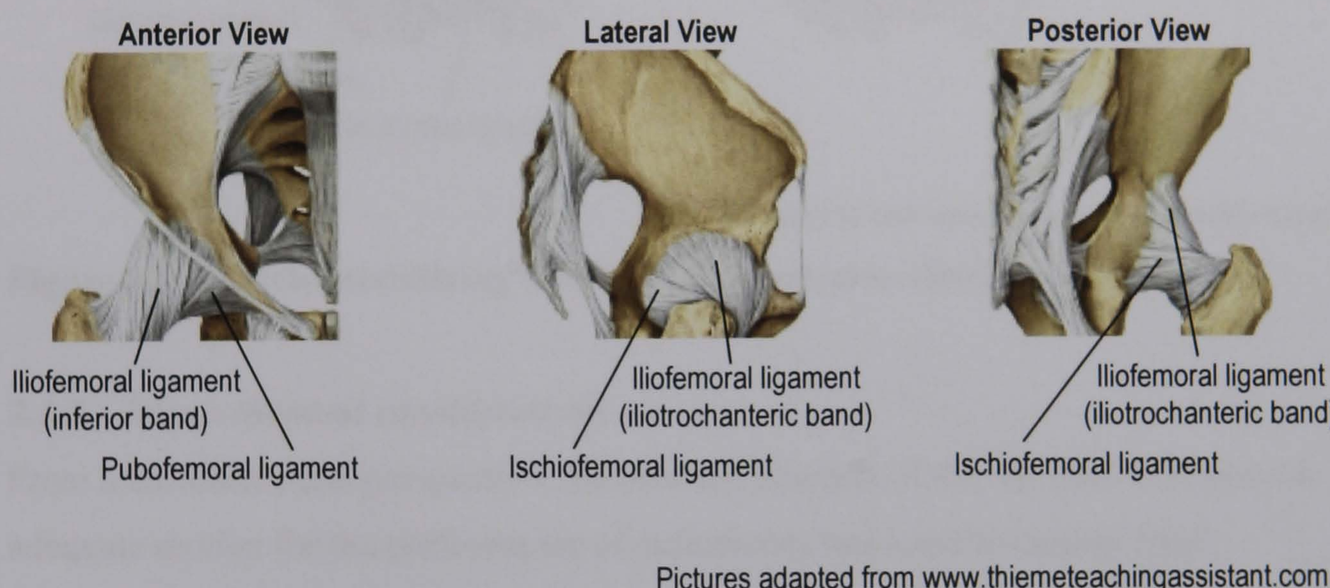


Figure 2.3: Extracapsular ligaments of the hip joint

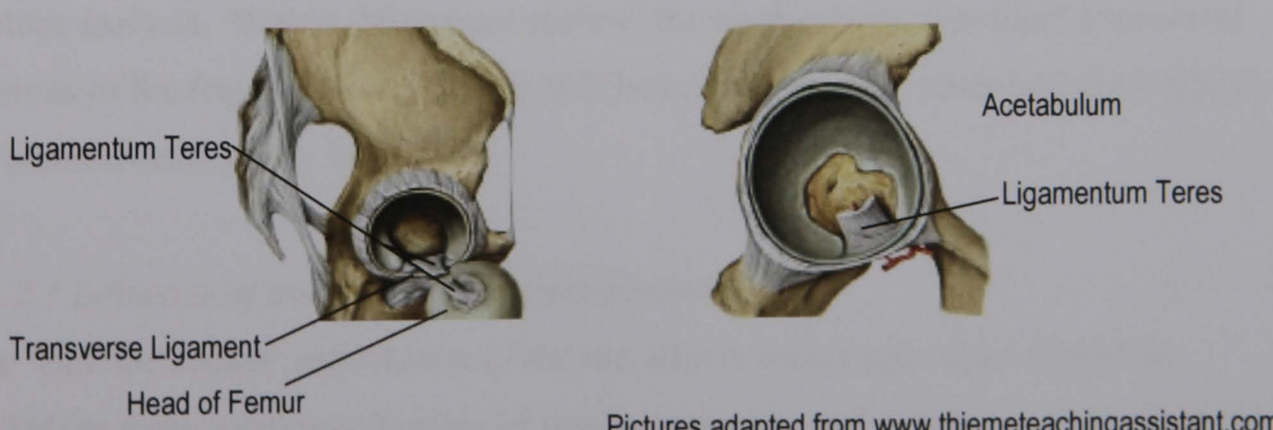
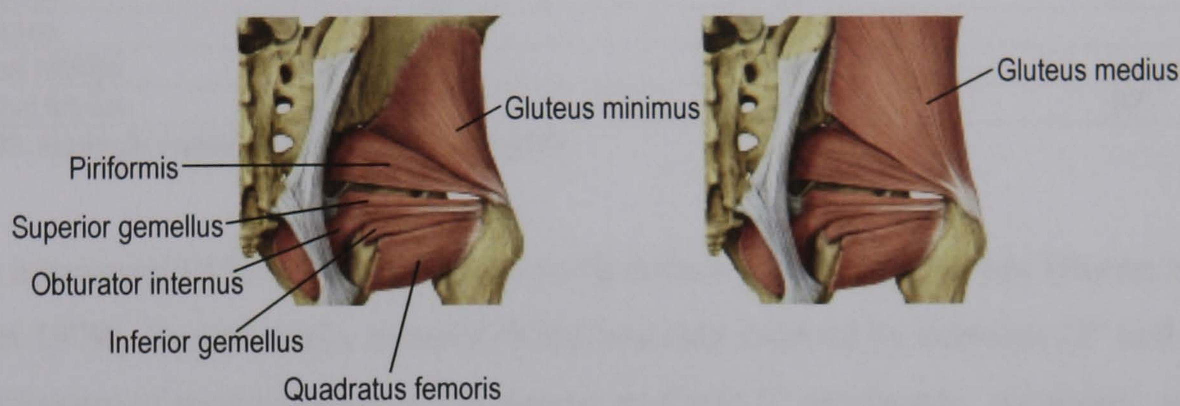


Figure 2.4: Intracapsular ligaments of the hip joint

Certain muscles around the hip joint act as dynamic stabilisers (Kapandji 1970, cited in Fagerson 1998), performing a similar role to that of the rotator cuff muscle group of the shoulder joint. These 'stabilising' muscles include piriformis, obturator externus (deep to obturator internus), gluteus medius and gluteus minimus (figure 2.5). Their line of action applies force in a direction which favours apposition of the joint surfaces, thus enhancing stability and contributing to joint loading and internal joint forces. Gluteus medius has role in maintaining equilibrium by postural holding and resisting moments about the hip (section 2.1.2.2) using eccentric muscle work (Cromerford and Mottram 2001).



Pictures adapted from www.thiemeteachingassistant.com

Figure 2.5: Muscles stabilising the hip joint (posterior view)

2.1.2 Biomechanical considerations

From a biomechanical perspective, the primary function of the hip joint is to provide adequate motion for the performance of ambulatory tasks and to control load transference form the pelvis to the femur. The technical aspects of hip joint biomechanics will be discussed in section 2.5 within the context of 3-dimensional motion analysis. Within the current section, the implications of several anatomical features of the femur and acetabulum will be considered with respect to their effects on biomechanics of the hip.

2.1.2.1 Influence of anatomy on hip joint kinematics

The 'ball and socket' articulation of the hip allows for a large range of motion (ROM) in three rotational degrees of freedom; flexion and extension, abduction and adduction and internal and external rotation (table 2.1). The variation in maximum angular measurements for the planar movements of the hip (table 2.1), is possibly due to the variations in the age of the subjects, the measurement techniques used and the resting positions of the joints from which the maximum range was recorded.

Table 2.1: Full active range of motion of the healthy hip joint

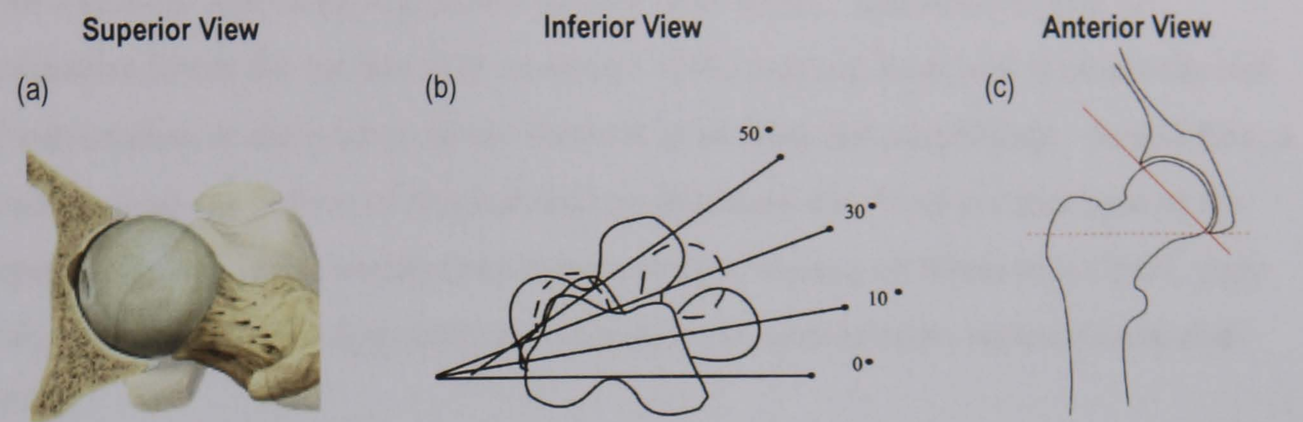
PLANAR MOVEMENT	Full active range of movement		Functional movement
	AAOS* (1965)	Boone and Azen (1979)	D'Lima et al (2000)
Flexion	113°	122.3° ±6.1°	90°-120°
Extension	28°	9.8° ±6.8°	-
Abduction	48°	45.9° ±9.3°	20°
Adduction	31°	26.9° ±4.1°	-
Internal rotation	45°	47.3° ±6.0°	-
External rotation	45°	47.2° ±6.3°	20°

*AAOS: American Academy of Orthopaedic Surgeons

The maximum ROM of the hip is known to reduce with increasing age (Boone and Azen 1979). For example, rotation ROM becomes reduced by between 15° and 20° by 20 years of age and continues to reduce by about 5° per decade. Similarly, about 10° to 15° of abduction tends to be lost in the first two decades, but appears to remain stable until the elderly years (Boone and Azen 1979). Despite the large range of full active movement of the hip, the angular range required for carrying out the various functional tasks of daily living is considerably smaller (Woolson et al 1985) and involves circumduction movement, featuring a combination of planar movements occurring simultaneously around the three joint axes (Fagerson 1998).

The physiological limit of hip ROM may be due to soft tissue tension generated in the surrounding ligaments and muscles or impingement of bone on bone when the femoral neck comes into contact with the acetabular rim (Sariali et al 2008). Variation in the maximum ROM achieved before impingement occurs is dependent on individual variations in bony architecture. The angle of anteversion of the femoral neck and the differential in the diameter of the femoral head and neck (head-neck ratio) are two important features of the anatomic femur which are directly related to hip joint kinematics (Sariali et al 2008).

The angle of anteversion of the femoral neck "describes the anterior rotation of the femoral neck relative to the anatomical axis of the femoral shaft" (Sariali et al 2008, page 371) or relative to the transtrochanteric axis of the femur (figure 2.6).



Pictures taken from www.thiemeteachingassistant.com, scieloprueba.sld.cu and wikimedia.org

Figure 2.6: Femoral anteversion (a, b) and acetabular inclination (c) (right side)

Greater angles of femoral anteversion may increase the angles of flexion (D'Lima et al 2000, Sariali et al 2008) and internal rotation motion (Sariali et al 2008) available at the hip before bony impingement of the neck on the acetabular rim, but may in turn reduce the range of extension, adduction and external rotation (D'Lima et al 2000). There is a complex relationship between the acetabular and femoral anteversion (Yoshimine 2006) and hence restrictions in motion will also depend on the anatomic orientation of the acetabulum, as defined by the inclination (figure 2.6) and anteversion of the acetabulum (Murray 1993), which are known to vary considerably between individuals (Sariali et al 2008).

The head-neck ratio of the femur is important in determining the amount of movement which can occur at the hip joint before impingement of the neck on the acetabular rim (Sariali et al 2008). Greater head-neck ratios (large diameter head and relatively smaller diameter neck) result in greater ROM before impingement occurs (Yoshimine and Ginbayashi 2002). At the point of bony impingement, a force acting to continue the hip motion beyond the physiological limits may potentially result in dislocation of the hip joint. The incidence of dislocation of the anatomic hip is inherently low due to the high conformity and net stability of the joint and is therefore mostly a result of traumatic injuries (Palastanga 1998). However, dysfunction of these aforementioned structures or anatomical differences in bony architecture may elevate the dislocation risk. An increased acetabular inclination angle (figure 2.6) results in a more vertical orientation of the acetabulum, consistent with a clinical sign of hip dysplasia (Ipavec et al 1999). Increased angles of inclination lower the surface area coverage of the femoral head and increase the risk of subluxation or dislocation of the femoral head from the acetabulum. In addition, a more vertical orientation of the acetabulum increases the "load per unit area of the superior aspect" of the acetabulum during weight bearing (Affatato et al 2004, page 400), predisposing the dysplastic-type hip joint to osteoarthritic wear (Crowe et al 1979).

2.1.2.2 Influence of anatomy on hip joint kinetics

During dynamic activities, the bony congruity of the joint surface and the acetabular labrum limit the magnitude of translation forces, while the ligamentous and muscular structures control rotational forces at the hip joint (Sariali et al 2008). Physiological features of the joint cartilage and synovium manage the residual forces and joint reaction forces that result.

In the healthy hip joint, both acetabular and femoral head surfaces are covered with articular, or hyaline, cartilage. The cartilage is considered to acts as a 'shock absorber' (Simon 1999), although it may be better associated with the distribution of load across the articular surface, where the underlying cancellous bone is responsible for energy absorption (Romanovskaya et al 1986). Cartilage is thickest in the major weight bearing (WB) areas including the superior lunate surface of the acetabulum and the anteromedial surface and superiomedial surface of the femoral head. The articular cartilage contains proteoglycans constrained within type II collagen. The proteoglycans retain water within the cartilage to maintain hydration and provide resilience, while the collagen provides tensile strength (Simon 1999). Load tolerance is enhanced by the hydrostatic pressure within the cartilage tissue (Macirowski et al 1994, cited in Fagerson 1998) allowing transfer of load to the subchondral bone (Simon 1999) and the low co-efficient of friction of the cartilage surface which minimises wear in response to translation forces (Sariali et al 2008). Fagerson

(1998) reported that the cartilage surface features small undulations and irregularities. These deepen and advance with age, which may contribute to age related wear of the natural hip joint, leading to osteoarthritic changes.

The synovial membrane of the hip joint is a dense, smooth connective tissue which produces synovial fluid to lubricate the joint. Lubrication is important for sliding motion within the joint (Hannouche et al 2005), to minimise friction and the resultant wear of adjacent joint surfaces (Silva et al 2005). Synovial fluid has been characterised as thixotropic (Silva et al 2005); it exhibits more fluid-like properties with joint motion and becomes semisolid during static joint positions, hence the fluid viscosity is a function of velocity of joint movement and shear strain rate (Schey 1996, cited in Silva 2005).

The anatomical structure of the femur is such that the head of femur is dissociated from the long axis of the bone by the femoral neck. The biomechanical significance of this feature is that it creates two femoral axis systems; the anatomical axis and the technical axis of the femur (figure 2.7).

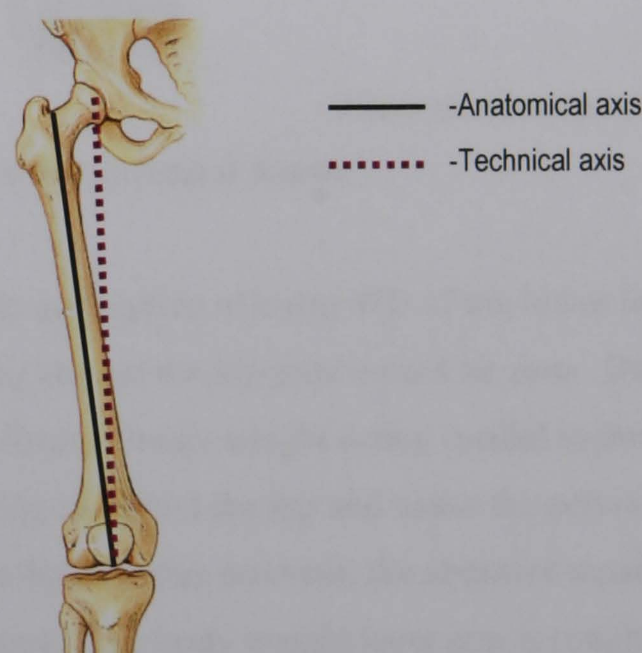
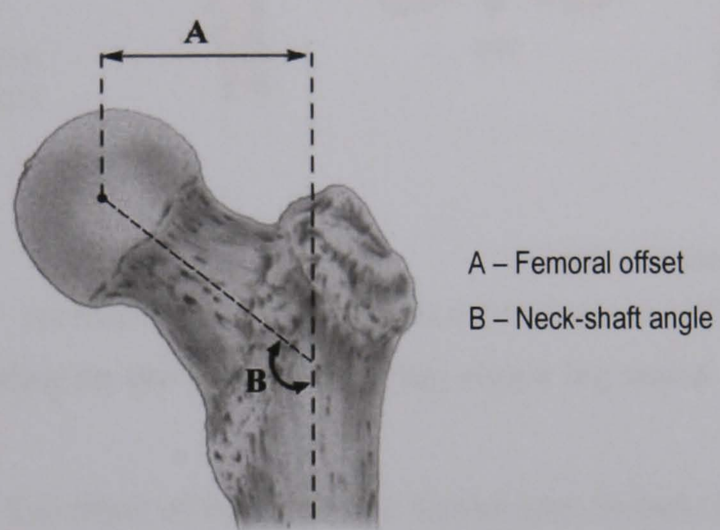


Figure 2.7: Anatomical and technical axis of the femur

The technical axis of the femur is consistent with a line connecting the hip joint centre and the knee joint centre and represents the axis through which forces act on the femur in the kinetic chain. The discrepancy between these two axes implies that during WB, the femoral neck is subject to a varus torque, and the trabecular bone modelling within the neck of femur is structured to reflect this, thus providing some resistance to this moment.

The femoral offset may be defined as the perpendicular distance from the centre of rotation of the femoral head to the line continuous with the anatomical axis of the femoral shaft (Charles et al 2004, McGrory et al 1995). The biomechanical significance of the femoral offset (figure 2.8) is that it defines the length of the lever arm by which the abductor muscle group operates.



Picture taken from Charles et al 2004

Figure 2.8: Anatomical features of the proximal femur

In balanced bilateral stance, with the assumption of equal WB of the lower limbs, the sum of the rotational moments acting around the hip joint would be zero. During single leg standing (figure 2.9), the force of body weight acting medial to the hip joint produces a large moment tending to adduct the hip and cause the pelvis to drop on the contralateral side. In order to balance this moment, the abductor muscles must produce an opposing abductor moment. The body weight lever arm is roughly 2.5 times greater than the magnitude of the abductor lever arm and hence, the abductors must produce a force 2.5 times greater than body weight to maintain equilibrium during a single leg stand (Fagerson 1998). If there is abductor muscle weakness then the torque produced by the abductor muscle group is reduced. Alternatively, if the femoral offset is reduced, resulting in a shorter abductor muscle lever arm, the force production of the abductor muscle group is reduced, leading to a reduction in the magnitude of the abductor moment.

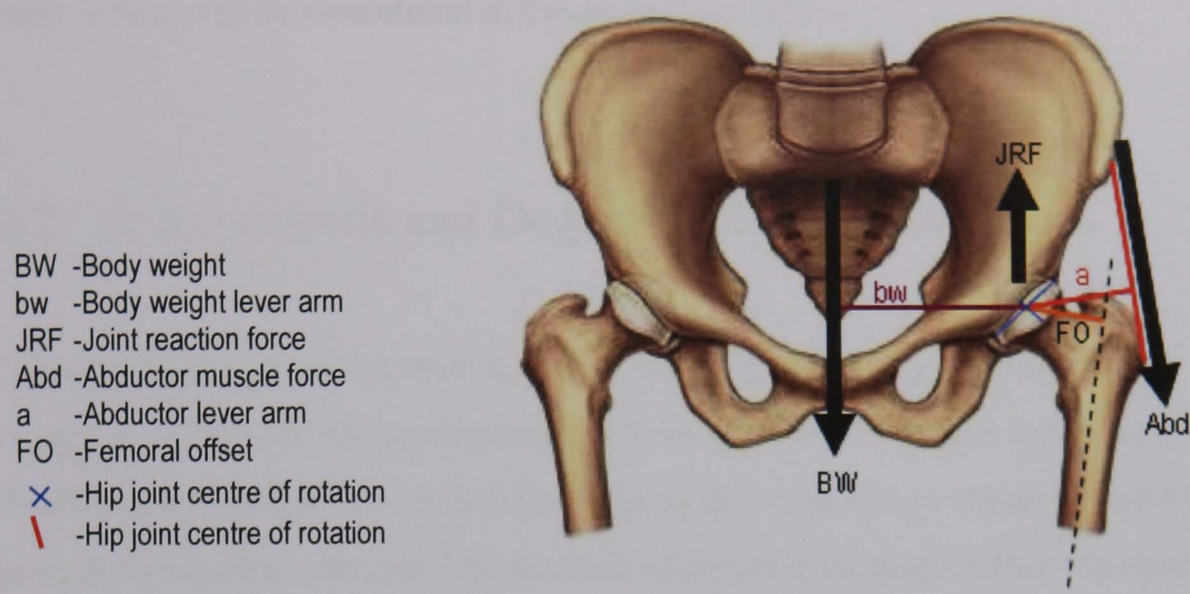


Illustration adapted from www.exac.com

Schematic adapted from Charles et al 2004, Fagerson 1998 and Sariali et al 2008

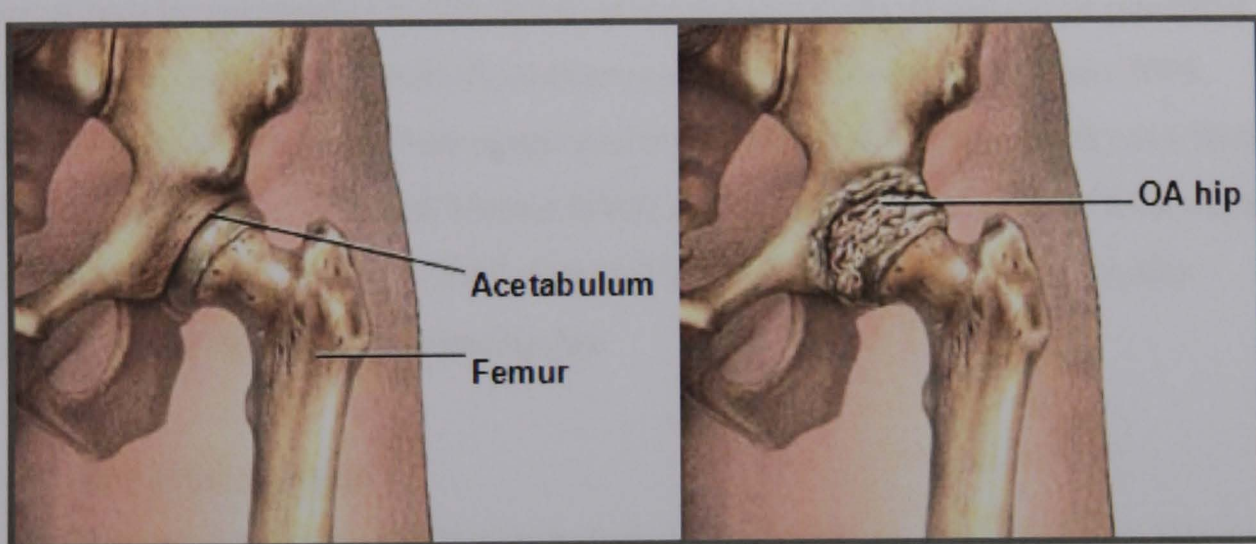
Figure 2.9: Forces acting on the hip joint during single leg stand

Mechanisms reducing the offset of the anatomic femur may include a naturally short femoral neck, a valgus neck-shaft angle (Charles et al 2004), an anteverted femoral neck (Sariali et al 2008) or a reduced leg length (Sakai et al 2002). Hence, abductor muscle dysfunction due to muscle weakness or a reduced femoral offset results in imbalance of the moments acting around the hip joint during single leg standing. To compensate, either a greater abductor muscle force is required to balance the force of opposing body weight, resulting in high joint reaction forces, or the moment due to body weight must be reduced. In the latter case, the body weight lever arm could be reduced by lateral movement of the centre of mass (COM) over the hip joint. The clinical characteristics of the resultant kinematics are representative of a Trendelenburg gait, including contralateral pelvic drop, limp or lurch during stance.

Orthopaedic literature commonly explains and predicts certain aspects of biomechanics of the hip joint with theories drawn directly from the relative position and alignment of anatomical structures (Charles et al 2004, Sariali et al 2008), with fewer studies supporting these theories with biomechanical data analysis. The anatomical features discussed in this section, and their relevance to biomechanics of the hip, have important implications for the pathological and the reconstructed hip joint, which will be considered in the sections to follow.

2.2 Osteoarthritis and Degenerative Joint Disease

Osteoarthritis (OA) is the most common pathology affecting synovial joints (Fagerson 1998) and the most prevalent form of arthritis (Hunter and Felson 2006). Traditionally considered a non-inflammatory disorder, OA is characterised by progressive destruction and deterioration of articular surfaces within the synovial joint (figure 2.10).



Pictures adapted from www.adam.com

Figure 2.10: The normal hip and the osteoarthritic hip

The prevalence of OA of the hip is greatest within the white European population, with between 7 and 25% of individuals over the age of 55 years having a clinical diagnosis (Lievense et al 2002). The economic cost of managing OA of the hip alone is significant. Although the pathology has long been recognised, the mechanisms causing OA remain poorly understood (Pelletier et al 2001), as are the existence of potentially modifiable risk factors, and hence the treatment options remain limited. At least a staggering 55,000 individuals per year (based on figures for 2006, National Joint Registry 2007) eventually require primary THR for degenerative hip disease, and the most common indication for THR is OA (Malchau et al 2002, Sariali et al 2008).

Primarily, OA affects the articular cartilage but progression to severe forms of the disease may lead to the involvement of all joint structures, periarticular muscle and ligament (Pelletier et al 2001). The specific mechanisms by which OA develops are debated but the aetiology is largely mechanical and biochemical in nature (Simon 1999) with some genetic element (Hunter and Felson 2006). Certain individuals may be predisposed to developing OA due to genetic abnormalities or altered structural integrity of their cartilage or subchondral bone due to metabolic or osteoporotic disease for example. Joint misalignment, poor bony congruence and muscle weakness are other structural features which may predispose an individual to mechanical initiation of OA (Hunter and Felson 2006). Those with healthy 'normal' tissues may be susceptible to OA due to activities involving excessive or repetitive joint loading, including obesity (Lievense et al 2002, Shakoor and Moisio 2004, Simon 1999), work related behaviours and traumatic joint injury. Obesity is a form of static loading (Shakoor and Moisio 2004) and is thought to accelerate wear of load bearing joints (Lievense et al 2002, Simon 1999). The incidence of OA is also positively associated with increasing age.

2.2.1 OA disease process

Degeneration of articular cartilage is initiated in response to mechanical loading and focal pressure of joint surfaces (Pelletier et al 2001). Loss of proteoglycans is followed by damage to the collagen framework of the cartilage, and subsequently a repair response is attempted. Inevitable failure of the repair action leads to further degradation and erosion of the cartilage leading to a reduction in its load bearing capacity (Simon 1999). Fragments of the cartilage matrix are released into the synovial fluid (Martel-Pelletier 2004) and a cellular response is triggered, featuring cytokine and chondrocyte activity (Simon 1999). The cellular response marks the initiation of the biochemical changes of the osteoarthritic process and is represented clinically by synovitis. Typically, this cellular activity is unsuccessful in modifying or reversing the structural changes, and instead it precipitates an increase in cartilage destruction by inflammatory mediators which consequently results in greater load on the subchondral bone. Changes in tissue stiffness due to subchondral sclerosis (Simon 1999) allow cracks to develop in the underlying subchondral bone (Fagerson 1998). The fissures created act as a cavity for synovial fluid to fill, precipitating the formation of local subchondral cysts. Compression of the weak bone under the continuing loading pattern results in collapse of the bone in peak weight bearing areas (Berme et al 1985, Fagerson 1998). There is progressive reduction in the joint space and loss of lubrication from synovial fluid in the joint cavity allowing greater friction and further abrasion of the cartilage and subchondral bone. There is a loss of subchondral tissue, necrosis and possible loose body formation in the joint cavity as a result of fragmentation of osteochondral bone. Further repair action within the joint initiates remodelling of bone and revascularisation of the residual cartilage. Proliferation of new cartilage occurs on the joint surface and at the peripheries of the joint (Berme et al 1985), although fibrocartilage is synthesised instead of articular (hyaline) cartilage (Simon 1999). This new cartilage is structurally unable to withstand the compressive joint forces and continued loading initiates osteoblast activity resulting in the growth of osteophytes around the joint (Fagerson 1998) and hypertrophic bone changes leading to subchondral bone plate thickening (Martel-Pelletier 2004).

Local inflammation is increasingly accepted as part of the OA disease process (Martel-Pelletier 2004, Pelletier at al 2001), while the absence of systemic inflammation is fundamental to the diagnosis of OA (Hunter and Felson 2006). Pelletier et al (2001) has formed a convincing argument for the role of synovitis in promoting local inflammation within the joint, leading to synovial hypertrophy and potentially causing the pathogenesis of structural changes which occur in advanced OA. Coinciding with the later stages of the cartilage destruction, thickening of the joint capsule is known to occur at the joint margins in response to 'low-grade' local inflammation (Simon 1999). Inflammation may affect extra-articular structures including tendon, bursa and ligament enthesis.

Swelling and deformity within the joint may place tension on intra-articular tissues and structures surrounding the joint. This causes laxity which reduces the stability of the joint. In addition, arthrogenic pain inhibition of muscles may reduce dynamic stability. The lack of joint stability may lead to abnormal sliding motion within the joint and edge loading leading to further progression of wear and precipitating the growth of osteophytes.

Osteoarthritis has associated neuromuscular and motor deficits. In response to the joint degeneration, proprioception becomes impaired (Sharma 1999) and muscle torques generated at the joint are reduced (Isobe et al 1998, Tanaka 1998). Muscle atrophy may develop secondary to disuse and arthrogenic pain inhibition, leading to uncoordinated movement and possible delayed muscle activation (Hassan et al 2001).

Given the multifactorial nature of osteoarthritis of the hip, the scope for variation in the extent of the disease and the number of structures involved is vast, and this must be considered when encountering patients following orthopaedic joint reconstruction. In particular, the impairment of extracapsular structures may continue following joint replacement, potentially affecting neuromuscular control or joint biomechanics.

2.2.2 Diagnosis, clinical and surgical implications of OA

Diagnosis of hip OA is done by the classification of clinical symptoms by their nature and severity and the support of clinical findings with radiographic assessment (Hunter and Felson 2006). The clinical symptoms of OA include pain, stiffness and related functional restrictions. Pain and discomfort is often local to the groin, with occasional referral of pain to the knee (Berme et al 1985). Pain generally increases with activity, due to WB and movement demands. Poor gliding and greater friction of the joint surfaces limits ROM and causes exacerbation of pain on movement. Synovitis and joint effusions may also be observed clinically, due to low grade inflammation (Simon 1999). Patients with OA experience early morning stiffness and stiffness following prolonged sitting or driving, which eases with between 20-30 minutes of activity (Simon 1999). Secondary muscle contractures and soft tissue shortening may be evident in severe presentations, and develop in response to reduced activity and limited ROM due to pain and stiffness.

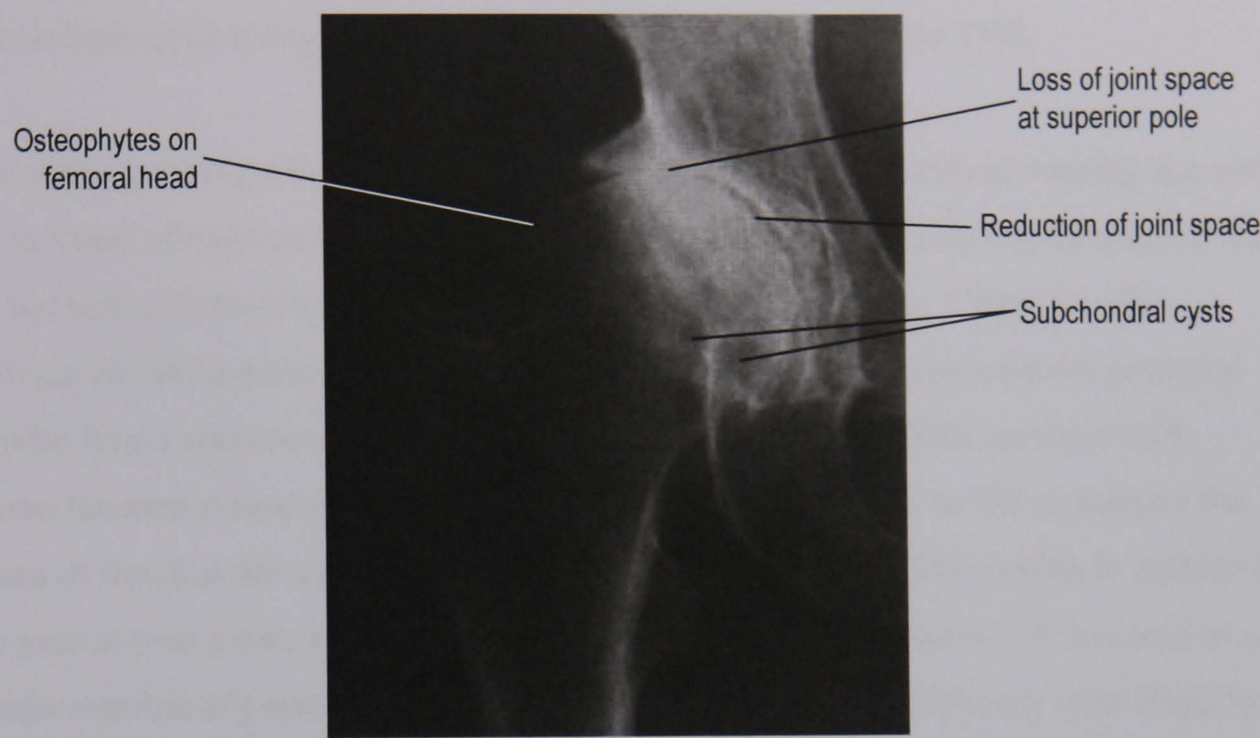


Figure 2.11: Radiograph of OA hip (Right)

Radiographic findings include narrowing of the joint space, often at the superior pole, which reflects the anatomic region of maximum WB (section 2.1.2.2). Sclerosis, osteophyte and cyst formation may be observed and in severe presentations there may be bony deformity (figure 2.11). Significant osteoarthritic changes must be present before the characteristic features of hip OA may be seen on 2-dimensional radiograph (Hunter and Felson 2006). Patients will however be asymptomatic at the initial stages of cartilage degradation as the cartilage tissue is aneural and does not stimulate a nocioceptive response. However, involvement of the synovium, subchondral bone or capsule will cause pain, and this is the reason why synovitis marks the initiation of the clinical stage of the disease (Martel-Pelletier 2004) following which radiographic changes may be observed.

Specific functional restrictions related to OA of the hip joint include limitation of walking distance (Berme et al 1985) and walking speed (Murray et al 1971), difficulty going up and down stairs (Hunter and Felson 2006), getting in and out of a chair, car or bath, tying shoe laces and stooping (Johnston and Smidt 1970, cited in D'Lima et al 2000). In general, following failure of medical and other non-surgical management, individuals with advanced OA of the hip who suffer significant pain and, or have severe functional limitations may be considered for THR.

Due to the growing elderly population, the prevalence of OA with increasing age and the increase of degenerative joint disease in the population (Burns and Bourne 2006), the number of individuals requiring THR surgery is increasing. There are also growing recommendations for earlier surgical intervention to optimise the potential activity levels achieved following THR (Hunter and Felson 2006), as those with severe functional restrictions pre-operatively have been shown to fail to achieve the return of function seen by those with less severe disabilities. This notion to replace a hip joint sooner rather than later and the increase in activity related OA has lead to a greater number of younger people requiring hip arthroplasty (National Joint Register 2007). Hence the age range of the population receiving THR is tending to diverge. These factors have prompted the orthopaedic community to adapt and modernise the design of the current total hip prostheses available, as well as the bearing materials and fixation methods used (Daniel et al 2004). Moreover, the growing awareness of issues such as cost-effectiveness of THR and the significant burden of revision surgery (Burns and Bourne 2006) have also driven a government and health community lead initiative to increase the use of cost-effective prostheses for primary arthroplasty. These factors will be discussed further and consideration will be made to the possible effect such factors may have on the functional outcomes of this younger population.

2.3 Total Hip Replacement

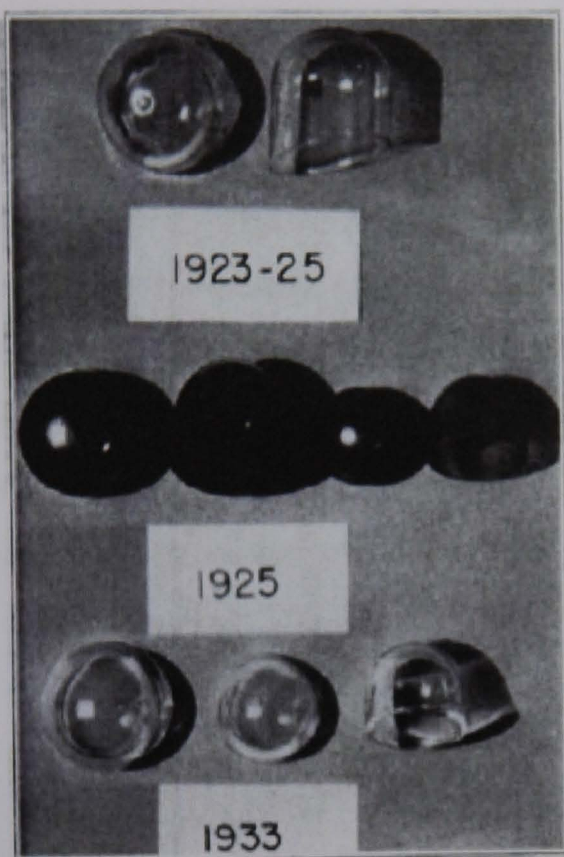
Total hip replacement (THR), or total hip arthroplasty (THA), is one of the most common of the major orthopaedic procedures (Heybeli and Mumcu 1999) and is viewed as one of the most successful (Gomez and Morcuende 2005, Sariali et al

2008). The aim of THR is to reduce pain, restore function, obtain stability of the joint (Affatato et al 2004) and improve health-related quality of life (Burns and Bourne 2006). THR involves the replacement of adjacent articular surfaces with new bearing surfaces. There is variation in the biomaterials and structure of the bearing components used for replacement arthroplasty, with contemporary prostheses having progressed with the advancement of technology and better understanding of tribology and biological factors limiting the longevity of the implant.

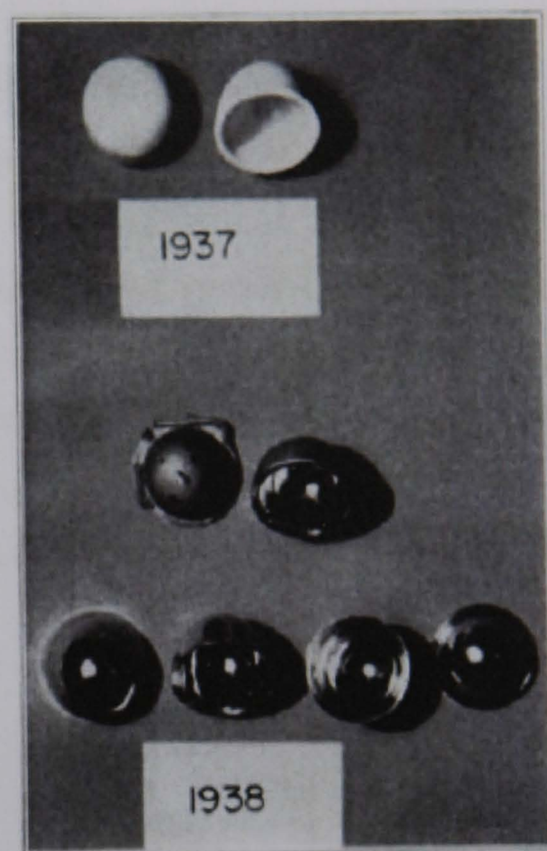
2.3.1 History and development of THR

The concept of THA was conceived in the 1890's and developed following the recognised shortcomings of the osteotomy arthroplasty and interpositional arthroplasty techniques used to treat ankylosed hips (Gomez and Morcuende 2005, Heybeli and Mumcu 1999). John Rhea Barton was the pioneer of trochanteric osteotomy in 1826, with the surgical objective of increasing motion of the hip, but this was done at the expense of stability and the gain in motion was of limited and temporary success. Interpositional hip arthroplasty involved a more modest removal of bone and the introduction of a remote or foreign material placed between the joint surfaces, such as muscle (e.g. tensor fascia lata), fibrous tissue, silver plates, gold foil, rubber sheets, magnesium or zinc, and the later, more popular option of pig bladder membrane. The purpose of the interposition material was to maintain motion at the osteotomy site and prevent bony re-growth and encourage cartilage regeneration (Heybeli and Mumcu 1999).

Marius Smith-Petersen created the mould arthroplasty prosthesis in 1923 consisting of a moulded interpositional material shaped to cover the femoral head, comprised of materials with less perishable and more inert properties (Smith-Petersen 1939, 1948). The materials ranged from glass in the initial design to vitallium in 1938 (figure 2.12). Vitallium is a cobalt-chromium alloy and provided greater durability than the other materials. Smith-Petersen implanted 500 vitallium interpositional mould arthroplasties within 10 years with good clinical results (Smith-Petersen 1948). Not only was the procedure the first interpositional arthroplasty to provide predictable results (Gomez and Morcuende 2005) it also demonstrated the first successful use of metal alloys for hip reconstruction. The vitallium prosthesis also bears similarity to the femoral component of the contemporary resurfacing prosthesis (section 2.3.5). Moreover, commentary on resurfacing arthroplasty often directly associates the evolution of the hip resurfacing concept from the Smith-Petersen's mould arthroplasty (Roberts et al 2005).



Evolution of the mold:
1923: **Glass.**
1925: **Viscaloid.**
1933: **Glass (pyrex).**



Evolution of the mold:
1937: **Bakelite.**
1938: **Unsuccessful and successful vitallium molds.**

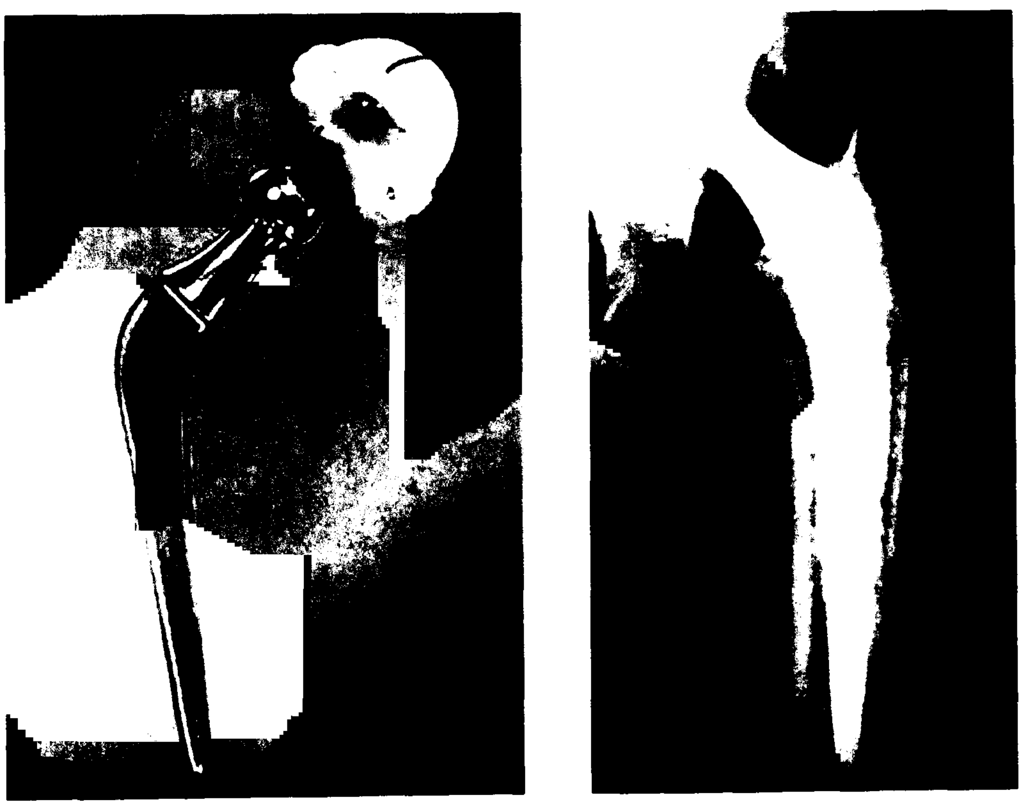
Pictures taken from Smith-Petersen 1939

Figure 2.12: Smith-Petersen Mould Arthroplasty

In contrast to interpositional arthroplasty, total joint arthroplasty involves the replacement of both bearing surfaces of a diarthrodial hip joint (Heybeli and Mumcu 1999), with fixation of the bearing materials to the joint surface. The first recorded THA was performed in 1891 by the German Professor Themistocles Glück using an ivory ball and socket joint (Gomez and Morcuende 2005, Heybeli and Mumcu 1999). The evolution in hip arthroplasty that followed was characterised by a range of bearing materials, designs and fixation methods which lead to the development of the conventional hip arthroplasty recognised today.

2.3.2 Conventional arthroplasty

The traditional hip arthroplasty, as it is known today, was introduced in the 1950's (Fagerson 1998) but significant developments in the design were made in the 1960's by Professor Sir John Charnley, regarded as the greatest contributor to the development of the modern hip arthroplasty. Charnley's main contributions were the introduction and popularisation of a THR with an ultra high molecular weight polyethylene (UHMWPE) socket articulating with a small diameter metal femoral head, initially stainless steel (Schmalzried and Callaghan 1999), and the use of acrylic cement for improved fixation of the components with reduced risk of post-operative sepsis (Heybeli and Mumcu 1999). Charnley believed that friction of the bearings was an important factor in the durability of the prosthesis, thus directing his selection of a small diameter femoral head to articulate against the polyethylene socket (section 2.3.3). His "low friction arthroplasty" provided excellent clinical results (Charnley 1972, Kavanagh et al 1994) and formed the basis of the conventional arthroplasty which is still in use today (figure 2.13).



Pictures taken from www.maitrise-orthop.com (a) and Hernández-Vaquero et al 2008 (b)

Figure 2.13: Charnley "low friction arthroplasty" prosthesis (a) at 22 years (b)

Consistent with the "gold-standard" low friction arthroplasty design (Wroblewski 2002, page 825), the latest generation of conventional THR prosthesis consists of a small diameter metal (typically cobalt chromium alloy) femoral head articulating on an UHMWPE acetabular socket (metal-on-polyethylene, M-PE). Multiple manufacturers market differing models of this standard THR and offer variable options for fixation of the components with regard to cemented or cementless fixation. The conventional M-PE THR has been extremely successful. Clinical observations report 10-year implant survival of 92.5% (±0.15) between 1992 and 2003 for primary total hip arthroplasty (http://www.jru.orthop.gu.se/). However, the success of the M-PE bearing combination is somewhat limited to the late middle-aged and elderly populations (Roberts et al 2005) and has been associated with high failure rates with long-term implantation or implantation in younger, more active individuals, particularly young males. Survival rates for males under 55 years have been reported as low as 70% at 10 years (Tipper et al 2005) and 33% at 16 years (Grigoris et al 2006) with conventional M-PE THR.

The main mechanism of failure of the conventional M-PE THR is osteolysis leading to aseptic loosening of the implant (Dumbleton et al 2002, National Joint Registry 2007). Mechanical wear of the bearing surfaces leading to the production of PE wear debris in and around the joint, triggers a foreign body cellular response which induces osteolysis (Agarwal 2004, Schmalzried and Callaghan 1999). Bone resorption occurs, leading to component loosening, failure of the prosthesis and eventual requirements for revision surgery (Affatato et al 2004, Dumbleton et al 2002). Osteolysis is a long-term failure mechanism but the volume of wear debris produced is thought to effect the initiation and rate of osteolysis, and therefore affect the rate of revision surgery (Dumbleton et al 2002). The National Joint Registry for England and Wales has published a 'crude revision rate' or 'revision burden' (number of revision THR as a percentage of the number of all primary THR, Malchau et al 2002) of 9% in 2004 (Burns and Bourne 2006) and 10% (n=5821) in 2006 (National Joint Registry 2007). With the economic cost of revision surgery exceeding £6000 per individual (Vale et al 2002) and the inferior outcomes associated (Burns and Bourne 2006), there is a drive to reduce the rate of revision surgery. In addition to reducing complication rates, the primary method of reducing the revision burden is by optimising the durability of the prosthesis to increase the longevity of the THR.

The age range of the population requiring THR has tended to diverge and the average age of these individuals is declining (Tipper et al 2005). Given the poorer survival rates of the conventional M-PE THR with younger individuals, there is increasing demand hip joint prosthesis with alternative bearing surfaces and improved prosthetic designs in order to enhance wear performance, avoid osteolysis and promote longevity of the implant, thereby prolonging function of the younger, more active individual.

2.3.3 Wear and total hip replacement

Wear is the "removal of material, with the generation of wear particles, that occurs as a result of the relative motion between two opposing surfaces under load" (Schmalzried and Callaghan 1999, page 115). Theoretically, there is potential for material wear to occur between any two moving components that share a contact surface area (Fagerson 1998).

Tribology is the science of wear, and there are several dependent factors which influence wear of a hip joint prosthesis, thus compromising its durability. The following sections will discuss the factors influencing wear, including the type of bearing material, friction and lubrication, and aspects of the bearing geometry. In addition to these factors, the nature and frequency of the activity of the individual also holds great importance. Studies analysing wear characteristics have observed an initial period of run-in wear, which is associated with relatively large volumes of wear debris production, followed by a longer-term period of steady-state wear (Hannouche et al 2005) which produces significantly lower volumes of wear debris and is more consistent. The kinematics and kinetics of the individual are dependent factors influencing wear as they describe the sliding behaviour of the bearing and the loading regime used (Tipper et al 2005).

2.3.3.1 Biomaterials and wear

The biomaterials used in contemporary THR designs include polymers, ceramics and metals. The mechanical and biological properties of each vary as does their behaviour when interacting with one another as a bearing in a prosthetic articulation.

The wear characteristics of UHMWPE may be enhanced by the use of certain manufacturing methods which improve the mechanical properties of the polymer. Cross-linking and annealing processes strengthen the PE microstructure and improve surface topography to reduce wear debris generation (Heisel et al 2003), although the resultant product remains suboptimal for young active individuals when contrasted with other bearing materials. Incorporated into a THR prosthesis, UHMWPE performs best as an acetabular component, as supported by a former cases of catastrophic wear when used as a femoral bearing (Roberts et al 2005). Metal or ceramic femoral components may be used to articulate with UHMWPE in contemporary designs. These are classified as hard-on-soft bearing combinations.

"All ceramics are not alike" (Hannouche et al 2005, page 63); alumina and zirconia have been used as biomaterials for orthopaedic implants, and have differing material properties. Zirconia (zirconium dioxide, ZrO_2) has relatively greater shock resistance but is otherwise less attractive as a bearing material, with poorer sliding properties and less biochemical stability, associating it with poor clinical results (Hannouche et al 2005). Aluminium oxide or alumina (Al_2O_3) is more desirable as a biomaterial. It is the hardest of all the THR materials (Schmalzried and Callaghan 1999) and almost biochemically inert, manufactured to achieve the highest state of oxidisation possible, making it thermodynamically stability. As a result, alumina is extremely biocompatible and induces minimal cellular responses, so is tolerated well by the body. Alumina ceramic is also hydrophilic which assists in lubrication of the prosthetic joint and features an excellent sliding capacity when articulating against itself or PE (Hannouche et al 2005). The combination of favourable tribologic properties and biocompatibility make alumina-on-alumina THR bearings very suitable for the young, active population requiring hip arthroplasty.

The preferred metal alloy for use as a biomaterial in THR is high carbon content cobalt chromium (CoCr), with some metal alloys also containing molybdenum (Mo). CoCr alloys are hard and offer corrosion resistance provided by the chromium component (Silva et al 2005). Carbon enrichment of the metal compounds during manufacturing makes these alloys even harder. Similar to the use of ceramics, metals have been popularised for use in THR to avoid the use of PE and the associated biological processes in response to PE wear debris. Metal-on-metal (MoM) articulations are however associated with other potential risks.

Metallic ions have been found in the serum and urine of subjects with MoM hip prosthesis at greater concentrations (times five) compared to control subjects (Silva et al 2005). Co and Cr ions are highly toxic, but at these sub-lethal quantities the effects are largely unknown (Tipper et al 2005). There have been low incidence reports of hypersensitivity syndromes which induce an immune response leading to osteolysis or to the recently recognised phenomena of aseptic lymphocytic vasculitis-associated lesions (ALVAL). ALVAL has been directly associated with the current 2nd generation MoM prosthesis, leading to symptoms of early groin pain due to a local reaction to the metal implant (Campbell et al 2008). There are also concerns that Co and Cr wear particles could induce DNA damage and cancer, depending on the amount of wear debris and the length of time they interact with human tissue. Metallic ions have also been shown to accumulate in the placenta, raising fears of teratogenicity, particularly with the lack of research determining the effects metal ions may have on foetal development (Silva et al 2005). For these reasons, which equate to potentially significant but largely unknown risks, some orthopaedic surgeons exercise restraint in the use of MoM arthroplasty bearings, regardless of the favourable tribology, and particularly in females of a child bearing age.

2.3.3.2 The effects of friction and lubrication on wear

Friction is the force resisting relative motion of the opposing joint surfaces. Fluid-film lubrication of the prosthetic joint reduces contact surface area and therefore friction between the bearing surfaces; Full fluid-film lubrication implies complete separation of the bearing surfaces and is favourable as the load is carried by the fluid thus minimising wear of the bearings (Silva et al 2005). However, the current biomaterials used in THR are "unable to produce a permanent lubricating film" (Rieker et al 2001, page 153) and often allow mixed-film lubrication, characterised by partial separation of the bearing surfaces, or boundary lubrication which features substantial surface contact (Tipper et al 2005). The ratio of film-thickness to surface-roughness (λ ratio) dictates the degree of joint lubrication and is dependant on the properties of the bearing material, the lubricating fluid, the bearing geometry and the surface topography (Silva et al 2005).

The high 'wettability' of alumina ceramic relative to polymers and metals, encourages excellent fluid-film lubrication of CoC bearings and reduces friction between the adjacent surfaces. The sliding properties of alumina are further improved by the low surface roughness achieved from modern machining and the high quality and purity of the ceramic used (Hannouche et al 2005), allowing for greater congruence of the components and minimal interruption of the fluid film. Metal alloys are known to have a poorer fluid affinity and operate a mixed-film lubrication mechanism (Silva et al 2005). In this case, achieving a low surface roughness of metal bearings is essential for improving sliding properties and reducing friction between MoM bearings. Sliding properties may be further enhanced depending on bearing geometry (section 2.3.3.3).

Several problems were associated with the first generation of metal-on-metal bearing, which were attributed to poor manufacturing and design of the prosthesis or faults with the surgical implant technique. The former casting method used for the production of metal bearings was associated with accelerated wear due to residual carbide asperities left on their surface which generated greater friction (Silva et al 2005) and interrupted the fluid-film. Most of the latest generation of MoM bearings undergo heated treatments to disperse carbides (Grigoris et al 2006) or are manufactured using a wrought-forging process (Metasul™), resulting in a harder bearing with a highly polished surface to enhance contact surface area and reduce surface roughness, providing superior wear performance (Roberts et al 2005, Silva et al 2005). Inherently, alumina is extremely hard, with a Young's modulus threehundred times greater than cancellous bone. It is therefore able to resist high compressive forces with negligible deformation and is highly scratch resistant, so maintains its low surface roughness. However, the brittleness of alumina ceramic implies a limited bending strength and low fracture toughness. Cracks which present in the alumina grow in response to stress and loading, and resultant implant failure due to ceramic fracture, which has been associated with early designs and with poor surgical implantation. Imperfections of metallic surfaces, on the other hand, have been shown to partially resolve with wear, highlighting the 'self-polishing' capacity of MoM bearings due to the ductile nature of the alloys (Rieker et al 2001).

The incidence of osteolysis related to second generation MoM implants is rare (Silva et al 2005). The few cases of osteolysis observed from CoC bearings, such as the Mittlemeier total hip system, have been related to the poor implant design and the large particle size produced from early alumina (Hannouche et al 2005). The smaller grain size associated with current surgical grade alumina can undergo macrophage phagocytosis and has not been associated with cellular response induced osteolysis.

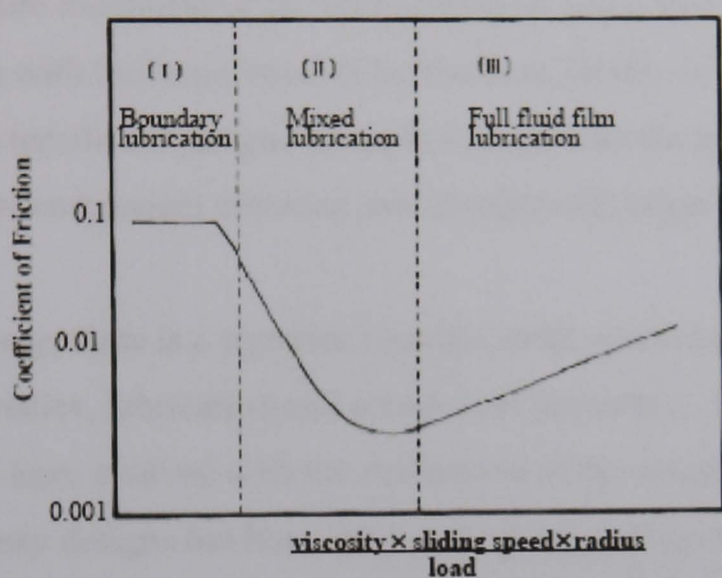
Alumina-on-alumina, has a low co-efficient of friction (0.09) when compared to metal-on-polyethylene (co-efficient of friction, 0.21), supporting the articulation of this bearing combination (Hannouche et al 2005). However, MoM bearings have a greater co-efficient of friction. This is about 2-3 times greater than M-PE articulations (Schmalzried and Callaghan 1999, Silva et al 2005). Nevertheless, retrieval studies have shown that volumes of wear debris resulting from conventional M-PE implants are significantly greater than that of MoM components (Sieber et al 1999, cited in Silva et al 2005) or CoC bearings (Bohler et al 2000, cited in Hannouche et al 2005). One of the few controlled studies directly comparing in vitro wear of MoM and CoC bearings showed that the running-in and steady state linear and volumetric wear rates were statistically similar for Metasul™ and Cerasul™ (alumina) products (Rieker et al 2001) although the ceramic particles are more inert. So, despite the higher co-efficient of friction and poorer lubricating mechanisms, MoM shows wear results comparable with CoC. The reasons for this may be explained by the component geometry.

2.3.3.3 Geometry of the bearings

The geometry of the bearings is a crucial factor in determining wear of THR implants (Roberts et al 2005). Aspects of the bearing geometry which have implications for material wear are mostly related to the diameter of the bearings.

Clearance is defined as the size of the gap between the two adjacent bearing surfaces at the equator (Silva et al 2005), or in other words, the difference in diameter of the acetabular cup and femoral head bearing surfaces. Manufacturing joints with appropriate clearance is essential for limiting material wear of the articular surfaces. A greater clearance results in a smaller area of contact between the bearing surfaces, which in turn increases the contact stresses at the pole and results in greater bedding-in wear (Tipper et al 2005). Reducing the clearance encourages fluid-film lubrication but if the diameter difference becomes too small, this results in equatorial contact (Silva et al 2005) which produces higher friction forces and torque on the prosthesis and the fixation, increasing wear and the risk of implant loosening. Optimal clearance creates an environment which encourages fluid film lubrication of the prosthetic joint. The magnitude of clearance classified as optimal is 100-150μm for MoM bearings (Roberts et al 2005) and 20-50 μm for CoC bearings (Hannouche et al 2005).

There is a greater frictional torque generated with larger diameter bearings (Roberts et al 2005, Schmalzried and Callaghan 1999), and this rule is fundamental to the concept of the low friction arthroplasty and contemporary THR. However, there is a complex "interplay" (Silva et al 2005) between all the factors influencing wear, and large diameter bearings can be effectively used in combination with a low wear material (Schmalzried and Callaghan 1999) such as modern metallic alloys or alumina. In addition, the combination of hard-on-soft bearings in the low friction arthroplasty operates with differing mechanisms of wear compared to the hard-on-hard bearing couples such as CoC and MoM. The hard-on-hard bearings demonstrate abrasive type wear, indicating the use of a "mixed lubrication regime" (Rieker et al 2001, page 153) and potentially intermittent full fluid film lubrication modes (Tipper et al 2005). The hard-on-soft articulation of M-PE has been found to use boundary lubrication (Smith et al 2001, cited in Tipper et al 2005) implying continuous surface contact and low lubrication, which may explain the high PE wear volumes relative to MoM and CoC bearings. Moreover, the conventional M-PE arthroplasty demonstrates wear rates which are proportional to the sliding distance, thus supporting the use of small diameter bearings to reduce the sliding distance and minimise wear (Tipper et al 2005). Optimal M-PE head diameters of between 20mm and 24mm have been suggested, within which the 22.225mm Charnley femoral components fall.



(I) – (III) correspond to λ ratio, where:

λ ratio ≤1 indicates boundary lubrication
λ ratio >1 and <3 indicates mixed lubrication
λ ratio ≥3 indicates full fluid-film lubrication

(Roberts et al 2005)

Figure taken from Tipper et al 2005

Figure 2.14: Stribeck curve describing modes of lubrication

As well as bearing friction, lubrication is also dependent on the viscosity of the fluid, the sliding speed, radius of the femoral head and the diametrical clearance (Tipper et al 2005). The interplay of these factors in influencing the mode of lubrication is effectively summarised in the Stribeck curve (figure 2.14). In vitro wear studies have shown that film lubrication improves with larger diameter bearings due to the greater sliding distance and sliding velocities resulting from the larger radius of the femoral head (Tipper et al 2005). Together with the appropriate clearance, the greater sliding velocity associated with large diameter bearings promote elastohydrodynamic action of the lubricant, which increases the fluid film thickness (greater λ ratio). Kinematic behaviour of the individual will influence the hydrodynamic lubrication of the bearing due to microseparation and squeeze film lubrication associated with in vivo motion of the joint (Rieker et al 2001). Contrary to the expectation of full fluid-film lubrication with alumina bearings, Scholes et al (1998) highlighted that the lubricating fluid in vivo contained protein as well as water which induced mixed lubrication (cited in Rieker et al 2001), confirming the categorisation of mixed lubrication with these hard-on-hard bearings.

Hannouche et al (2005) highlighted that "contrary to the M-PE couple, the alumina-on-alumina combination does not favour small head size" (page 65). However, the use of large diameter CoC THR is limited, due to the reduced compression strength and fracture toughness of the thin acetabular liners that would be required to articulate with the large heads (Cuckler et al 2004). In contrast, metal implants have adequate tensile and fatigue strength to cope with the requirements of a thin acetabular component allowing use of relatively larger diameter components.

In summary, there is a complex multifactorial relationship between material wear characteristics, lubrication and component geometry. The understanding of these concepts have evolved with the realisation of the mechanisms of failure of previous arthroplasty designs but have shaped the future of hip arthroplasty to more adequately meet the functional needs of the younger more active population. Given the equivalent wear performance of the existing MoM and CoC bearings, it may be more appropriate to base the choice of hip arthroplasty on other design features and aspects of function promoted by the implant design.

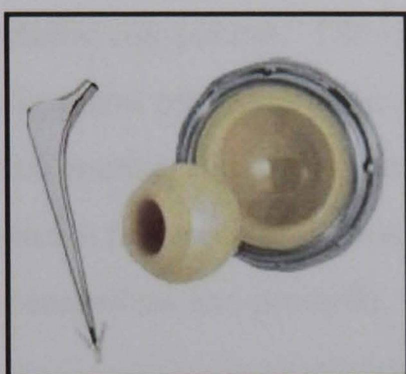
2.3.4 CoC and MoM for the young and active patient

For the young and active patient requiring THR, there are several prosthetic options available, combining the use of a low wear bearing combination in a standard stemmed THR or a hip resurfacing arthroplasty design.

2.3.4.1 Ceramic-on-ceramic THR

Boutin introduced alumina-on-alumina bearing THR during 1970 in France. Amongst observations of very low wear behaviour (Hamadouche et al 2002, Jazrawi et al 1999), early results revealed several cases of fracture of the bearing surface

(Hannouche et al 2005, Rieker et al 2001) related to suboptimal quality and reproducibility of the manufacturing. By 1977, production of alumina with greater purity, porosity and smaller grain size was achieved, improving the mechanical properties of the bearing. By machining components with an initial crack length smaller than 100μm, the fatigue limit was avoided, achieving greater fracture toughness. Nevertheless, difficulties persisted in the methods of obtaining stable component fixation, including attaching the femoral head to the stem and gaining a stable acetabular bone-ceramic interface (Hannouche et al 2005), and so CoC THR was largely abandoned in the 1980's (Rieker et al 2001). Introduction of the Morse taper acetabular design provided a significant advancement in achieving a stable fixation of the acetabular shell and interest was revived in the 1990's.



Trident™ ceramic acetabular system, ceramic head from Stryker-Howmedics-Osteonics and Exeter stem

Figure 2.15: Ceramic-on-Ceramic Trident™ THR prosthesis

The alumina-on-alumina Trident™ THR system produced by Stryker (figure 2.15) is one of the leading CoC THR in the UK, meeting the requirements for quality and reproducibility of materials and design. The Trident™ ceramic acetabular system is a clinically established press-fit acetabular fixation providing optimal stability. The ceramic insert is taper-locked into a titanium shell, which is covered with hydroxyapatite to promote bony in-growth onto its roughened surface, adding to the stability to the fixation. The ceramic femoral head is mounted onto a metal stem of the surgeon's choice depending on desired fixation method and modularity options. In line with best practice, the surgeons cited in this thesis implemented cement fixation (Malchau et al 2002) using an Exeter femoral stem component. The alumina femoral head features a larger diameter (28-36mm) than conventional Charnley prosthesis, which, in addition to the favourable effects on wear (section 2.3.3.3), allows increased ROM and joint stability, which reduce the risk of impingement and dislocation (D'Lima et al 2000, Kluess et al 2007, Yoshimine 2006).

2.3.4.2 Metal-on-metal THR

Large diameter MoM articulations are available for use in standard stemmed THR systems. However, resurfacing arthroplasty offers potentially significant advantages over the stemmed designs.

2.3.5 Resurfacing arthroplasty

Resurfacing arthroplasty of the hip describes a total replacement procedure with preservation of the anatomic neck of femur (figure 2.16). Femoral osteotomy partially removes bone from, and reshapes the head of femur allowing acceptance of a cemented metallic component. The metaphyseal stem of the femoral component is designed for alignment purposes (Amstutz and Le Duff 2006) and has minimal influence in load bearing and stress transfer (Roberts et al 2005). Contemporary hip resurfacing systems feature hybrid fixation methods, implying that the acetabular component is cementless and press-fit, while the femoral component is cemented.

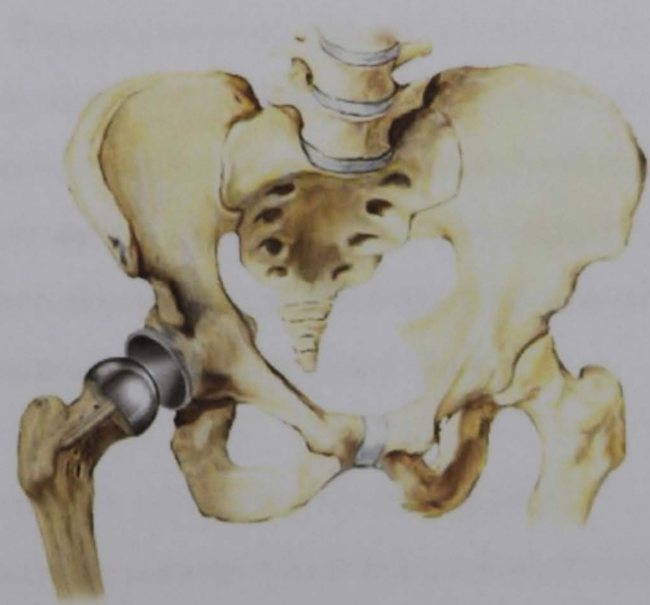


Illustration taken from www.fredphillips.co.nz

Figure 2.16: Resurfacing arthroplasty

Current orthopaedic practice advocates the use of resurfacing arthroplasty as an alternative to standard stemmed THR for the orthopaedic management of arthritic hip disease (Amstutz et al 1998, NICE 2002, Roberts et al 2005). The in vivo success of current resurfacing designs is dependent on an excellent surgical technique, using sophisticated surgical instrumentation, and careful patient selection (Amstutz and Le Duff 2006). Recommendations indicate that resurfacing is suitable for the young active individual who is likely to require more than one episode of hip joint reconstruction in their lifetime and older individuals who partake in activities which may otherwise subject a conventional M-PE THR to excessive, accelerated wear and early failure (Amstutz et al 1998).

The conservation of femoral bone stock has highlighted the potential for greater ease of revision surgery, allowing conversion to a stemmed THR which may provide "increased durability after revision" (Amstutz et al 1998, page 172). In this respect, resurfacing arthroplasty is sometimes regarded as 'buying time' until standard THR is required (Amstutz et al 1998, Tipper et al 2005), although fundamentally, resurfacing and standard THR share the same indications and both are considered as primary arthroplasty procedures.

Doubts over the ease of revising a resurfacing procedure have been highlighted (Grigoris et al 2006). Potential revision difficulties relate to the acetabular component. Excessive removal of acetabular bone stock to accommodate the large diameter bearings reserves minimal reaming depth and residual bone stock to accept a revision cup (Amstutz and Le Duff 2006), and may require significant bone grafting. Surgeons have also expressed anecdotal fears over potential difficulties in removing the highly secure press-fit acetabular shells.

2.3.5.1 History and evolution of resurfacing arthroplasty

Resurfacing arthroplasty is a concept which has evolved directly from the Smith-Petersen interpositional mould arthroplasty (Amstutz and Le Duff 2006, Roberts et al 2005) (Section 2.3.1). Charnley developed the first total hip resurfacing prosthesis in the early 1950's (figure 2.18). A Teflon-on-Teflon bearing was used which resulted in failure due to poor wear performance despite his recognition of the low friction properties of the Teflon couple. Maurice Muller followed in 1967 with a cementless

MoM resurfacing design, coinciding with the debut of his stemmed MoM THR. Both produced excellent short- and long-term results (Roberts et al 2005) but the shift toward use of M-PE articulations, in the wake of the successful low friction arthroplasty, discouraged use of these MoM components. Following suit, cemented M-PE resurfacing designs were introduced by various individuals in the 1970's (figure 2.17), including Paltrinieri and Trentani, Furuya, Capello and Tanaka, but more famously, Freeman, Wagner and Amstutz (Amstutz et al 2008). The poor clinical results were multifactorial but mostly related to osteolysis. Despite experimentation with alternative bearing combinations such as ceramics, resurfacing continued to produce poor results. Consequently the resurfacing procedure was abandoned in most countries and institutions during the 1980's (Grigoris et al 2006, Roberts et al 2005) only to be succeeded by a second generation resurfacing design with MoM bearings.

In hindsight, the modes of failure of 1st generation resurfacing arthroplasty were related to the poor wear properties and manufacturing of the materials, the sub-optimal prosthetic designs and fixation methods, and the surgical techniques, which were crude and lacked the appropriate instrumentation for accurate and repeatable implantation (Amstutz and Le Duff 2006). Secondary to osteolysis, femoral neck fracture was a common cause of failure (Amstutz et al 1998, Roberts et al 2005). Notching of the femoral neck occurred secondary to under-sizing of the femoral head, in order to spare acetabular bone stock and reduce frictional torque, and the use of extreme valgus positioning of the femoral component, attempted to minimise tension and sheer stress at the head-neck junction (Grigoris et al 2006). Trochanteric osteotomy or varus femoral component positioning may have contributed to fracture in the absence of neck notching (Grigoris et al 2006) by weakening the head-neck junction and subjecting the neck to large torques during WB.

Continuation of THARIES implantation during the 1980's by Amstutz at UCLA, went some way to demonstrating that the resurfacing procedure itself was sound. The relative success of the THARIES prosthesis (figure 2.17) was attributed to the correct selection of patients, implementation of specifically designed surgical tools to standardise the procedure and the use of improved surgical techniques (section 2.4.3). Cylindrical and champhered reaming of the femoral head with the use of a pin guide to centre the femoral component were significant advances in the femoral head preparation which reduced the incidence of failure due to femoral neck fracture (Amstutz et al 1998). Amstutz and colleagues (1998) published reasonable short-term results but poor long-term survivorship (43% at 15 years) of their prosthesis. The THARIES experience confirmed that failure was mostly due to osteolysis, and with the current understanding of wear mechanisms, it is clear that the 1st generation M-PE resurfacing components structured as large diameter bearings provided a high wear producing THR model (Grigoris et al 2006, Roberts et al 2005). Meanwhile, the Müller MoM resurfacing prosthesis had demonstrated good clinical results at up to 25 years (Müller 1992, 1995 cited in Grigoris et al 2006, Roberts et al 2005).

2.3.5.2 Renaissance-2nd generation resurfacing arthroplasty

Precipitated by the development of the Metasul™ bearing (extremely low wear, high-carbon content, wrought-forged Co-Cr alloy) in 1988 by Weber and Sulzer, Wagner introduced the second generation of MoM resurfacing arthroplasty in 1991 (Grigoris et al 2006). Simultaneously, but independently, McMinn produced a cast Co-Cr alloy MoM resurfacing prosthesis (figure 2.17). While the Wagner design was poor and few were implanted, the McMinn resurfacing (by Corin) underwent successive design adaptations, mainly related to the fixation characteristics, and experienced more clinical use. In 1992, the cementless fixation method was replaced by a cemented design, in an attempt to prevent loosening, only to result in cement debonding at the cup-bone interface (Grigoris et al 2006). This lead to the introduction of the hybrid fixation method in 1994 (Roberts et al 2005) and from the original McMinn prosthesis, the Cormit and Birmingham Hip Resurfacing (BHR) prostheses were developed in 1997. Amstutz' design of a MoM resurfacing prosthesis from 1993 (Amstutz and Le Duff 2006) lead to a similar implant design in 1996, which debuted as the first of the new generation on hip resurfacing prosthesis to be marketed (Roberts et al 2005).

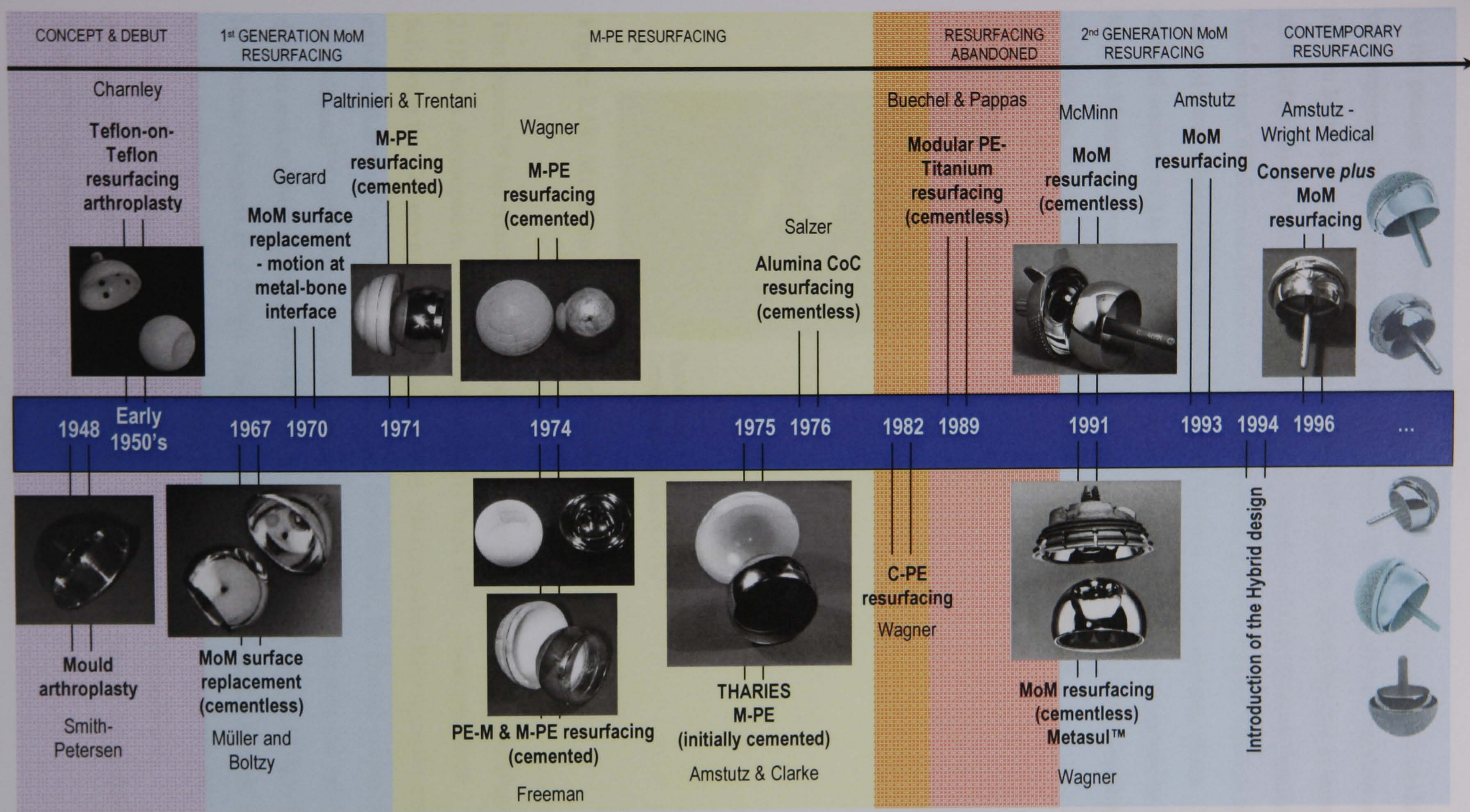


Figure 2.17: Timeline of the evolution of resurfacing arthroplasty

At the present date, most major industrial orthopaedic manufacturers have models of contemporary MoM hip resurfacing on the market. The contemporary models all feature large diameter MoM bearings with hybrid fixation, but differ due to their metallurgy, macro and micro geometry and the specifics of their fixation features. The longest duration of clinical use, and correspondingly, the longest duration of clinical data, relates to the BHR by Smith and Nephew. Measurements of the clinical success of hip resurfacing have so far used revision rates as opposed to Kaplan Meier Survivorship indexes. The National Joint Register (2007) reported a revision rate of 1.8% between 2003 and 2006, based on BHR data.

Durom™ hip resurfacing prosthesis,
Centerpulse Orthopaedics Ltd

Figure 2.18: Metal-on-metal Durom™ resurfacing prosthesis

The Durom™ hip resurfacing prosthesis by Centerpulse Orthopaedics Ltd (figure 2.18) was developed in 1997 and used clinically from 2001. Conceptualised following the Conserve *plus*, BHR and Cormet, the designers drew upon the pre-existing shortcomings to optimise the design. Specifically, the Durom prosthesis uses Metasul™ technology. The wrought-forged Co-Cr is harder than the cast designs and machine polished for a low surface roughness and improved fluid film lubrication (Grigoris et al 2006). The diametrical clearance is standardised at 150μm and the acetabular component is 4mm thick representing the lowest threshold thickness for bone conservation while balancing compressive strength and deformation resistance. The bearing diameter of the femoral head is larger than the Trident™ CoC design, ranging from 38-60mm.

2.3.5.3 Resurfacing arthroplasty-Contending with the standard

While consistently accepted as a significant development in anatomical reconstruction of the hip, resurfacing arthroplasty is yet to provide long-term clinical data to support its use. Grigoris et al (2006) claimed that "hip resurfacing potentially offers the ultimate bone preservation and restoration of function in appropriately selected young patients" (page 95). The limitations of acetabular bone preservation have already been highlighted; the claims of restored of function are attractive and have been detailed by multiple authors. The claims of greater ROM and restoration of normal biomechanics (Amstutz et al 1998) have been largely unsupported with evidence to date. This thesis aims to determine the validity of these perceived advantages of hip resurfacing and quantify them in relative terms with standard stemmed THR, which in contrast is clinically established and well supported.

2.4 Anatomical reconstruction and surgical procedure

Testament to the failure of earlier THR and resurfacing designs, the surgical techniques for implantation are critical in minimising prosthetic wear (Affatato et al 2004, Roberts et al 2005), improving the longevity of the implant (Piriou et al 2007) and optimising ROM and function (Charles et al 2004, Kluess et al 2007) of the patient. Specifically, the surgical approach (Fagerson 1998), restoration of pre-disease soft-tissue tensioning (Asayama et al 2005, Charles et al 2004, Sariali et al 2008) and implant positioning (Affatato et al 2004, D'Lima et al 2000) are critical variable factors, which require standardisation for a comparative study of standard and resurfacing arthroplasty.

2.4.1 Surgical approach

The aim of the surgical approach is to gain adequate exposure of the appropriate anatomical structures to allow accurate anatomical reconstruction of the hip and achieve optimal success of the procedure post-operatively. The two common approaches used are the anterior and the posterior approach. The anterior approach, also termed the lateral or anterolateral approach, often requires trochanteric osteotomy and reattachment, which has potential negative effects for gluteal muscle function post-operatively. Failure or inappropriate reattachment of the greater trochanter or scarring of the muscles may impede rehabilitation, while reattachment of the abductors in a lengthened position may increase dislocation risk (Fagerson 1998). The main advantage of the posterior, or posterolateral approach, is the preservation of the abductor mechanism (Fagerson 1998). Nevertheless, the posterior approach is associated with higher dislocation rates: three times that of the anterior approach (Morrey 1992, cited in Fagerson 1998). The geometrical structure and design of both the Durom™ resurfacing arthroplasty and Trident™ CoC stemmed THR is such that they minimise the mechanical risk of dislocation. Moreover, the greater soft-tissue tension of the younger population reduces the inherent risk of dislocation in the current subject group. The risk of dislocation is reduced further by accurate positioning of the prosthetic components by the surgeon and adherence to post-operative precautions by the patient. The posterior approach was used by the surgeons concerned with this study for both hip resurfacing and standard THR.

2.4.2 Surgical procedure using posterior approach

The posterior approach involves an incision through the iliotibial band (ITB) and a possible incision through gluteus maximus. Detachment of the short external rotators (piriformis, superior and inferior gemelli and obturator internus) at their insertion to the femur is followed by excision of quadratus femoris and obturator externus. Retraction of gluteus medius and minimus then reveals the posterior capsule and capsular ligaments, allowing ligament detachment and full capsulotomy to be performed. The femoral head is then dislocated and bony preparation of the acetabulum and femur is performed in order to fit the prosthetic components.

2.4.3 Differences in surgical procedure for THR and hip resurfacing

Despite the more modest removal of bone required for resurfacing arthroplasty, the surgical procedure is more complex and requires greater anatomical exposure, more complicated instrumentation and a slightly greater duration of operating time than the standard THR procedure. A larger incision is used to cut through gluteus maximus, medius and occasionally minimus, depending on the ease of exposure of underlying bony anatomy. Detachment of most of the gluteus maximus fibres from the insertion at the iliotibial band and gluteal tuberosity is often required. This has possible implications for the recovery of function in these muscles. Femoral head preparation involves careful circumferential and champhered reaming (figure 2.19) to accept the femoral head prosthesis, without exposing underlying cortical bone at the head-neck junction, and therefore avoid the risk of femoral neck fracture.

Figure taken from www.castleortho.com

Figure 2.19: Femoral head preparation for resurfacing arthroplasty

The CoC standard stemmed THR follows a similar surgical procedure to that of the conventional arthroplasty. The excision of soft tissue is less extensive than the resurfacing procedure but the removal of femoral bone more extensive; including the entire head and surgical neck. A cavity is reamed in the shaft of the femur to accept the stemmed femoral component.

Acetabular preparation and press-fitting of the components is similar for both arthroplasty types, although the resurfacing procedure requires more extensive removal of more bone relative to the standard THR socket reaming, so to accommodate the larger diameter acetabular component.

2.4.4 Differences in anatomical reconstruction for THR and hip resurfacing

Implant positioning is critical for accurate hip joint reconstruction, and is the surgeon's opportunity to restore the anatomy and biomechanics of the hip joint to the optimal level achievable (Sariali et al 2008). The position and orientation of the components is directly related to the availability of hip motion before impingement occurs and the stability of the joint. Accurate implant positioning also favours optimal muscle mechanics and soft tissue balance (Charles et al 2004).

2.4.4.1 Application of anatomical reconstruction to hip joint kinematics

The position and orientation of the acetabular cup and the size of the femoral head are the two main factors influencing range of motion following anatomical reconstruction of the hip. Impingement following hip arthroplasty describes the physiological limitation in angular motion of the hip joint due to contact of the acetabular rim (bony or prosthetic component) on the femoral neck (anatomical neck with hip resurfacing or prosthetic neck with stemmed THR) (Malik et al 2007). Placement of the acetabular component may be described in terms of the angles of inclination and anteversion (Murray 1993), or also the depth of seating of the cup (Kummer et al 1999). Many authors have studied the optimal cup orientation which allows essential and optimal ROM before impingement (D'Lima et al 2000, Kummer et al 1999, Yoshimine 2006, Yoshimine and Ginbayashi 2002). A summary describing recommendations of component placement for anatomical reconstruction is featured in appendix 1. The significance of having adequate ROM before prosthetic impingement occurs is not only important for kinematic purposes, but also to avoid subluxation, dislocation, wear and component loosening. When impingement occurs, the centre of rotation shifts from the centre of the femoral head to the impingement site. If further motion continues, this motion acts to lever the femoral head out of the acetabulum (Kluess et al 2007), causing subluxation of complete dislocation of the joint. Contact stress at the impingement site may cause material damage (Kluess et al 2007, Yoshimine and Ginbayashi 2002) or significant sheer forces at the acetabular cup-bone interface (Kluess et al 2007) which may lead to material failure and component loosening.

In addition to optimal positioning, the MoM resurfacing and CoC stemmed THR have design features which limit the likelihood of the negative consequences of impingement. Large diameter bearings reduce dislocation risk by allowing greater ROM before impingement occurs (D'Lima et al 2000, Yoshimine 2006). The alumina liner of the Trident acetabular component has a protective rim which prevents impingement of the alumina insert and the femoral stem during extremes of ROM, thus avoiding ceramic fracture and limiting rotational torque and shear at the interface.

2.4.4.2 Application of anatomical reconstruction to hip joint kinetics

Piriou et al (2007) claimed that the ultimate goal of THR surgery is to "restore the optimal proximal femoral anatomy in order to re-establish the function of the abductors, to allow mobilising without a limp and to reduce the risk of instability" (page 216). In maintaining the anatomic neck of femur, resurfacing arthroplasty is thought to result in optimal restoration of normal anatomy and a "more precise" biomechanical reconstruction of the hip (Girard et al 2006, page 721). There is significant agreement that resurfacing results in accurate restoration of 'normal' anatomy (Girard et al 2006, Mont et al 2007, Roberts et al 2005, Silva et al 2004) and superior restoration of anatomy and function compared to the standard THR (Daniel et al 2004, Girard et al 2006, Mont et al 2007).

The prosthetic stem of the THR femoral component replaces the natural offset of the femoral neck, potentially changing the magnitude of the offset and the anteversion of the prosthetic neck (Asayama et al 2005), which has a direct affect on the length of the abductor lever arm (section 2.1.2.2). A reduced offset, or under-restoration of the femoral offset, may compromise abductor muscle function by reducing the length-tension relationship of the muscle group and, consequently, the ability to generate force, and may result in limp, muscle fatigue and reliance on walking aids (Asayama et al 2005, Berme et al 1985, Charles et al 2004). The resultant hip JRF may become excessive (figure 2.9), and promote early implant wear (Asayama et al 2005, Sakalkale et al, cited in Charles et al 2004). In addition, resultant laxity in the surrounding tissues may predispose the hip to reduced stability and greater dislocation risk (Asayama et al 2005). In contrast, an increased offset may be favourable to abductor muscle function and soft-tissue balance, and may possibly increase abduction ROM (McGrory et al 1995), but, in turn, could have the negative effect of increasing the bending moment subjected to the femoral component and increasing the stress at the femoral shaft, including the bone and cement interfaces

(McGrory et al 1995, Piriou et al 2007). Appendix 2 summarises the methods of manipulating femoral offset during anatomical reconstruction and the associated effects which may influence function.

The standard reference for guiding anatomical reconstruction is given by the geometry of the 'normal' contralateral hip on X-ray (Girard et al 2006, Loughead et al 2005, Silva et al 2004) and X-ray templating of the arthroplasty components on the hip to be reconstructed (Charles et al 2004). In addition to accurately restoring femoral offset, restoration of the patient-specific centre of rotation of the femoral head and leg length are also important for function (Girard et al 2006). The stemmed component appears to better compensate for leg length discrepancies observed before surgery, whereas the hip resurfacing has been associated with more limited ability to alter limb length (Loughead et al 2005) due to the reduced ability to lengthen the offset or alter the anteversion and neck-shaft angle of the femur. However, resurfacing is often viewed as advantageous, as it results in a more anatomical form of restoring leg length (Loughead et al 2005, Silva et al 2004) and avoids complications associated with inaccurate restoration of leg length with stemmed components (Girard et al 2006).

Acetabular reconstruction is similar for both procedures. Although there is greater acetabular reaming required for the resurfacing procedure, the relatively larger diameter femoral head of the resurfacing component is unlikely to result in medialisation of the centre of rotation of the femoral head (Girard et al 2006) relative to the stemmed THR (Silva et al 2004).

2.4.5 Selection of arthroplasty type

The orthopaedic surgeon considers individuals who are young and active for both the MoM resurfacing and the stemmed CoC THR. The low wear rates and expected longevity of both prosthesis imply their suitability to those who may out-live or 'out-run' the conventional hip arthroplasty. Individuals aged sixty-five years and younger are generally considered for MoM resurfacing (NICE 2002) or CoC stemmed THR. As for the appropriate selection of prosthesis for the individual, this decision is more complex. Given the potential merits of both procedures and lack of research directly comparing their use, the surgeon has limited argument for the clinical decision making process. However, certain eliminating factors rule out the use of resurfacing for certain anatomical presentations. The presence of cysts and necrosis of bone renders the resurfacing unsuitable due to the weakening of the bone which is required to support the femoral component. Bony deformation may have altered the thickness of the femoral neck. If the neck thickness is similar to or greater than the inner diameter of the femoral head prosthesis then resurfacing would be undesirable, as reaming would expose bone on the femoral neck. These scenarios may precipitate fracture of the femoral neck and therefore stemmed THR is the probable option.

Due to metallurgy, young females of childbearing age may not be considered for resurfacing depending on the surgeons weighting of the risks associate with metal components and the patient-surgeon informed decision process. Older or post-menopausal females may not be considered for resurfacing due to the weakening of bone associated with osteopenia (Mont et al 2001).

Overall there are potentially small differences in the structural and soft-tissue reconstruction resulting from resurfacing and standard THR. Based on the minimal anatomical differences observed during radiographic comparisons, existing literature has predicted an outcome of minimal kinematic and kinetic differences between the procedures, to the extent of suggesting that the existing differences are "likely to be clinically insignificant" (Loughead et al 2005, page 165), yet claims that resurfacing results in better function and ROM compared to standard THR continue. The greatest differences between the resurfacing and standard THR procedures and designs, relate to the preservation of the neck and greater diameter of the femoral head with hip resurfacing arthroplasty. As the preservation of the femoral neck is thought to provide superior anatomical reconstruction and biomechanics, the resultant patient function may display superior kinetics compared to patients with standard THR; as the greater diameter femoral head is thought to, among other benefits, improve ROM, the functional kinematics of the resurfacing patient may be superior to that of the patient with standard THR. The assessment of these claims beyond radiographic analysis has been scant. The limited literature concerning these claims will be considered in the following sections, reviewing motion analysis studies relating to hip arthroplasty.

2.5 Motion Analysis

Modern human motion analysis may be achieved by using stereophotogrammetry. This is a method of motion capture which acquires quantitative information about the musculoskeletal system during locomotor activities, often by simultaneous collection of kinematic and kinetic data. Data can be processed to analyse specific parameters of gait and other motor tasks and comparison of parameters can highlight deviations from normal function or relative differences in motion characteristics of test subjects.

Motion of the subjects is recorded within a defined capture volume, associated with a specified global reference frame. The body is represented as a series of segments, interlinked by joints, allowing relative movement of the segments. These segments are considered as rigid bodies within a kinetic chain (Cappozzo et al 2005). Three-dimensional motion analysis records intersegmental joint motion in three degrees of freedom (DOF) with each rigid body segment assigned a 3-dimensional axis system, originating at the joint centre. Wu et al (2002) have defined a standardised co-ordinate system for the lower limb joints, as recommended by the International Society of Biomechanics (ISB).

During motion analysis, the limb segments are defined by a given marker system, which assigns a technical reference frame to the body segment. Anatomical landmarks associated with the segment are identified and the position of the various joint centres is calculated by standardised methods. Connecting one joint centre to the next generates a line representing the technical axis of the underlying bone, and creates the basis for the anatomical frame, which is representative of a subject-specific model. Anatomical and technical frames are known as local reference frames (Cappozzo et al 2005). Continuous recording of the position and orientation of local reference frames in relation to the global reference frame of the laboratory, allows analysis of the body motion relative to the environment. Description of the relative movement of adjacent segments to one another allows joint kinematics to be analysed, and the use of integrated force plates to provide ground reaction force (GRF) data allows kinetic analysis such as the calculation of intersegmental joint moments by inverse dynamics (Bell et al 1989). The practical components of motion analysis will be covered in detail in chapter 4.

Stereophotogrammetric errors can be attributed to calibration, filtering and errors in marker positioning (Chiari et al 2005). The latter is the greatest source of error, and may be due to inaccurate marker positioning, movement between the marker on the skin surface relative to the underlying bone and marker flickering or complete loss of marker tracking. Skin movement artefact creates a positional discrepancy between the bone and the skin surface (Leardini et al 2005), which may result in errors in the calculation of angle and moment values related to the associated joint. These errors can be minimised by using clusters of markers to identify the segment and a pointer system to identify the anatomical landmarks instead of direct skin marker placement (Della Croce et al 2005). Secure fixation and appropriate placement of marker clusters limits the motion of the clusters relative to muscle movement and prevents displacement of the cluster with respect to the bone.

Accuracy in locating the hip joint centre (HJC) is important for precise kinematic and kinetic analysis (Bell et al 1989, 1990, Cereatti et al 2007, Leardini et al 1999, Seidel et al 1995). Due to the depth of the hip joint, identification of the true HJC may only be done using radiographic analysis (Crowninshield et al 1978), but amongst difficulties with conversion of planar X-rays into accurate 3-D coordinates and correction for magnification, such X-ray exposure is unnecessary and potentially harmful (Bell et al 1989, Seidel et al 1995). Hence for practical and ethical purposes, methods of estimating the HJC location are preferred for motion analysis. Two recommended methods exist; the predictive approach and the functional approach.

The predictive approach uses a set formula, derived from anthropometric data of the pelvis, to calculate the position of the HJC from palpable anatomical landmarks (table 2.2). The 3-D position of the HJC may be expressed as a percentage of the distance between the anterior superior iliac spines (ASIS), located at a given percentage distance distal, medial and posterior to the ASIS. This method was introduced by Tylkowski et al (1982, cited in Bell et al 1989) and has been validated by Bell et al (1989, 1990). The predictive method has been criticised as the founding anthropometric data was based on pelvic measurements from a relatively small number of adults, latterly males, and from cadaver specimens (Della Croce et al 2005). However, high repeatability between individuals and genders has been demonstrated (Bell et al 1989, Seidel et al 1995) and low errors in predicting the true radiographic HJC when compared to other methods (Bell et al 1990, Leardini et al 1999) (table 2.2).

Table 2.2: Comparison of hip joint centre estimation methods

Method	Description of method	Advantages	Disadvantages	Error (cm) Bell et al 1990	
				Mean	SD (±)
Predictive -by Tylkowski et al 1982, adapted by Bell et al (1990)	30% distal, 14% medial and 19% posterior to ASIS (where 100% = inter-ASIS distance)	Usable Accurate Reliable Repeatable Simple	Not subject specific Subject to additional marker errors	1.90	1.2
Predictive -by Andriacchi et al 1982 (cited in Bell et al 1990)	1.5-2cm distal to the midpoint of a line between the ASIS and pubic symphysis and an unspecified distance from the greater trochanter	Accurate in 2-D Simple	Poor reliability Not subject specific Subject to additional marker errors Requires pubic symphysis marker (poor visibility & potentially socially inappropriate for subjects)	3.61	1.2
Functional	Regression analysis applied to kinematic data of multiple large oscillations of the thigh segment to predict COR	Subject specific Usable with appropriate software program	Subject to additional marker errors Requires large oscillation angles for accurate prediction	3.79	1.9

Information sourced from Bell et al 1989, 1990; Della Croce et al 2005

The functional approach (Cappozzo 1984, cited in Seidel et al 1995) regards the HJC as the functional pivot point for movement between the femur and the pelvis, and uses regression equations to calculate the "co-ordinates of the centre of mutual rotation" on a subject-specific basis (Della Croce et al 2005, page 228). Physically this requires large excursions of hip joint motion, measured by movement of the thigh segment relative to the pelvis, in order to accurately detect the pivot point (Bell et al 1990, Leardini et al 1999). Errors in determining the position of the pivot point increase as the performed ROM decreases and consequently, may not be reliable for estimating the HJC of individuals with ROM restrictions (Seidel et al 1995) or an inability to perform the required motor task during motion analysis.

When estimating the HJC location, "no method is entirely satisfactory" (Bell et al 1989), or entirely accurate. Although the functional approach is recommended for subject-specific location of the HJC (Della Croce et al 2005) it may not be suitable or accurate enough for use on individuals with pathology or those recovering function following hip resurfacing or THR. The greater usability, consistency and accuracy of the predictive method modified by Bell et al (1990), may provide the optimal method of HJC estimation for motion analysis using hip arthroplasty subjects.

2.6 Gait Analysis

Level walking gait is characterised by sequential alternate movement of the lower limbs to provide support and propulsion, allowing forward progression of the COM (Whittle 2002). The gait cycle is classified as the time interval between two successive heel contacts, of the same foot (Murray et al 1964), for example, from the heel strike of the right foot to the following heel strike of the right foot. During the cycle, each limb has a stance phase and a swing phase, accounting for approximately 60% and 40% of the gait cycle respectively.

2.6.1 Features of normal gait

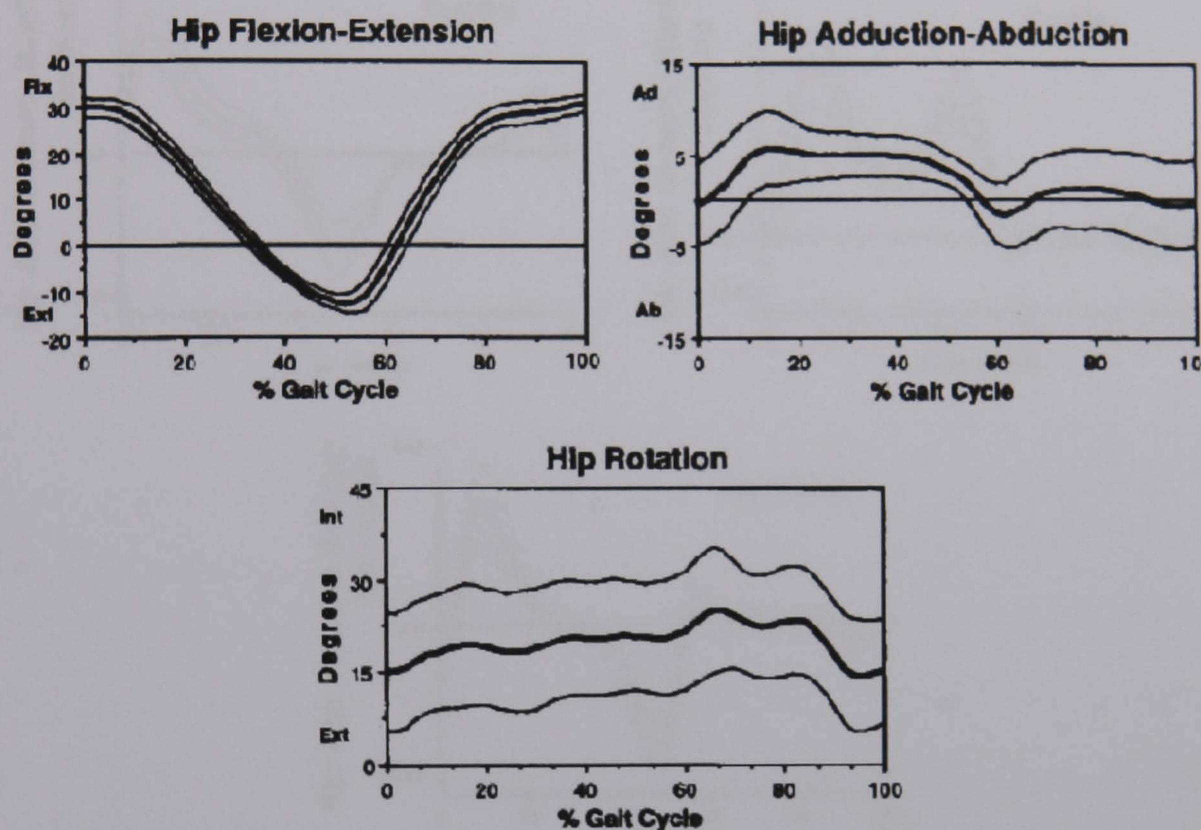
When considering the 'normal' population, gait characteristics are often assumed to be symmetrical (Loizeau et al 1995). In such case, assessment of an individual with unilateral hip arthroplasty would therefore allow existing dysfunction to be clearly identified from an objective comparison of operated and non-operated limbs. However, differences between left and right limbs of able bodied subjects have been identified (Loizeau et al 1995) from temporospatial, muscle power and muscle energy parameters, although these differences were subtle and not statistically significant. Such small differences may be due to limb dominance and discrepancies in the habitual motion patterns from left to right limbs, although the authors did not allude to such reasons nor did they document, or control, the dominant lower limb of their able bodied subject group. Other secondary factors causing biomechanical differences between limbs may include relative muscle weakness and reduced proprioception (Madsen et al 2004).

Gait velocity is regarded as a useful and reliable indicator of functional ability and has been frequently used as an outcome measure in motion studies (Andriacchi et al 1977, Aminian et al 2004, Eng and Winter 1995, Kadaba et al 1989, Murray et al 1964). Normative values for level gait velocity have been observed at a mean of 1.6m/s in the young population (age 19-26 years; Eng and Winter 1995) and 1.19m/s (Riley et al 2001) to 1.3m/s (Kadaba et al 1989) for the young to middle-aged population (age 18-40 years). Gait velocities do not appear to vary due to gender (Riley et al 2001).

Certain kinematic and kinetic characteristics have consistently been observed during 'normal' gait. These may be used as a reference for contrasting with the motion characteristics found following THR. Most studies assessing gait kinematics and kinetics of normal, pathological or arthroplasty subject groups, have concentrated on sagittal plane motion, although the importance of 3-D motion analysis has been stressed, particularly for hip joint analysis (Eng and Winter 1995).

2.6.1.1 Kinematics of normal gait

The functional ROM required during gait differs significantly to the full active or passive ROM of the hip joint. James et al (1994) claimed that only 30% of full (passive) sagittal plane motion was used during level walking, yet the peak extension angles involved closely represent the maximum anatomical extension ROM (Murray et al 1964). The mean 'normative' range of sagittal hip motion during gait is consistently reported at 30-35° of flexion, occurring at the mid-swing phase, and 10-15° of extension, before foot-off (Johnston and Smidt 1969, Kadaba et al 1989, Murray et al 1964, Whittle 2002). Coronal and transverse motion have been documented less, although Kadaba et al (1989) performed a repeatability study on normal adults aged 18 to 40 years, and the angular values may be observed in figure 2.20. Murray et al (1964) found that there was excellent consistency of sagittal gait kinematics and temporospatial features within the normal population, and minimal variability due to height or age (age range 20-65).



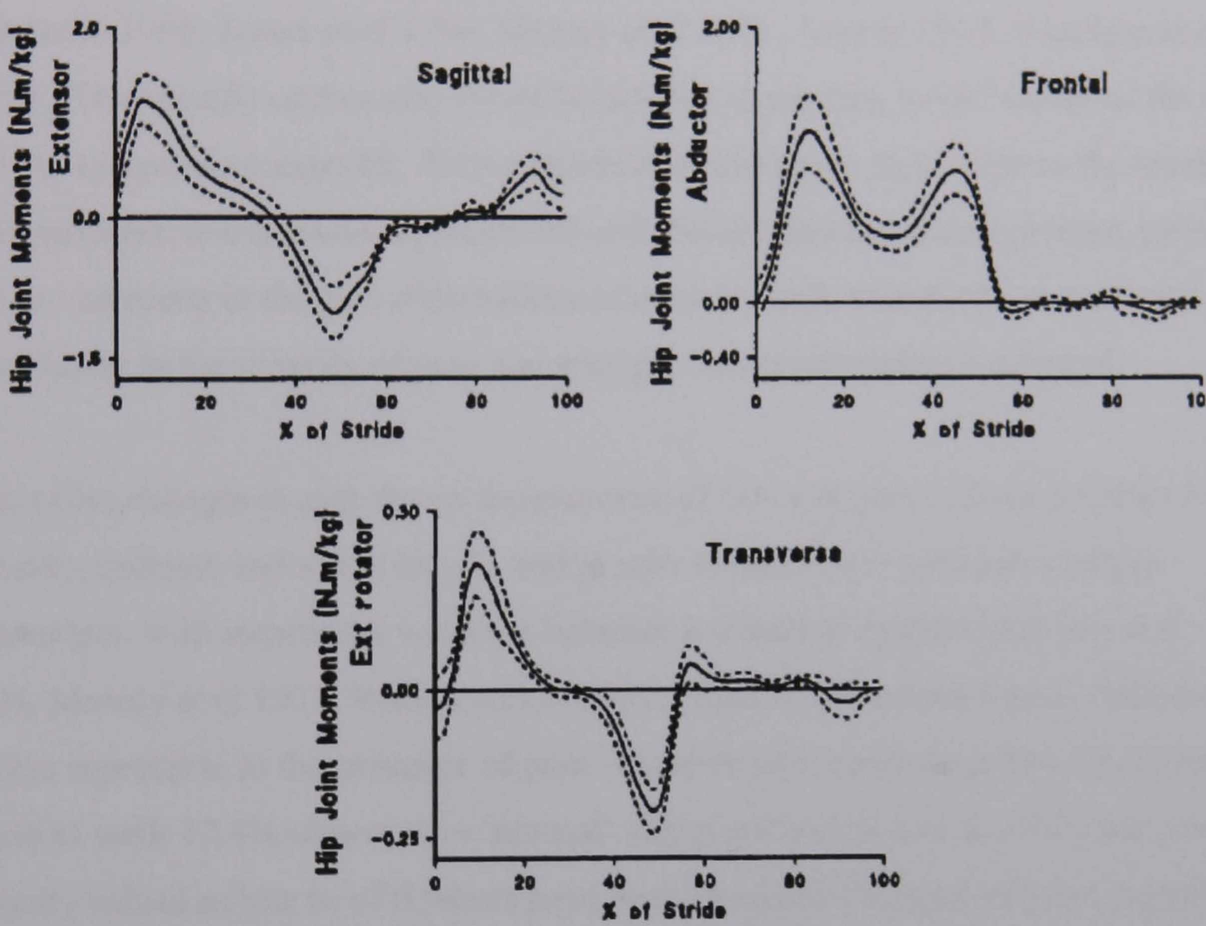
Figures taken from Kadaba et al 1989

Figure 2.20: Normative kinematic data for the hip during level gait

2.6.1.2 Kinetics of normal gait

The vertical GRF produced from force plate data provides a vector representation of the resultant force acting in response to foot contact during gait and other functional activities of daily living (ADL). Characteristics of the reaction force profile are known to vary with gait velocity, and with pathology or following THR (Andriacchi et al 1977, Riley et al 2001), and correspondingly the moment profiles also vary.

Several studies have examined the kinetic characteristics of 'normal' gait (Eng and Winter 1995, Kadaba et al 1989), commonly using the parameter of peak external or internal moments. From heal contact to mid-stance, the GRF vector passes anterior to the hip joint, producing an external flexor moment, which is opposed by an internal hip extensor moment. This is followed by an external extensor moment and an opposing internal flexor moment as the GRF vector moves posterior to the hip joint at the initiation of double-stance (Tanaka 1998, Whittle 2002). Figure 2.21 illustrates internal hip moment profiles representing normal gait.



Figures taken from Eng and Winter 1995

Figure 2.21: Normative kinetic data for the hip during level gait

Despite the early findings of Murray et al (1964), certain motion parameters of 'normal' gait do appear to be velocity-dependent (Perron et al 2000). With increasing age, Tanaka (1998) found that gait velocity, peak acceleration and peak flexor moments reduced, while sagittal hip angles and extensor and abductor moments did not differ significantly across the age range of a middle-aged 'normal' population (41-77 years, mean 60.8 years). The existence of age-related variations in outcome parameters implies a need for controlling the age range of individuals involved in a comparative study. When considering the cause of the reduced peak flexor moment observed, it may be that the reduced gait velocity and acceleration were responsible, as a direct correlation between these findings was observed. Perron et al (2000) did however agree with Tanaka (1998) that no correlation or variation in the abductor moment was observed with change in gait velocity.

2.6.2 Gait and osteoarthritis

Osteoarthritis of the hip results in significant gait abnormalities (Aminian et al 2004, Isobe et al 1998, James et al 1994, Murray et al 1971, Tanaka 1998, Watelain et al 2001). The specific causes may be multifactorial depending on the extent of the OA and the symptoms presented. Biomechanical disturbances may relate to the level of pain, stiffness, compensation patters and soft tissue adaptations that present, giving rise to variations in the gait observations of subjects with unilateral osteoarthritis due to variation in the severity of pain and antalgic movement patterns adopted.

Generally, changes in gait due to the presence of OA and pain include a reduced gait velocity, cadence and stride length, and greater variability in symmetry of gait parameters, with significant variation between successive cycles (Aminian et al 2004, Murray et al 1971, Watelain et al 2001). Gait characteristics may, therefore, be less repeatable in the presence of pain. Patients with early stage OA have been found to walk 12.4% slower than 'normal' subjects (Watelain et al 2001) and mean velocity values as low as of 0.94m/s have been observed for men with OA (aged 32-79 years; Murray et al 1971). The sound limb tends to develop a shorter stride, conducive with a shorter duration of stance on the osteoarthritic limb which occurs secondary to the reduced weight-bearing capacity of the joint (section 2.2). Pain inhibition is responsible for a reduced ROM at the hip (Madsen et al 2004), specifically a reduction in the peak extension angle at late-stance (Murray et al 1971). Reduction in stride length of the sound limb is also related to the common lack of extension ROM available at the OA hip joint due to flexion contracture of the hip joint (Tanaka 1998). However, individuals with unilateral hip OA have demonstrated significantly greater hip flexion angles ($p<0.01$) on the unaffected side compared to the affected side which is believed to be compensatory (Tanaka 1998).

Kinetic parameters of the OA hip during gait have also shown significant differences to 'normal'. Vertical GRF curves during gait have demonstrated a plateau (Kelly 1984, cited in James et al 1994) and a lower rate of loading when compared to the contralateral 'normal' hip, indicative of a possible attempt to control the joint loading in order to minimise pain. Reduced sagittal hip moments, 52% below 'normal', have been observed at push-off (Watelain et al 2001). Coronal plane kinetic abnormalities have been suggested but were mostly reported as positive Trendelenburg signs as opposed to objective values (Watelain et al 2001).

Individuals with painful hip joints may use walking aids. These may partially relieve pain by reducing the loading on the pathological joint, as demonstrated by the reduced peak vertical ground reaction forces during gait (Murray et al 1972). Compensatory mechanisms may also be adopted in other joints, including increased lumbar spine motion and pelvic rotation, to assist forward progression of the moving limb (Vogt et al 2003).

Given the variations in the presentations and pain behaviour of subjects with varying severities of OA and the overall variability in biomechanics, the measure of pre-operative outcomes may not provide an informative or valuable form of comparison between resurfacing and standard arthroplasty groups. Nevertheless, the importance of pre-operative similarities between the groups has been recognised for the purpose of controlled, homogenous group comparisons.

2.6.3 Gait following THR

Literature supports the use of hip arthroplasty in providing successful reduction of pain, restoration of function and joint stability (Perron et al 2000, Loizeau et al 1995) and improvement in quality of life (Madsen et al 2004). Studies investigating the restoration of joint biomechanics have found that, among kinematic and kinetic improvements, THR increases the stride length, cadence and velocity of gait (Murray et al 1972, Perron et al 2000, Tanaka 1998), and improves the symmetry of the stance and swing phases (Isobe et al 1998, Madsen at al 2004). However, improvements have been found to be below normal levels for comparisons to the contralateral limb and with age-matched control groups (Madsen et al 2004, McCrory et al 2001, Perron et al 2000, Tanaka 1998). Hence THR is not likely to lead to full return of function (Perron et al 2000). In any case gait characteristics "may not return to normal for several years" (Madsen et al 2004, page 44) and it is believed that these residual impairments may "contribute to the persistence of locomotor disabilities" (Perron et al 2000, page 505).

Independent walking without aids tends to be achieved within 6 months (Perron et al 2000), although variation in this milestone may to be related to the age of the study group, the motivation and co-morbidities of the individual and the rehabilitation protocol adhered to. Gait velocity following hip replacement tends to remain below 'normal' values. Perron et al (2000) presented findings of gait velocity values up to 25% below normal at between 2 and 4 years following surgery.

2.6.3.1 Kinematic assessment following hip arthroplasty

The angular excursion of hip flexion-extension movement (total joint movement in the sagittal plane) increases following THR but remains below normal values (Murray et al 1971, Perron et al 2000). In particular, the peak extension angle of the operated hip remained lower than that of the non-operated hip joint and the extension values found in the 'normal' population (Perron et al 2000), which may be due to remaining hip flexion contractures (Hurwitz et al 1997). Hip flexion contractures of up to 10° may persist for up to 1 year post-operation (Olsson 1986, cited in Perron 2000). Perron et al (2000) attributed these flexion contractures to be the primary reason for a loss of hip extension at terminal stance, reduced by 59% compared to healthy age matched subjects, thus limiting the length of stride of the contralateral sound limb and reducing sagittal hip moments and energy generation for push-off.

However, other factors may act to limit extension ROM. These may include resistance from remaining anterior capsule tissue, features of the prosthetic components or their alignment relative to the bone, which may cause impingement (section 2.1.2.1). Alteration of the abductor lever arm with stemmed femoral components may also contribute to the reduced flexion-extension range, due to a corresponding change in the muscle action (Andriacchi et al 1980). Madsen et al (2004) compared post-operative function following anterior-lateral (A-L) and postero-lateral (P-L) surgical approaches and discovered a smaller sagittal plane ROM during gait following the A-L (34.0°) compared to the P-L approach (39.4°). The greater sagittal motion following the P-L approach was attributed the to suture repair of the posterior capsule as opposed to the anterior capsule, which restricted extension motion less.

Although most studies have concentrated on sagittal plane motion, a reduction in hip movement in the coronal and transverse planes has also been observed following THR (Hattori et al 1983). Perron et al (2000) reported that there was greater external rotation on the operated side relative to the sound limb following hip arthroplasty.

2.6.3.2 Kinetic assessment following hip arthroplasty

Kinetic parameters have been shown to become more symmetrical post-THR (James et al 1994). Although reduced abductor, flexor and extensor moment values have been observed for OA hips during gait, values for THR joints (12-85 months post-operatively) have been found to be statistically similar to that of the contralateral sound joint and did not differ significantly from subjects without THR (Tanaka 1998). However, in earlier studies by Tanaka (1993, cited in Perron et al 2000) decreased hip flexor, extensor and abductor moments were reported as residual impairments, continuing in the short-term following THR. It would appear, then, that these kinetic parameters stabilise following 1 year post-surgery. Perron et al (2000) found that the peak extensor moment in a female THR group was 20% less than that of women in the 'normal' healthy group.

The hip abductor moment remains below normal values following THR. Perron et al (2004) observed that at the end of weight acceptance during gait, the peak abductor moment was 15% less in the THR group compared to an age and sex matched 'normal' group. By way of compensating, greater movement of the COM over the THR limb may occur during stance, thus reducing the BW lever arm (figure 2.9) and resulting in less force production demands on the abductor muscles to provide balancing torques (Sliwinski et al 2004, Madsen et al 2004).

Hip moments in the transverse plane have been reported infrequently. The comparative study of healthy females to those with THR by Perron et al (2000) observed a significant reduction in the peak external rotator moment during the mid-stance phase of gait following THR.

The correlation between ROM and moment outcomes has been assessed (Tanaka 1998). THR subjects with hip extension angles greater than 0° were associated with significantly higher internal flexor moment values during gait than those with fixed hip flexion contractures. This finding is of some importance as the internal flexor moment is proportional to the propulsion during the early swing phase and is crucial for the acceleration of the trunk (Tanaka 1998).

To date, most biomechanical studies of hip arthroplasty have assessed individuals with conventional THR. Although there is a great improvement in biomechanical function of gait following the conventional arthroplasty, there appear to be residual deficits. These residual gait deviations may be directly related to the difference in mechanics of the pre-diseased hip joint, differences established as a result of the disease process, differences due to the structure of the prosthesis and its associated alteration in structural anatomy of the joint, or may be associated with the altered muscle function or a loss of proprioceptive acuity (Kelly 1984, cited in James et al 1994).

2.6.4 Factors influencing abductor muscle function

Efficient functioning of the abductor muscle group is of great biomechanical importance (section 2.1.2.2, figure 2.9). Abductor dysfunction may be represented by reduced external abductor moments, reduced isometric muscle strength and a positive Trendelenburg sign.

The abductor moment has been shown to correlate well with abductor strength (Tanaka 1998) and reduced abductor muscle strength may be observed clinically with a positive Trendelenburg sign. Downing et al (2001) demonstrated a close positive correlation between isometric abductor muscle strength and the Trendelenburg test. Tanaka (1998) found that the post-operative muscle strength was significantly lower in the operated hip compared to the non-operated hip and that of 'normal' subjects. Greater post-operative abductor muscle strength has also been observed following the P-L approach compared to the A-L approach (Gore et al 1985). The Trendelenburg sign has been shown to improve by three months post-THR and more so by 12 months, possibly due to the "loss of the effect of pain inhibition" (Downing et al 2001, page 218). Abductor and extensor muscle strength in particular have been highlighted throughout literature for their important role in locomotor recovery. Better regarded is the importance of abductor muscle strength; less documented is the importance of the extensor muscles, which were of significance in the conclusions of Perron et al (2000), who recommended extensor muscle strength training to optimise recovery of function.

There may be others co-existing factors benefiting abductor muscle function such as the return of a more 'normal' movement pattern, including the reduction of compensatory patterns established pre-operatively, greater activity levels or improved proprioception and neuromuscular control. In addition the various factors related to anatomical reconstruction of the hip (femoral neck offset, varus/valgus alignment of the stem, the diameter/depth of the acetabular component and leg length) may also influence abductor muscle function post-operatively (Downing et al 2001).

2.6.5 Study of outcomes following hip resurfacing arthroplasty

There are few studies examining functional outcomes following resurfacing arthroplasty compared to the relatively large body of literature concerning standard THR. The lack of outcomes following hip resurfacing is reflective of the relatively shorter lifespan of the current generation resurfacing prosthesis and the poor results and low numbers of implants associated with the first generation design. Second generation resurfacing has only excelled in popular use since the late 1990's and consequently only short-medium term data is available (Back et al 2005).

Nevertheless, there still remains a lack of objective data documenting the post-operative performance of those with a hip resurfacing. Most studies have focussed on clinical indicators and subjective outcomes from questionnaires or have objectively reviewed radiographs for outcomes to rate the anatomical reconstruction of the hip, allowing speculation on the effects on function (Loughead et al 2005). Radiographic critique to predict biomechanics and function is valid to some extent, but studies of objective functional measures are required to validate expected outcomes derived from radiographic reviews and practically evaluate movement and function following hip resurfacing.

2.6.5.1 Resurfacing arthroplasty-Clinical indicators and subjective data

Daniel et al (2004) undertook a long-term retrospective analysis of individuals with OA having had a hip resurfacing (McMinn hybrid hip resurfacing and BHR) via posterior surgical approach. They assessed survival rates, Oxford Hip Scores and activity levels of 384 patients at a range of post-operative follow-up times, from 1.1 to 8.2 years and found promising results. A survivorship of 99.8% was observed, which was significantly better than comparative data from the Swedish Hip Arthroplasty Register (2000) showing a survivorship of 80.5% at 10 years. The survivorship was also better than that from data of conventional arthroplasty at 8 years, although the age mismatch may negate the power of this comparison, as the better outcome may be related to the relative youth of the resurfacing subjects (<55 years) rather than the type of arthroplasty used. Of the patients without contralateral hip OA or co-existing pathology, Daniel et al (2004) observed Oxford Hip Scores of between 12 and 24 (out of a possible 60, indicating the worst outcome) and no subjective report of severe pain or other functional limitation. They also used the modified UCLA Activity Level Scale and found that all employed individuals returned to their previous jobs, including those involving heavier work activities. Accounting for leisure activities, 81.3% of their entire subject group was rated as 'very active'. Those of a lower level of functional activity were said be limited by pain in another joint.

Back et al (2005) conducted a prospective analysis of 230 hips following resurfacing arthroplasty assessed using Harris Hip Scores, SF-12 scores and the Oxford Hip Score in addition to radiographic assessment and revision rates. The authors claimed that their patients had an excellent return to function. Radiographs revealed minimal implant loosening or femoral neck resorption and only two hips were revised (one early femoral neck fracture and one episode of component loosening). Subjective scores also represented good or excellent outcomes.

Mont et al (2001) compared the use of limited femoral resurfacing (hemi-resurfacing arthroplasty) with THR for younger patients with femoral head osteonecrosis. Harris Hip Scores (88 points and 93 points) and Kaplan-Meier survival analyses (90% and 93%) were carried out and no statistically significant difference was found between femoral resurfacing and THR. However, it was noted that the limited resurfacing group maintained a higher level of activity. Although a total resurfacing prosthesis was not used, this study represented the first known comparative investigation with standard THR.

Nevertheless, there remains a lack of literature featuring objective functional outcome measures while including comparison of standard stemmed THR and second generation hip resurfacing. Given the significant differences in the design of these arthroplasties and the surgical procedure involved, it is possible that there may be differences in motion analysis outcomes when compared individuals with resurfacing and standard arthroplasty.

2.6.5.2 Resurfacing arthroplasty-Objective comparisons with standard THR

Gore et al (1985) compared pre- and post-operative isometric abductor muscle strength ROM and gait velocity between individuals (n=20) with first generation resurfacing (THARIES, Wagner, Indiana components; mean age 55 years) and standard THR (mean age 62 years). They found that the resurfacing group had greater muscle strength and gait velocity but this was also found pre-operatively, and hence, differences were attributed to the age mismatch between groups.

Only one known experimental design has compared gait characteristics following modern standard stemmed THR and modern resurfacing arthroplasty. Mont et al (2007) compared radiographic and motion analysis outcomes, using velocity, angle and moment parameters during level walking, for patients with the Conserve *Plus* hip resurfacing (n=15) and standard stemmed THR (n=15). The authors concluded that the resurfacing group demonstrated superior function, as defined by faster walking speeds and larger abductor and extensor moment values (table 2.3), which were statistically similar to data from a 'normal' sub-group. In addition, the sub-optimal results found for standard THR group were statistically similar to subjects of an osteoarthritic sub-group.

Comprehensive data showing 3-dimensional hip angles and moments was not presented in the study by Mont et al (2007), but as suggested from previous literature, the extensor and abductor muscle function is of high importance following hip arthroplasty, so perhaps the most clinically significant findings were selected for reporting. No difference was found for angle or moment parameters for the knee and ankle joints. The authors did state that the standard THR group and OA sub-group "walked with increased hip abduction compared with the resurfacing group" (page 105), which was classified as a deviation from normal. Poor transparency in the figures presented makes these statements difficult to verify, with only data for the OA (5-8° abduction) and normal (6-7° adduction) sub-groups reported. The statement of significance suggests that the difference in coronal plane gait angles between the arthroplasty groups was not significant ($p<0.21$) which contradicts the concluding statements, and suggests kinematic similarities following resurfacing and stemmed THR. Hence, based on the available data, the differences between arthroplasty groups were related to velocity and moment outcomes alone. Given the greater gait velocity of the resurfacing group and the positive correlation between gait velocity and moment values (Riley et al 2001), it may be that the greater hip moments of the resurfacing group were related to the faster walking speeds adopted.

No correlation analysis was performed to support this relationship, but if this were the case, the use of kinetic data to classify that resurfacing resulted in superior function may have been subject to a type II error and must be judged with care.

Table 2.3: Comparison of resurfacing and standard arthroplasty during gait

PARAMETER	Resurfacing Group (mean, SD, range)	Standard THR Group (mean, SD, range)	Difference (p-value)
Gait Velocity (m/s)	1.26 ±1.8 0.99-1.56	0.96 ±1.3 0.74-1.28	<0.001
Abductor Moment (Nm/kg)	0.78 ±0.1 0.54-0.98	0.63 ±0.2 0.15-0.81	0.21
Extensor Moment (Nm/kg)	1.05 ±0.3 0.68-1.57	0.71 ±0.2 0.35-1.17	<0.001

Information sourced from Mont et al 2007

Other variables within the study by Mont et al (2007) lacked sufficient control. The stemmed THR prosthesis and manufacturer were not named and control of the implant type was not documented. The surgical approach (anterolateral) and rehabilitation protocols were controlled, although the assessment session was not consistent across all subjects (mean 13 months; range 6-15 months post-op). Previous studies have observed changes in objective outcomes up to 1 year following THR, so the variation in the follow-up may have introduced errors and large variations in the results.

Overall, although Mont et al (2007) have produced the first objective and functional related study comparing resurfacing arthroplasty and standard THR, their study appears to have several flaws and potential areas of bias which may compromise the power of their conclusions. Nevertheless, their study does support the expected outcome that resurfacing results in superior functional outcomes. The benefits of resurfacing were attributed to the "closer approximation to the normal proximal femoral anatomy" (page 106) and larger femoral heads which were thought to be beneficial in maintaining the abductor and extensor lever arm.

The study by Mont et al (2007) and other THR investigations have consistently used the study of level gait analysis to yield the outcomes for their comparison. However, the use of other ambulatory tasks has been recommended, such as stair climbing (James et al 1994), which provides a better understanding of the 'environment specific adaptations' (Nadeau et al 2003) concerned with recovery post-arthroplasty.

In order to confidently affirm the functional benefits of resurfacing over THR, a better controlled, comparative study is required, which re-assesses gait parameters and incorporates more challenging ambulatory tasks.

2.7 Analysis of Stair Negotiation

2.7.1 Significance of stair negotiation

Like level walking, stair climbing is a commonly performed locomotor task used regularly during activities of daily living. It has been likened to tasks such as negotiating obstacles, and hence requires the use of other learned motion patterns in order to respond to the stairs environment safely (Nadeau et al 2003). Stair negotiation also requires greater muscle strength and co-ordination than level walking. Given the greater demand of the task, those with pathology or having had joint replacement may find this task more challenging (Riener et al 2002), and hence, analysis of stair negotiation may highlight greater differences in functional outcomes between resurfacing and standard THR groups than level gait analysis.

Biomechanical aspects of stair negotiation have not been studied as much as those of level gait, although more literature has emerged in the last two decades (Nadeau et al 2003). Nevertheless, this literature features studies of a high quality and highlights some interesting comparisons between these two locomotor activities.

2.7.2 Differences with level walking

Gait analysis, as it is studied, tends to be characterised by reciprocal walking in a straight line, carried out at a consistent velocity of forward progression, to obtain a representation of steady state walking. Stair negotiation is more complex. Firstly, it can be divided into two sub-tasks of stair ascent and stair descent. It also combines the analysis of level walking to reach the stairs, which is complicated by deceleration at the foot of the staircase in order to alter to the stair environment and ascend. Ascent is followed by further forward progression by accelerating into level walking or decelerating and turning in order to descend the stair case. Further to this, stair ascent and descent may be reciprocal or non-reciprocal and might involve the use of handrails.

2.7.2.1 Temporospatial parameters

Authors studying stair negotiation have tended to simplify the assessment and study one or more sub-units of the task. Several reports have assessed groups of healthy individuals during level walking and stair negotiation, allowing comparison of the task performance as a means of characterising biomechanical aspects of the motion involved.

Temporospatial analysis characterises stair ascent with a relatively longer step cycle than level walking. This ascent cycle has a proportionally greater swing phase and a lower cadence compared to level walking (Nadeau et al 2003). The greater swing phase was also observed by earlier studies (Livingston et al 1991, cited in Nadeau et al 2003) and may be attributed to the greater time required to clear the foot over the intermediate step.

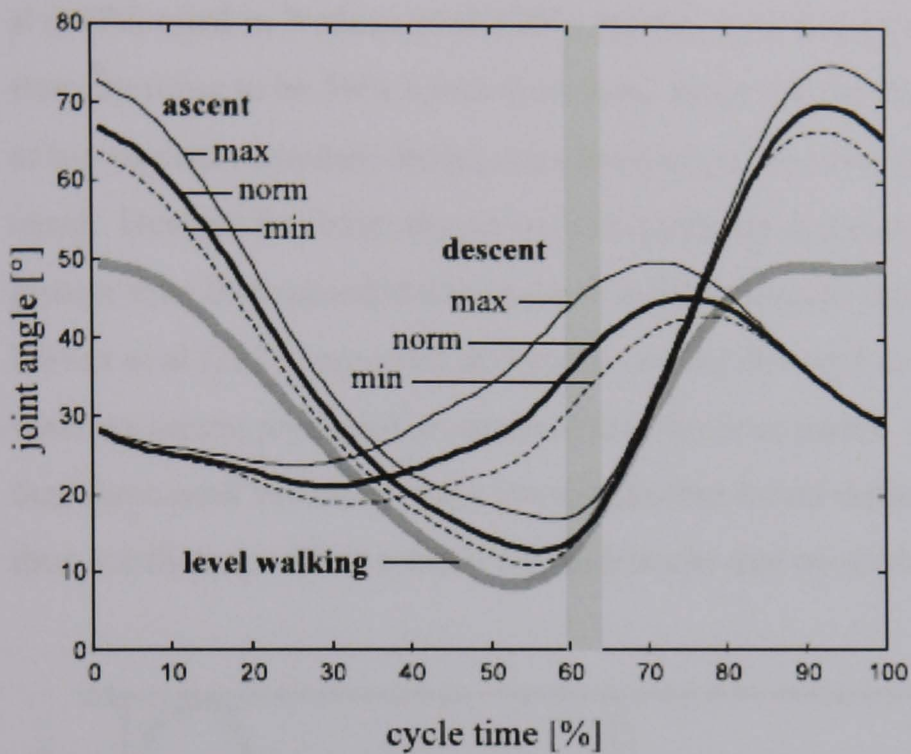
Contrary to these findings, Riener et al (2002) described conflicting results; a relatively greater stance phase for stair climbing compared to level walking. However, Nadeau et al (2003) speculated that this difference may have been due to inconsistencies in the step dimensions and stair design. The main variable featured in their study was the inclination of the slope (table 2.4) and the dimensions of their stair case designs were well documented. Yet at all inclinations, even the 'normal' 30° inclination which best compared with other studies, their percentage stance phase was consistently higher than that reported for level walking (63.6% stance for 30° slope compared to 61.1% stance for gait).

When directly comparing the modes of stair negotiation, there appear to be relatively greater duration step cycle for ascent compared to descent (Protopapadaki et al 2006, Riener et al 2002). The percentage proportion of swing and stance phase was statistically similar for ascent and descent, with stance accounting for 60.74% and 60.45% of ascent and descent respectively (Protopapadaki et al 2006), which was similar to the percentage stance phase reported graphically by Nadeau et al (2003) for ascent. Protopapadaki and colleagues did not however examine level walking, so phase comparisons to gait cannot be derived from their results.

2.7.2.2 Sagittal plane kinematics and kinetics

In the comparison between stair ascent and level walking, all studies agree that the greatest kinematic and kinetic differences for the sagittal plane were observed at the knee. Early studies showed that there was a greater magnitude of knee flexion and larger moments about the knee joint during stair ascent compared to level walking (Andriacchi et al 1980, Nadeau et al 2003).

However, greater sagittal plane ROM has been observed for all the lower limb joints during stair ascent compared to level walking. An increase in hip flexion angles of 15-20° have been observed during stair ascent (Andriacchi et al 1980, Livingston et al 1991, cited in Protopapadaki et al 2006) with the maximum hip flexion occurring during the swing phase (Protopapadaki et al 2006) for both ascent and descent. The maximum hip flexion angles have also been shown to increase with increasing inclination of the staircase (Riener et al 2002) for both modes of stair negotiation. Protopapadaki et al (2006) reported an interesting sagittal plane analysis of 'normal' stair ascent and descent for a young healthy population aged 18-39years. However, without an additional comparison to level walking, their analysis is largely descriptive, although they have drawn some biomechanical comparisons between the ascent and descent modes. They found that greater hip flexion was required during ascent with flexion values of 65° during stair ascent and 40° during descent. Although these results were not directly compared with level walking, Riener et al (2002) did compare all three modes of ambulation; level walking, stair ascent and descent. They found that descent required a smaller sagittal plane angular excursion than level walking and that stair ascent exhibited a larger excursion, due to the greater flexion angles (figure 2.22).



Includes comparison of hip angles on 3 different stair inclinations:
'norm'-Normal step inclination (30°)
'max'-Maximum step inclination (42°)
'min'-Minimum step inclination (24°)

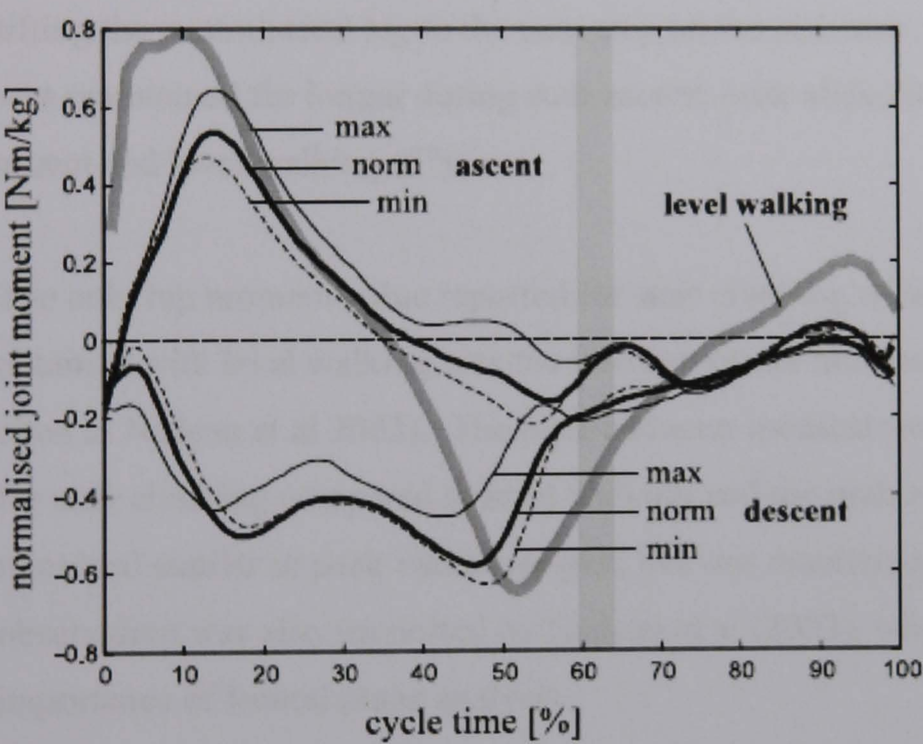
(Riener et al 2002)

Figure taken from Riener et al 2002

Figure 2.22: Comparison of hip flexion angles during walking, stair ascent and descent

The maximum flexion angles found by Protopapadaki et al (2006) for ascent and descent were similar to those Riener et al recorded for the minimum inclination stair set-up. Interestingly, when comparing the step dimensions, those used by Protopapadaki and colleagues were best matched to the normal inclination stair case inclination. Hence Riener et al (2002) found slightly greater sagittal hip angles overall. The discrepancies between studies may be due to methodological differences or a mismatch in the subject demographics. Riener et al used a narrower age band of slightly younger subjects (n=10; 24-34 years, mean 28.8 years) and they were all male, whereas Protopapadaki et al used a larger, mixed gender cohort (n=33; 18-39 years, mean 28.1 years).

Differences in external joint moment values between level walking and stair ascent have been reported. Nadeau et al (2003) found that although the external hip flexor moments were similar for walking and stair ascent, the flexor moment peaked for a longer duration during stair climbing. In contrast, the external extensor moments of the hip were significantly lower during stair ascent than those found during level walking, even with the emergence of a second peak during early swing. Kirkwood et al (1999, cited in Nadeau et al 2003) agreed, reporting peak extensor moments during stair climbing to be 59% lower than those observed for level walking. The analysis of hip sagittal moments during stair descent on the other hand, shows a conflicting trend. Descent has been shown to yield peak hip extensor moments up to 1.5 times greater than that recorded during level walking (Andriacchi et al 1980). Indeed, Riener et al (2002) agreed that descent largely featured external extensor moments whereas ascent produced an external hip flexor moment. However, they also found that these peak moments were lower than that found during walking (figure 2.23), thus conflicting with the study by Andriacchi and co-workers.



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'min'-Minimum step inclination (24°)

(Riener et al 2002)

Figure taken from Riener et al 2002

Figure 2.23: Comparison of external hip flexor moments during walking, stair ascent and descent

2.7.2.3 Frontal and transverse plane kinematics and kinetics

Although 3-D motion analysis was conducted in most of the studies of stair negotiation, the authors have predominantly reported sagittal plane outcomes. This gap within the literature was identified by Nadeau et al (2003) and they were one of the first to report comprehensive frontal plane analysis. Moreover, their study was one of the first normative stair analysis studies to use a middle aged population (41-70 years, mean 53 years) so features as the best comparison to the cohort of the current study. In addition, Kirkwood et al (1999, cited in Nadeau et al 2003) reported frontal plane and transverse plane kinetics. Although their study was less well matched for age similarities, their healthy group partially resembles that of the current study (55-75 years). Nevertheless, biomechanical results for frontal and transverse plane analysis during stair ascent and descent remain limited.

While walking is characterised by hip adduction during stance, Nadeau et al (2003) highlighted a relatively greater angle of abduction during the stance phase of stair ascent. This observation is likely due to the relatively greater activity in the hip abductors, for the purpose of tilting the contralateral pelvis upwards to assist in lifting the contralateral leg to the next step on the staircase. Although hip abduction was maintained for longer during stair ascent, peak abduction angles were similar for ascent and level walking (5°).

The only hip moment value reported for stair climbing which is higher than that obtained with level walking was the internal rotator moment (Kirkwood et al 1999, cited in Nadeau et al 2003). The peak adductor moment was significantly less (61%) for stair climbing compared to level walking and the peak abductor moment remained similar to peak values for gait, but was maintained for longer. This observation was also supported by Nadeau et al (2003), who emphasise the importance of frontal plane analysis.

In summary, most literature focuses on the differences between gait and stair ascent and comparisons to stair descent are few. Although the most significant biomechanical differences between these modes have been observed for the knee joint, there remain some important discriminating outcomes for the hip during stair negotiation. On the whole, relative to level gait, stair ascent shows greater sagittal hip angles and temporal differences in the 3-D moment profiles, whereas stair descent may show relatively greater peak moments but reduced hip angles.

Although the study by Nadeau et al (2003) was useful in providing normative data for frontal plane angles and moments, they only considered stair ascent in their comparison with walking. This gap within the literature was studied by Riener at al (2002); the only normative study known to have compared level walking and stair descent using angle and moment outcomes. It seems that stair descent in particular may be a valuable mode of ambulation to study in order to identify differences between resurfacing and stemmed arthroplasty. Although biomechanical differences were not identified between those with resurfacing and stemmed THR for level walking (Mont et al 2007), perhaps the assessment of stair negotiation, in particular stair descent, may highlight differences due to the greater possible biomechanical demand of the task.

2.7.3 Critique of stair climbing simulation

There is variation throughout the literature regarding the laboratory set-up used to allow simulation of stair ascent and descent for motion analysis. It is important to consider these variations in the stair design when evaluating results from other studies as it has been recognised that certain parameters may be influences by the staircase characteristics (Livingston et al 1991, cited in Nadeau et al 2003, Riener et al 2002).

Table 2.4: Literature review of staircase dimensions and characteristics

Step Characteristics	Andriacchi et al 1980	Nadeau et al 2003	Protopapadaki et al 2006	Riener et al 2002
Number of steps	3	4	4	5
Step height (mm)	210	170	180	*170 (138, 170, 225)
Tread depth (mm)	255	260	285	*290 (310, 290, 250)
Slope inclination (°)	38°	33°	-	*30° (24°, 30°, 42°)
Width of steps (mm)	580	1070	-	-
Extended walkway	No	2.44m from top step	Turn area	Turn area
Force platform	Ground and 1st step	Ground, 1st and 2nd steps	2nd step	Steps 1, 2 and 3
Handrails	Left side during ascent	No, only as safety barrier at rear of top platform	-	-

* Riener et al 2002-Height and tread values are given for a 30° inclination

Several studies have used a staircase set-up with 4 steps (Nadeau et al 2003, Protopapadaki et al 2006). Most studies report the various features of the staircase including step height or riser, tread depth or width, and occasionally the slope of the staircase. These are summarised in table 2.4. The requirement of a walkway leading up to the staircase and an extended walkway attached to the top step has also been recognised (Nadeau et al 2003) to allow natural changes in acceleration and turning patterns without being restricted by the environment.

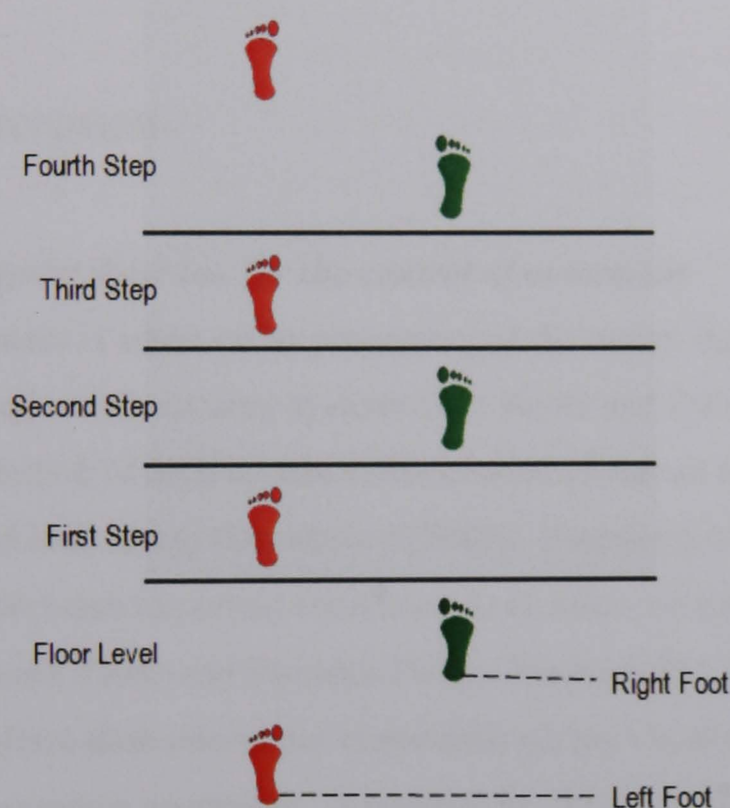


Figure 2.24: Reciprocal stair gait pattern on a four step stair case

It would appear logical that a minimum of 4 steps is required for simulating steady-state stair ascent and descent. Stepping performance on the floor level and the top step may not be considered as steady state stair climbing as these levels are associated with acceleration and deceleration of forward progression and do not impose the same restricted step area associated with other successive step levels. Considering reciprocal ambulation on a four step staircase, the first foot (e.g. left) moves from the floor (far proximity to the stairs) to the 1st step, then from the 1st to the 3rd step, and from the 3rd to meet the contralateral (right) foot on the 4th step. The other foot (e.g. right) leaves the position on the floor (near proximity to the step) and moves to the 2nd step, then moves from the 2nd to the 4th step (figure 2.24). Kinematic data of the whole step cycle must then represent limb movement from and to a step on the stair case. Nadeau et al (2003) collated their kinetic data from the second step to use for the comparison of walking as they recognised that steady state stair ascent had been achieved once the second step was reached. Although stair descent was not evaluated in the study by Nadeau et al (2003), the same principal would apply to the descent cycle, where stance on the second step best represents steady state descent.

2.8 Proprioception

2.8.1 Proprioceptive function for the control of movement

Control of movement is achieved by processing of the neural input from visual, vestibulospinal and somatosensory systems (van Hedel and Deitz 2004). The individual contribution of each system to the control of human movement may vary between tasks and individuals (Drouin et al 2003). Proprioception is a sensory modality which provides important contributions to adaptive voluntary and reflexive control of movement (Dietz and Duysens 2000). Functionally, the integration of proprioceptive information into motor commands occurs via neural pathways that are independent of conscious awareness. Nevertheless, the conscious perception of body position and movement will be dependent on sensory feedback from proprioceptors (Gandevia et al 1992 cited in Cromerford and Mottram 2001, Hall et al 1995).

Proprioception facilitates neuromuscular control (Blackburn et al 2000), including balance and postural stability (Benvenuti et al 1999), by regulating the appropriate timing and amplitudes of muscle activity (Allum et al 1998), coordinating activation of appropriate muscle synergies and accurate agonist/antagonist interactions, and governing the correct combination of muscle forces and duration of activity required for these purposes (Cromerford and Mottram 2001, Riemann and Lephart 2002). Proprioception compensates for deviations in stance and gait (Blackburn et al 2000) by providing quick and accurate feedback to the CNS, which responds accordingly by adapting motor control programs to assist in static and dynamic equilibrium (Riemann and Lephart 2002). Proprioceptive acuity is therefore essential for the control of motion during gait and other ambulatory tasks. Proprioception is known to deteriorate with advancing age and become dysfunctional in the presence of pain or secondary to trauma (Glencross and Thornton 1981, Hassan et al 2001, Koralewicz and Engh 2000). Pain causes alterations in the sensory neural processes which inhibits the function of somatosensory receptors.

Although no single receptor system works in isolation, the various receptors responsible for proprioception and postural control may be categorised:

- Load receptors, (golgi tendon organs (Duysens et al 2000) within muscles) required for feedback of kinetic information (Dietz and Duysens 2000);
- Mechanoreceptors (within tendons, joint capsules, ligaments and muscles: Pacinian corpuscles, Merkel's discs, Ruffini corpuscles and free nerve endings) required for end of range afferent feedback of kinematic information (Burke 2004, Hassan et al 2001, Shakoor and Moisio 2004) and respond to constant or slowly applied pressure, motion or rotation (Koralewicz and Engh 2000);
- Stretch receptors (within muscles: golgi tendon organs and muscle spindles) required for mid-ROM afferent feedback of kinematic information (Dietz and Duysens 2000, Hassan et al 2001) and respond to muscle lengthening.

Proprioceptive function of a joint is required for the control of joint loading (Shakoor and Moisio 2004). Altered biomechanics, and 'pathological' loading of a joint, can lead to the cartilage and subchondral bone degeneration associated with OA pathogenesis (Shakoor and Moisio 2004). OA results in articular deafferentation and resultant alteration in the kinaesthetic sense and position sense of the joint (Lephart et al 1998). Conversely, impaired proprioception is also believed to promote degeneration in OA joints (McNair and Heine 1999). Hence it is unclear whether proprioceptive deficits relate to the cause or effect of the osteoarthritis process.

Clinically, following THR there is disruption of some of the receptors thought to be responsible for joint proprioception, due to capsulotomy, weakness and pain inhibition of muscles, inequality of leg length, alteration of the abductor lever arm, restriction of the ROM and altered patterns of weight bearing (Nallegowda et al 2003). This implies that the removal or disruption of receptors from in and around the joint is likely to be detrimental to proprioception of the hip joint following arthroplasty. While it has been claimed that hip arthroplasty patients have "no hip joint proprioception" (McCrory et al 2001, page 108) other authors claim that patients with artificial joints have normal proprioception (Burke 2004). To investigate these claims, a review of the literature surrounding joint proprioception was conducted.

2.8.2 Proprioceptive function of a joint

The role of proprioception in coordinating movement and postural control of the body as a whole is a complex and multifactorial sensory neural processes. More specifically, studies have investigated the role of proprioception local to a joint, and have highlighted that the two main aspects of proprioceptive acuity concerned are the sense of joint position and the sense of joint motion (Jerosch and Prymka 1996, Pap et al 2000). Both somatosensations are closely related (Barrack et al 1984a, 1984b).

2.8.2.1 Joint position sense and motion detection sense

Joint position sense (JPS) is "the ability to evaluate subjectively the position of a limb in space" (Grigg et al 1973) by passively or actively measuring an individual's ability to reproduce a joint position or subjectively interpret the position of their limbs. This is frequently termed joint reposition sense. JPS may be divided into three methodological categories including interpretation of active JPS, passive JPS or visual analogue scale (VAS) scoring of a joint position. Active JPS involves active movement of the joint to reproduce a pre-selected angle, given either by active or passive joint positioning by the operator; whereas with passive JPS, the subject aims to stop the operator when the pre-selected angle is reached during passive movement of the joint. During VAS scoring, the subject gives an estimation of the angle at which the joint is positioned. All methods measure the accuracy of interpreting the position of the joint, which is compared to the actual position to give an error angle.

Table 2.5: Literature review of velocity of motion used in joint lower limb MDS studies

Study	Joint	Condition	Aims	Subject information	Methodology	Conclusions	Velocity of motion
Attfield et al (1996)	Knee	TKR	Examine effects of soft-tissue (ST) balance with TKR surgery	n=51, tested pre-op, 3 & 6 months post-op; n=12 control subjects; 25-40 years	Air cast splint; sitting; rig pulley system; start positions 10°, 25°, 40°	↑ MDS threshold with ↑ starting angle; improved proprioception following TKR	**0.5°/s** (No reason given)
Barrack et al (1984a)	Knee	Ballet dancers	Determine the immediate effect of training on MDS of the knee	As Barrack et al (1984b); control group did not warm up	As (1984b) with training and stretching before testing (also tested JPS)	Dancers had significantly reduced threshold MDS angles (better MDS), less accurate JPS	**0.5°/s** (No reason given)
Barrack et al (1984b)	Knee	Ballet dancers	Determine effect of training on MDS of the knee; dancers thought to have better MDS	n=12, 5 male, 7 female, average age 25 years; age matched control group	Air cast splint on foot and ankle; blindfolded; reclined sitting position; rig with starter motor, 10 reps	Dancers more consistent MDS and reduced threshold MDS angles; no gender difference; Training improves knee MDS	**0.4°/s** (No reason given)
Grigg et al (1973)	Hip	Unilateral THR	Assess effect of THR with capsulotomy on proprioception/ investigate mechanisms responsible	n=16, n=10 at 2 weeks post-op, n=6 at 4-8 months post-op; age unknown	Air cast splint full leg; rig with starter motor; start angles unknown	More precision in MDS on control limb, but difference not significant; no difference kinaesthesia due to THR	**0.6°/s** (Threshold velocity at which 50% of movements detected)
Hall et al (1995)	Knee	Hyper-mobility syndrome	Assess whether Hypermobility syndrome affects proprioception	n=10 hypermobile subjects, age 23-29 years; n=20 controls, age 21-40 years	Air cast splint; white noise; stepping motor rig; 5°/ 30° start angles; 2 directions	No difference with gender or with flexion/ extension; Greater detection in hypermobility group and at end ROM (5° trials)	**0.4°/s** (No reason given)
Karanjia and Ferguson (1983)	Hip	Unilateral THR	Assess effect of THR with capsulotomy on proprioception/ investigate mechanisms responsible	n=10; n=5 at 2-4 weeks post-op, n=5 up to 2 years post-op; age unknown	0.5° magnitude Flexion/ extension; full leg air-cast splint; manual/no rig; multiple start positions	Faster motion stimuli gives more consistent detection of motion; no difference in MDS due to THR	**0.6°/s & 2°/s** (No reason given Referred to Grigg et al (1973))
Lephart et al (1996)	Knee	Gymnasts	Compare kinaesthetic patterns in dominant knee of gymnasts and non-gymnasts	n=15 gymnasts; n=30 non-gymnasts, both age 17-23 years	Air cast splint; blindfolded, white noise; sitting, start angle 45°; motorised rig	Gymnasts lower threshold angles; training has enhanced proprioception	**0.5°/s** (No reason given)
Koralewicz and Engh (2000)	Knee	Unilateral OA knee	Determine difference in proprioception between OA and 'normal' knees	Severe OA group, n=117, mean age 68 years; age-matched normal group	Blindfolded, earphones; to identify motion direction; compare group/leg	Reduced proprioception of OA knee; 'normal' knee also affected by reduced proprioception	**0.5°/s** (to test knee capsule mechanoreceptors)
Pap et al (2000)	Knee	Unilateral TKA	TKA effect on proprioception due to removal articular surface	n=15; 4-6 years post-op; age 56-73 years	Air cast splint; white noise, dark glasses; stepping motor rig; Start angle 45°	Significantly higher threshold MDS angles on replaced knees compared to 'normal' knees	**0.6°/s** (No reason given)

Joint motion detection sense (MDS), or kinaesthesia is determined by the ability to detect passive motion of a joint. This is commonly quantified by measuring the threshold for detection of passive motion at a joint (Lephart et al 1998). The joint is moved passively at a standardised angular velocity and the subject acknowledges when they detect the onset of motion. The threshold angle is determined by quantifying the amount of motion occurring before subjective detection. A standardised velocity of motion is administered as MDS is known to vary with the velocity of the motion stimulus (Karanjia and Ferguson 1983). A wide range of angular velocities of motion have been used throughout the literature (table 2.5), from 0.4°/s to 2°/s, although a velocity of motion no more than 1°/s has been recommended (Koralewicz and Engh 2000). Slow motion stimuli are considered desirable for MDS tests as they minimise stretch receptor stimulation and allow subjective appreciation of limb position and limb movement (Lephart et al 1998).

Most studies assessing proprioceptive function of a joint have concentrated of the ankle, knee and shoulder joints, with fewer looking at the lumbar spine, metocarpalphalangeal joints, elbow and importantly, the hip joint. Of the lower limb studies, investigations of knee joint proprioception are most prevalent.

2.8.2.2 Proprioception studies of lower limb joints

Proprioceptive acuity of the knee joint has been assessed in highly trained gymnasts compared to relatively inactive control groups, using a passive knee motion stimulus (Lephart et al 1996, table 2.5). The gymnast group had a significantly lower threshold of MDS compared to the control group. The authors concluded that training enhances knee joint proprioception which is reflected by greater co-ordination and balance in the trained group. Proprioceptive training has been demonstrated to enhance agonist/antagonist inter-muscular communication (Gollhofer 2002), which has also been supported by earlier studies using a cohort of ballet dancers. Barrack et al (1984a, 1984b) found that the ballet dancers had reduced knee joint threshold MDS angles when compared to controls (n=12 per group). The same group were tested using a JPS protocol and found to have significantly poorer accuracy in knee joint reposition sense. These results appear to be conflicting, demonstrating better proprioception as determined by MDS tests and poorer proprioception as determined by JPS tests. The discrepancy of the outcome may have been due the exercise carried out by the dancers before testing (table 2.5) which potentially affected their muscle physiology. The JPS test may not have been sensitive enough to identify proprioceptive differences as the MDS test did. Although JPS and MDS are closely related, the test protocols may investigate slightly different aspects of proprioception (Benvenuti et al 1999) and challenge different neural mechanisms (Barrack et al 1984a) which may be a reason for the differences observed.

Injury of the knee which disrupts the anterior cruciate ligament is known to be detrimental to knee joint proprioception as demonstrated by both MDS and JPS testing (Jerosch and Prymka 1996). Joint laxity secondary to exercise muscle fatigue (Lattanzio and Petrella 1998) or hypermobility syndrome (Hall et al 1995, table 2.5) may result in impairment of knee joint proprioception.

Sharma (1999) reported reduced proprioception in osteoarthritic knees and a negative correlation between proprioception and severity of the OA. When assessing the association between proprioception and functional activity, a positive correlation has been noted between proprioception and stair walking time, but not with gait outcomes or postural stability tests (Sharma 1999). This is one of the few reports associating joint proprioception with functional outcomes.

The influence of knee joint replacement on proprioception has been studied (Attfield et al 1996, Lattanzio and Petrella 1998). Attfield et al (1996, table 2.5) found better proprioception in replaced knees that had been corrected for soft tissue balance, implying the importance of achieving the appropriate soft-tissue tension for optimising sensory neural mechanisms of the joint. Although a total knee replacement (TKR) involves disruption of many of the knee joint receptors, no significant difference in knee joint proprioception has been found between the operated and non-operated knees (Barrack et al 1983, cited in Sharma 1999, Barrett et al 1991, cited in Lattanzio and Petrella 1998). As proprioception was maintained following TKR, the authors concluded that the muscle spindles located in muscles local to the knee joint were the primary receptors responsible for knee joint proprioception as opposed to intra-articular joint receptors.

Ankle injury has been associated with reduced proprioception due to partial deafferentation of mechanoreceptors in the joint (Lephart et al 1998). The ankle joint has also been the focus of research relating joint proprioception to postural control, supporting theories of the ankle strategy. The ankle strategy is believed to play a primary role in the control of postural sway (Blackburn et al 2000). Nallegowda et al (2003) argued that the hip strategy is less efficient than the ankle strategy, while Allum et al (1998) showed that the receptors of the ankle joint may only shape the final output of the muscle synergy as opposed to triggering balance corrections. Triggers responsible for the lower limb control and balance corrections are instead believed to be initiated at a more proximal level, at the hip and trunk. Proprioceptive triggers mediated by receptors at the hip joint have been shown to initiate early in the postural response and are direction specific (Allum et al 1998). Hence, although a minority of studies have investigated hip joint proprioception, this may be an important function.

2.8.2.3 Proprioception studies of the hip joint

Beyond neurology studies and theoretical proposals relating to the hip joint, little research has looked at the hip joint proprioception. Nallegowda et al (2003) found that following THR, patients did not have a proprioceptive deficit but the test protocol in use featured functional activities, such as gait assessment and static and dynamic balance tests. MDS and JPS were not assessed in this study.

Mendelsohn et al (2004) looked at the effect of rehabilitation on hip and knee joint proprioception following hip fracture. Active reproduction of three passively positioned hip flexion angles (15°, 30° and 60°) was tested and error angles compared between the fracture limb and the control limb. There was no significant difference between error values of each limb and no correlation was made between results for JPS and functional measures, including the timed get up and go test, Berg balance score and gait speed. Although the subject numbers appeared sufficient (n=30), the protocol for measuring JPS was flawed and poorly controlled. Hip flexion was coupled with knee flexion, resulting in multiple joint movements. Muscle activity was against gravity, introducing an additional subject variable of muscle strength. Differences in muscle strength between subjects and fatigue were not identified as a variable factor to be controlled. Moreover, at no point did the author reveal the classification of the hip fracture or the method of management of the fracture.

The influence of type of fracture management on hip joint proprioception has been assessed by Ishii et al (1999, 2000). The initial study compared subjects having hemiarthroplasty and THR with age-matched control subjects, and showed no significant difference in proprioceptive function between the surgical groups or the controls. The test protocol involved active reproduction of active flexion and abduction hip angles, performed in a standing position and error values were compared between groups. The same protocol was used to compare groups with hemiarthroplasty and femoral nailing to a control group (Ishii et al 2000). Although there was a trend towards poorer joint angle reproduction in the hemiarthroplasty group compared to the femoral nailing group, the results were not significant. The authors concluded that maintenance of the femoral head does not benefit hip joint proprioception in the elderly hip fracture group and capsulotomy and femoral head replacement was not detrimental to active JPS. Some fixation of the lumbar spine was provided as a by-product of wearing the hip potentiometer brace but the knee was not fixed and sensory cues, such as visual and auditory input or sensory feedback provided by foot contact on the floor, were not controlled. Twelve individuals participated in each group but all were at a different post-operative phase, which resulted in dissimilarity in groups before testing. Overall, the poor attention to methodological considerations may have reduced the strength of the conclusions.

Grigg and colleagues (1973, table 2.5) considered the influence of unilateral THR (Müller prosthesis) on proprioception by using various proprioceptive test methods, including passive and active JPS testing with VAS scoring and passive threshold MDS testing. The VAS scoring used arbitrary units rather than assigning units of degrees of motion. All movement was performed using a mechanical rig and hip abduction motion, comparing operated and non-operated limbs. Two cohorts were tested; ten subjects at two weeks post-surgery and six subjects at between four and eight months post-op, but not all were tested using all three protocols.

For the MDS trials (table 2.5), the lower limb was passively and mechanically abducted (constant stimuli) at an angular velocity of 0.6°/s and the subject pressed a cut-off switch when the onset of motion was detection. For the majority subjects in the early post-op group (n=9 of 10), there was an increase in the threshold of motion detection on the operated limb, but this result was only statistically significant for 3 subjects. Of the longer term post-operative subjects (n=6), five had a significantly greater threshold of detection on their operated hip, but the average difference in threshold detection between limbs was less compared to the short-term group. This might suggest that motion detection sense improves over time following THR. Results were presented for the 10 short-term follow-up subjects and these showed mean detection angles of 1.22° (range 0.48-2.78°) for the operated hip compared to 0.56° (range 0.13-1.32°) for the non-operated hip. On the whole, the results show that there is no significant deficit in MDS following THR.

The JPS protocol used by Grigg et al (1973) involved passive movement of the limb at a velocity of 2.0°/s and subjective estimation of the magnitude of abduction motion. There was a tendency for THR subjects to overestimate the angular displacement, however, when comparing the passive and active position sense of the operated and control limb, no significant difference was observed for short and long-term post-operative subjects.

The apparatus used by Grigg et al (1973) was the first of its kind, and similar rigs have been constructed to investigate knee joint MDS since then (Hall et al 1995, Barrack et al 1984). However, Grigg et al (1973) failed to limit potential errors by neglecting to fix the pelvis, limit visual and auditory stimulus, and accurately estimate and align the HJC with the fulcrum of the lever arm supporting the lower limb. Restriction of visual and acoustic feedback allows individuals to rely more on proprioceptive feedback for the control of movement (van Hedel and Dietz 2004) and detection of movement. Vibration of the starter motor may have provided extra sensory cues, indicating to onset of motion of the lever arm during the MDS trials.

Karanjia and Ferguson (1983, table 2.5) concluded that passive motion detection threshold is minimally affected by THR. They assessed 10 patients with unilateral Anfranc-Turner THR (n=5 at 2-4 weeks post-surgery, n=5 within 2 years of surgery). The limb was manually (passively) moved by the operator, providing repeated (n=20 repetitions) 0.5° increments of flexion or extension from three different starting positions. Subjects indicated when the felt motion and what direction their limb moved in. The authors concluded that there was minimal deficit in MDS following THR, with no significant difference in detection of motion on the operated and control limbs. Two test velocities were used (0.6°/s and 2°/s) and the faster motion velocity induced a greater number of correct detections. There was no difference in the sub-groups due to post-operative phase and the initial position of the limb was thought to be 'unimportant' (page 656). The sample group used in this study was small and the post-operative phase varied widely. The MDS method used was also more subjective than that of Grigg et al (1973) with qualitative documentation of the ability to detect motion or not as opposed to recording quantitative threshold angles.

To summarise, joint proprioception may be influenced by a number of variables including physical training, injury or trauma, fatigue, joint laxity and soft tissue tension and OA. Research investigating proprioception and joint arthroplasty remains inconclusive but suggests that joint proprioception is not deficient following arthroplasty. Research relating proprioception to functional activity is scarce and it is not known whether MDS or JPS measures correlate with balance or motion analysis outcomes. Currently there does not appear to be a suitable measurement tool for quantifying joint proprioception of the hip and literature supporting any one measurement method is also inconclusive.

2.8.3 *Measuring proprioception*

Literature surrounding the measurement of proprioception shows many variations in the protocols used (Drouin et al 2003) and it is well recognised that no single tool can accurately measure all facets of proprioception, such as aspects of postural control mechanisms or balance (Benvenuti et al 1999). Functional tests such as single leg stand (Blackburn et al 2000) have been used to test proprioception and experiments assessing postural sway have found that sway is inversely proportional to strength and proprioception (Hassan et al 2001), where greater postural sway correlates with an overall derangement in postural control mechanisms (Benvenuti et al 1999). Nevertheless, balance tests do not focus on the relevant aspects of proprioception or concentrate on proprioception of an individual joint.

Patients with THR and hip osteoarthritis often perceive an awareness of leg length inequality (Burke 2004), possibly secondary to pain or altered proprioceptive awareness. Anecdotally, surgeons have also remarked that patients often feel heaviness or increased physical effort in their hip replacement limb following THR. It is possible that these sensations are related to altered proprioceptive feedback following hip joint reconstruction. Moreover, in the clinical environment, it is often informally reported that patients with resurfacing arthroplasty report less sensory disturbance post-operatively and commonly attribute a 'natural feel' to their new hip joint (Loughead et al 2005). If proprioceptive differences are responsible for the sensory alterations reported, the relatively favourable reports following hip resurfacing imply that proprioception may be better preserved following resurfacing arthroplasty. It may, therefore be appropriate to determine whether there are differences in proprioception following resurfacing and standard THR by using a satisfactory method of measuring proprioception of the hip joint.

Grob et al (2002) support the opinion that MDS and JPS tests describe different facets of functional proprioception. JPS tests have been shown to be more accurate in weight-bearing conditions (Lattanzio and Petrella 1998), which would incorporate the load receptors of the joint and joints demonstrate superior proprioception when weight-bearing, as demonstrated by reduced errors in reposition sense (Drouin et al

2003). However, in order to evaluate reposition sense in weight-bearing conditions, other variables become involved such as balance, muscle strength and joint stability, so the operator cannot be sure of isolating JPS or attributing the results to this aspect of proprioception alone. Ishii et al (1999, 2000), who favoured active JPS testing, claim that passive techniques are unreliable and are associated with high tolerance intervals. However, given that passive techniques involve a more selective stimulation of receptors (Lephart et al 1998) passive tests seem more suitable in terms of validity and control of variable factors.

Kinaesthesia (sensation of joint movement) is a specialised form of tactile sensation (Blackburn et al 2000), tested by threshold MDS testing. MDS tests have been found to be more reliable and consistent in their results, with better sensitivity (Gorb et al 2002) and provide "a more objective measure of proprioception than (joint position) reproduction on a VAS does" (Koralewicz and Engh 2000, page 1582). MDS involves a passive change in muscle length, believed to be detected by muscle spindles (Barrack et al 1984) or mechanoreceptors (Koralewicz and Engh 2000). The receptors responsible may depend on the joint range tested, and end ROM tests have shown greater kinaesthetic sensitivity with reduced threshold MDS angles (Hall et al 1995, table 2.5). MDS protocols are classified as dynamic tests (Jerosch and Prymka 1996) and may correlate better with dynamic functional outcomes than JPS tests. Kinaesthesia is also responsible for the sensation of force, effort of workload, and heaviness perceived during a motor task (Cromerford and Mottram 2001). Relating to subjective reports from hip arthroplasty subjects, a kinaesthetic test may seem appropriate.

Pap et al (2000) recommended that MDS tests should also record the failure to detect motion and, in contrast to the comments of Karanjia and Ferguson (1983), the appreciation of the direction of the motion is also regarded with importance (Hall et al 1995, Koralewicz and Engh 2000). As mentioned, the velocity of the motion stimulus is a key variable in measuring the outcome of threshold MDS. The range of methodologies applying to lower limb MDS tests (summarised in table 2.5) demonstrate the variables requiring control during kinaesthetic joint assessment.

Considering the aspects of hip joint proprioception highlighted in published literature, it appears that a controlled kinaesthetic MDS test may be the most suitable method of quantifying hip joint proprioception and comparing hip proprioception following resurfacing and standard THR. Further consideration of the aspects involved in designing such a protocol and overcoming the shortcomings of previous methods will be considered in chapter 3.

2.9 Study Aims and Hypotheses

There is significant information to support the expectation that there may be function and activity related differences discriminating between individuals with resurfacing arthroplasty and those with standard stemmed THR. Given the growing popularity of hip resurfacing, yet the lack of objective data supporting its use, the need for a better understanding of the key differences between resurfacing and standard THR has been identified (Loughead et al 2005).

The aims of the current research were to:

1. Identify whether differences exist in the kinematics of the hip joint of individuals having had a resurfacing arthroplasty compared to those with a standard stemmed THR.
2. Identify whether differences exist in the kinetics of the hip joint of individuals having had a resurfacing arthroplasty compared to those with a standard stemmed THR.
3. Identify whether proprioceptive function, as determined by hip joint motion detection sense, would be impaired following hip arthroplasty and determine whether differences exist in the proprioceptive hip function of individuals having had a resurfacing arthroplasty compared to those with a standard stemmed THR.

The potential functional differences may be related to differences in the structural design of the implants and the resultant anatomical and potential biomechanical differences resulting from their implantation. Variation in the surgical procedures, in terms of the extent of the soft-tissue involvement and the resection of bone may also induce differences in kinematic and kinetic outcomes, which may be short-term impairments that could improve with soft-tissue healing. Balanced consideration of the literature surrounding resurfacing and standard THR and the theoretical influence of anatomical differences, it seems indicated to predict that hip resurfacing has several merits, potentially surpassing the functional provision of the standard stemmed arthroplasty. These include the potential for greater functional ROM resulting from the relatively larger diameter femoral head bearing and improved abductor muscle function from the more anatomic offset resulting from femoral neck preservation associated with hip resurfacing. The relatively conservative reports of sensory abnormalities following hip resurfacing may suggest that resurfacing arthroplasty results in better restoration of proprioception or greater preservation of receptors governing proprioception of the hip joint.

Therefore, the following expected outcomes may be presented:

1. Individuals with hip resurfacing have greater hip motion during functional activities than those with standard THR, as determined by larger hip angles during walking and stair climbing. Resurfacing hips show fewer differences in hip angle values compared to the non-operated hip than standard THR hips.
2. Individuals with hip resurfacing demonstrate superior biomechanics during functional activities than those with standard THR, as determined by greater external intersegmental hip moments during walking and stair climbing. Resurfacing hips show fewer differences in hip moment values compared to the non-operated hip than standard THR hips.
3. Proprioception is impaired following hip arthroplasty but improves as post-operative time increases. Those with hip resurfacing have relatively less impairment of proprioception compared to those with standard THR.

The following chapters will outline the tools and methods used in order to test these hypotheses.

3 Design of Motion Detection Sense Protocol

3.1 Design Aims

3.1.1 Purpose

A motion detection sense (MDS) protocol was designed to test kinaesthesia of the hip joint, in order to objectively quantify the ability of an individual to detect passive motion of their hip. The protocol was therefore joint specific. Integral to conducting the protocol was the use of a validated test rig, to facilitate mechanically movement of the lower limb, and the execution of a controlled test procedure allowing measurement of the threshold angle of motion required for subjective detection of the movement stimulus. The protocol was intended to allow the measurement of kinaesthetic sense in two orthogonal planes of motion, to qualify the direction specificity of hip joint kinaesthesia.

Ultimately, the MDS test protocol was designed to allow the effect of hip arthroplasty on hip joint proprioception to be determined. In addition, potential differences in proprioceptive hip function between individuals with resurfacing arthroplasty and standard stemmed THR was to be assessed.

3.1.2 Previous designs

Similar protocols have been used in previous studies. The most recent of these studies, by Karanjia and Fergusson (1983), omitted the use of a mechanical test rig within their protocol. Passive limb motion was achieved by manually moving the limb. While the subject maintained a side lying position, the operator supported the test limb, encased within an air-cast splint, then flexed or extended the hip joint from the initial limb position to provide sagittal plane motion. Small increments of movement (0.5°) were provided at 0.6°/s and 2.0°/s. A linear potentiometer was attached to a mechanical arm and strapped over their lateral thigh of the subject, in order to quantify angular motion, which was monitored using feedback from an oscilloscope. This allowed the operator to control the velocity of the motion. Three test conditions were used. From initial positions of 15° or 45° of flexion, the limb was moved into a greater angle of flexion and from 15° extension, the limb was moved into a greater angle of extension (figure 3.1). The subjects indicated whether or not they felt the movement and communicated the perceived movement direction.

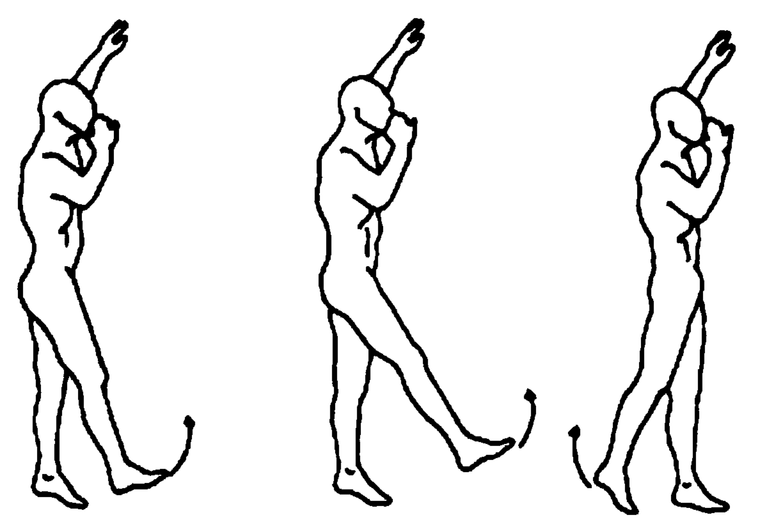


Figure 3.1: MDS methodology by Karanjia and Ferguson (1983)

This method of moving the limb had several limitations. Firstly, regardless of the level of training and practice of the operator, the velocity of the limb movement and smoothness of movement were likely to be subject to inconsistencies. Variation in the quality of the motion stimulus may have occurred during the limb excursion of individual trials or over repetitions of trials within or between subjects, potentially compromising the reliability of the data obtained. Potential errors from the operator reading the analogue output of the oscilloscope may have affected the velocity of motion. Secondly, the subject was positioned in side lying without any support or fixation method. Rotation of the trunk or pelvis may have caused a semi prone or supine position to be adopted. In this situation, the motion may have featured a combination of motion in multiple orthogonal planes as opposed to motion limited to the sagittal plane. Consequently, poor control of the subject's position during the procedure may have generated errors in the measurement of the threshold MDS angles.

The study conducted by Grigg et al (1973) used a mechanical rig to generate motion of the subject's test limb. This was done by strapping the subject's lower limb, encased in an air-cast splint, onto the mechanical arm of the test rig. The HJC of the subject was aligned with the pivot point of the mechanical arm. Subjects were positioned in semi-supine lying and the test limb was abducted. Angular motion of 0.6°/s was generated by a starter motor and hip angles were measured using a potentiometer, positioned at the pivot point of the rig. The initial limb position was not defined by the authors. Once the motion commenced, the subject marked the instance of motion detection by pressing a cut-off switch and the angle at the point of detection of the movement was recorded as the threshold MDS angle.

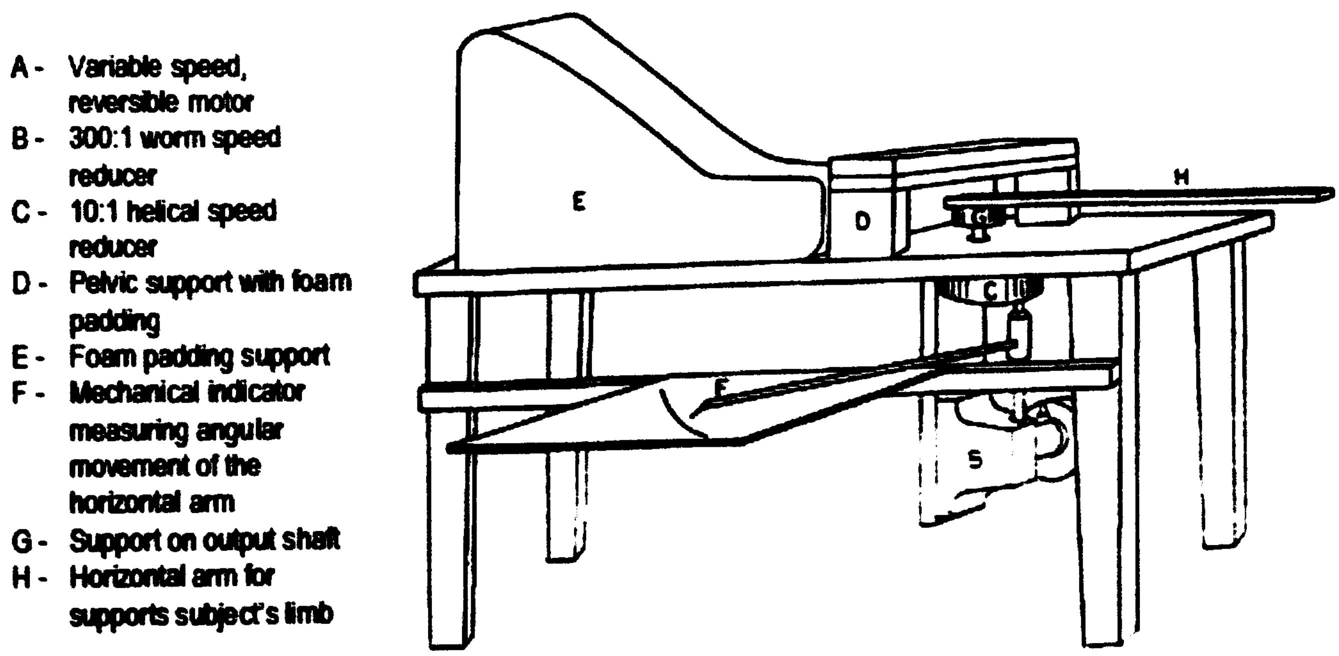


Figure taken from Grigg et al 1973

Figure 3.2: "Mechanical Stimulator" test rig by Grigg et al (1973)

Although the use of a mechanical rig improved the reliability of this study, there were certain features of the protocol that may have generated inaccuracies in the results, by either introducing errors into the measurement of threshold angles or adversely affecting the subject's ability to detect the movement of their limb.

Accurate alignment of the HJC with the pivot point of the rig would have been essential for accurate measurement of the threshold MDS angles. Malalignment would have resulted in under- or over-estimation of these angles, as measured from the analogue mechanical indicator. The exact method of aligning the HJC with the pivot point of the rig was not documented by the authors. In addition, although the knee was splinted and immobilised, Grigg and colleagues failed to immobilise the pelvis. It was possible that the position of the pelvis may have been disrupted due to movement of the subject's upper body or free lower limb (non-test limb). This would have resulted in malalignment or further malalignment of the hip joint centre with the pivot point of the rig. The authors did mention, however, that "the position of each patient was monitored visually and visible changes in position were corrected as observed" (Grigg et al 1973, page 1018). Nevertheless, this remains a poorly controlled feature of the protocol.

Further errors may have arisen from the method of reading the start angles and detection angles from the mechanical indicator, particularly as the threshold detection angles were small in value, yet measured and recorded to two decimal points of a degree. Although the starter motor controlled the velocity of limb movement, the vibration of the motor may have provided the subjects with additional sensory input. Sensing vibration or audible noise on initiation of the starter motor may have pre-empted subjects to the start of the limb motion by providing preparatory cues as to when the motion was about to start. On the other hand, some subjects may have been distracted by the noise or vibration and their ability to detect the motion may have been affected as a result.

In contrast to Karanjia and Ferguson (1983), Grigg et al (1973) used motion in the coronal plane to provide the test stimulus, as opposed to motion in the sagittal plane, and tested one direction of motion (abduction and not adduction). Karanjia and Ferguson found no significant difference in MDS between flexion or extension motion trials. Overall, the differences in the protocols do make the results of these studies difficult to compare (section 2.8.2). Comparison of the results is further limited by the use of different forms of motion stimulus, featuring small increments of motion (0.5°) by Karanjia and Ferguson (1983) and continuous motion stimuli by Grigg et al (1973). The use of continuous motion stimuli allows objective threshold angles to be measured as opposed to binary qualification of 'detection' or 'failure to detect'. No published study, to date, has examined threshold MDS during two different planes of motion, using the same protocol and method of motion stimuli.

3.1.3 Design requirements

Following an appraisal of the existing MDS protocols and their test apparatus, the shortcomings of the existing designs were considered. The physiology surrounding the mechanisms of MDS were also considered before designing the protocol and test rig for the current study. The resulting design specifications will be discussed in the sections to follow, and are summarised in table 3.1a.

3.1.3.1 Neurological and physiological considerations

When testing an individual's ability to detect hip motion, a kinaesthetic sensory input must be given. However, the theoretical method of providing this sensory input without providing other forms of sensory input is impossible. The aim was therefore to provide the sensory stimulation while limiting additional sensory input as much as possible. With the predominant sensation of motion of the lower limb, the operator could be sure that the subject detected a change in the position of their limb as opposed to the following:

- change in cutaneous sensation
- audible cues suggestive of change in limb position
- visual cues suggestive of change in limb position

Additional cutaneous input would be limited by ensuring the skin of the moving limb was not subjected to a change in light or crude touch, stretching of the skin or vibrations. When positioning the subject, consideration was to be made to their overall comfort and areas of pressure sensation originating from contact of the apparatus supporting the test limb. Most importantly, during the movement of the limb, effort was to be made to avoid brushing of the skin against the supporting surfaces or surrounding apparatus, movement of loose clothing, stretching of the skin or excessive or concentrated areas of uneven pressure at the skin generated from the motion stimulus.

Audible cues could be eliminated by creating a silent test protocol. This would therefore require another mechanism to passively move the test limb rather than a starter motor. Eliminating visual cues could be done by blindfolding the subjects. Karanjia and Ferguson (1983) asked their subjects to keep their eyes closed but blindfolding was regarded as more reliable.

Repeatable positioning of the patient was of great importance. Standardised positions would be planned in order to avoid placing physiological structures on stretch or promoting joint positions at the end range of joint motion. As there may be variability in the soft-tissue viability of the hip arthroplasty subject group, limiting between-subject variability in positioning may be better achieved by providing the motion stimulus at mid-ROM positions of the hip joint.

During their study, Karanjia and Ferguson (1983) used a test condition which appears to be unsound in terms of physiological and neurological aspects. The extension test condition (figure 3.1) used an initial limb position of 15° of extension. Normal physiological hip extension in adults is limited to 12.8 ±5.4° (Boone and Azen, 1979). The initial limb position therefore seems unfeasible, particularly as individuals with hip arthroplasty often characteristically have hip flexion contractures on their operated limbs. It must be assumed that the initial limb position and consequent extension motion possibly combined hip extension, anterior pelvic tilt and lumbar extension, which

Table 3.1a: Design specifications for MDS apparatus

Design Specification	Feature meeting specification	Justification for use
Minimising additional sensory input	Weighted pulley system to provide passive motion stimuli	Minimise vibration and noise stimuli
	Flexible straps instead of rigid struts for connecting the limb to the motion beam	Avoid changes in light or crude touch, brushing or stretching of the skin originating from contact with the supporting apparatus during the motion
	Subjects must wear lycra shorts	To avoid additional cutaneous sensory stimuli from movement of loose clothing
	Subjects wore eye masks	To eliminate visual stimuli during testing
Ensuring motion stimuli was isolated to the hip joint	Subjects wore a PROM brace	Used to immobilise the knee joint to ensure motion was only produced at the hip joint
	Pelvis clamps attached to rig	To immobilise pelvis
Mechanism to support and move limb	Horizontal beam, fixed to a frame on the plinth	Stable structure from which to support the test limb
	Horizontal beam with pivot axis at proximal end and wheels at distal end	Allow horizontal movement of the beam in 2 planes
	Horizontal beam with pivot axis which can be moved along the width of the plinth	Moveable pivot axis to allow the pivot axis of the beam to be aligned with the hip joint centre for simultaneous beam and limb movement and minimal adverse cutaneous stimuli
Mechanism to control the velocity of limb movement	Viscous damper in series with the pulley system	Settings to be altered to account for the variation in limb mass between subjects to allow a consistent velocity of 0.4°/s
Mechanism to measure limb movement	Electrogoniometer	Measure angular displacement of the hip joint during limb movement
Large enough dimensions for subjects to be positioned appropriately	Rig dimensions: Plinth W 860mm, L 2200mm; Bridges H 529mm, W 846mm; Beam L 800mm	To allow for variation in subjects' anthropometry and be suitable for individuals of varying heights and body proportions while supine and side-lying
Easy to mount and dismount	Proximal bridge could be opened to allow subject access	Subject safety and comfort, to avoid 'climbing into' the rig
Easy for operator to use rig	Plinth at user friendly height (840mm)	Meeting manual handling requirements of the operator

implies that this MDS condition was not conducted on a joint specific basis. Moreover, extension from this position would certainly be classed as end range motion, in contrast to the mid-range stimulus of the other test conditions (flexion from 15° or 45°). Therefore the detection mechanism or receptors tested may have differed between test conditions. The validity of comparing the direction specificity of MDS by this protocol was therefore questionable. Moreover, air-cast splinting of the limb would have fixed the knee joint in full extension, which may have had negative effects in the protocol used by Karanjia and Ferguson (1983). It is likely that adverse stretch may have been placed on the hamstring muscle group during both of the flexion motion conditions, particularly flexion from 45°. Threshold sensation of movement during these trials may have been triggered early or with greater ease due to heightened activation of the stretch receptors within the hamstring muscles. The subject group tested may have had varying degrees of flexibility of their hamstrings, so motion detection may have been a function of the flexibility of the group. With application to the current protocol design, it was felt that the flexion MDS trials should be carried out with the knee in a small amount of flexion to offload the hamstrings from full stretch.

3.1.3.2 Mechanical requirements

The test rig was to be designed as a stable structure with a horizontal mechanism suitable for passively supporting and moving right or left lower limbs. The supporting structure needed to be adaptable, allowing limb support while the subject lay in both supine and side lying positions, for testing MDS during abduction and flexion motions respectively. The horizontal support arm had to be able to move the limb to generate a motion stimulus, rotating at a pivot axis which aligned with the HJC, ensuring that the lower limb and the support arm moved in synchrony and therefore limited skin stretch and abnormal pressure on the limb. Given the potential variation in anthropometry between prospective subjects, the rig had to be suitable for individuals of varying heights and body proportions. Therefore, the pivot axis of the moving support arm had to be adjustable in order to ensure it would align with the subject's HJC regardless of body proportion. Similarly, the mechanism for attaching the subject's limb to the support arm needed to be adjustable to suit all limb circumferences and lengths.

As the limb moved, the movement had to be of a smooth quality: controlled and without vibration to avoid additional sensory cues. Therefore, the pivot point bearing had to produce continuous movement with minimal friction and the mechanism generating motion of the leg was to be smooth and silent. The mobile support arm also had to allow adequate range of motion to allow the movement stimulus to continue long enough for the subjects to sense the motion and indicate their detection of the motion, thus following the continuous motion stimulus method used by Grigg et al (1973).

3.1.3.3 Positional aspects-Freedom and restraints

The test rig was to provide adequate surface area for all subjects to lie in a comfortable position. Supine positioning of the subject was to be used while testing MDS during hip abduction. While supine, the hip was to adopt an anatomically neutral position (in the coronal and sagittal planes) from which it would be abducted. This was consistent with a mid-range of abduction-adduction ROM, which was believed to be used in the protocol by Grigg et al (1973). The rig was to be suitable to restrain the starting position and the movement of surrounding joints, but permit sufficient freedom of hip abduction motion while running the trials to allow detection of the movement.

A side lying position was to be adopted by the subject while testing MDS during hip flexion. The horizontal arm would support the limb so it was anatomically neutral in the coronal plane (0° abduction) with 30° of flexion in the sagittal plane. From this position, the hip would be flexed. This was consistent with mid-range of sagittal hip ROM and featured as the median of the two initial flexion angles used by Karanjia and Ferguson (1983). The rig was to be suitable to restrain this starting position and the movement of surrounding joints, but allow sufficient freedom of flexion motion of the hip while running the trials.

Consideration was made as to whether to control the axial rotation angles of the hip (transverse plane). As most hip arthroplasty subjects have limitation of rotation range of motion, it was decided that rotation would not be pre-determined and restrained or controlled between subjects. It was felt that if the subject had their limb positioned in adverse angles of rotation, they may experience increased sensitivity of motion which would result in early detection of motion and reduced threshold angles. This may occur secondary to firing of stretch receptors or mechanoreceptors. Therefore standardising the rotation of the limb may result in a type II error and could be avoided by positioning hip joint rotation as the subject is comfortable.

In both supine and side lying positions, the test limb would be restrained at the appropriate starting angle of the hip. An additional requirement of the rig was to ensure when the MDS trials took place, that all the movement was occurring at the hip joint alone, as opposed to the surrounding joints. Therefore, it was required that the joints above and below the hip joint – the lumbar spine and the knee joint – were immobilised. Movement of the knee or spine may alter the position of the hip joint centre, disrupting the quality of the limb movement. Freedom of these joints may also lead to additional movement and hence false detection of hip movement. The rig therefore had to have attachments to brace the pelvis and the knee joint required a splint to immobilise it.

When testing MDS during flexion trials, effort had to be made to avoid overstretching of the hamstring muscles which may result in early detection due to the activation of stretch receptors or mechanoreceptors (section 3.1.3.1). Therefore the knee was to be maintained in 20° of flexion to offload the hamstrings as they cross the knee joint. Hence, the bracing mechanism required for the knee joint had to be flexible; allowing bracing at 0° knee flexion during the abduction MDS trials, and bracing at 20° knee flexion during the flexion MDS trials. An air cast splint would therefore be inappropriate as it would fully extend the knee joint.

3.1.3.4 Comfort and safety

Comfort of the subjects was of great importance, not only for maintaining ethical standards but also to ensure that they did not become distracted from their role of detecting the limb movement.

The test rig had to be designed with consideration to safety aspects, particularly as the hip replacement subject group may not be as agile as those without physical restrictions. Therefore the rig would need to be easy to mount and dismount. While on the rig, any sharp or rigid edges would be covered with padding.

3.1.3.5 User requirements

Setting up and running the protocol needed to be efficient. The rig was therefore required to be user friendly, efficient and easy to operate. All inter-connections and accessories had to be easy to assemble. Positioning the subject in the rig had to be comfortable for the operator and in line with current manual handling guidelines. The rig therefore had to be level with the operator's waist height.

The operator also needed a mechanism of controlling and monitoring the velocity of the limb motion, allowing the velocity to be set and controlled for each subject.

3.2 Design Features

The test rig was designed and constructed in accordance with the requirements outlined, following consideration of neurological and physiological aspects, mechanical requirements, positioning, comfort and safety of the subjects and the needs of the user.

3.2.1 Basic structure

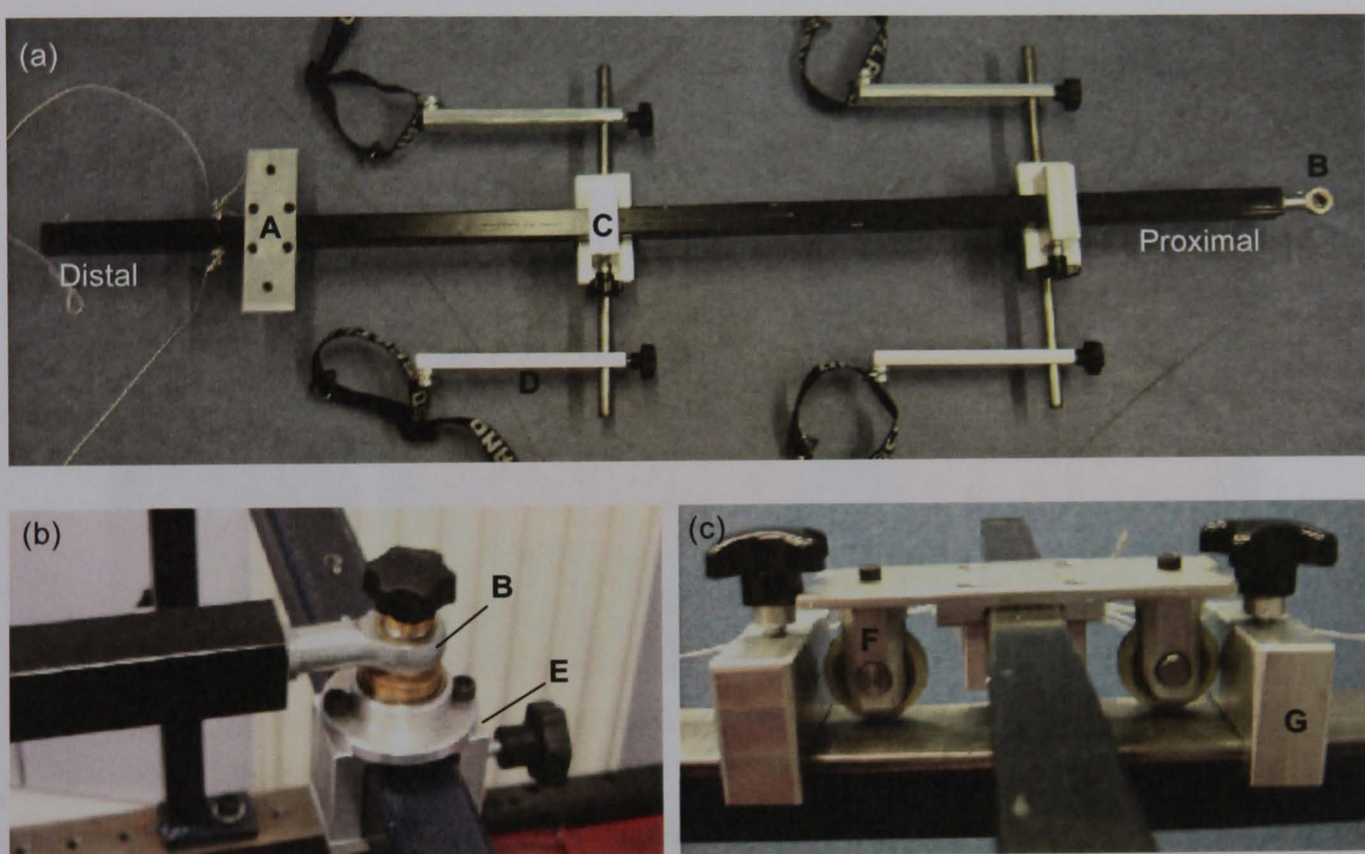
The test rig apparatus was constructed upon a steel framed plinth (H 840mm, W 860mm, L 2200mm). The plinth was mobile but could be fixed in a stationary location. It featured two steel lateral borders which were continuous with the distal half of the plinth and featured a series of bolt holes 100m apart. Two metal bridge structures were bolted onto the lateral rims of the base frame of the plinth, one distally at the foot of the plinth and another proximally, fixed midway relative to the length of the plinth (figure 3.3). The surface onto which the subjects were positioned was lined with foam padding and covered with a wipe-able material.

Figure 3.3: Basic structure of the MDS rig, opened for subject access

The proximal bridge structure was hinged 340mm above its fixture, allowing the structure to be released from the contralateral bolt fixation and opened to allow access on and off of the plinth (figure 3.3). A small set of steps were used to allow subjects to climb up to the level of the plinth without risk to themselves. As the lateral rim of the plinth was steel, this was covered by a padded fitting as the subjects climbed on or off of the rig. When on the rig, the position of the subject's waist would correspond with the level of the proximal bridge structure.

3.2.2 Suspension of the limb

In order to suspend the test limb, a beam (figure 3.4a) was attached between the proximal and distal bridges. This was connected to the proximal bridge using a low friction ball bearing which was mounted on a sliding fixture (figure 3.4b), allowing the pivot point of the beam to side along the horizontal support and align with the HJC of the subject beneath (figure 3.6).



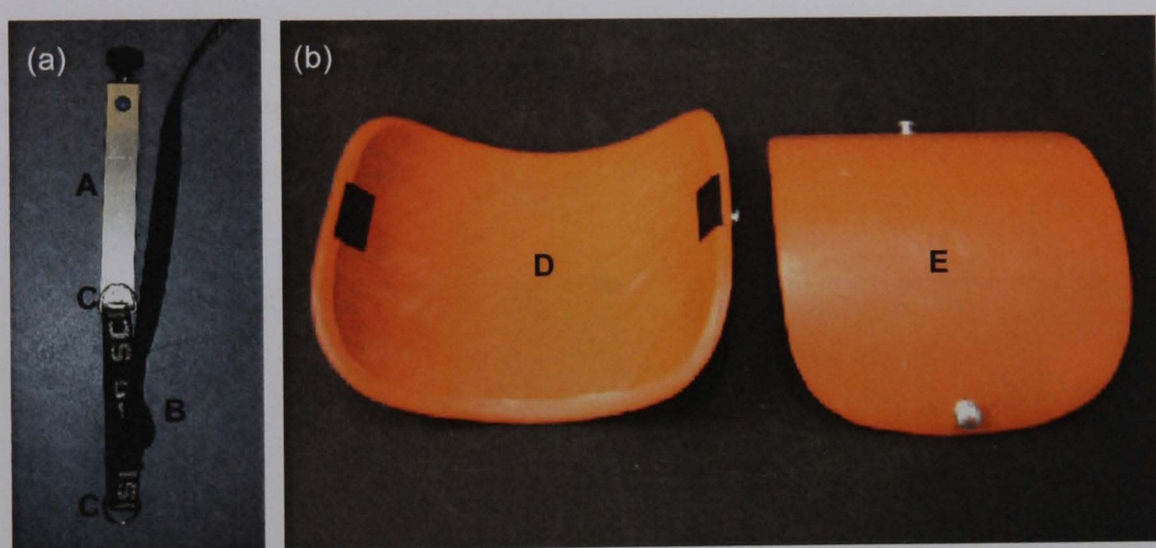
A-Location of runners; B-Low friction ball bearing; C-Sliding attachments, D-Metal connectors, E-Sliding fixture of the pivot point; F-Runners; G-Clamps fixing position of distal beam

Figure 3.4: Limb suspension mechanism-(a) Beam, (b) Proximal attachment, (c) Distal attachment

The distal end of the beam was mounted on runners allowing the beam to slide right or left around the pivot point. The runners consisted of two polyethylene wheels (figure

3.4c). The horizontal support of the distal bridge was fitted with a steel platform along its full length, on which the polyethylene wheels could roll, maintaining a low friction coefficient.

Two sliding attachments were positioned between the pivot point and runners of the beam (figure 3.4a). These featured two steel struts which projected horizontally from the main body of the attachments. Secured onto the horizontal struts were aluminium connectors, which dropped vertically towards the limb below (figure 3.5a). Two plastic crescent shaped cradles (figure 3.5b) were used to support underneath the thigh and shank segments of the test limb, and adjustable straps were used to connect the cradles to the beam via the connectors.



A-Rigid metal connectors; B-Adjustable strap attachments; C-'D' clips; D-Limb cradle-smooth contact surface; E-Limb cradle-underside with attachments for 'D' clips.

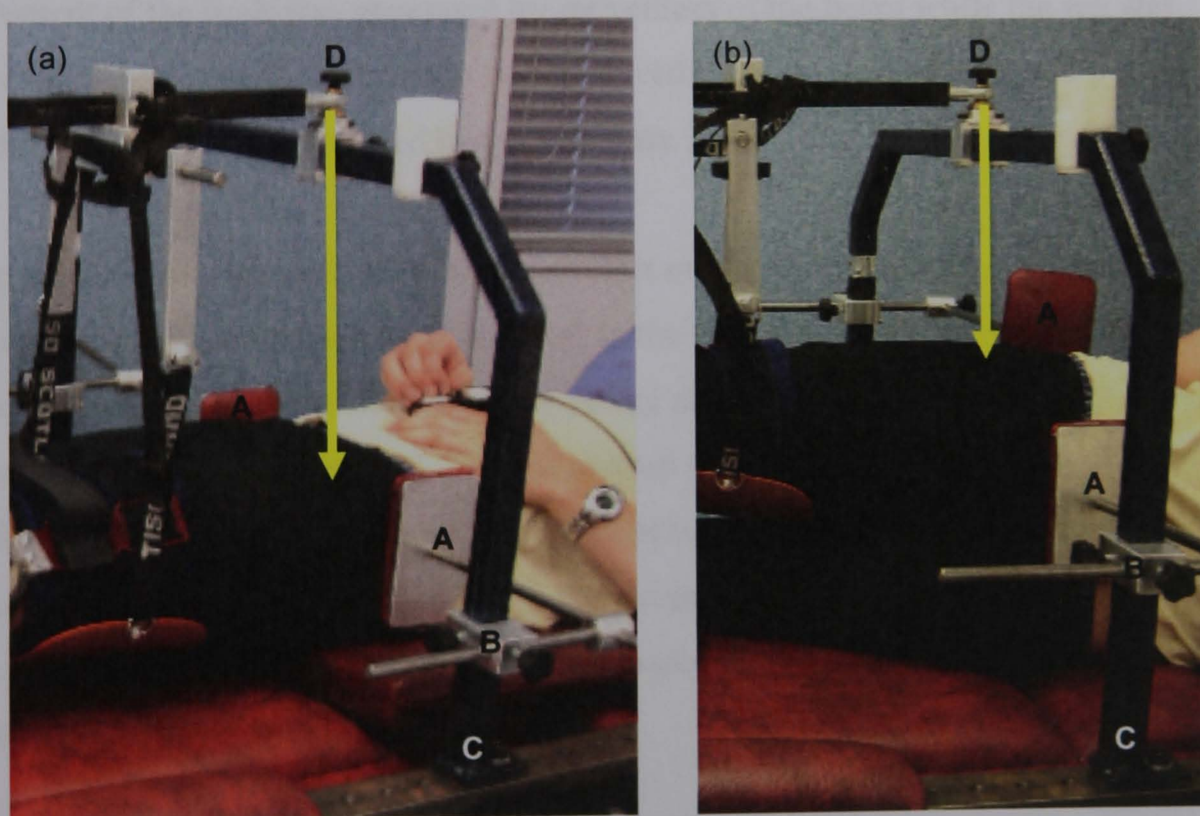
Figure 3.5: Limb suspension attachments-(a) connector attachments, (b) cradles

The flexibility of the strap connecting the cradles to the rigid construct allowed pendular support of the limb. This had various advantages. Firstly this feature was more compatible with the inertial properties of the limb, so that as the surrounding support (cradles) accelerated, the limb accelerated at the same rate, thus preventing the limb from receiving adverse pressure from the surrounding support structures which may have pre-empted the subjects in detecting the movement of their limb.

The protocol relied on the subjects relaxing fully as the tests were performed. Pendular suspension of the limb was also considered advantageous as it allowed the operator to better monitor the state of relaxation of the limb and provide feedback to the subject, encouraging relaxation.

3.2.3 Limiting motion to the hip joint

It was desirable to restrict movement of the joints above or below the hip joint and maintain a standardised position of the subject during the trials. To achieve this, padded clamps were used to brace the bony prominences of the pelvis. These were placed in firm contact with the iliac crest bilaterally while the subject was positioned supine (figure 3.6a), and against the anterior superior iliac spine (ASIS) and sacrum while the subject was positioned in side lying (figure 3.6b). Positioning of the pads was carried out using feedback from the subject to ensure that the stable fixation was achieved without compromising comfort.



A-Padded clamps; B-Attachments for padded clamps; C-Proximal bridge; D-Pivot axis (arrows point to HJC landmark)

Figure 3.6: Pelvis fixation in (a) supine and (b) side lying positions

The knee was immobilised by using a passive range of motion (PROM) brace (Townsend), which comprised of four padded Velcro bands and a separate hinged metal strip which attached onto the pads with adjustable Velcro straps (figure 3.7). The hinge had an analogue numerical indicator highlighting the angle of the knee joint once the brace was attached. This could be adjusted and locked to restrict mobility to a set range with upper and lower limits of motion or restrict the joint to a fixed angle. The PROM brace would be fixed for 0º for the abduction trials and 20º for the flexion trials.

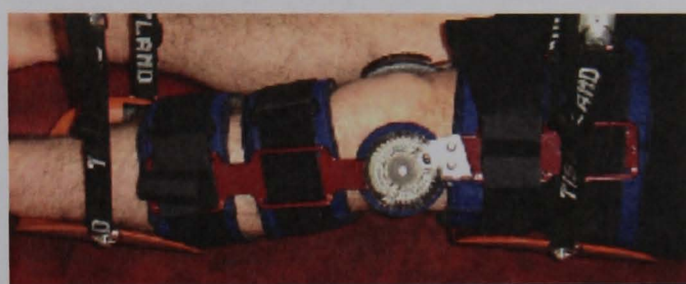
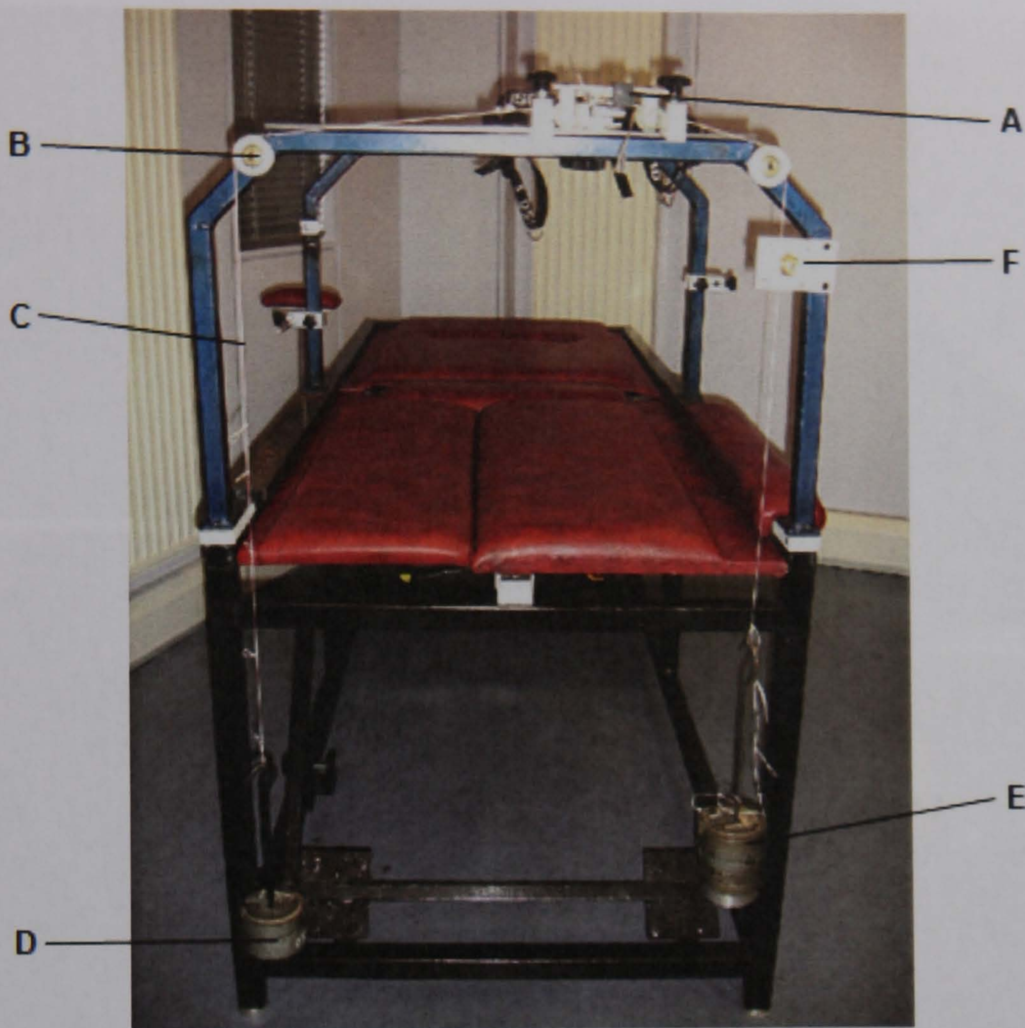


Figure 3.7: PROM brace (supine position, 0° knee flexion)

3.2.4 Movement of the limb

Movement of the limb was generated by movement of the beam while the leg was suspended. Movement of the beam was achieved by using a fixed rope and pulley system with the beam connected in series (figure 3.8).

Strong, two-cord rope was connected to eyelets on the lateral surfaces of the distal beam, and tensioned around pulley turns attached to the distal bridge on the rig. A 4.68kg mass was suspended from one end of the pulley and a 3.40kg mass suspended from the other. The difference in mass (1.28kg) was used to drive the beam, and the test limb, in the desired direction of movement. This mechanical movement was used to provide the motion stimulus to the hip joint. The use of a rope and pulley system allowed an efficient method of generating the motion stimulus without providing vibratory or audible sensory input.



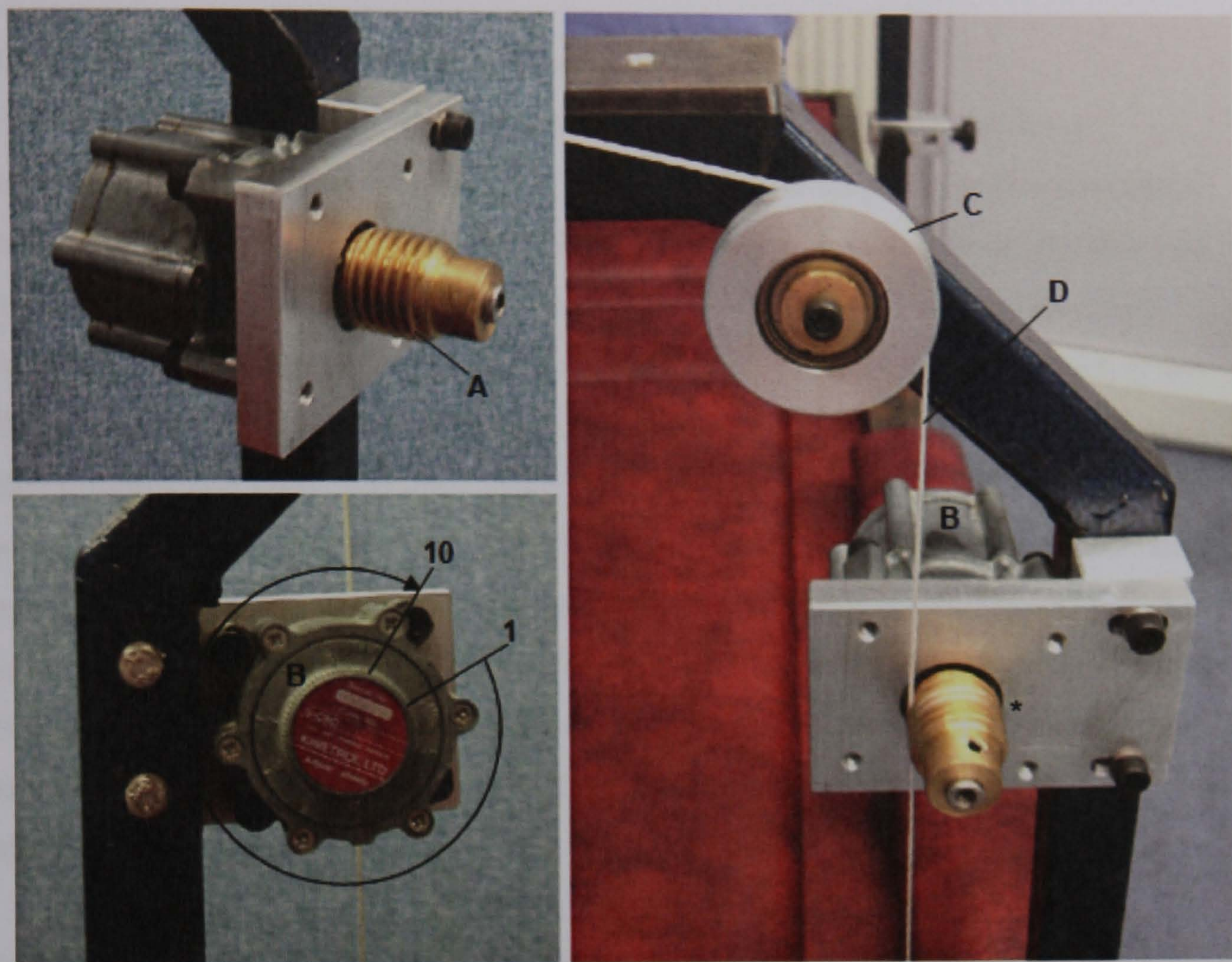
A-Distal end of beam; B-Pulley turn; C-Pulley rope; D-Balancing weight; E-Driving weight; F-Viscous damper

Figure 3.8: Pulley System

3.2.5 Control of limb movement

As the threshold detection of movement is known to vary with the velocity of the motion stimuli (Grigg et al 1973), it was critical to control the velocity of the limb motion to maintain repeatability of between and within subject measurements. Without restrictions the pulley system would drive the beam at a high velocity motion. The desired velocity of motion was less than 1.0°/s (Koralewicz and Engh 2000; section 2.8.2; table 2.5).

A viscous damper (Kinetrol Ltd) was used to control the velocity of the motion of the beam (figure 3.8 and 3.9). The viscosity settings of the viscous damper could be altered in order to control the velocity of the motion.



A-Coil groove of viscous damper; B-Analogue viscosity adjustment (arrow indicating increasing viscosity); C-Pulley; D-Rope (*wound twice around grooves of the viscous damper)

Figure 3.9: Viscous damper

Increasing the viscosity setting on the damper (figure 3.9), for example, decreased the velocity of motion of the beam, and vice versa. This also allowed the operator to control for the variation in limb mass between subjects. A heavier subject with a relatively large lower limb mass would require a higher viscosity setting due to the additional mass and inertia driving the limb at a greater velocity, in contrast with subjects of a lower mass who would require a relatively lower viscosity setting on the damper.

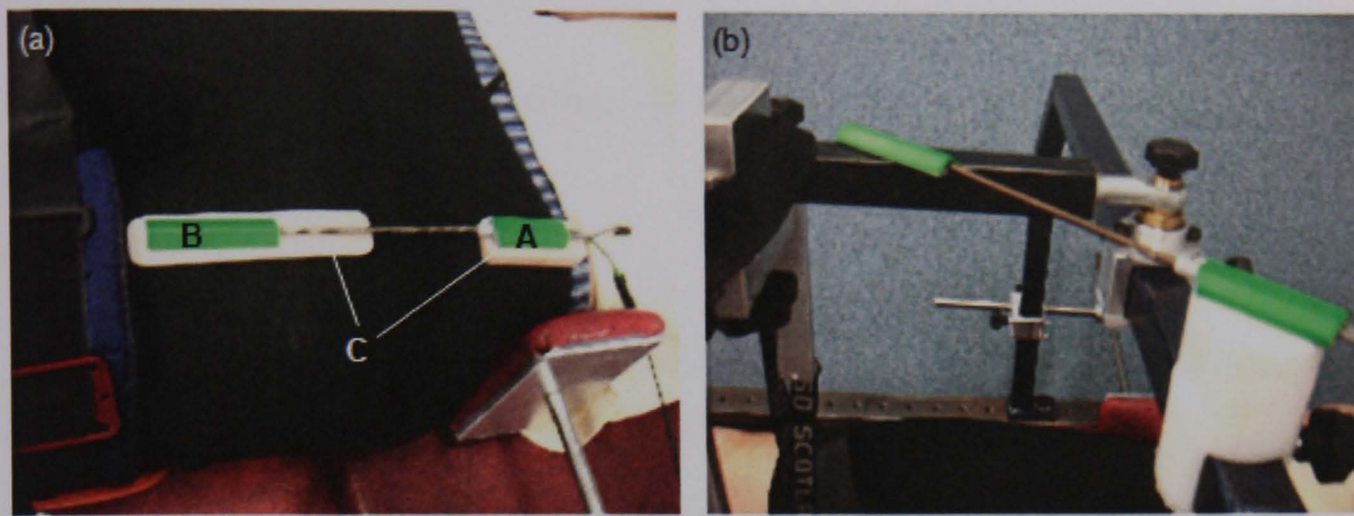
As standard practice, the rope was wound twice around the coil groove of the viscous damper (figure 3.9). Practice trials were used to calibrate the viscosity setting on the damper to a level which achieved limb movement at the desired velocity.

3.2.6 Measuring the amplitude of movement of the limb

The pivot point of the beam was best aligned with the HJC of the subject lying underneath. This allowed the fulcrum of the beam and the subject's limb to be aligned in the y-axis, thus facilitating smooth motion of the lower limb around the hip joint and limiting the possibility of excess pressure or contact with the supporting suspension structures. Achieving the correct alignment was done by estimating the position of the HJC according to the predictive methods highlighted in section 2.5. HJC estimation in supine (abduction trials) was done using the recommendations of Bell et al (1990), where the HJC was predicted to be 30% and 14% of the inter-ASIS distance distal and medial to the ASIS, respectively. HJC estimation in side lying (flexion trials) was done using the recommendations of Andriacchi et al (1982, cited in Bell et al 1990), where the HJC position was estimated to be deep to the GT (table 2.2, section 2.5). The operator aligned the pivot axis of the beam with the estimated HJC by viewing the rig from above.

In the study by Grigg et al (1973), the authors measured the threshold angles by calculating the amplitude of motion occurring at the pivot point of the rig. As mentioned previously, this may have introduced errors in the measurement as a result of malalignment of the HJC with the pivot point of the rig. Therefore, in the current study the protocol was designed with the intention of measuring the threshold detection angles from a device over the hip joint, in order to measure true hip motion.

To achieve this, a flexible electrogoniometer (Biometrics Ltd) was positioned over the hip joint (figure 3.10a). The electrogoniometer was positioned so that it crossed the hip joint, with the location of the HJC estimated in accordance to the protocol defined by Bell et al (1990) and Andriacchi et al (1982, cited in Bell et al 1990), as documented above. To ensure the electrogoniometer crossed the hip joint, the operator passively moved the lower limb, while it was suspended, and ensured that the proximal leg of the electrogoniometer was fixed and that the distal end moved with the thigh segment.



A-Proximal leg of electrogoniometer; B-Distal leg of electrogoniometer; C-Plastic panels

Figure 3.10: Electrogoniometer placement for measuring (a) hip joint and (b) beam movement

To ensure the electrogoniometer over the hip joint did not move on the surface of the subject's clothing, plastic panels were used to mount the electrogoniometer in position. These were made of rigid polyethylene and attached over lycra shorts worn by the subject using double sided sticky tape. Rectangular panels 135mm in length were used for the distal leg of the electrogoniometer. These were positioned over the thigh ensuring the panel and electrogoniometer were flat and not sloping with the contour of the thigh. The proximal panels (75mm) were positioned medial to the ASIS. Various proximal panels were available for use, varying in structure from flat to wedged panels in order to accommodate the lower abdominal or lateral hip contour of the subject. The wedged panels comprised of flat panels with a foam wedge glued to its underside, providing a gradient along the length or the width of the panel. Panels with a positive gradient along their width ranged from 15°, 25°, 35° and 45°, and panels with a positive gradient along their length ranged from angles of 5° and 10°. These panels were for use in supine and side lying positions respectively and ensured the position the proximal leg of the electrogoniometer maintained a horizontal position, continuous with the distal leg.

3.3 Protocol Design

The rig, with the aforementioned design features was integrated into the full MDS protocol.

3.3.1 Protocol Overview

All subjects participating in the MDS protocol were to carry out abduction and flexion trials on both right and left lower limbs, giving rise to four test conditions:

- Left Abduction
- Right Abduction
- Left Flexion
- Right Flexion

Each test condition featured ten repetitions of the motion stimulus, as recommended by Grigg et al (1973), to which the subject responded by pressing a switch on the instance they became aware of the motion stimulus. After pressing the switch, the subjects were to indicate verbally which direction their limb had moved in, by voicing a predetermined verbal indicator (table 3.1). The limb was then returned to the starting position, ready for the next motion stimulus trial. The subject was also asked to press the switch when they detected the motion of their limb returning to the start position (i.e. adduction or extension of the hip), although this was not recorded by the operator as the beam was reversed manually and the velocity and smoothness of the movement was not controlled within the standardised mechanical means. Hence, this was a false trial.

Table 3.1: Motion detection terminology

Hip Movement	Purpose	Description given	Verbal Indicator
Abduction	Motion stimulus	Motion away from the other (contralateral) limb	'Away'
Adduction	Returning motion (false trial)	Motion towards the other (contralateral) limb	'Towards'
Flexion	Motion stimulus	Motion bringing the knee forwards	'Forwards'
Extension	Returning motion (false trial)	Motion bringing the knee back	'Back'

The false trials provided several benefits. By this means, the subjects were tested not only on their ability to detection motion of the limb, but also the direction specificity of their motion detection sense. The use of the false trials also encouraged the subjects to maintain their level of awareness and served to vary the stimulus sufficiently to avoid them becoming too familiar to the actual test stimulus. To enhance this further, the velocity of the returning motion was varied randomly by the operator and the amplitude of the movements and the sequence of the direction stimuli were also varied by, for example, repeating the abduction stimulus twice before returning to the starting position. MDS data was however only acquired for the first motion stimulus occurring from the starting position; all other trials were false and not recorded or analysed.

The threshold MDS angle was determined by calculating the hip angle at the point of motion detection, minus the hip angle recorded before the motion commenced.

3.3.2 Recording angular motion

The output from the electrogoniometers was amplified (x10000) and sampled in real-time via analogue-digital conversion using a National Instruments Data Acquisition Card (DAQCard™-6036E). Figure 3.11 outlines the circuit set up.

The output was displayed in LabView (National Instruments, version 8) using a custom written virtual instrument (VI). This allowed the angle and angular displacement of each goniometer to be viewed and systematically saved for later data management. The electrogoniometers were calibrated and scaled within LabView allowing the raw amplified signal to be output in degrees. Calibration was checked and repeated every 6 months.

The electrogoniometer positioned between the beam and the proximal bridge (figure 3.10b) was responsible for measuring the angular displacement of the beam. The electrogoniometer positioned over the hip joint (figure 3.10a) was responsible for measuring the angular displacement of the lower limb. Threshold MDS measurements were calculated from the electrogoniometer positioned over the hip joint as this was a true measure of hip motion. The electrogoniometer over the beam was used primarily for quality control, to cross reference the output profile of angular displacement over time with that from the electrogoniometer over the hip joint. The operator could therefore identify any anomalies of the hip motion that may be due to active movement, involuntary twitching of the limb etc. Also, mismatches in the gradient of the profiles (angular displacement by time) could suggest that the rotation centre of the beam and the hip were misaligned which was undesirable as this may generate additional cutaneous pressure on the limb. Real-time monitoring of the electrogoniometer signals therefore allowed alterations in the set-up to be made during the practice trials.

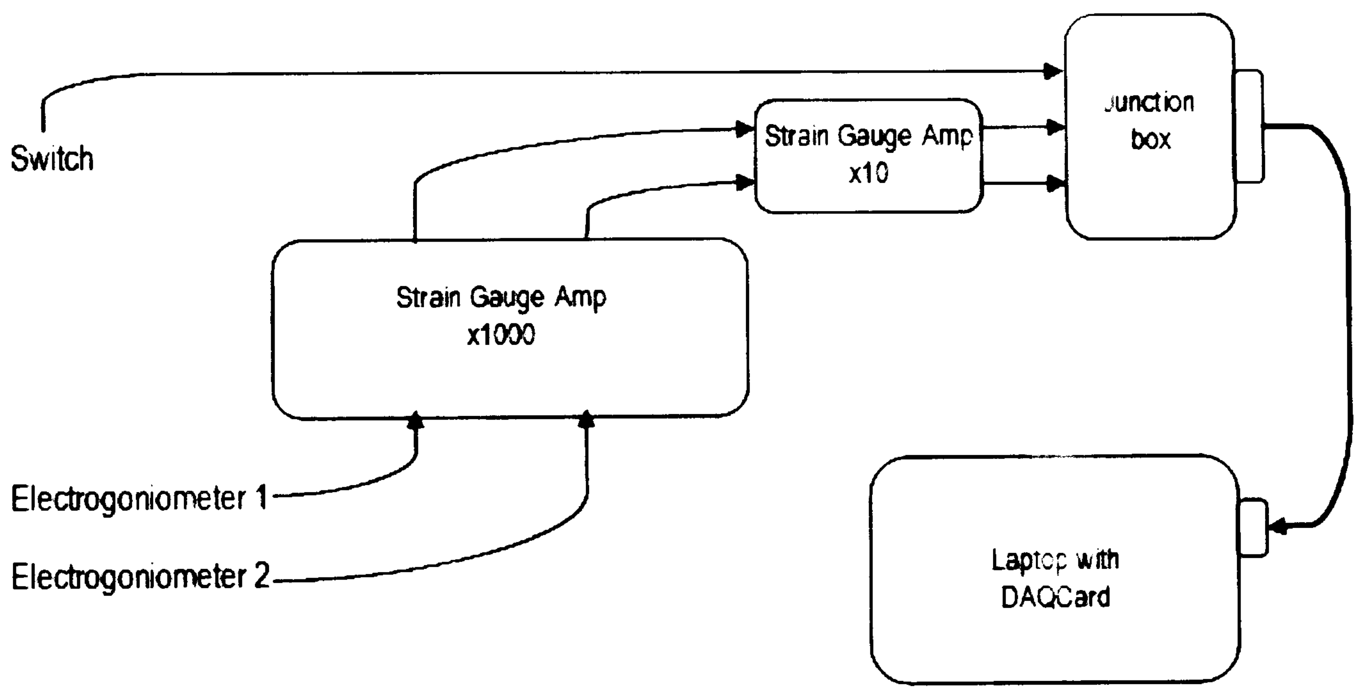


Figure 3.11: Circuit set up

3.3.3 Recording angular velocity

The LabView VI was written to allow the angular velocity to be output. This was done by differentiating the angular displacement over time and plotting the output to display the angular velocity against time. During the practice trials, the operator adjusted the viscosity settings of the damper while monitoring the angular velocity output. When an angular velocity of 0.4°/s was achieved (section 3.4), the test stimulus was administered in the given conditions. As the trials were ongoing, the velocity was monitored both in graphical format and from a numerical indicator, to ensure consistency.

3.4 Pilot Testing

3.4.1 Validation of the MDS protocol

Initial trials were conducted in order to validate the test rig and to validate and standardise the procedure encompassing the test protocol. Pilot subjects were used to provide feedback regarding the comfort of the test procedure and helped towards minimising additional sensory input. Minor alterations to the design, such as optimising the shape of the thigh and calf cradles etc, were accomplished following pilot trials to improve the test procedure.

Measurement of the threshold angles for detection of motion were similar to the values obtained by Grigg et al (1973) at between 0.5° and 1.5°, confirming the validity of the procedure.

3.4.2 Selection of the angular velocity of MDS stimulus

Given the range of angular velocities previously used within literature (section 2.8.2.1), pilot trials were conducted to determine the most appropriate velocity of motion for the test stimulus. Five 'normal' subjects without hip arthroplasty, neuromusculoskeletal pathology or sensory problems completed three test sessions (3x10 trials) of abduction and flexion motion of their dominant lower limb. Individual results were averaged and the results of all subjects were then averaged.

These trials showed that velocities of 0.2-0.6°/s were similar in terms of the mean threshold detection angles obtained (figure 3.12). At 0.8°/s, the mean detection angle was significantly larger. This was thought to be due to the beam having moved too quickly, allowing a larger angular displacement of the limb before the subject reacted to press the switch. At this velocity, the angles were a function of reaction time rather than MDS.

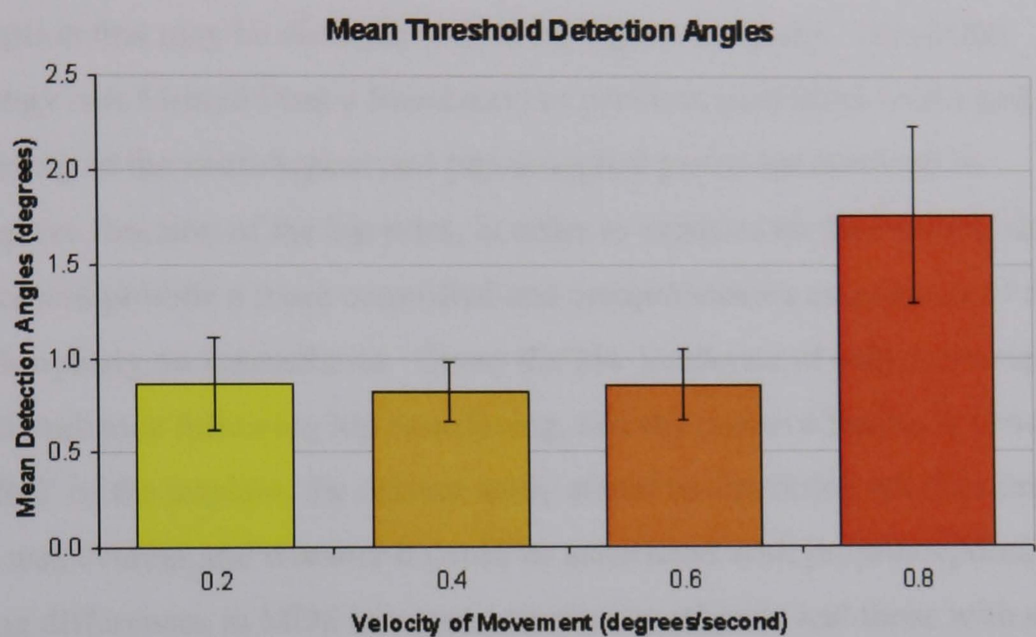


Figure 3.12: Graph of mean threshold angles as a function of velocity of motion

As threshold MDS angles at 0.4°/s had the lowest standard deviation, indicating the lowest mean within-subject variability, and featured as the median of the most appropriate velocity values, 0.4°/s was chosen as the velocity at which the test stimulus would be administered. A motion stimulus of 0.4°/s has been used in previous knee MDS trials (Barrack et al 1984b, Hall et al 1995), where the stimulus had provided sufficient sensitivity to highlight group differences in MDS threshold angles (table 2.5).

3.4.3 Reliability of the MDS protocol

Pilot studies have shown that by varying the viscosity setting on the pulley system, the velocity of motion of the test limb varies linearly (R^2=0.9888) and follows an inverse relationship. Test-retest reliability was also ensured, with good consistency in the set up techniques of the operator and the threshold values obtained showed high repeatability for within-subject pilot trials.

In summary, in order to test proprioception of the hip joint, a test procedure quantifying threshold motion detection, in response to continuous passive hip motion stimulus, was selected and incorporated into a test protocol using a validated test rig. The protocol tested kinaesthesia, the ability to detect motion, which is an important facet of joint proprioception that may be disrupted following hip arthroplasty. This MDS methodology was formed from a foundation of previous published works and an understanding of the neurological and physiological processes involved in proprioceptive function of the hip joint, in order to improve on the existing test procedures and provide a more controlled and comprehensive assessment of the effects of hip arthroplasty on kinaesthesia. Given the low incidence of subjective reports of sensory disturbance following hip resurfacing, and the positive feedback concerning the 'natural feel' of the implant, the current study aimed to determine whether this anecdotal evidence was evident and whether it could be associated with proprioception, by identifying differences in MDS between resurfacing subjects and those with standard THR. As a supplementary aim, the protocol may be used to determine differences in proprioception, as measured by MDS, between the operated and non-operated control limb of subjects with unilateral hip arthroplasty. By doing so, evidence of impairment of proprioception of the operated limb may indicate that intra-capsular mechanisms are responsible for kinaesthesia of the hip joint, and that hip arthroplasty removes or damages receptors required for subjective awareness of motion. Moreover, this may have associated implications for function, or may be responsible for any functional deficits persisting following hip arthroplasty.

The following chapter will detail the entire methodology used for data collection.

4 Methodology

4.1 Study Design

This investigation follows an age-matched subject design with the independent variable of the prosthetic design and orthopaedic intervention carried out. The dependent variables for assessment include functional outcome parameters from gait and stair climbing activities and the threshold for detection of hip motion for the two subject groups, in order to determine whether there are differences between subjects with resurfacing arthroplasty and standard stemmed THR.

4.2 Subject Recruitment

The sample group was formed from a cohort of patients having had unilateral hip replacement surgery at Glasgow Royal Infirmary between November 2005 and December 2007 by one of two consultant orthopaedic surgeons (RI and AS). Surgical intervention included either a MoM hip resurfacing procedure (Durom™ Hip Resurfacing, Centerpulse Orthopaedics Ltd, Zimmer, Switzerland) or a CoC hybrid bearing THR (Trident®, Stryker® Howmedica Osteonics, UK) with a cemented Exeter stem (Exeter™, UK). An additional cohort of individuals with unilateral hip resurfacing was also recruited from the Southern General Hospital in Glasgow. This group included individuals having had a Durom™ prosthesis implanted by an additional consultant orthopaedic surgeon (DM) between July 2007 and August 2007. All three orthopaedic surgeons were experts in their field, had trained together and conducted similar surgical techniques. The expertise of the surgeon was quantified in terms of the length of time spent as a consultant surgeon (RI: 15 years; AS: 5 years; DM: 5 years) and the number of resurfacing arthroplasty procedures carried out within this term (RI: 30; AS: 30; DM: 350) (information current as of December 2008).

The planning of the surgical procedure and the selection of the prosthesis was decided between the patient and surgeon, assisted with advice from the surgeon regarding the patient-specific suitability of each arthroplasty option, based upon the existing knowledge and experience of the surgeon. Therefore, subjects were not randomly allocated into either surgical group, but allocated according to the selected prosthesis and resultant surgical intervention. Random allocation of subjects into the arthroplasty groups was considered unethical due to the greater precautions and contraindications associated with the resurfacing procedure which deemed it unsuitable for certain patients (section 2.4.2).

Potential participants awaiting hip arthroplasty were initially filtered for suitability using information gathered from hospital case notes. Patients were informed about the study during their pre-operative assessment visits, and formally invited to take part within 6 weeks following surgery, having reviewed the post-operative status of the patient with regard to aspects such as complications of surgery and confirming the final prosthesis selected for implantation. Eligibility to participate ran in accordance with the criteria outlined in figure 4.1.

Individuals with a monoarticular hip joint disorder and unilateral arthroplasty (Charnley category A) were selected in order to allow for within-subject comparisons of the operated limb relative to the 'normal' limb. This within-subject comparison facilitated the assessment of deviations from the normal as a result of the arthroplasty while controlling for the individual variations that act as a limitation for between-subject comparisons. However, the nature of OA dictates the high prevalence of bilateral joint pathology. Hence, the incidence of subjects with purely unilateral joint disease was extremely low. Having recognised this from preliminary monitoring of patients at Glasgow Royal Infirmary, the inclusion criteria were widened to allow patients with 'mild' contralateral hip pain or pathology to take part (appendix 3). The severities of lower limb and spinal pathologies were graded according to visual analogue scoring and radiographic evidence of abnormalities in bony architecture were graded by the surgeon.

Inclusion Criteria

- Pre-operative monoarticular OA or degenerative disease of the hip
- Surgery classification as primary, unilateral hip joint arthroplasty
- Aged between 40-60 years at the time of surgical intervention
- Prosthetic implant type was Durom resurfacing or CoC Trident THR
- One of two orthopaedic surgeons performed the CoC stemmed THR
- One of three orthopaedic surgeons performed the resurfacing procedure
- Informed consent was provided

Exclusion Criteria

- Bilateral OA or degenerative disease of the hip joint
- Moderate to severe pain or joint disease affecting other lower limb joints
- Moderate to severe pain or pathology affecting the lower back
- Injury or trauma affecting back and lower limbs 6 weeks preceding test intervals
- Other neuromusculoskeletal disorders affecting ambulation
- Dependence on walking aids
- History of epilepsy and seizures
- Severe, unstable cardiovascular disease
- Somatosensory disorders
- Surgical intervention modified intra-operatively

Figure 4.1: Inclusion and exclusion criteria for the recruitment of subjects

Pain, pathology or trauma affecting the back or other lower limb joints is known to influence biomechanics of gait and stair climbing (James et al 1994, Madsen et al 2004, Tanaka 1998, Vogt et al 2003) and pain, arthritis, previous injury and neurological or somatosensory disorders directly influence proprioception (Hassan et al 2001, Lephart et al 1998, Shakoor and Moisio 2004). The co-existence of any one of these factors in addition to unilateral hip arthroplasty may limit the ability of the operator to attribute the presenting motion abnormalities and proprioception findings to the orthopaedic intervention. Hence individuals with co-existing neuromusculoskeletal abnormalities were excluded from the study.

The MoM resurfacing procedure and the CoC stemmed THR are generally administered to adults under the age of sixty-five (section 2.4.2). Those over sixty-five years may be offered this intervention if they are particularly active and have a low incidence of co-morbidities. However, this wide potential age range (18 to 65 years) was narrowed in order to limit the influence of age variables when comparing differences between the surgical groups (Kadaba et al 1989, Tanaka 1998). A minimum age of forty years was selected, thus eliminating individuals aged eighteen to thirty-nine who may have been otherwise appropriate for inclusion. Nevertheless, numbers falling into this younger age range were low and their inclusion may have introduced unnecessary variability within- and between-groups. The maximum age was restricted to sixty years to correlate with the age generally accepted as the minimum age above which an individual may be classed as elderly (http://www.who.int/healthinfo/survey/ageingdefnolder/en/index.html). Hence, the age range of forty to sixty years (Isobe et al 1998) was considered to provide sufficient control of age related variables and cover the median age range of those considered for both types of arthroplasty.

Individuals with epilepsy, seizures and unstable cardiovascular disease were excluded in order to limit risk to the patient during the test procedure. The threshold MDS test required subjects to lie horizontally in supine and side lying positions. Recognising that these positions alter haemodynamics, this test procedure may have presented a risk to such individuals.

Finally, any modification of the surgical procedure during theatre justified subject exclusion, as patients within this category were exposed to a combination of surgical methods. This was typically observed for patients who had planned to undergo the resurfacing procedure but were converted to a stemmed THR or a large diameter MoM stemmed THR due to findings on open examination of the subject's bony anatomy (section 2.4.5).

Interested subjects falling within the specified criteria provided informed consent in order to participate in the study. Ethical approval was granted by the NHS Ethics Committee, North Glasgow Trust and the University of Strathclyde Ethics Committee and the department for Research and Innovation.

4.3 Instrumentation

4.3.1 Subjective instruments

For subjective data collection, the Oxford Hip Questionnaire was used to gather information regarding the functional outcome of individuals. This questionnaire has been validated (Dawson et al 1996, 2001) and is used routinely in the clinical institutions where the orthopaedic surgery was conducted. The Oxford Hip Score is a 12-item patient-centred questionnaire targeting aspects of pain, function and ability to conduct various activities of daily living (appendix 4). For each item, or question, there are 5 alternative responses, which are graded in severity from 1 to 5 points. The total scoring values vary from 12 to 60, where a score of 12 represents the best outcome.

A record of complications and patient satisfaction was also compounded in order to assess the influence of psychological factors on the outcomes. A satisfaction questionnaire (appendix 5) was used to determine the subjects' attitude towards the surgery and its effects.

Subjects also rated their subjective level of pain and stiffness on visual analogue scales (VAS). Each VAS comprised of a 10cm line with two given extremes, titled 'no pain/stiffness at all' (0cm) and 'the worst pain/stiffness ever imaginable' (10cm) (figure 4.2). The subjects marked the VAS with a vertical line and the level of pain or stiffness was calculated by measuring the position of the subject's mark between these two extremes.

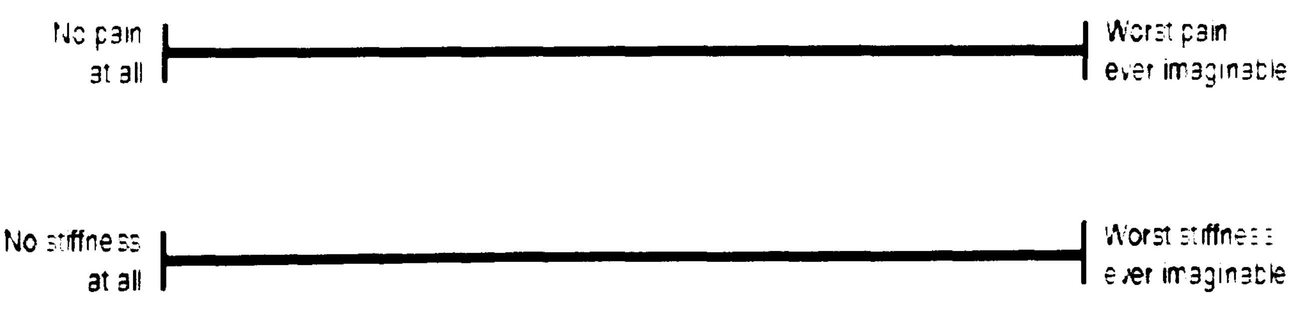


Figure 4.2: Visual analogue scales recording subjective pain and stiffness

Subjective data was stored on a laptop computer using OrthoWave® software programme (Version 5.04), allowing patient records and outcomes to be filled together under password protection.

4.3.2 Kinematic analysis

A VICON 612 motion analysis system (Oxford Metrics Ltd., UK) with eight infra-red cameras (figure 4.3) was used to collect optoelectronic photogrammetric data for biomechanical evaluation. Data was sampled at a rate of 120Hz, capturing motion and analogue data simultaneously.

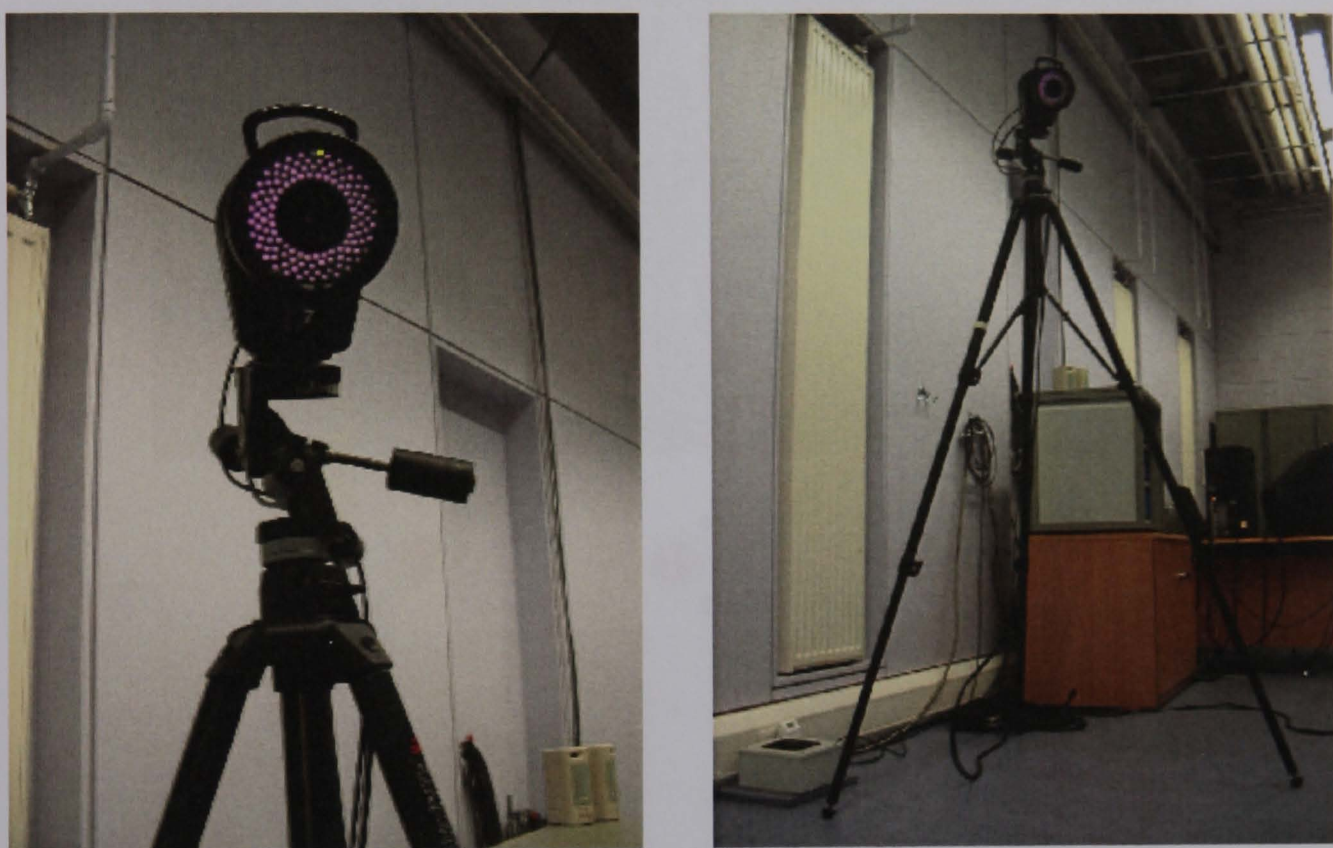


Figure 4.3: Vicon camera and adjustable tripod stand

A capture volume of approximately 4.8 m^3 (length 2.0m-breadth 1.2m-height 2.0m) was created. Appropriate camera settings for gait analysis were adopted, dictating a set focal length (infinity) and lens aperture (≤f4). The cameras were positioned high upon adjustable tripods (≥1.5m) and tilted downwards in order to optimise the visualisation of markers without cameras visualising one another, thus reducing residual errors in marker identification and reconstruction. The relative camera placement within the lab is shown in figure 4.4. Cameras positioned at the extreme ends of the capture area (cameras 1, 4, 5, 8) were positioned highest with a bias for collecting marker position data when the subjects were at the top of the stairs, and the four side cameras were lower, ensuring that the distal limb markers were visualised.

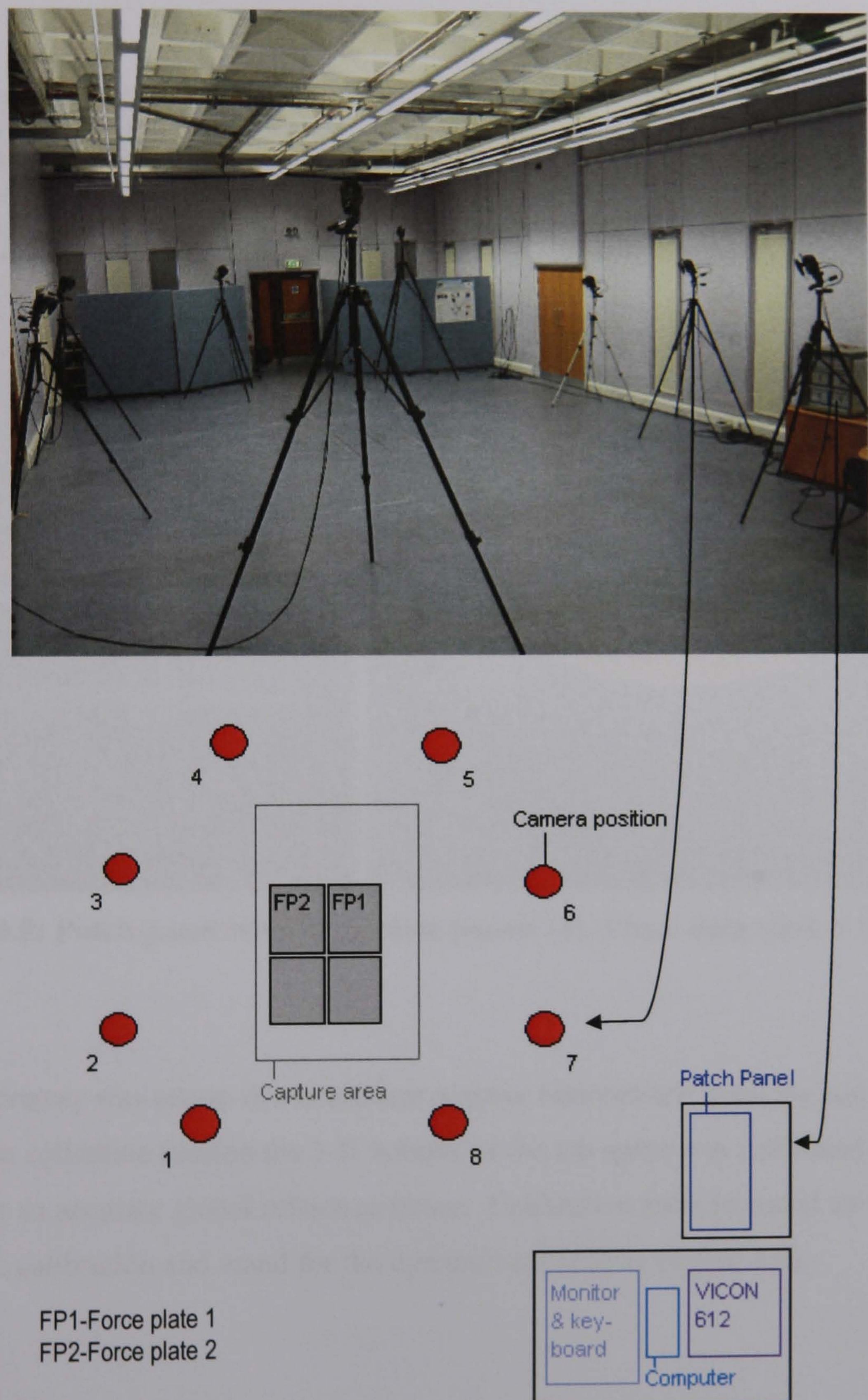


Figure 4.4: Camera position within the lab space

The acquired video data was channelled through a patch-panel, processed by the VICON 612 data station and then visually represented using Vicon Workstation software (version 4.4) on a Dell (Precision 650) computer (figure 4.5). Data was saved continuously using the Workstation software (c3d, enf, tvd and vad files) for further processing and analysis.

Figure 4.5: Patch-panel with force plate panels (a), Vicon data station (b) and PC (c)

The laboratory was set-up in a consistent manner between test sessions and prior to each data collection session the 3-D volume of the lab space was calibrated to establish an accurate global reference frame. Calibration tools included an L-frame for static calibration and wand for the dynamic calibration (figure 4.6).

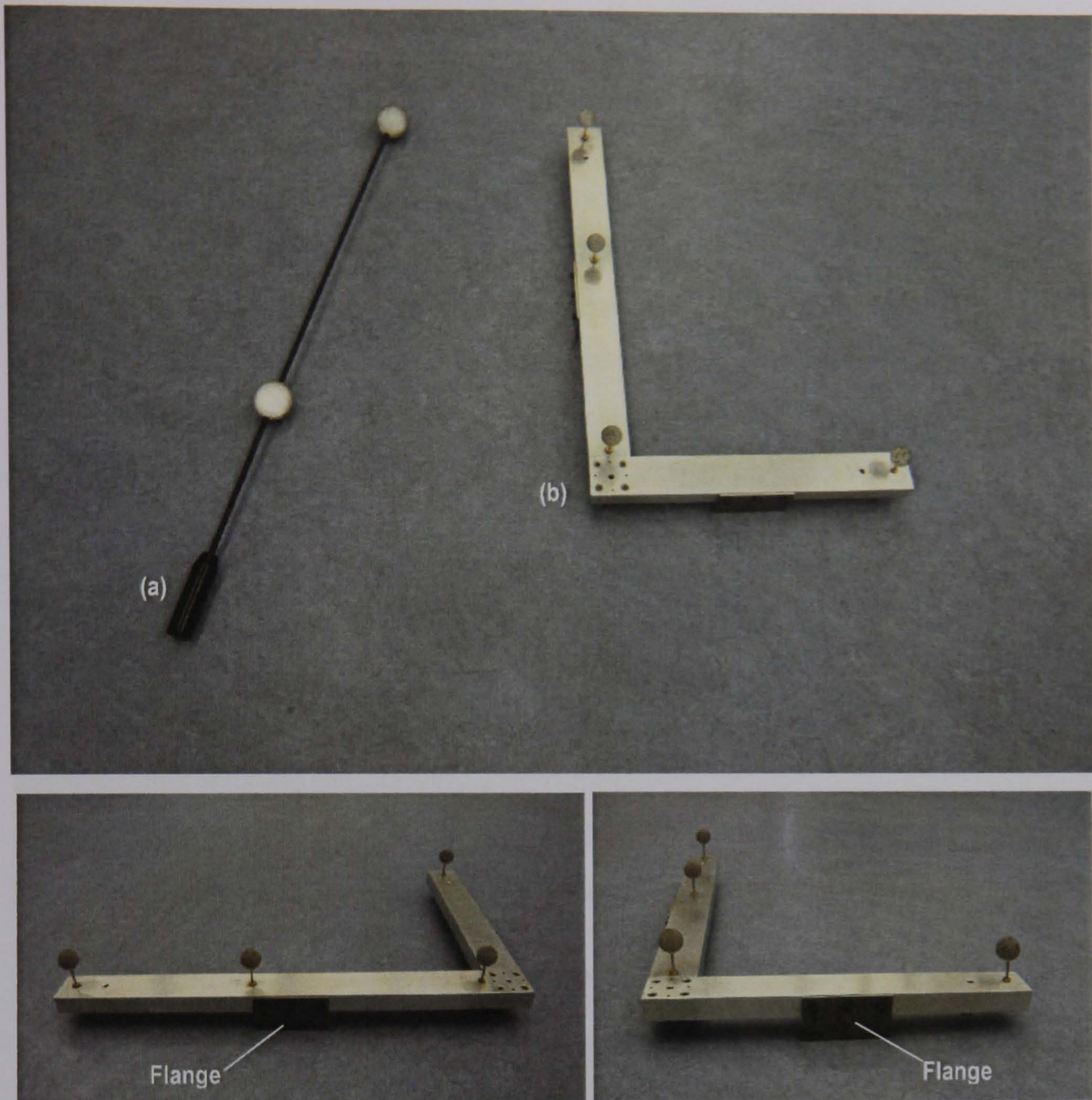


Figure 4.6: Calibration Instruments

(a) Dynamic calibration wand
(b) L-frame (with flanges)

Spherical retroreflective markers (diameter 14mm, figure 4.7) were used to mark bony prominences and limb segments using individual markers and clusters of markers respectively. Marker clusters were formed by an arrangement of four markers set at a fixed distance and mounted on a thermoplastic moulded oval structure (figure 4.7). Clusters were placed on the corresponding body segments using a Velcro attachment on a neoprene cuff. The pelvic cluster was formed by fixing a Velcro strap around the pelvic rim and attaching markers, separated at a distance in excess of 200mm.

A smaller calibration wand was used for pointer trials (Cappozzo et al 1995, Della Croce et al 2005), which were conducted in order to define and reference specific bony landmarks to their corresponding segments. The pointer calibration wand (figure 4.8) was formed by a pointed tip at one extreme and the centre of a spherical marker at the other. Another marker was situated 270mm from this marker and 123mm from the point, which produced a vector representation on the Workstation software. By knowing the position of the vector in space, the point 123mm from the marker closest to the tip of the pointer, in line with this vector could be assigned as a virtual point, marking a given anatomical landmark.

Figure 4.7: Individual 14mm markers (a) and marker clusters for thigh (b) and shank (c) segments

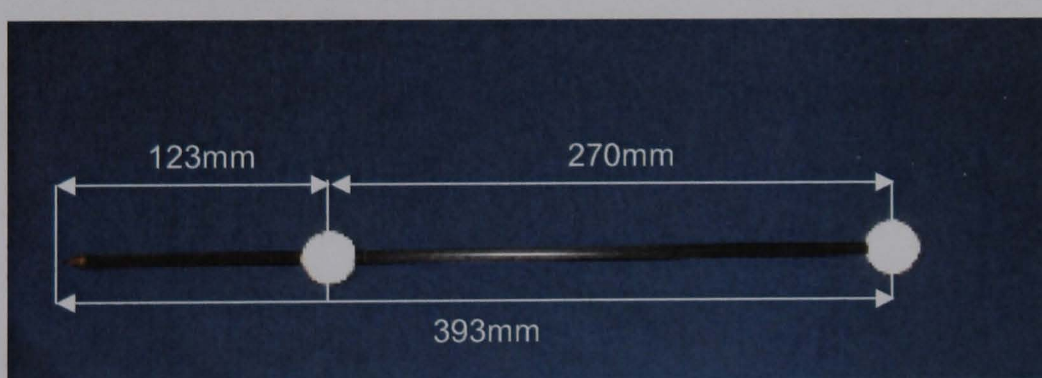


Figure 4.8: Pointer for anatomical landmark identification

This data was used to construct an anatomical reference frame representing the underlying bony structure, developed using a BodyBuilder code (Vicon software version 3.5). The development of anatomical reference frames will be discussed in more detail.

4.3.3 *Kinetic analysis*

Two Kistler force plates (Kistler Instrumente AG, Switzerland) were used to measure ground reaction forces during the motion activities (figure 4.9). These force plates were embedded in the floor within metal frames, allowing each to be completely independent of the other. The surface for force application was level with the surrounding floor space. The outputs of the force plates were zeroed following each trial to remove residual electronic outputs. Data acquired from the two force plates (F_x, F_y, F_z; M_x, M_y, M_z) were collated via channels on the patch panel and presented with the kinematic data in the Workstation application of the PC.

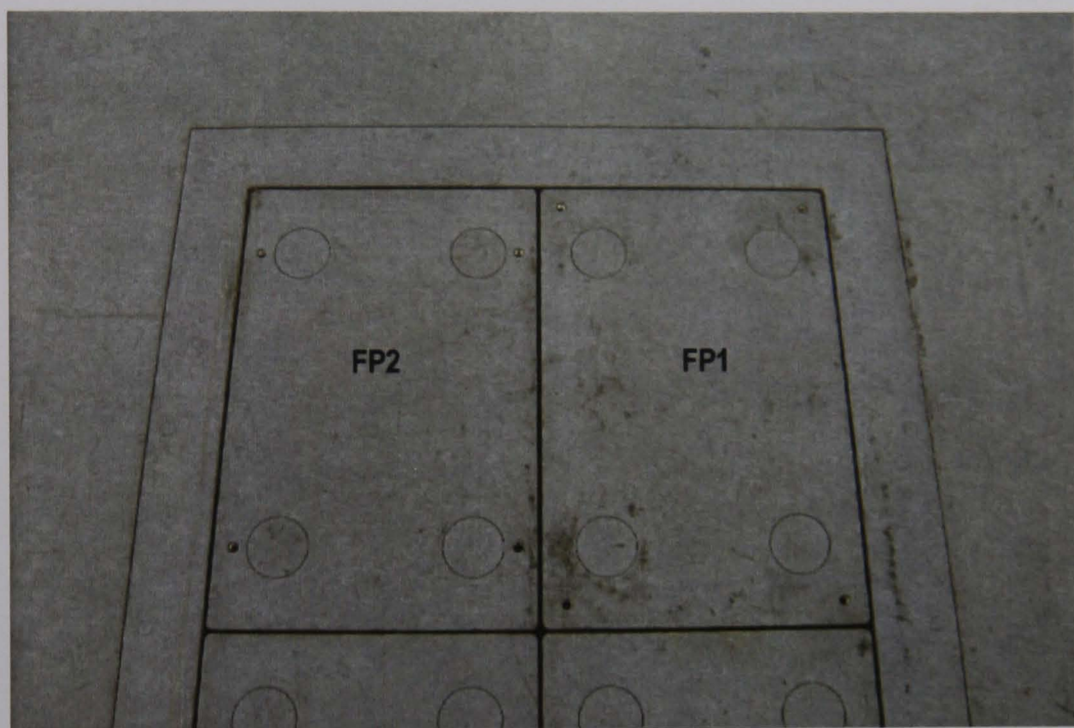


Figure 4.9: Adjacent Kistler force plates

FP1-Force plate 1
FP2-Force plate 2

4.3.4 *Tools for functional analysis*

A custom built staircase with an instrumented force plate step was used for combined kinetic and kinematic analysis during stair climbing. The instrumented step featured as the second step in a series of four constructed by bolting two individual step platforms (figure 4.10) onto force plates one and two. The two step platforms of the second step were therefore structurally independent from one another and from the remaining staircase apparatus, which was built around the force plate step. Hence only contact with the instrumented step resulted in a kinetic profile during data collection.

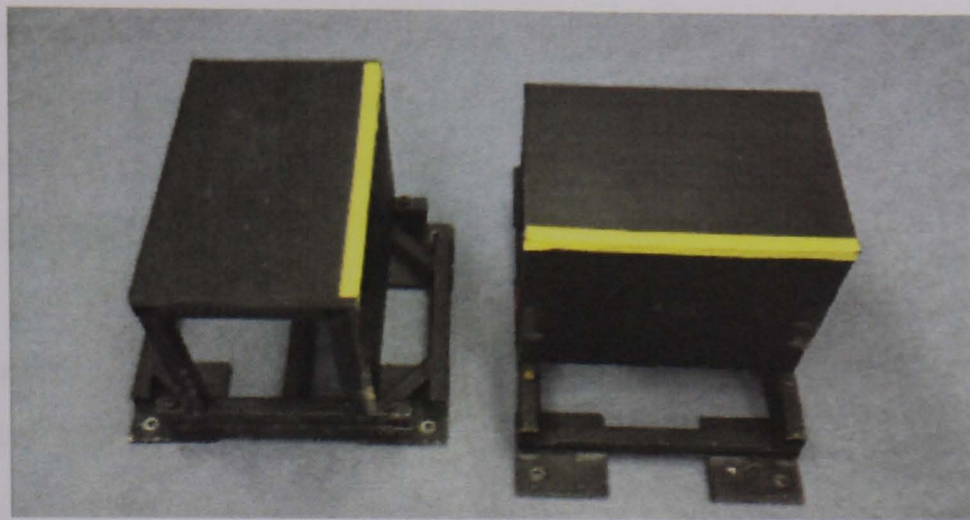


Figure 4.10: Force plate steps

Figure 4.11: Instrumented staircase

The staircase was constructed with standard step dimensions (step height 185mm, tread depth 280mm), similar to those cited in published literature (Protopapadaki et al 2006, table 2.5) and featured bilateral handrails and a platform for turning on the fourth step (figure 4.11).

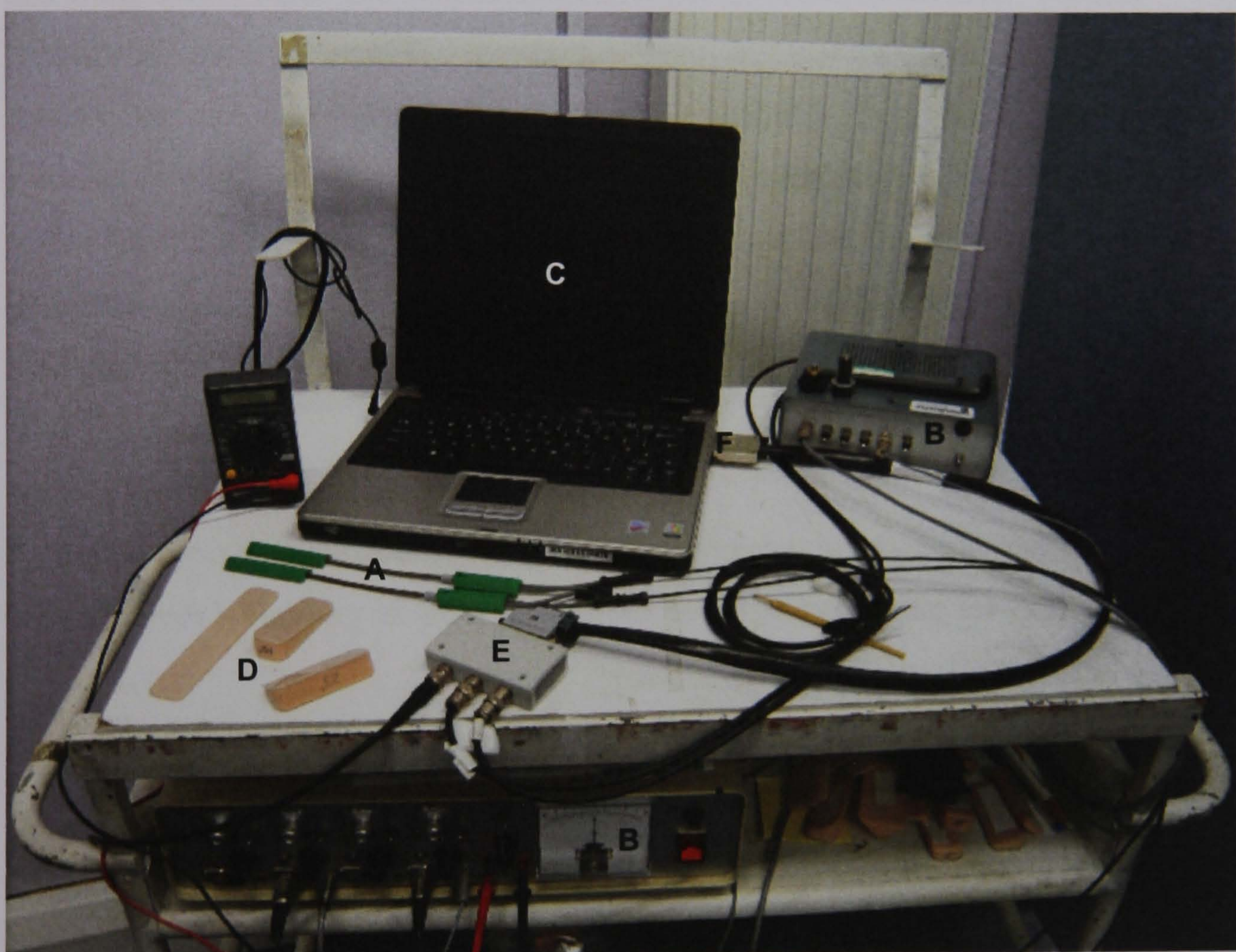
4.3.5 Proprioception rig

A rig was designed and built for the analysis of hip joint proprioception (figure 4.12, Chapter 3), using the outcome of threshold MDS of the hip joint, by measuring the change in angle at the hip joint from the point of initiation of passive hip motion to the point of subjective detection of the movement stimulus. The rig was approved by the Bioengineering Unit Safety Committee (University of Strathclyde).

A-Pulley system; B-Limb cradles; C-Limb suspension beam, D-Sliding pivot point of beam, E-Padded wedge; F-Padded pelvis clamp; G-Eye mask; H-Viscous damper

Figure 4.12: Proprioception rig

The various design features of the rig are explained in depth in Chapter 3. Associated instruments included the viscous damper (Kinetrol Ltd.), used to control the velocity of the beam movement, and subsequently the velocity of movement of the lower limb; the electrogoniometers (Biometrics Ltd.), used to measure the change in ROM at the hip joint; strain gauge amplifiers, used to amplify the electrogoniometer output; and a lap top computer (RM AL51, England) with National Instruments Data Acquisition Card (DAQCard™-6036E) and LabView software (National Instruments, version 8), used to display the angular displacement and the angular velocity of motion (figure 4.13). The circuitry set up of the electronic instruments is illustrated in figure 3.11 of the design chapter.



A-Electrogoniometers; B-Strain gauge amplifiers; C-Laptop computer, D-Plastic panels, E-Junction box; F-DAQ Card

Figure 4.13: Overview of MDS instruments for use with rig

For all participants in the study, the same instrumentation was used, in order to ensure inter-subject reliability.

4.4 Procedure

Subjects were systematically involved in data collection over a 31 month period (March 2006-September 2008) by attending two test sessions at 3 months and 12 months following surgery (Tanaka 1998) within the Bioengineering Unit, University of Strathclyde. The duration of each session was three hours.

4.4.1 Preparation and baseline measures

Following informed consent, subjects were provided with cycling shorts in preparation for functional assessment and MDS testing. Baseline outcomes of height and weight were taken and contraindications and precautions checked. All test procedures were explained in a standardised fashion before running practice trials or actual trials, and the subjects were given opportunities to ask questions. All ambulatory tasks were carried out in bare foot (James et al 1994, Heller et al 2001, Protopapadaki et al 2006, Riener et al 2002, Riley et al 2001) and were self-paced (Aminian et al 2004, Eng and Winter 1995, Perron et al 2000, Protopapadaki et al 2006, Riley et al 2001).

4.4.2 Kinematic analysis

Prior to subject data collection, various procedures were carried out in the biomechanics laboratory in order to calibrate the instrumentation for use.

4.4.2.1 Creation of the global co-ordinate system

The eight cameras were set up for data capture in the biomechanics laboratory. A global co-ordinate system was created by calibrating the pre-defined capture volume using the calibration tools (figure 4.6). This method involved static capture of the L-frame while it was positioned in the centre of the capture volume with the flanges fixed in the gap between the force plates, around force plate one (figure 4.15). The L-frame referenced the origin of the lab (x, y, z = 0, 0, 0) to the corner of force plate one and defined the direction of the orthogonal axes of the global co-ordinate system (figure 4.14, figure 4.15). The global frame definition was consistent with the International Society of Biomechanics (ISB) recommendations (Wu and Cavanagh

1995; Wu et al 2002) and is widely accepted for use in the biomechanics community (Cappozzo et al 2005). Static calibration was also used to determine accurate magnitudes of distance within the capture space.

Global Co-ordinate System: Axis Definition

Origin	Corner of force plate one
x-axis	Pointing forwards (anterior), corresponding with the mean direction of progression of the locomotor activities
y-axis	Pointing vertically upwards, perpendicular to the x-axis
z-axis	Pointing to the right, perpendicular to the plane created by the x-axis and y-axis

Figure 4.14: Axis definition for the global co-ordinate system

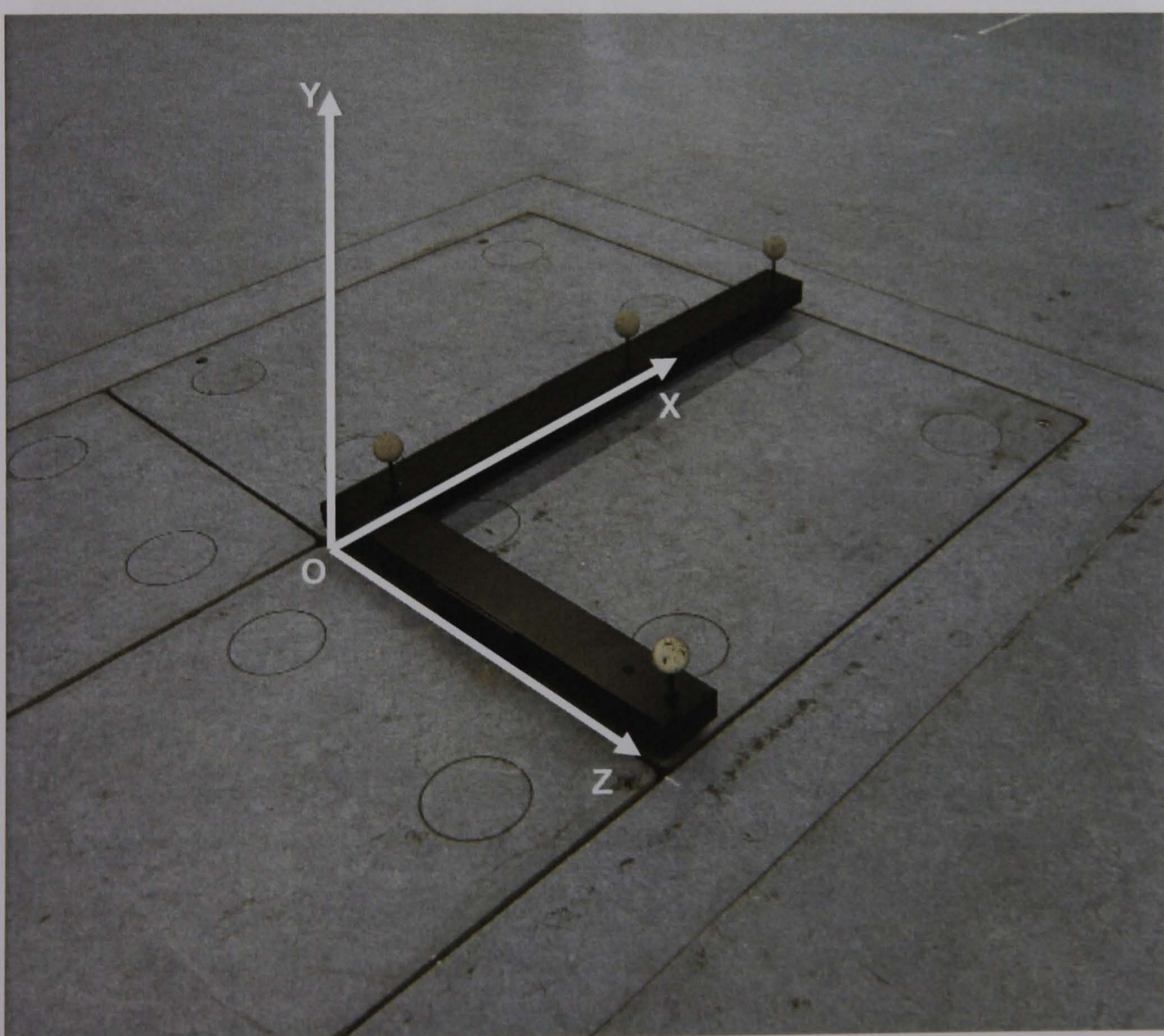


Figure 4.15: L-frame position for static calibration (global axis system)

Dynamic calibration was then conducted by moving the calibration wand through the entire capture volume where subjects would be performing the locomotor tasks, ensuring the extremities of the volume were covered, particularly the height for stair climbing trials (figure 4.16). The velocity of the wand motion was matched to the expected movement velocity of the subject group. This component of the calibration procedure was done in order to calibrate the camera parameters for the capture of dynamic marker data.

Figure 4.16: Dynamic calibration of the capture volume

4.4.2.2 Marker set

The marker set used is outlined in table 4.1, describing the positions of the markers on their corresponding segments, from proximal to distal, and the anatomical landmarks identified by or derived from the marker positions. Markers were consistently applied to every subject while they maintained their natural anatomical standing position with their feet approximately hip width apart.

Table 4.1: Marker set used for kinematic analysis

Segment	Marker placement	Anatomical landmark identified	Method of identification
Pelvis	Pelvic cluster 4 markers attached to waist band in arbitrary positions	Right ASIS Left ASIS Right PSIS Left PSIS	Pointer trials referring landmarks to the pelvic cluster
Thigh	Right thigh cluster placed in an arbitrary position on distal right thigh Left thigh cluster placed in an arbitrary position on distal left thigh	Right lateral femoral epicondyle Right medial femoral epicondyle Left lateral femoral epicondyle Left medial femoral epicondyle	Pointer trials referring landmarks to the thigh clusters
Shank	Right shin cluster placed in an arbitrary position on distal right shin Left shin cluster placed in an arbitrary position on distal left shin Right lateral malleoli Right medial malleoli Left lateral malleoli Left medial malleoli	Right lateral malleoli Right medial malleoli Left lateral malleoli Left medial malleoli	Direct markers referred landmarks to the shin clusters and later removed after static calibration trial
Foot	Calcaneus 1st metatarsal head 5th metatarsal head	Calcaneus 1st metatarsal head 5th metatarsal head	Direct markers

ASIS-Anterior superior iliac spine
PSIS-Posterior superior iliac spine

Velcro straps were attached to the pelvic, thigh and shank segments, and clusters of 4 retroreflective markers attached (figure 4.7). Strap attachments were tight enough to prevent slip during the movement tasks, but comfort was ensured. Four non-aligned markers were selected for use on clusters as opposed to 3 in order to give redundancy if one marker was occluded from camera vision.

Superficial bony prominences were identified by standardised palpation methods (Cappozzo et al 1995, Della Croce et al 2005, Norton and Olds 1996) and the specific points were selected as they were "identifiable in a repeatable fashion" (Cappozzo et al 2005) therefore limiting within- and between-subject variability. The pointer calibration trials were carried out while the subject maintained a quiet anatomical standing position within the capture area. By pointing to the anatomical landmarks, the position vector of the points were identified and referenced to the corresponding cluster and body segment using a BodyBuilder code (appendix 6).

The application of direct skin markers was done using double-sided hypoallergenic adhesive tape. The use of direct markers instead of pointer trial identification was conducted for identifying anatomical landmarks which were subject to minimal skin movement artefact.

In the sections to follow, the marking of each body segment will be described individually, from proximal to distal, and the corresponding technical (marker cluster technical frame) and anatomical (bony embedded) reference frames defined. As with the global co-ordinate system, the convention used for all local co-ordinate systems was consistent with the standard convention of right-handed orthogonal triads recommended by Wu and Cavanagh (1995). The definition of joint co-ordinate systems (JCS) was based on the standardised proposal presented by Cole et al (1993) which was derived from the earlier knee JCS definitions reported by Grood and Suntay (1983). This involves definition of one rotation axis embedded in the proximal segment ($\hat{e}_1$), coincident with the z-axis of the proximal anatomical reference frame, a second rotation axis embedded in the distal segment ($\hat{e}_3$), coincident with the y-axis of the distal anatomical reference frame, and a third axis ($\hat{e}_2$), which is a floating axis and is "normal to the two fixed body axes" (Cole et al 1993, page 345). Where JCS are defined, the clinical interpretation of rotation about the axes of the JCS will also be defined.

4.4.2.3 Waist segment and pelvic co-ordinate system

Definition of the pelvis during kinematic analysis was done by attaching a cluster of four markers around the pelvis. The cluster was formed by attaching an elastic strap around the lower waist of the subject, superficial to the innominate bones, and positioning the markers at least 20mm apart using double-sided adhesive tape, as shown in figure 4.15. As there were differences in soft tissue proportions of the subject group, variations in the relative position of the waist band between subjects existed. This ensured that the band was in an optimum position on a subject specific basis, thus limiting the likelihood of the band slipping up or down during the locomotor activities.

Figure 4.17: Waist segment marker cluster (anterior and posterior aspects)

The position of the four waist cluster markers defined the technical frame of the pelvis relative to the underlying bone. Pointer trial calibration of the ASIS and PSIS positions (table 4.1, figure 4.18) relative to the pelvis technical frame formed the basis for defining the anatomical frame of the pelvis. The origin of the pelvis co-ordinate system (PELF) was fixed at the midpoint between the ASIS (Cappozzo et al 1995), as shown in figure 4.19. The orthogonal axis system of the pelvis co-ordinate system is described in figure 4.20.

Figure 4.18: Pointer trial calibration of anatomical landmarks of the pelvis (right ASIS and right PSIS)

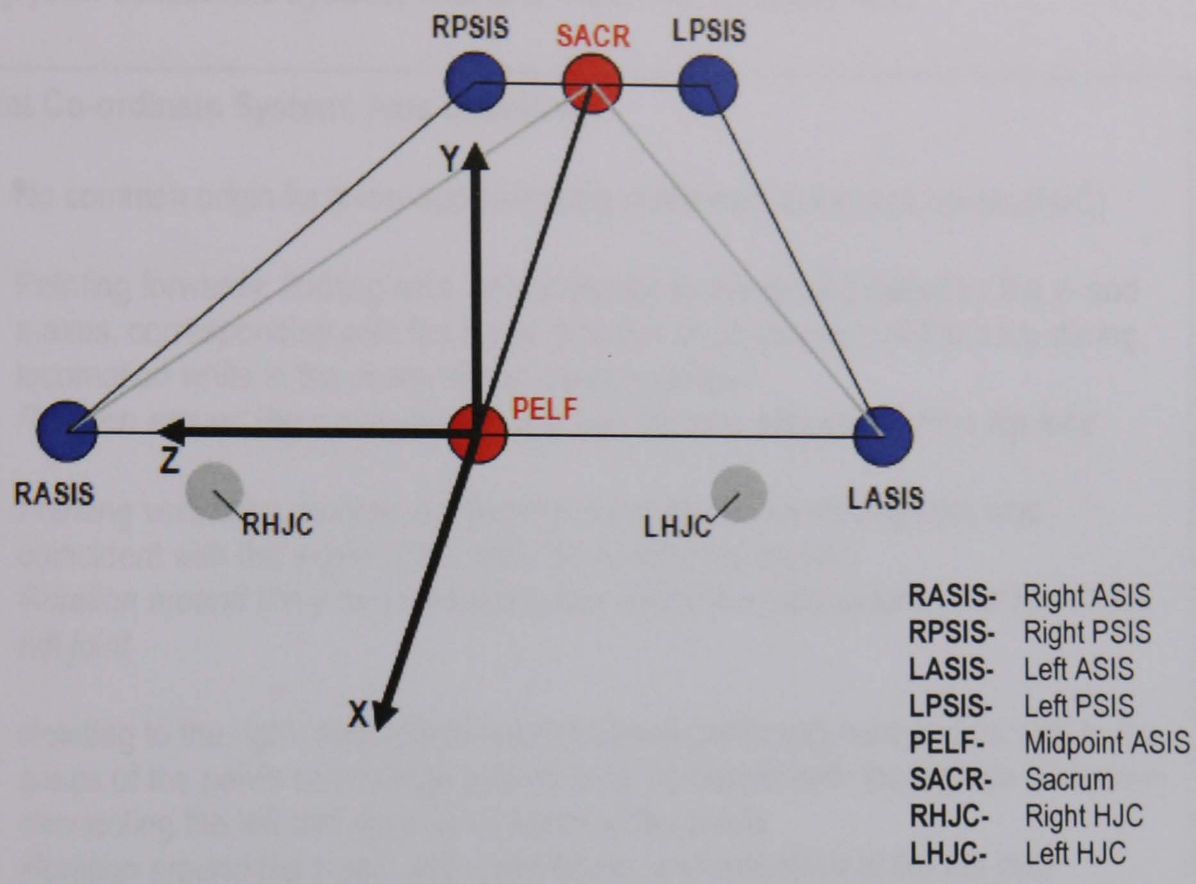


Figure 4.19: Pelvis anatomical frame and co-ordinate system

Pelvis Co-ordinate System: Axis Definition

Origin Mid-point between the ASIS (PELF, figure 4.19)

x-axis Pointing forwards (anterior), corresponding with the mean direction of progression of the body during locomotion and consistent with the plane defined by a line connecting the SACR (mid-point between the PSIS) and PELF (mid-point between the ASIS) points

y-axis Pointing vertically upwards (cephalad), perpendicular to the plane created by the x- and z-axis

z-axis Pointing to the right, perpendicular to the x-axis and consistent with the attitude of the line connecting the left to right ASIS points

Consistent with the pelvis anatomical frame definitions by Cappozzo et al 1995

Figure 4.20: Axis definition for the pelvis co-ordinate system

The position of the hip joint centre of right and left limbs was derived from a mathematical algorithm developed by Bell (1990) based on anthropometric data used to generate a generic offset from the right and left ASIS respectively (section 2.5).

The hip joint centre was synonymous with the point that may be classed as the origin of the hip joint coordinate system, which is described in figure 4.21.

Hip Joint Co-ordinate System: Axis Definition	
Origin	No common origin for axes; approximately positioned at hip joint centre (HJC)
x-axis	Pointing forwards, floating axis, perpendicular to the plane created by the y- and z-axes, corresponding with the mean direction of progression of the body during locomotion while in the anatomically neutral position
$(\hat{e}_2)$	*Rotation around the x-axis describes adduction and abduction of the hip joint*
y-axis	Pointing vertically upwards, embedded in the distal femoral segment and coincident with the y-axis of the femoral co-ordinate system
$(\hat{e}_3)$	*Rotation around the y-axis describes internal rotation and external rotation of the hip joint*
z-axis	Pointing to the right, embedded in the proximal pelvis segment and parallel to the z-axis of the pelvis co-ordinate system so is consistent with the attitude of the line connecting the left and right ASIS points of the pelvis
$(\hat{e}_1)$	*Rotation around the z-axis describes flexion and extension of the hip joint*

Figure 4.21: Axis definition for the hip joint co-ordinate system

4.4.2.4 Thigh segment and femoral co-ordinate system

Figure 4.22: Thigh segment marker clusters (anterior and lateral aspects)

The technical reference frame for the thigh segment was established by positioning the thigh marker clusters on the distal thigh, superficial to the surface area with least muscle bulk, just superior to the knee (figure 4.22). This position was optimal for limiting movement of the marker cluster relative to the soft tissues during locomotion, allowing the marker cluster to best represent movement of the underlying femoral bone. Anterolateral orientation of the clusters optimised visibility of the markers during the ambulatory tasks.

Figure 4.23: Pointer trial calibration of anatomical landmarks of the femur (right lateral and right medial epicondyle)

Femoral Co-ordinate System: Axis Definition	
Origin	Knee joint centre (KJC)
x-axis	Pointing forwards, corresponding with the mean direction of progression of the body during locomotion and perpendicular to the plane created by the y- and z-axis
y-axis	Pointing vertically upwards, perpendicular to the z-axis and embedded in the line connecting the KJC and the HJC
z-axis	Pointing to the right, embedded in the line connecting the medial and lateral epicondyle points of the femur

Figure 4.24: Axis definition for the femoral co-ordinate system

Pointer trial calibration of the position of the femoral epicondyles (figure 4.23) relative to the thigh technical frame formed the basis for defining the anatomical reference frame of the femur. The origin of the femoral co-ordinate system was the knee joint centre (KJC). This point was determined by calculating the midpoint between the lateral and medial femoral epicondyles. The orthogonal axis system of the femoral co-ordinate system is described in figure 4.24.

Knee Joint Co-ordinate System: Axis Definition	
Origin	No common origin for axes; approximately positioned at KJC
x-axis	Pointing forwards, a floating axis, perpendicular to the plane created by the y- and z-axes, corresponding with the mean direction of progression of the body during locomotion while in the anatomically neutral position
($\hat{e}_2$)	*Rotation around the x-axis describes adduction and abduction of the knee joint*
y-axis	Embedded in the distal shank segment, corresponding the y-axis of the tibial co-ordinate system Pointing vertically upwards
($\hat{e}_3$)	*Rotation around the y-axis describes internal rotation and external rotation of the knee joint (shank segment relative to the thigh segment)*
z-axis	Embedded in the proximal thigh segment, corresponding to the z-axis of the femoral co-ordinate system Pointing to the right, embedded in the line connecting the medial and lateral epicondyle points of the femur
($\hat{e}_1$)	*Rotation around the z-axis describes extension and flexion of the knee joint*

Figure 4.25: Axis definition for the knee joint co-ordinate system

The knee joint co-ordinate system was defined in a similar manner to the femoral co-ordinate system, although the rotation axes were based on a non-orthogonal reference frame, following the standardised joint coordinate systems proposed by Cole et al (1993). Differences in the axis definition were applied in order to accurately describe kinematics of one segment relative to the other in a clinically applicable manner (Cole et al 1993). In this case, motion of the distal shank (child) segment relative to the proximal femoral (parent) segment describes kinematics of the knee joint. The knee joint co-ordinate system is defined in figure 4.25 and definitions are consistent with standard conventions (Cole et al 1993, Wu and Cavanagh 1995).

The rotation axes of the knee joint co-ordinate system may be visualised in figure 4.26.

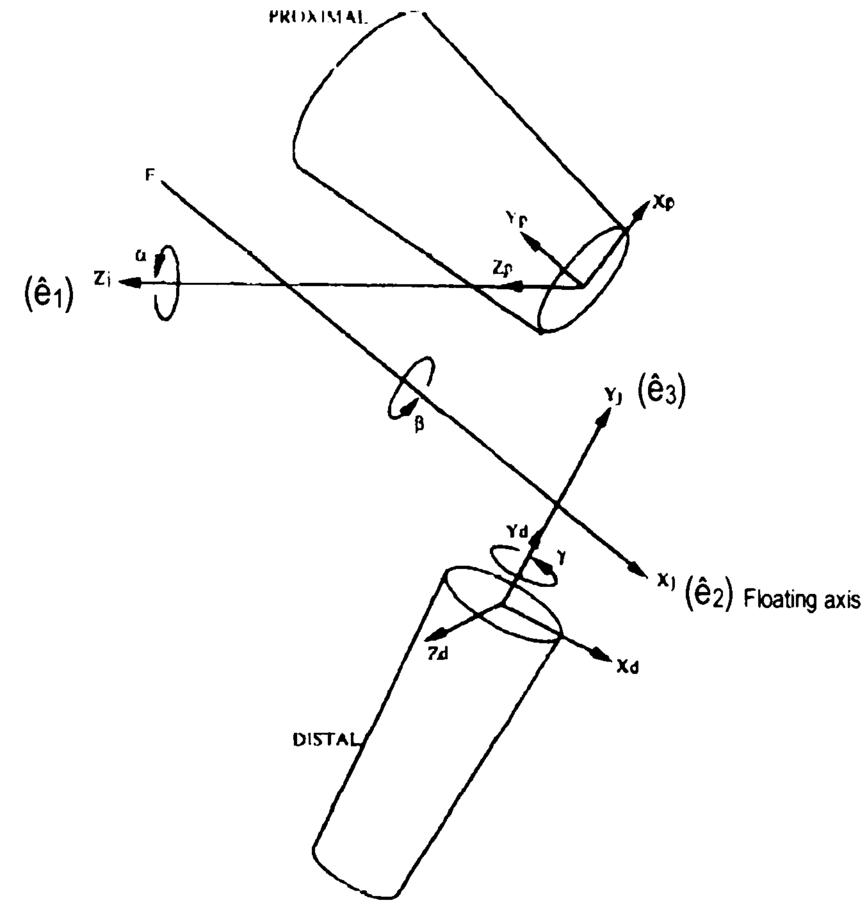


Figure adapted from Wu and Cavanagh (1995)

Figure 4.26: Joint co-ordinate system for the knee joint

4.4.2.5 Shank segment and tibial co-ordinate system

The technical reference frame for the shank segments were also marked with clusters of 4 markers. Similar to the thigh, the shank marker clusters were positioned with an anterolateral orientation and distal to the muscle bulk, below the triceps surae muscle belly (figure 4.27). This position optimised visualisation of the markers and limited movement of the marker cluster due to soft tissues displacement during locomotion.

The most prominent tip of the lateral and medial malleoli were identified with skin mounted markers (figure 4.27) and associated with the shank clusters of the ipsilateral limb using the BodyBuilder code. In order to achieve this, one static standing trial capturing all relevant markers (4 shank cluster markers and malleoli markers) was required. The malleoli markers were then removed from the subject.

This procedure was similar to the pointer trial method, but had the advantage of reducing the data capture time.

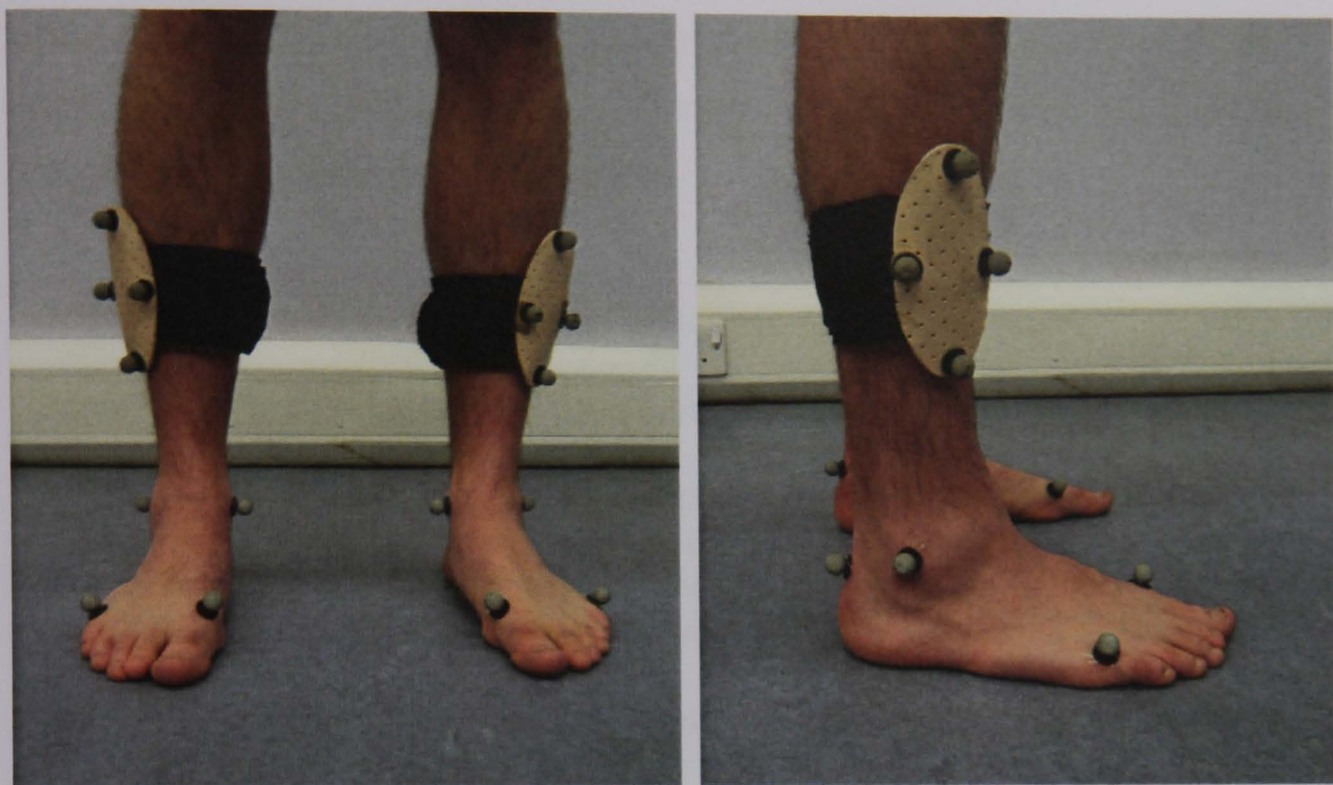


Figure 4.27: Shank segment marker clusters and malleoli markers

Although the shank segment consists of both the tibia and fibula, the model of the underlying bone has been simplified and hence, the anatomical reference frame associated with the shank technical frame will be referred to as the tibial reference frame.

Tibial Co-ordinate System: Axis Definition	
Origin	Ankle joint centre (AJC)
x-axis	Pointing forwards, corresponding with the mean direction of progression of the body during locomotion and perpendicular to the plane created by the y- and z-axis
y-axis	Pointing vertically upwards, perpendicular to the x-axis and embedded in the line connecting the AJC and the KJC
z-axis	Pointing to the right, embedded in the line connecting the medial and lateral malleoli points of the tibia

Figure 4.28: Axis definition for the tibial co-ordinate system

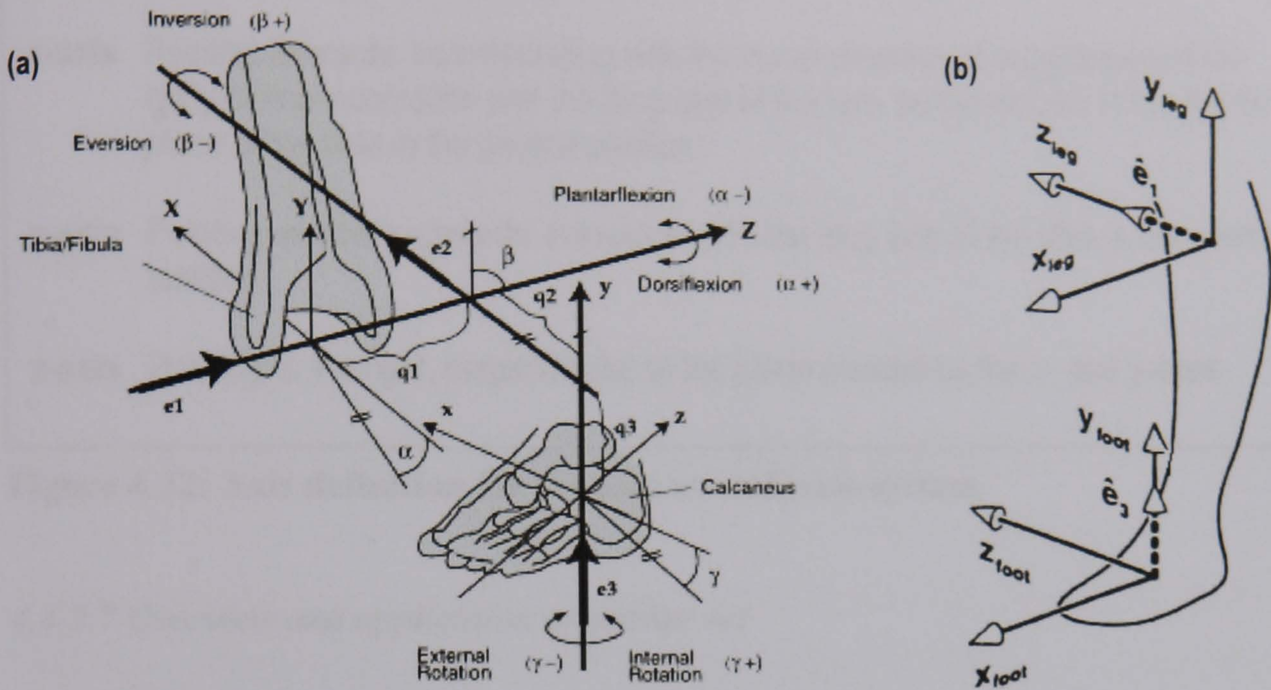
Identification of the malleoli relative to the shank cluster formed the basis for defining the anatomical frame of the tibia. The mid-point between the medial and lateral malleoli markers defined the ankle joint centre (AJC) and was taken as the origin of the tibial co-ordinate system. The orthogonal axis system of the tibial co-ordinate system is described in figure 4.28.

Ankle Joint Co-ordinate System: Axis Definition	
Origin	No common origin for axes; approximately positioned at AJC
x-axis	Pointing forwards, a floating axis, perpendicular to the plane created by the y- and z-axes, corresponding with the mean direction of progression of the body during locomotion while in the anatomically neutral position
($\hat{e}_2$)	*Rotation around the x-axis describes inversion and eversion of the ankle joint*
y-axis	Embedded in the foot (distal) segment, corresponding to the y-axis of the foot co-ordinate system Pointing vertically upwards
($\hat{e}_3$)	*Rotation around the y-axis describes internal rotation and external rotation of the ankle joint (foot segment relative to the tibial segment)*
z-axis	Embedded in the tibial (proximal) segment, corresponding to the z-axis of the tibial co-ordinate system Pointing to the right, embedded in the line connecting the medial and lateral malleoli points of the tibia
($\hat{e}_1$)	*Rotation around the z-axis describes dorsiflexion and plantarflexion of the ankle joint*

Figure 4.29: Axis definition for the ankle joint co-ordinate system

The ankle joint co-ordinate system was defined in a similar manner to the knee joint co-ordinate system. In this case, motion of the distal foot (child) segment relative to the proximal tibial (parent) segment described kinematics of the ankle joint. It must be mentioned, that the description of kinematics of the "ankle joint" had been used to describe the combined motion of the talocrural and subtalar joints, derived from the ankle joint co-ordinate system. The ankle joint co-ordinate system is defined in figure 4.29 and definitions are consistent with standard conventions (Cole et al 1993, Wu et al 2002).

The rotation axes of the ankle joint co-ordinate system may be visualised in figure 4.30.



Figures taken from (a) Wu et al (2002) and (b) Cole et al (1993)

Figure 4.30: Joint co-ordinate system for the ankle joint

4.4.2.6 Foot segment and foot co-ordinate system

The foot segment was identified by the 1st and 5th metatarsal (MT) markers, positioned on the most prominent part of the MT head, close to the joint line, and the calcaneal marker, positioned over the most prominent point of the calcaneus (figure 4.31). From this triad of markers, the anatomical co-ordinate system of the foot was derived. The origin was located at the ankle joint centre and the axes defined in figure 4.32. Definitions follow the recommended ISB conventions (Wu et al 2002).

Figure 4.31: Foot markers (lateral, anterior and posterior aspects)

Foot Co-ordinate System: Axis Definition

Origin Coincident with the AJC in the neutral anatomical position

x-axis Pointing forwards, corresponding with the mean direction of progression of the body during locomotion and the long axis of the foot; perpendicular to the frontal plane of the tibia in the neutral position

y-axis Pointing vertically upwards, coincident with the long axis of the tibia in the neutral position

z-axis Pointing to the right, perpendicular to the plane created by the x- and y-axes

Figure 4.32: Axis definition for the foot co-ordinate system

4.4.2.7 Overview and application of marker set

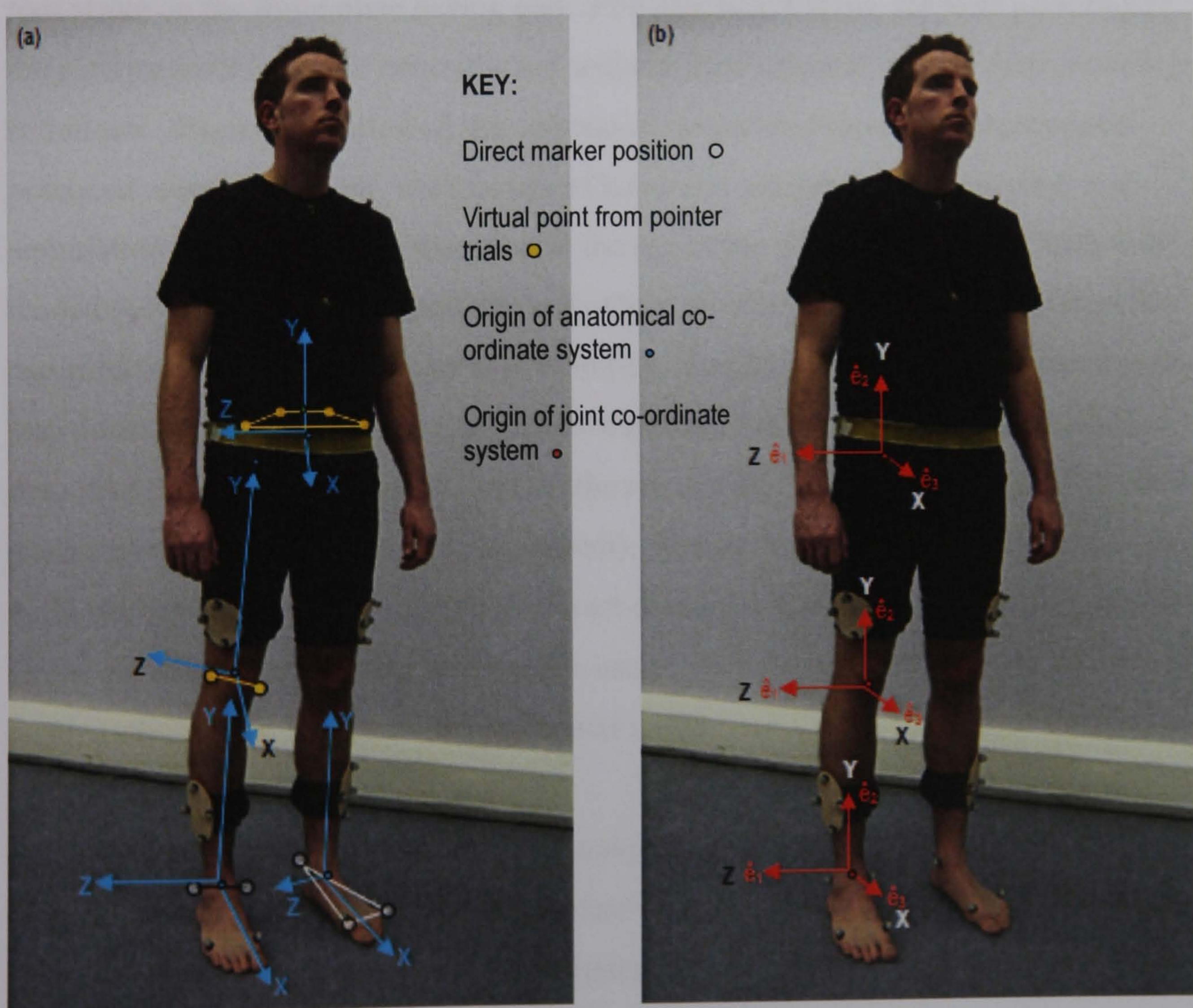


Figure 4.33: Overview of kinematic modelling method; (a) anatomical reference frames and (b) joint co-ordinate systems

Following data capture, the markers and marker clusters were labelled in Workstation and data was than processed using a BodyBuilder code (appendix 6). Virtual points were then constructed, representing the calibration points and joint centres. These points are overviewed in figure 4.33a and the aforementioned anatomical and joint co-ordinate systems superimposed in figure 4.33b. Figure 4.33 shows a unilateral representation of the co-ordinate systems used, although a bilateral model was used in practice.

4.4.3 Functional activities

The subjects were asked to perform three everyday tasks to the best of their ability, at their natural pace, including level walking, stair ascent and stair descent. Three trials were repeated for heel strike on both right and left limbs. The subjects conducted the trials from a given starting position, which was manipulated as required to achieve foot strike on the force plate during gait. For stair negotiation, subjects performed the activity as they would naturally and without instruction to modify their motion or technique. These trials allowed the operator to establish subject preferences and practiced motor behaviour, such as use of handrails, reciprocal or non-reciprocal ambulation patterns, turning direction at the top of the stairs and finally which foot tended to lead during ascent and descent. Once an adequate number of trials were captured with the subject's natural pattern (e.g. 3 right foot strikes on the force plate step during ascent), providing the subject was able, the stair trials were modified at the operator's request in order to obtain the remaining trials required (e.g. 3 left foot strikes on the force plate step during ascent). Hence three trials of data were collected for both left and right foot contact during gait, stair ascent and stair descent. Other qualitative information such as the use of walking aids and behaviours such as turning direction at the top of the stairs was noted.

In addition to the main functional tasks, subjects were asked to perform a Trendelenburg test. The Trendelenburg test was performed by standing on one leg for up to 10 seconds, while the pelvic alignment was monitored by the operator and the kinematics recorded. The Trendelenburg test was positive if the subject's pelvis dropped to the side of the unsupported limb during the single leg stand. indicating abductor muscle group weakness on the standing limb side (Downing et al 2001).

4.4.4 Subjective assessment

Following the motion analysis protocol, the subjects were provided with a physical rest period during which they were asked to complete the Oxford Hip Questionnaire (appendix 4) and Satisfaction Questionnaire (appendix 5). Questions and answer options were read to the subjects and the operator checked the correct box corresponding to the subject's response.

4.4.5 Proprioception

The subjects mounted the motion detection sense rig (section 3.2. figure 3.3 and figure 4.12) and adopted a supine lying position with a foam wedge beneath their hips to lift the buttocks and prevent contact of the posterior thigh with the surface of the plinth (figure 4.34). A PROM brace (figure 3.7) was applied to the first test limb, which was randomly selected for assessment before the other limb. The hinge of the brace was fixed in the neutral limb position to maintain a neutral knee position in full extension. The limb suspension beam was then fitted with the rotation point (figure 3.4) over the hip joint centre of the test limb. Padded clamps were placed flush with the subject's iliac crest to immobilise the pelvis bilaterally (figure 3.6). The limb was suspended in the desired position and plastic panels attached over the subject's shorts using double sided tape (figure 3.10). The electrogoniometers were then attached over the panels, again with double sided tape.

Once the operator was satisfied with the initial starting position of the limb, the electrogoniometers were zeroed. For abduction motion detection sense trials, measured in supine, the starting position of the test limb ensured that the hip was in zero degrees of abduction. For flexion motion detection sense trials, measured in side-lying, with the foam wedge removed, the starting position was standardised at 30 degrees of flexion. These starting positions were checked using a hand held goniometer, with reference to the anatomical neutral.

Standardised instructions were given (appendix 7) and the subjects were then blindfolded. Up to five practise test trials were administered. The practice trials were used to check subjects understanding of the task, that the subject was relaxed, to familiarise the subject with the procedure and to check or modify the velocity of the limb movement (section 3.3.3).

Actual trials for data collection were repeated 10 times (Grigg et al 1973). At random time intervals, the operator released the beam from the starting position without giving forewarning. The subject pressed a switch when they sensed the movement of their limb, and afterwards indicated which direction they perceived the movement to have taken place (table 3.1). Following motion detection, the operator stopped the beam movement, again at a random time interval. Therefore the timing and amplitude of beam movement were randomised between and within trials. This was done in order to limit the possibility of a learning effect. The amplitude of the beam movement never exceeded 15° of displacement from the starting position.

This protocol was repeated for abduction and flexion movements of both hips. This created four test conditions:

- Right hip abduction
- Left hip abduction
- Right hip flexion
- Left hip flexion

The order in which the test conditions were presented was randomised to eliminate bias from a practice effect. For flexion trials (figure 4.35), the PROM brace was adjusted to 20° of knee flexion before the subject assumed a side-lying position. This was to offload any stretch on the hamstring muscles which might otherwise pre-empt motion detection by eliciting a heightened sensory response triggered by stretch receptors in the muscle or mechanoreceptors (section 3.1.3.1). For stabilising the pelvis in side-lying, the clamps were placed over the uppermost ASIS anteriorly and the sacrum posteriorly. Figures 4.34 and 4.35 show the subject positioning used during abduction trials (supine) and flexion trials (side lying) respectively.

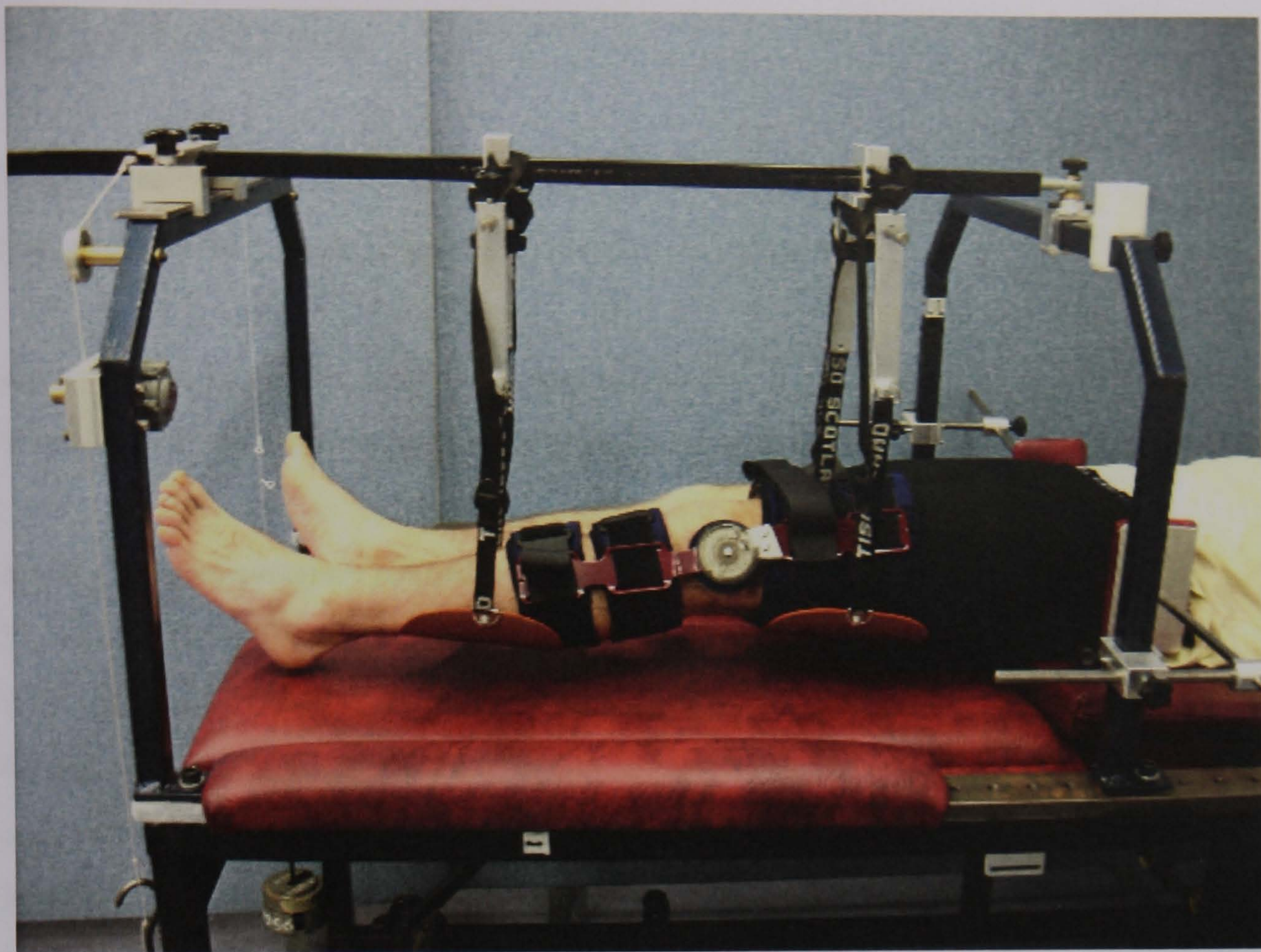


Figure 4.34: Subject positioning and set up for abduction MSD trials

Figure 4.35: Subject positioning and set up for flexion MSD trials

4.5 Pilot Studies

Trials were conducted to validate the design of the proprioception rig and test instruments and operator repeatability for all test protocols.

4.5.1 Proprioception rig validity and reliability

Validity and repeatability of the MDS protocol is documented in section 3.4.

4.5.2 Motion analysis reliability tests

To test the suitability of the cluster or direct marker system for the pelvis, a study was conducted with the subject wearing both marker sets. The results showed that during the functional movements, the ASIS markers were frequently lost from camera visibility, particularly on the stair trials, during turning and due to occlusion by arm swinging and the handrails of the staircase. The virtual ASIS and PSIS points from the pointer trials were compared to the skin marker points. The direct skin marker position was more inconsistent, with the distance between points varying. Given that this movement may cause greater error in the hip joint centre location, the pelvis cluster method was favoured. Virtual points associated with the pelvis cluster were more consistent, with the variation of the distance between the anatomical landmarks consistently less than 2mm, thus supporting the current use of pointer trials with cluster technical frames as opposed to direct skin markers (Cappozzo et al 1995).

Chapter 5 outlines the methods involved in the analysis of the kinematic and kinetic data collected during the motion assessment and the proprioception data obtained from the threshold MDS protocol.

5 Data Analysis Methods

5.1 Outline of Data Management

This chapter concerns the techniques and methods used to refine the data from their raw format into a meaningful numerical format. The various stages of the data management will be explained and justified for the processing of subjective outcomes and both motion analysis and proprioception objective data outcomes.

The application of statistical analysis will also be outlined for all data comparisons. These include intra-subject comparisons, such as comparison of outcomes between operated and control limbs and same subject temporal comparisons between 3 and 12 month data sets; and inter-subject comparisons within and between the surgical groups at each data collection session and inter-subject comparison between the 3 and 12 month data sets.

5.2 Disclosure and Management of Confidentiality Issues

Data acquired from the subjects and their associated clinical institutions were stored by secure methods in order to maintain confidentiality. Electronic data was stored on a PC with single user password protected encryption. Paper format data was converted to electronic format where possible. Remaining paper information, such as consent forms, were filed and stored in a locked cabinet. Hardware backup of electronic data was stored in a similar secure fashion.

Anonymity of subject data was maintained by replacing names and addresses with coded reference names.

5.3 Management of Subjective Data

The collection of subjective data, including Oxford Hip Scores (OHS), satisfaction questionnaires and visual analogue pain and stiffness scales were done by marking laminated charts with non-permanent marker. Immediately following the data collection session, the information was coded into numerical format. For example, individual questionnaire items of the OHS and satisfaction questionnaire were numerically scored and total scores calculated. Scoring the VAS results (figure 5.1) was done by ruler measurement (mm) of the pain or stiffness scales. Subjective levels of pain or stiffness were recorded exact to 1 decimal place.

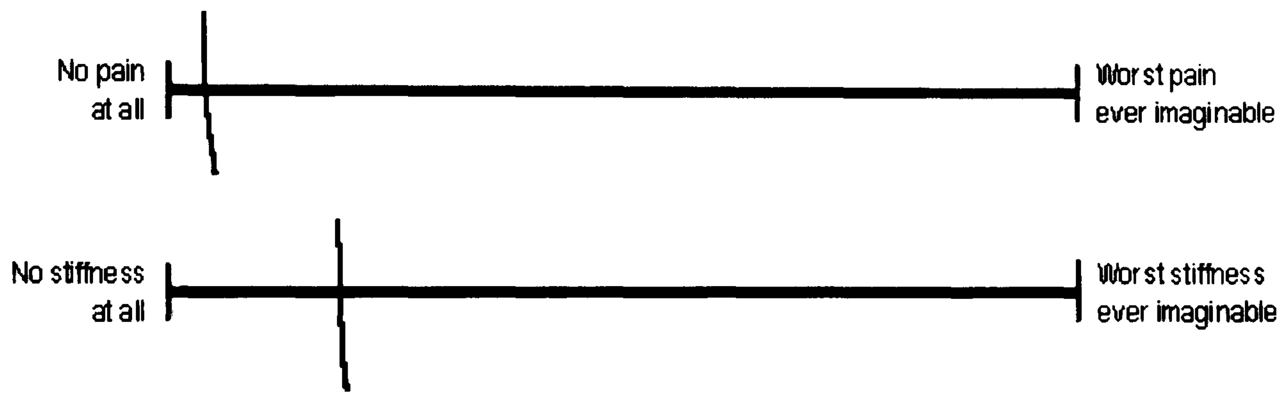


Figure 5.1: Example of subjective pain and stiffness VAS results

For all subjective measures, the results were averaged for each surgical group and then compared. Further to this, for the questionnaires, the individual item scores were also compared between the surgical groups.

5.4 Management of Motion Analysis Data

5.4.1 Processing of markers and motion trajectories

The raw motion analysis data was manipulated and processed using Vicon Workstation software (version 4.4). The trials were firstly cropped to omit frames without marker data. Markers were then labelled using a custom written marker set file (table 4.1) and the temporal events of the motion tasks were marked using the standard Workstation temporal markers. These marked the 'foot strike' and 'foot off' events for right and left lower limbs. The frame numbers of these events were also noted and were used at a later stage of the analysis to identify the beginning and end of stance phases and step cycles specific to a given data set.

Identifying the frame number at which the temporal events occurred was done by two methods. The first method was to identify 'foot strike' and 'foot off' from the force plate data. The 'foot strike' event was given by the frame number at which the ground reaction force increased above zero; the 'foot off' event was given by the frame number at which the ground reaction force returned to zero. However, not all temporal events of the gait cycle involved foot contact with a force plate, such as the consecutive 'foot strike' which followed toe off from the force plate or events marking motion of the contralateral limb. In such situations, a second method was used to identify these temporal events which involved analysis of the acceleration profiles and sagittal translation of the foot segment. For instance, 'foot strike' was defined as the point at which the foot segment decelerated and ceased forward translation of the heel marker; 'foot off' was defined as the point at which the foot segment accelerated and began forward translation of the 1st metatarsal marker.

Following the marker labelling and identification of temporal events, the marker trajectories underwent visual inspection and 'snagging'. The 'snagging' process ensured that there were no missing markers and erratic marker trajectories which would cause errors in reconstructing the position data of the joint centres following BodyBuilder processing. Such instances may cause spikes and errors in the angle and moment outputs. The 'snagging' process involved several steps. Firstly the 'gap fill' function was used to replace markers which were missing for up to 15 frames (0.125s). The 'gap fill' was limited to 15 frames as filling large gaps was known to increase the risk of incorporating errors into reconstruction of the position of the missing marker.

Providing there were at least 3 visible markers on each marker cluster during any one frame, there was sufficient kinetic data to recreate the anatomical point related to the cluster and hence recreate the anatomical frame to generate a 'stick figure' representation of the subject. Marker clusters were visually checked at the time of data collection, but on exceptional cases where there were less than 3 cluster markers for short durations (> 15 frames and <30 frames), a second 'snagging' method was used. This allowed the processor to use either of the remaining two markers on the rigid cluster and 'copy' the pattern of movement of a chosen remaining marker to reconstruct the position and orientation of the missing marker. Given that the markers of a common cluster were at a fixed distance and orientation from each other, this is a valid method of replacing missing markers when required. Nevertheless, this was limited to short durations in order to minimise the potential of introducing errors.

The final 'snagging' method used was a technique designed to 'snip trajectories' where they became erratic and replace the erratic section of the marker trajectory with a corrected smooth trajectory. Such erratic trajectories (figure 5.2) may have been produced if a limited number of cameras (less than 3) were in view of that particular marker during that instance. The method of replacing the erratic marker follows the 'gap fill' principal described above, therefore only short durations (≤15 frames) of erratic marker trajectories were replaced by this method.

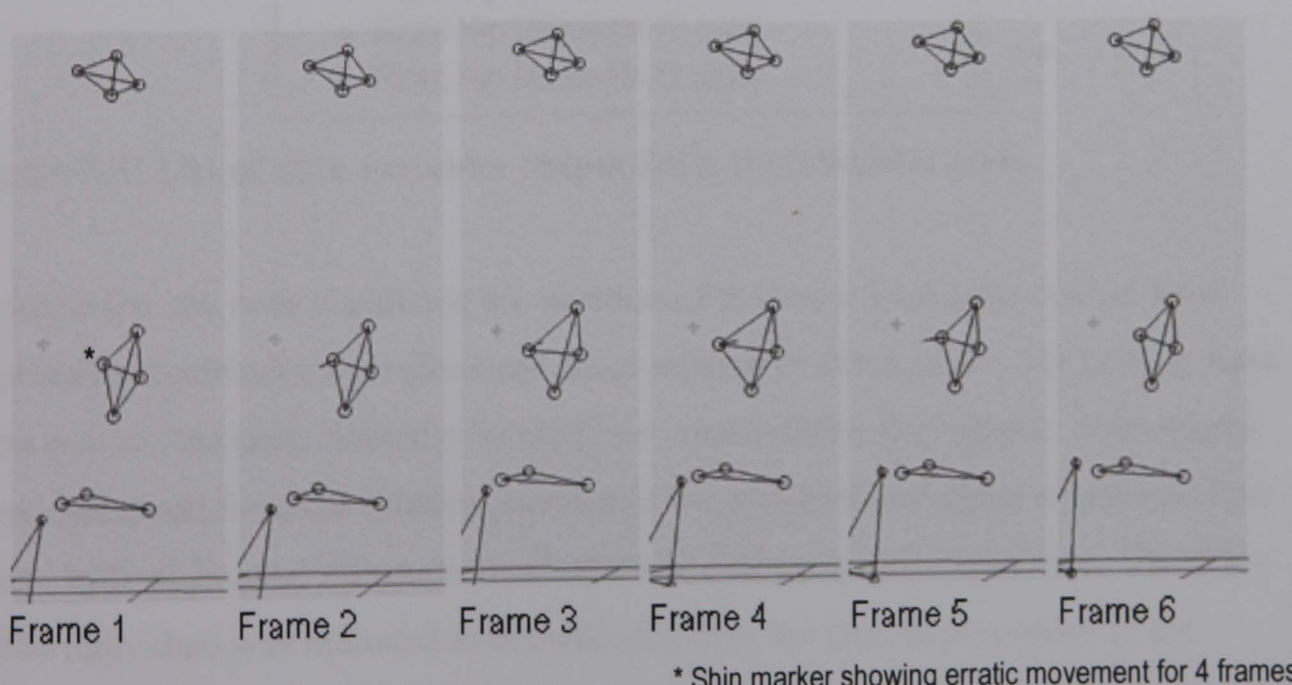


Figure 5.2: Erratic marker trajectory

Following the visual inspection and 'snagging' of markers, further smoothing of marker trajectories was done using a Woltring filter which was inherent in the Workstation software and was applied to the data while running the BodyBuilder code. The marker data was filtered using a MSE filter value of 15, as recommended for gait analysis (Peters et al 2009, Woltring 1985, 1991). Following this, the filtered kinematic data was processed using the BodyBuilder code.

5.4.2 BodyBuilder code

The BodyBuilder code (appendix 6) was ultimately used to output 3-dimensional angles and external inter-segmental moments of the right and left hip joints (Figure 5.3). The method by which this is achieved is outlined in section 4.4.2.

Raw BodyBuilder Output-List of Variables

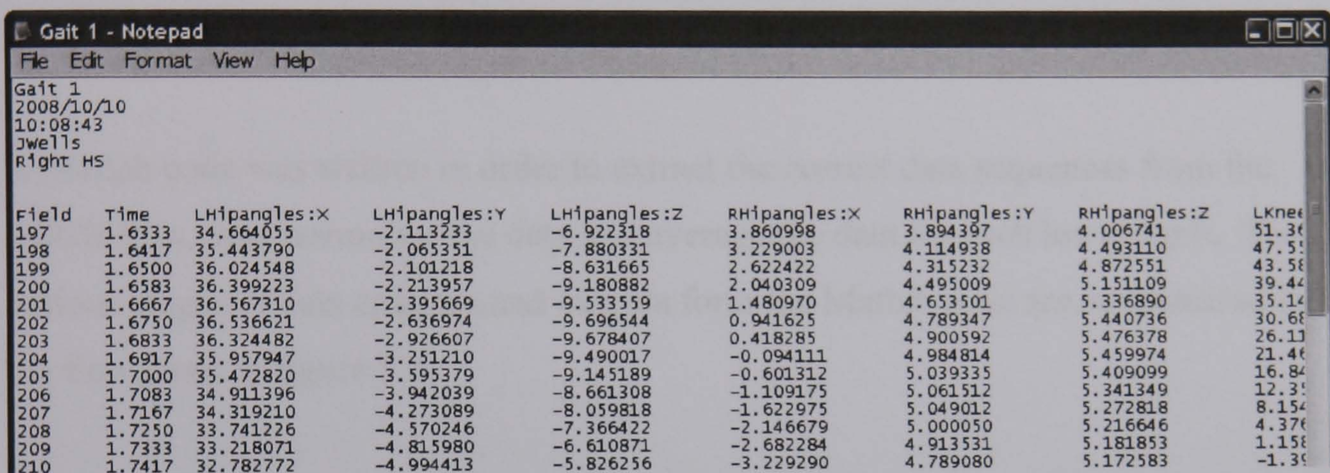
- Left Hip Angles (X axis)
- Left Hip Angles (Y axis)
- Left Hip Angles (Z axis)
- Right Hip Angles (X axis)
- Right Hip Angles (Y axis)
- Right Hip Angles (Z axis)
- Left Hip Moments (X axis)
- Left Hip Moments (Y axis)
- Left Hip Moments (Z axis)
- Right Hip Moments (X axis)
- Right Hip Moments (Y axis)
- Right Hip Moments (Z axis)

Figure 5.3: List of data variables output from BodyBuilder code

In summary, the code calculated the position of the lower limb joint centres from anatomical landmarks and referenced these relative to the markers. By joining these joint centres, the code created a 'stick figure' representing the subject. Joint angles were calculated from the relative position of the proximal and distal segments of the joint. Inter-segmental moments were calculated using inverse dynamics. The mass of the individual was included in the analysis, thus the inertial properties of the segments were accounted for within the BodyBuilder MACRO moment calculations. The BodyBuilder code used within the current study was based on a generic lower limb model file (appendix 6) used for standard motion lab assessment at the University of Strathclyde. Minor modifications to the code were performed in order to suit the marker set chosen for the current study.

5.4.3 Management of data output from BodyBuilder code

Data were output from the BodyBuilder code in ASCII format. These files contained a column of frame numbers, with their corresponding time value, and columns of data with the appropriate output in 3-dimensions (figure 5.3, figure 5.4).



Gait 1 - Notepad

File Edit Format View Help

Gait 1
2008/10/10
10:08:43
Jwells
Right HS

Field	Time	LHipangles:X	LHipangles:Y	LHipangles:Z	RHipangles:X	RHipangles:Y	RHipangles:Z	LKnee
197	1.6333	34.664055	-2.113233	-6.922318	3.860998	3.894397	4.006741	51.3
198	1.6417	35.443790	-2.065351	-7.880331	3.229003	4.114938	4.493110	47.5
199	1.6500	36.024548	-2.101218	-8.631665	2.622422	4.315232	4.872551	43.5
200	1.6583	36.399223	-2.213957	-9.180882	2.040309	4.495009	5.151109	39.4
201	1.6667	36.567310	-2.395669	-9.533559	1.480970	4.653501	5.336890	35.1
202	1.6750	36.536621	-2.636974	-9.696544	0.941625	4.789347	5.440736	30.6
203	1.6833	36.324482	-2.926607	-9.678407	0.418285	4.900592	5.476378	26.1
204	1.6917	35.957947	-3.251210	-9.490017	-0.094111	4.984814	5.459974	21.4
205	1.7000	35.472820	-3.595379	-9.145189	-0.601312	5.039335	5.409099	16.8
206	1.7083	34.911396	-3.942039	-8.661308	-1.109175	5.061512	5.341349	12.3
207	1.7167	34.319210	-4.273089	-8.059818	-1.622975	5.049012	5.272818	8.15
208	1.7250	33.741226	-4.570246	-7.366422	-2.146679	5.000050	5.216646	4.37
209	1.7333	33.218071	-4.815980	-6.610871	-2.682284	4.913531	5.181853	1.15
210	1.7417	32.782772	-4.994413	-5.826256	-3.229290	4.789080	5.172583	-1.3

Figure 5.4: Example of raw ASCII data output from BodyBuilder code

The first frame in the exported data sequence was usually greater than 1 due to the trial having been cropped to the appropriate length. In the ASCII file example featured in figure 5.4, a level gait trial with right foot strike on the force plate was cropped to start at frame (or 'field') number 197, which was when the full 'stick figure' of the subject became visible as the subject fully entered the capture volume. Hence within the data sequence was a data sub-set relating to:

- Stance phase
  - -from frame number corresponding to 'foot strike'
  - -to frame number corresponding to 'foot off'
- Gait cycle
  - -from frame number corresponding to 'foot strike'
  - -to frame number corresponding consecutive 'foot strike'

Having identified the frame numbers at which the relevant temporal events occurred (specific to each trial), the data sub-set of interest could be extracted from the ASCII file and analysed further. The data extraction was done using Matlab software. Hip moment variables were analysed by extracting the stance phase data sub-set. Hip angle variables were analysed by extracting the gait cycle data sub-set. This method was consistently applied for both level gait trials and stair gait during ascent and descent trials. For any given motion task (level gait, stair ascent gait and stair descent gait) there were 6 trials to be analysed. Of the 6 data sets (ASCII files) corresponding to these 6 trials, 3 contained data for left lower limb stance phase and gait cycle, and 3 contained data for right lower limb stance phase and gait cycle.

A Matlab code was written in order to extract the correct data sequences from the ASCII files, time normalise the data and average the data for each lower limb. The various stages of data analysis and outputs form the Matlab code are summarised in the flow chart in figure 5.5.

Raw data ASCII files (n=6)
Contained in a folder with Matlab Code

↓

Code applied to ASCII files using Matlab software

↓

Code requests frames numbers for each trial (ASCII file)
e.g. 'input start frame of trial 1'; 'input stop frame of trial 1' etc...

↓

Code extracts correct data sub-set within ASCII file, e.g. Right Hip Angles X, Y and Z
Data is extracted for all 6 trials (ASCII files) according to frame number input

↓

Code time normalises data to 100% of gait cycle
by interpolation of data sub-set to 100 points for right and left lower limbs

↓

Data is averaged for right and for left lower limbs

↓

Average and peak values are output in Matlab software and figures generated

Figure 5.5: Flowchart of Matlab processes in application to motion analysis data

The average data set for right and left lower limbs is output over 100 points. Time normalisation of the values is used to facilitate intra-subject comparison of data over repeated trials and inter-subject comparisons of the angle or moment outputs. The average and peak values which are output from the Matlab code are then saved in excel spreadsheets.

In addition to outputting values, the Matlab code also produced graphs plotting the 3-dimensional angle or moment waveforms against time for all 3 trials of each lower limb, the average and standard deviation of the 3 trials and the average profiles for operated and control limbs for the subject of interest.

The data saved in excel spreadsheets was also used to graph and calculate average values for inter-subject comparisons. Averages were collated for both standard THR and resurfacing surgical groups and then compared.

5.4.4 Additional data outputs

The mean velocity of motion was recorded for the step cycles during level gait, stair ascent and stair descent. This data were sourced from the Workstation software by plotting the velocity of displacement of the PELF point in the X-axis and noting the average velocity. Velocity of motion was recorded from the PELF point (origin of the pelvis segment) for several reasons. Firstly, it was a favourable point as it shared a close proximity with the estimated COM position of the body, therefore was assumed to best represent full body velocity of motion. Secondly, the PELF point was in the midline of the body. During the movement tasks described, displacement was in the X-axis of the global co-ordinate system, corresponding to the direction of progression of the subject within the lab. Points in the midline provided better accuracy in measuring full body displacement in the X-axis, as they are subject to relatively less Y-axis (up and down) and Z-axis (side-to-side) displacement.

5.5 Management of Proprioception Data

5.5.1 Raw motion detection sense data

Data collected during MDS testing was visualised in real time. This allowed real time inspection of the data for smoothness of motion and alteration of the viscous damper to achieve the correct velocity of limb motion (section 4.4.5). The data was collected using a custom written virtual instrument (VI) in LabView (version 8.0), developed specifically for the MDS test procedure. The VI collected 3 raw channels of data as shown in figure 5.6. The first channel displayed data from goniometer number 1 (angle, degrees), positioned on the limb; the second from goniometer number 2 (angle, degrees), positioned on the beam of the proprioception rig; the third channel displayed data from the event marker switch (voltage, volts). All channel data were synchronised and plotted against time (seconds).

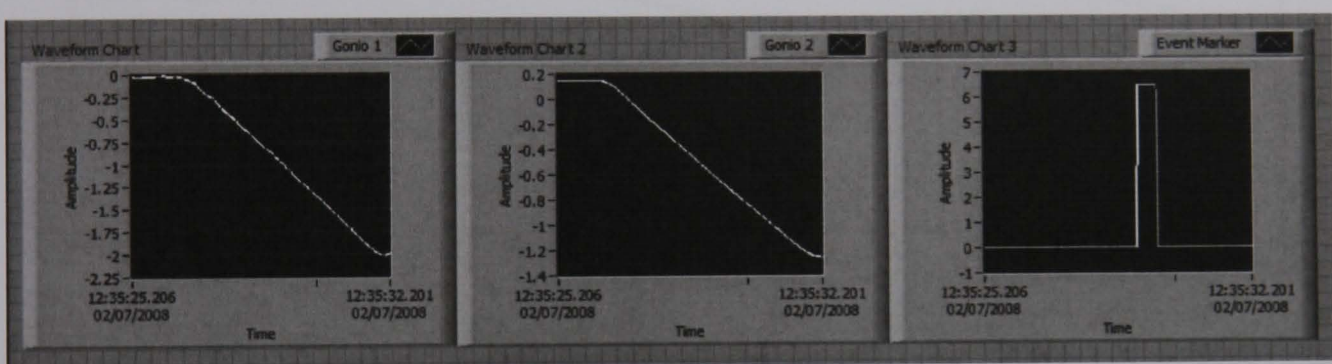


Figure 5.6: Example of real time 3-channel raw MDS output from LabView

The raw data from all three channels were automatically saved following each trial within a LabView instrument file. This file could be opened and saved as an ASCII file, displaying a column of time and angle values for each of the goniometer channel outputs and a column of time and voltage for the event marker channel.

5.5.2 Management of motion detection sense data

Ten trials were collected for each of the 4 movement conditions. All raw data were processed using a standardised data management process. The ASCII files produced from the LabView VI were processed using Matlab. A custom developed code produced figures displaying multiple channels in graphical format. The first displayed all 3 channels of data on a single graph as shown in figure 5.7.

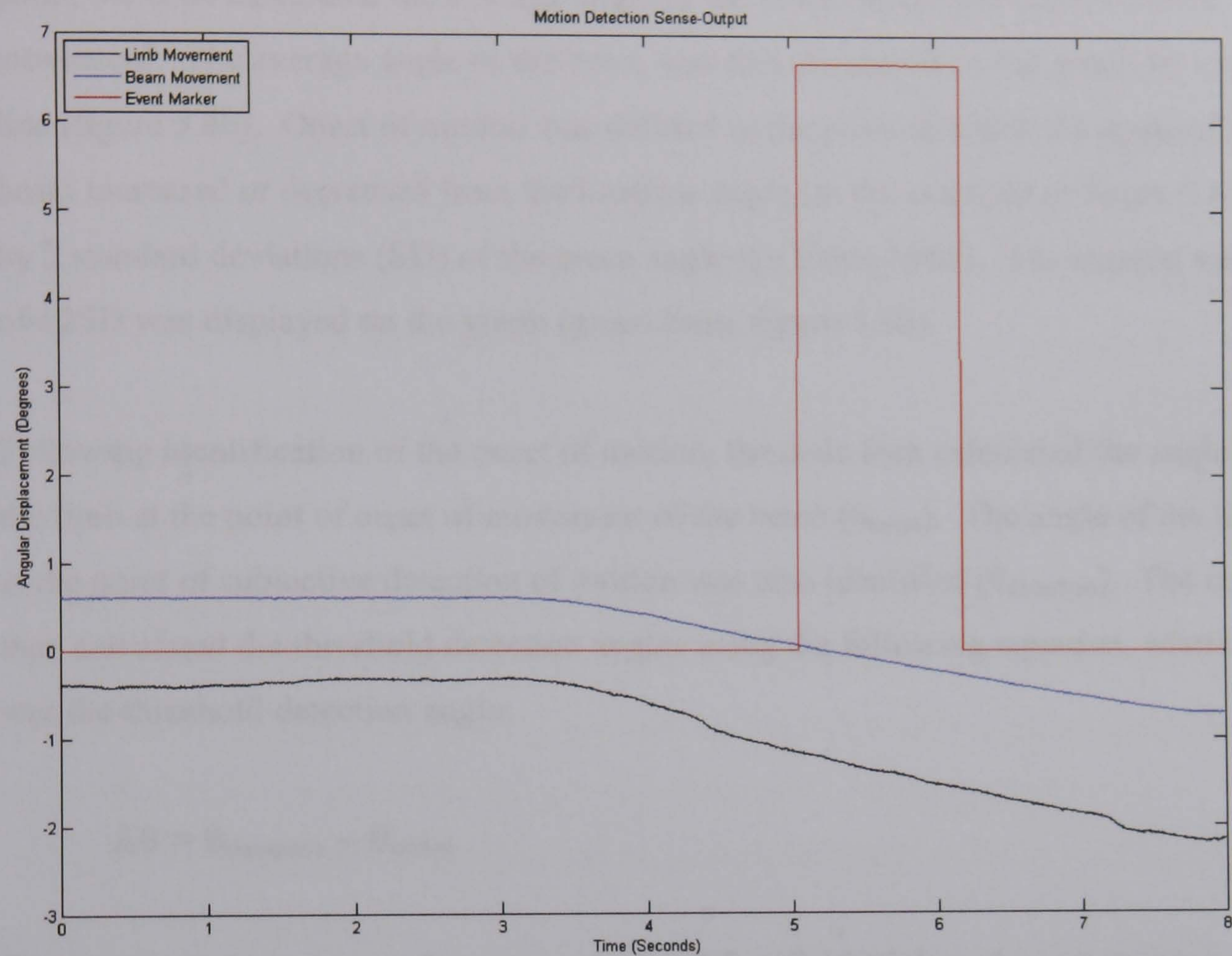
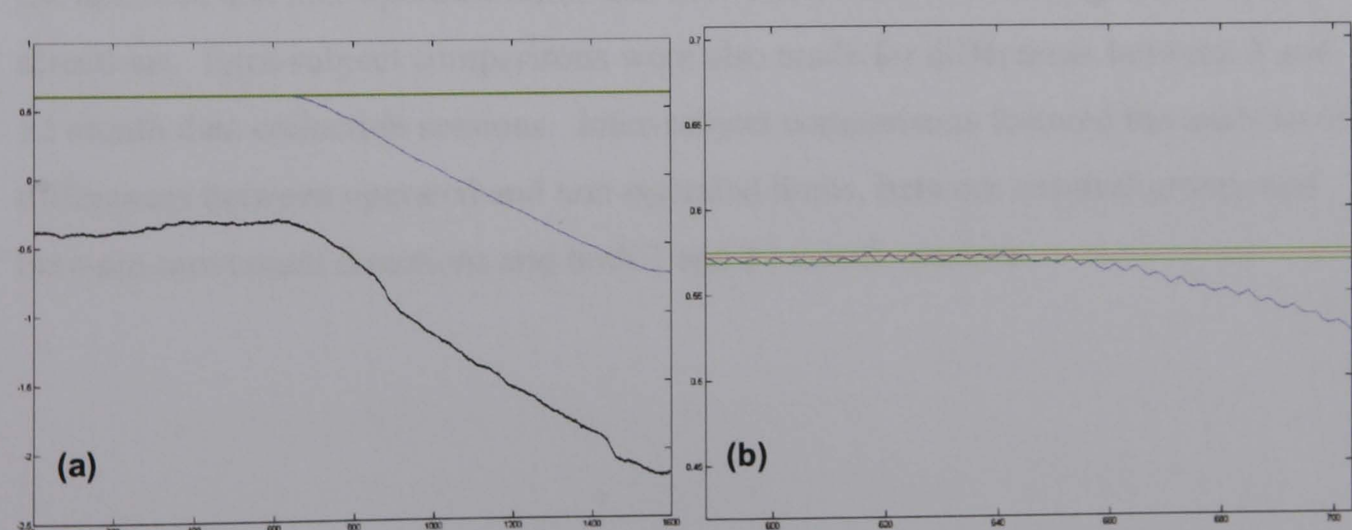


Figure 5.7: Example of 3-channel MDS data



(a): Figure created by code to identify onset of motion of the beam

(b): Magnified view of beam profile following identification of onset of motion of the beam (red: mean angle at onset; green: 2SD)

Figure 5.8: Goniometer output and identification of onset of movement

The second produced a graph of the goniometer outputs, displaying movement of the limb and the beam (figure 5.8). On this graph, a moveable cursor was used to select the point at which the beam movement appeared to commence. On selecting this point, the code calculated the average angle of the beam before the onset of beam movement. The average angle of the beam was then displayed on the graph by a red line (figure 5.8b). Onset of motion was defined as the point at which the angle of the beam increased or decreased from the baseline angle (in the example of figure 5.8) by 2 standard deviations (SD) of the mean angle (Di Fabio 1987). The angular value of ±2SD was displayed on the graph (green lines, figure 5.8b).

Following identification of the onset of motion, the code then calculated the angle of the limb at the point of onset of movement of the beam (θ_{onset}). The angle of the limb at the point of subjective detection of motion was also identified ($\theta_{detection}$). The code then calculated the threshold detection angles using the following equation, where $\Delta\theta$ was the threshold detection angle:

$$\Delta\theta = \theta_{detection} - \theta_{onset}$$

The threshold detection angles were calculated for all 10 trials and average values were calculated for each movement condition. Outcomes were compared between the operated and non-operated limbs and between flexion and abduction movement directions. Intra-subject comparisons were also made for differences between 3 and 12 month data collection sessions. Inter-subject comparisons featured the analysis of differences between operated and non-operated limbs, between surgical groups and between movement directions and both 3 and 12 month sessions.

5.6 Statistical Methods

5.6.1 Subjective data statistics

Descriptive statistics were applied to the subject data, reporting frequencies, ranges, means and standard deviation of the means for the overall subject group and for the data according to subject group. Descriptive similarities were contrasted between subjects and between the arthroplasty groups to determine the homogeneity of the sample. Variables included the surgeon, reason for arthroplasty, age, gender, height, body mass, activity variables and employment, which were noted from patient records and informal conversation. Formal subjective data recorded pain, stiffness, satisfaction and OHS.

The results of the questionnaire data were analysed using a repeated measures ANOVA, testing the fixed effects and interaction effects between the THR groups and the within-subject variance over time for repeated measures. Generalised linear model ANOVA's were used to test three repeated measures of the OHS (pre-operative, 3 months and 12 months) and two repeated measures for the satisfaction scores (3 and 12 months).

The visual analogue scores were also assessed for variability due to arthroplasty type and variability over repeated measures (3 and 12 months) using a generalised linear model ANOVA.

Finally, the interaction effects of the subjective scores were examined by performing a correlation analysis on the data for all subjects.

5.6.2 Statistical methods for motion analysis data

The motion analysis data outcomes of hip angles and external moments during the three ambulatory tasks were assessed. Specifically the dependent variables of mean peak values of the 3-D angles and moments for each subject were assessed for variance according to the within-subject variables (limb: operated and control; and time: 3 and 12 months) and the between subject variable of arthroplasty type (resurfacing and standard THR). The dependent variables were assessed within groups according the mode of ambulation and a general linear model 2-by-2-by-2 (2*2*2) repeated measures ANOVA was applied to the data to assess the effects of the within- and between-subject variables and the existence of interaction effects.

Following this analysis the effect of additional between-subject variables were assessed, such as the effects due to surgeon, age and gender for example.

5.6.3 Statistical methods for proprioception data

The MDS data was tested for normality and t-tests, or equivalent non-parametric tests (Mann-Whitney) were used to test within-subject variability between the operated to the control limb (paired t-test), within-subject variability between 3 and 12 months (paired t-test) and between-subject variability due to arthroplasty type (2-sample t-tests).

The chapters to follow will present the results of the data recording and data collection sessions.

6 Descriptive Statistics and Subjective Data

The result of the study will be divided into three parts, forming three chapters, which will report the independent variables relating to the sample group (chapter 6) and present the dependent variables measured during the objective tests (motion analysis in chapter 7; proprioception in chapter 8).

This chapter accounts for the characteristics of the sample group and will present descriptive characteristics and the results of the subjective tools used during data collection.

6.1 Formation and Refinement of the Sample Group

6.1.1 Recruitment

Between November 2005 and December 2007, the case records of 144 patients within the study age range (40-60 years) who were due to have THR underwent triage review pre-operatively. Of these patients, 56 were excluded immediately from invitation to the study due to co-morbidities within their past medical history, such as a history of other joint arthroplasty, previous orthopaedic surgery affecting the lower-limbs, previous lower-limb fractures or complicated multiple-joint pathology, such as rheumatoid arthritis. Of the remaining 88 patients, 36 (40.9%) were recruited to the study in response to verbal invitation by the researcher. The residual 52 patients were excluded from the study either pre-operatively or post-operatively due to the reasons documented in table 6.1.

The majority of subjects (61.5%) invited to take part in the study were unable to participate due to evidence of moderate to severe OA of the non-operated control hip. This excluding factor was discriminated by X-Ray diagnosis determined by the surgeon and/or subjective report of 'moderate' or 'severe' joint pain determined by the patient. A small amount of patients were planned to have resurfacing arthroplasty intervention pre-operatively but were converted to a standard THR intra-operatively (5.8%). This was either due to the presence of cysts or bone necrosis affecting the femoral head which were not fully apparent from pre-operative X-ray assessment (n=2) or due to notching of the femoral head during the reaming process (n=1). The post-operative complication which eliminated 2 subjects from the study was femoral nerve palsy in both cases. Two patients who had complained of mild intermittent low back pain pre-operatively experienced a continuation and exacerbation of their symptoms post-operatively and could not participate in the study.

Table 6.1: Reason for exclusion from study

REASON FOR EXCLUSION	Number of Patients	% of Total Excluded
Moderate to severe OA in contralateral hip	32	61.5
Intra-operative conversion of arthroplasty	3	5.8
Post-operative complications	2	3.8
Lower back pain	2	3.8
Work commitments	6	11.5
Distant geographical location	2	3.8
Refused	4	7.7
Unable to contact	1	1.9
TOTAL	52	100%

Inability to take part in the study due to work commitments was the second most common reason for exclusion. This reflected aspects of the employment status of some of the sample population, who generally received no pay for time away from work and could not afford to, or chose not to take leave for such voluntary scientific purposes. Of the 4 subjects who refused to take part, their reasons were generally not stated, however, one subject refused participation on the basis that she was not comfortable wearing shorts in the presence of others (researcher and laboratory assistant). Finally, due to a change in contact details the researcher and associated clinical staff were not able to contact one of the subjects following discharge from hospital and he/she was consequently withdrawn form participation in the study.

6.1.2 Initial sample group selected for the study

Following recruitment and informed consent, 36 patients (n=22 standard THR; n=14 resurfacing arthroplasty) entered the study and underwent all or part of the test protocol. Omissions from the data collection will be explained further during this chapter and in the following results chapters in the context of the relevant objective assessment method.

The 3 month data collection session highlighted 5 subjects from the 36 tested who were either not homogenous with the overall sample group or displayed character traits which retrospectively lead to ambiguity whether they continued to fit within the inclusion criteria of the study. Hence these subjects became outliers within the overall group (table 6.2).

The first category of outliers within this group of 5 related to subjects who were reliant on walking aids (n=2). Verbal pre-assessment communication with these subjects revealed that they required the use of a crutch or walking stick outdoors only. However, their ambulation indoors was unsteady and appeared to lack confidence and there was an impression that the subjects were attempting to appease the researcher which lead to an implied exaggeration of their abilities. In addition to this were subjects (n=2) who showed a similar behaviour in relation to their pain. This second category of outliers insisted their pain was within the thresholds of the study criteria ('mild' hip pain acceptable, appendix 3) but during the test session showed signs and symptoms of 'moderate' to 'severe' pain which got worse with time and physical effort. These patients showed obvious pain behaviours which lead to them being withdrawn from the 3 month data collection session before completion. Specifically, following the motion analysis tests, one subject could only complete two of the proprioception test conditions and the other subject withdrew prior to starting the proprioception test protocol. Finally, the last subject was highlighted as an outlier due to the display of an unusual cognitive and behavioural affect. Despite full consent and insistence on her desire to take part in the study, this subject showed erratic patterns of physical motion during the motor tasks which were often associated with pauses or verbal outbursts during the trials, leading to a lack of repeatable data being collected. Additionally, aspects of the motion analysis tests revealed fundamental differences in the patterns of motion of these 5 subjects, such as a non-reciprocal 'step-to' stair gait pattern combined with use of the handrails, which could not be directly compared with the remaining group.

Table 6.2: Standard THR subject exclusions from 3 month data analysis

REASON FOR EXCLUSION	Number of Patients	% of Total Excluded
Reliant on walking aids	2	40
Pain (greater severity than 'mid') during testing	2	40
Cognitive and behavioural issues	1	20
TOTAL	5	100%

Therefore these outliers skewed the homogeneity of the initial sample group. All five outliers were subjects with a standard THR. The exclusion of these subjects appears to neglect the intention-to-treat principle (http://www.cochrane-net.org/openlearning/html/mod14-4.htm), however, application of the initial exclusion criteria (figure 4.1) would have caused all but one of the subjects (with cognitive and behavioural issues) to be excluded, but the indications for exclusion were not revealed until during the test sessions. Removal of these outliers was desirable in order to better balance the homogeneity of the functional abilities within and between the groups. Exclusion of these subjects had the effect of reducing the standard THR subject group from 22 to 17 subjects. Since the resurfacing arthroplasty group had a total of 14 subjects, the removal of the 5 functional outliers from the standard group did not adversely affect the balance of the subject numbers between groups.

6.1.3 Final sample group

The final study group (n=31) was considered to be a homogenous sample with a similar threshold of functional activity. All subjects were comfortable and accustomed to walking without aids, indoors and outdoors. Most subjects, with the exception of 2 with standard THR, were able to perform stair negotiation with a reciprocal stair gait pattern (exception n=1) and without support from the handrails (exception n=1).

The total sample group reduced in numbers by 12 months to 20 participants (figure 6.1), but the group ratio improved with equal numbers of subjects within each group (n=10 standard THR; n=10 resurfacing) at 12 months. The reasons accounting for the reduction in subject numbers are documented in table 6.3. These mainly related to the bilateral nature of the subjects' hip pathology by the late follow-up stage.

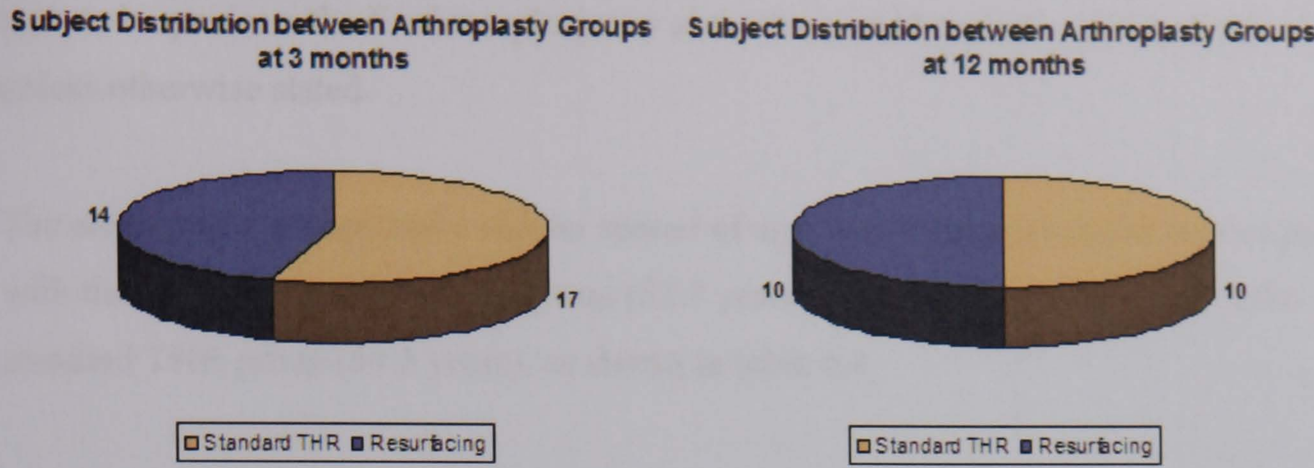


Figure 6.1: Subject distribution according to arthroplasty group and assessment phase

Table 6.3: Reasons for reduced follow-up at 12 months

REASON FOR SUBJECT DROP-OUT	Number of Patients	% of Total Drop-out
Arthroplasty of contralateral hip	3	27.3
Pain ('moderate' or 'severe') in contralateral hip	3	27.3
Work commitments	2	18.1
Poor general health	1	9.1
Refused (self-conscious in clothing for testing)	1	9.1
Moved abroad	1	9.1
TOTAL	11	100%

Hence the study group represented a non-randomised, controlled group. The non-operated limb allowed within-subject controlled comparisons. No additional control group without hip arthroplasty was assessed and subjects could not be randomised to the arthroplasty groups due to the ethical issues surrounding consent for hip arthroplasty and the additional risk factors associated with the hip resurfacing procedure (section 2.4.5). Given the relatively low numbers recruited to the study, analysis of the data was not performed according to a paired design.

6.2 Descriptive Characteristics of the Sample Group

6.2.1 Descriptive data of the sample group

Subject characteristics of the 31 participants (n=17 standard THR; n=14 resurfacing arthroplasty) were compared within and between their arthroplasty groups. The data reported represents the final sample group at the 3 month data collection session unless otherwise stated.

The arthroplasty groups had a similar spread of ages and were of a similar mean age, with the resurfacing arthroplasty group (52.7 years) being slightly younger than the standard THR group (54.3 years), as shown in table 6.4.

Table 6.4: Age data for sample group

AGE (years)	Standard THR	Resurfacing	All
Max	60	60	60
Min	44	42	42
Mean	54.29	52.71	53.58
SD of Mean	5.19	5.66	5.38

The gender distribution within the arthroplasty groups is displayed in figure 6.2, showing that the standard THR group had a greater proportion of females compared to males at both 3 and 12 months, while the resurfacing group had a greater proportion of males compared to females at both 3 and 12 months. Nevertheless, at 3 months the numbers of male subjects were well balanced between the arthroplasty groups and both groups has a similar balance of male-to-female ratios at 12 months.

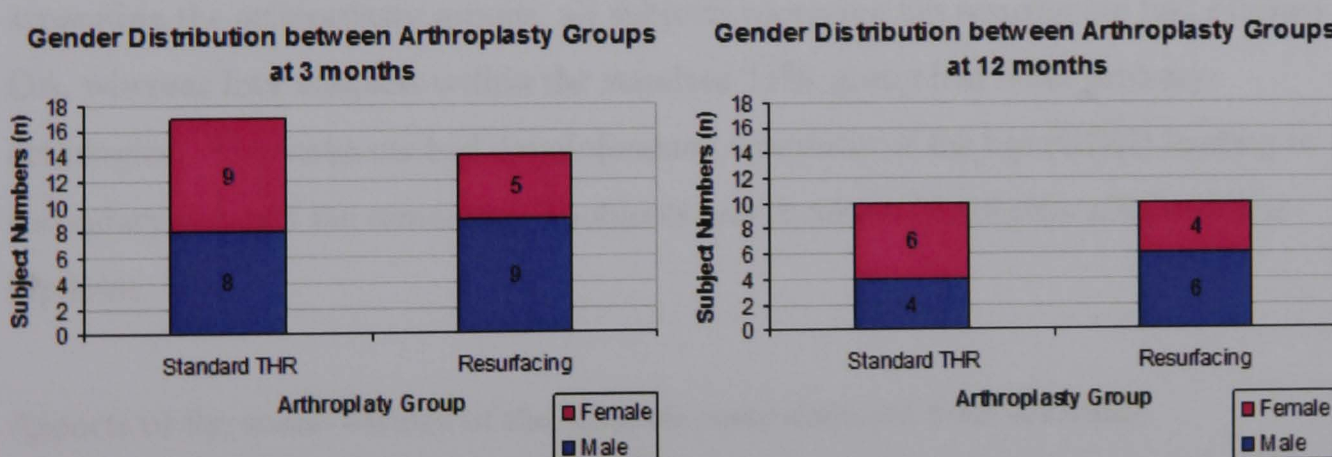


Figure 6.2: Gender distribution according to arthroplasty group at 3 and 12 months

The arthroplasty groups were also compared for height and body mass characteristics of their respective subjects. The height characteristics of the groups were similar, with mean heights of 1.67m (±0.07m; range 1.57-1.81m) for the standard THR group and 1.71m (±0.10m; range 1.54-1.87m) for the hip resurfacing group.

Body mass was recorded at both 3 and 12 months post-op. The standard THR group had a greater spread of body mass within the group (range 59.5-137.7kg; mean 85.8kg ±24.03) compared to the resurfacing group (range 52.5-91kg; mean 77.74kg ±11.68). The distribution of body mass in the standard THR group was positively skewed due to several outliers with a relatively large body mass. Without these outliers, the body mass characteristic were similar between the groups. Nevertheless, normalisation of the moment data relative to body mass eliminated the effect of this dissimilarity of between group data on the outcomes.

Hence, aside from a slight gender bias, the arthroplasty groups displayed similar subject characteristics. The between-group variability of independent variables derived from the subject demographics was therefore thought to be minimal.

6.2.2 Aetiology and social characteristics of the sample group

When assessing the reasons for orthopaedic intervention, the majority of the sample group (87.1%) required hip arthroplasty for primary OA of the hip. Moreover, when separating the arthroplasty groups, all subjects receiving hip resurfacing had primary OA, whereas four subjects within the standard THR group had other primary aetiologies. Two subjects had developmental dysplasia of the hip (DDH) leading to secondary OA and the remaining 2 subjects had rheumatoid arthritis affecting one hip joint.

Aspects of the social history of the subjects were deduced from informal conversation and hospital case notes to form an impression of the activity levels of the subjects. Most subjects were employed within a job and returned to work shortly after 3 months post-op, as advised by clinical guidelines. Some subjects had early retirement and some were unemployed. The employment status of subjects was uninfluenced by the orthopaedic event in all cases. The spread of employment characteristics of the total sample is shown in figure 6.3, specific to the arthroplasty groups.

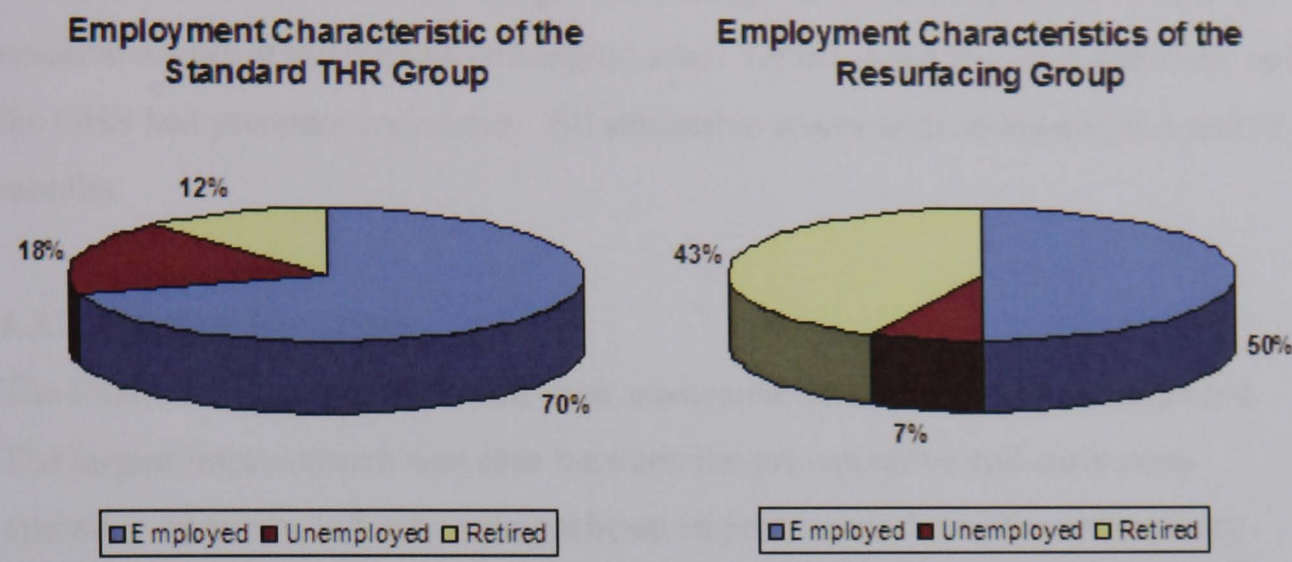


Figure 6.3: Employment distribution according to arthroplasty group

Figure 6.3 shows that the resurfacing group had a greater proportion of retired subjects than the standard THR group and the standard THR group had a relatively greater proportion of unemployment within their subject group.

A formal recording tool quantifying social and leisure activity of the subjects was not employed during the test procedure. However, the conversational information gathered from the subjects was used by the researcher to retrospectively score a UCLA scale and quantify the relative activity levels of the arthroplasty groups. The UCLA scoring system (appendix 8) was adopted as a diverse spread of activities (modes, frequency) were conveyed during informal conversation. For example, some subjects were very sedentary in their activity, with minimal or no leisure activity and their core activity conducted at work (if applicable); whereas some subjects were very active during work and/or leisure and one even conducted extreme sports (skiing) on a relatively frequent basis. Overall, the results showed that the resurfacing group tended to be more active (mean UCLA score of 8 ±1.5) than the standard THR group (mean UCLA score of 5 ±1.1). Indeed several subjects in the resurfacing group were accustomed to regular hill walking and the subject who skied was also in the resurfacing group.

6.3 Results of the Subjective Data Collection

Pre-operative subjective data was gathered using the Oxford Hip Score (OHS) by research nurses at the respective hospital sites. Of the subjective data gathered, only the OHS had preoperative scores. All subjective scores were measured at 3 and 12 months.

6.3.1 Oxford Hip Score

The total OHS improved between each assessment phase, as shown in figure 6.4. The largest improvement was seen between the pre-operative and early post-operative intervals, indicating a significant improvement due to the arthroplasty intervention. The improvement continued to a lesser extent between 3 and 12 months ($p=0.382$), implying some mild improvement in subjective functional scores with post-operative healing, as detected by the OHS.

The results of the OHS were similar between the arthroplasty groups at the pre-operative stage and at the 3 month post-operative test interval. At 12 months, there appeared to be a trend difference between the groups for the total OHS, with the resurfacing group scoring lower, indicating a better outcome. Statistical analysis showed that the between-group differences were not statistically significant at any assessment phase ($p=0.925$).

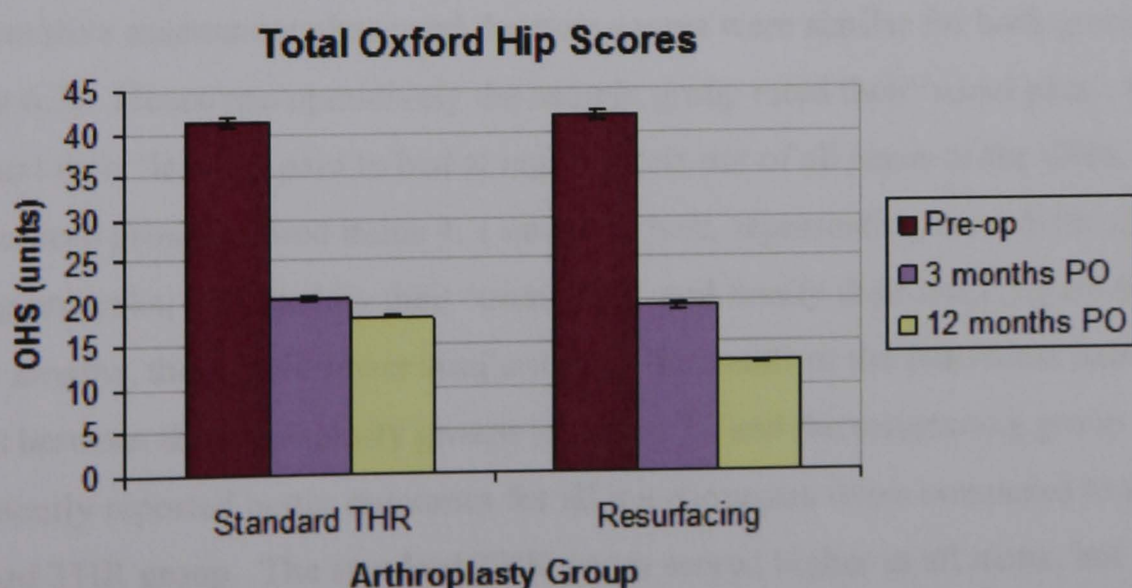


Figure 6.4: Results of total OHS according to arthroplasty group

Figure 6.4 also shows a reduction in the between-subject variability of the total OHS from pre-operative scores to 3 months scores and from 3 months to 12 months, as indicated by the smaller standard deviation bars of the histogram at each consecutive assessment phase.

Table 6.5: OHS results summary

OHS SCORE CATEGORY	Pre-op STHR	Pre-op RA	3 months STHR	3 months RA	12 months STHR	12 months RA
Item 1	4.9	4.5	1.9	2.1	1.6	1.0
Item 2	2.9	3.3	1.8	1.5	1.5	1.2
Item 3	3.4	3.2	1.6	1.4	1.7	1.1
Item 4	3.4	3.4	2.3	2.2	1.6	1.2
Item 5	2.9	2.8	1.9	1.5	1.9	1.0
Item 6	2.4	2.7	1.4	1.2	1.2	1.0
Item 7	2.9	2.9	1.6	1.5	1.4	1.2
Item 8	3.1	3.5	1.4	1.6	1.3	1.0
Item 9	4.2	4.1	1.9	2.1	1.4	1.0
Item 10	3.4	3.7	1.5	1.4	1.4	1.0
Item 11	3.8	3.6	1.6	1.4	1.7	1.0
Item 12	3.9	4.2	1.6	1.3	1.6	1.0
Sum	**41.30**	**41.90**	**20.45**	**19.28**	**18.30**	**12.70**
Max	4.86	4.50	2.31	2.21	1.90	1.20
Min	2.43	2.70	1.38	1.21	1.20	1.00
Mean	3.44	3.49	1.70	1.61	1.53	1.06
SD of mean	0.67	0.57	0.27	0.35	0.20	0.09

STHR: Standard THR group
RA: Resurfacing Arthroplasty group

The full data set for the OHS results is shown in table 6.5. Assessing the individual items of the OHS showed that both groups scored items 1, 9 and 12 highest at their pre-operative assessment phase and the item scores were similar for both groups (figure 6.5). Hence pre-operatively the sample group rated their 'usual pain', their limp and their 'level of pain in bed at night' worst out of all items of the OHS. At 3 months both groups scored items 4, 1 and 9 highest, representing most difficulty putting on socks, followed by their 'usual pain' and finally their limp (figure 6.6). By 12 months, there were fewer similarities in the trends of the individual item results between the arthroplasty groups (figure 6.7), and the resurfacing group consistently reported better outcomes for all questionnaire items compared to the standard THR group. The standard THR group scored higher in all items, but highest for items 5, 3 and 11 (doing household shopping, getting in and out of a car and pain limiting work/housework respectively). The resurfacing group scored highest for items 2, 4 and 7 equally (washing and drying their body, putting on socks and climbing a flight of stairs). Overall, there was little carryover in the scoring patterns between 3 and 12 months for either group, and little trend similarity in the scoring between the groups at 12 months.

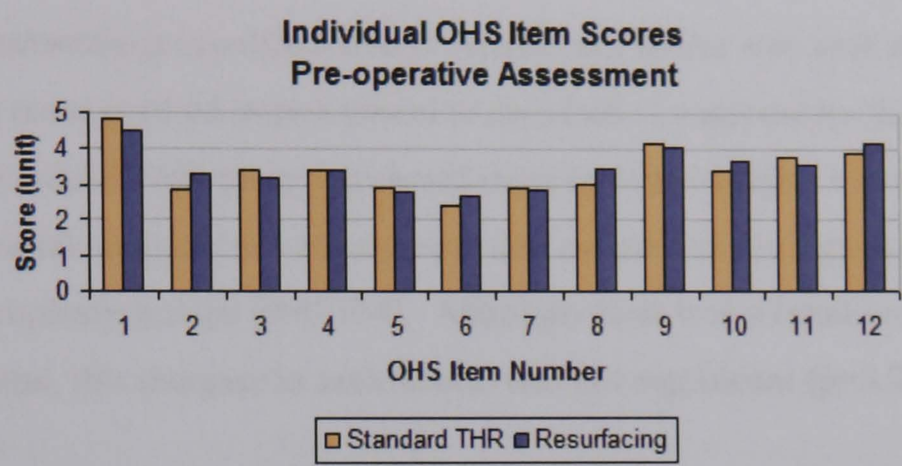


Figure 6.5: Comparison of pre-op individual item OHS results between groups

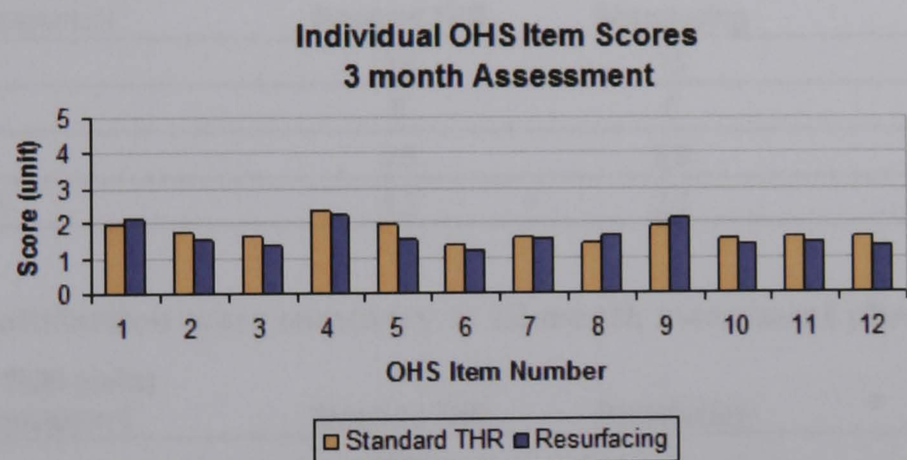


Figure 6.6: Comparison of 3 month individual item OHS results between groups

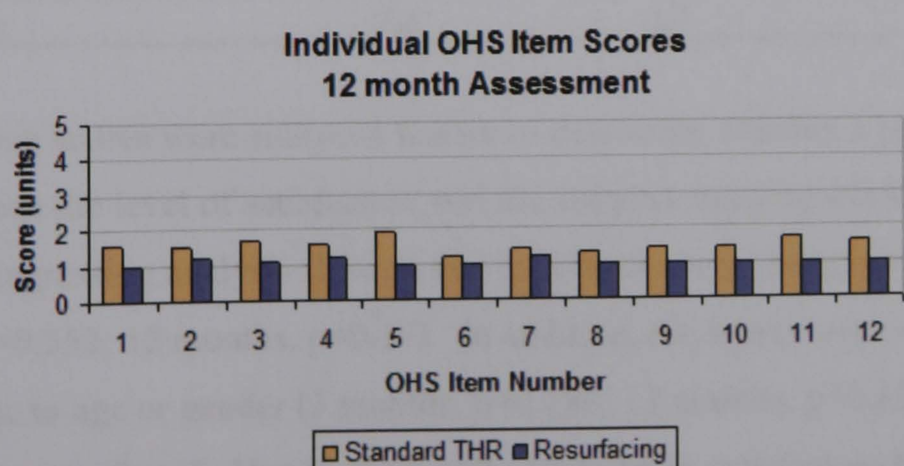


Figure 6.7: Comparison of 12 month individual item OHS results between groups

6.3.2 Satisfaction Score

The results of the satisfaction scores at 3 months and 12 months post-op are shown in tables 6.6 and 6.7 respectfully. The standard THR group tended to score slightly higher at both post-operative assessment phases, implying least satisfaction with their surgery and the associated outcome.

The greatest between-group difference in satisfaction scores was seen at 12 months, following the more marked improvement in satisfaction outcome by the resurfacing group. The standard THR group expressed more between-subject variability in their scores and overall, analysis of variance revealed no statistically significant difference between arthroplasty groups (p=0.374). Although there was a trend improvement in scores with time, this increase in satisfaction was not significant (p=0.369).

Table 6.6: Satisfaction score summary at 3 month assessment phase

SATISFACTION (units) 3 month assessment	Standard THR	Resurfacing	All
Max	19	15	19
Min	5	5	5
Mean	9.5	8.9	9.2
SD of Mean	4.2	2.7	3.5

Table 6.7: Satisfaction score summary at 12 month assessment phase

SATISFACTION (units) 12 month assessment	Standard THR	Resurfacing	All
Max	16	8	16
Min	5	5	5
Mean	7.8	5.4	6.6
SD of Mean	3.2	1.0	2.6

The satisfaction scores were analysed further to determine whether a relationship existed between the level of satisfaction and the surgeon carrying out the orthopaedic procedure. Regression analysis showed that satisfaction was independent of surgeon (3 months, p=0.552; 12 months, p=0.17). In addition, the scores were assessed for variability due to age or gender (3 months, p=0.188; 12 months, p=0.153) and again, no relationship was found. Hence, the small variations in satisfaction between groups were related to factors other than the operating surgeon, age, gender or the arthroplasty type. The late (12 month) satisfaction scores had a close relationship with body mass (p=0.06) and the early (3 month) satisfaction scores correlated positively with employment history (p=0.042). Hence, those who were retired had greater satisfaction levels at 3 months, and those who were employed scored the lowest satisfaction level at 3 months.

6.3.3 Visual analogue scores

The visual analogue pain and stiffness scores were recorded and averaged according to arthroplasty group. A summary of the pain and stiffness scores at 3 and 12 month assessment phases are shown in table 6.8 and 6.9 respectively.

The range of hip pain scores and the mean pain score was similar between the groups at 3 months (table 6.8). All subjects showed a significant improvement in the pain scores by 12 months post-op (p=0.026), although this improvement was most marked for the resurfacing group. Hence, by 12 months the standard THR group had relatively greater subjective pain scores. Nevertheless, there was no significant difference in pain levels between the arthroplasty groups with repeated measures statistical analysis (p=0.762).

Table 6.8: Visual analogue pain scores

	3 months			12 months		
PAIN VAS (units)	STHR	RA	All	STHR	RA	All
Max	7.1	7.4	7.4	2.9	0.5	2.9
Min	0.0	0.0	0.0	0.0	0.0	0.0
Mean	1.5	1.4	1.4	0.7	0.1	0.4
SD of Mean	1.7	2.0	1.8	0.9	0.2	0.7

Table 6.9: Visual analogue stiffness scores

	3 months			12 months		
STIFFNESS VAS (units)	STHR	RA	All	STHR	RA	All
Max	7.9	5.9	7.9	5.7	2.5	5.7
Min	0.0	0.2	0.0	0.0	0.0	0.0
Mean	2.6	2.1	2.4	1.8	0.6	1.2
SD of Mean	2.1	1.5	1.8	1.9	0.8	1.6

On average, hip stiffness (table 6.9) scored higher than hip pain and showed less improvements with time (no significant difference with repeated measures, p=0.375).

Again, the standard THR group showed the poorest outcome with greater stiffness scores relative to the resurfacing group. Overall, there were no significant differences between the arthroplasty groups for subjective stiffness levels ($p=0.908$).

Pain and stiffness VAS scores did not show any variability due to surgeon (pain, $p=0.554$; stiffness, $p=0.833$), gender or age. There was a close relationship between body mass and pain at 3 months ($p=0.087$) and between body mass and stiffness at 12 months ($p=0.071$). As with the satisfaction scores, there were positive correlations with stiffness levels and employment status (3 months, $p=0.01$; 12 months $p=0.02$), with greater subjective stiffness in the retired population.

To summarise, the current chapter has detailed the descriptive characteristics and subjective data relating to the study population, and has established the homogenous nature of the sample group. It can therefore be highlighted, that the sole dependent variable differentiating the sample group members was the type of unilateral hip arthroplasty used to reconstruct their hip joint: a standard CoC THR or a MoM resurfacing arthroplasty. In the following chapters, the objective data outcomes which were used to test the difference between the sample group members due to arthroplasty type are presented.

7 Motion Analysis Results

In this chapter, the results from the motion analysis trials will be presented, examining the main data outcomes of angles and moments of the hip joint. The results will be systematically presented for each functional task; hence angle and moment data will be reported for level walking, stair ascent and then stair descent.

Within the category of each functional motion task, the mean and mean peak data values will be presented and compared to test the influence of the following subject variables:

- Limb (within-subject comparison of operated or control limbs)
- Time (within-subject comparison of 3 and 12 month post-op data sets)
- Arthroplasty type (between-subject comparison of standard THR and resurfacing arthroplasty)

Aspects of the temporospatial characteristics of the motion tasks and the subjects' preferential motion patterns will also be discussed. Finally, relevant correlation analysis will be performed. Results obtained from the functional tasks will be contrasted and the relative value of each task as an assessment tool will be considered.

7.1 Subject Characteristics

The sample group corresponding to the motion analysis results was identical to that of the preceding results chapter (chapter 6). There were several omissions to sub-sets of the data which will be highlighted.

7.1.1 Subject exclusions

One subject of the 31 included for objective assessment had to be excluded from the final stair negotiation analysis. The exclusion was implicated as this subject did not perform stair ascent and descent with a reciprocal gait pattern. As this resulted in a double support phase and shorter stepping cycle, the stair gait characteristics were not comparable with the remaining subject group, and were therefore excluded from the analysis. This subject belonged to the standard THR group and returned for the 12 month assessment phase. Hence subject numbers for the standard THR group were reduced to 16 at 3 months and 9 at 12 months for stair negotiation analysis.

One other subject also used the handrails to provide support during stair negotiation. The subject using handrail support was not excluded from the analysis as this was thought to reduce the subject numbers below the threshold of the minimum required for fair statistical comparison.

7.1.2 Descriptive assessment of motion

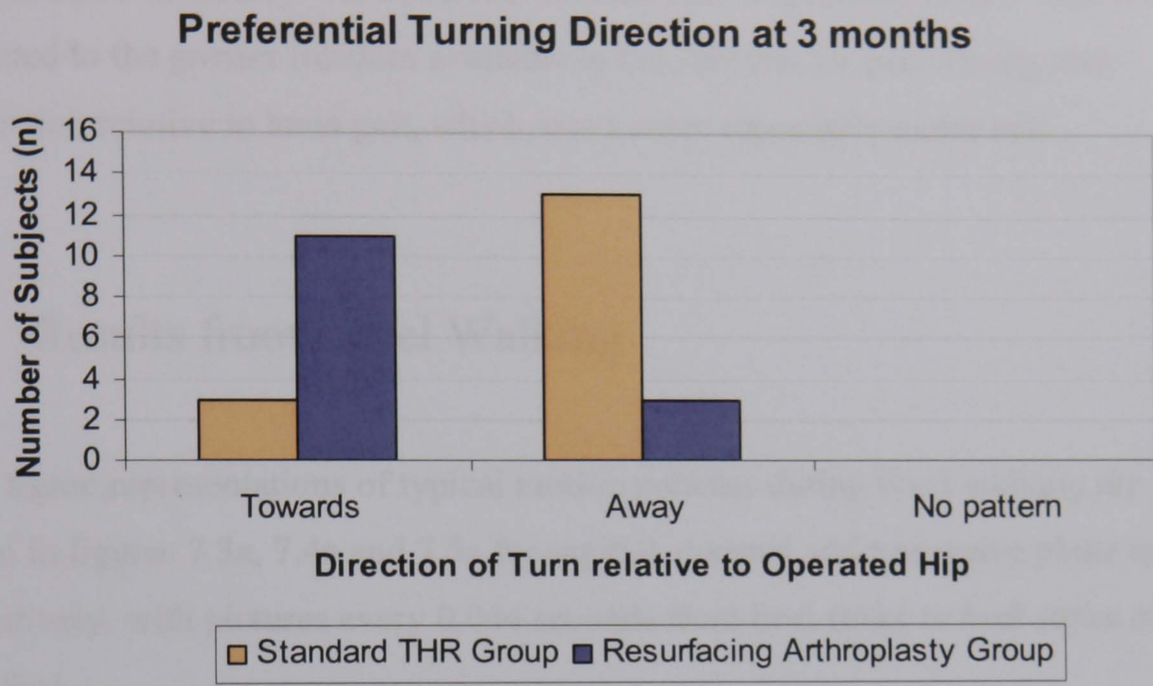
General observations were made during the motion tasks which highlighted certain subjects from the 'normal standard' of the overall cohort. Particularly evident at 12 months, and more common to subjects in the resurfacing group, was a tendency to descend the staircase with their body angled so they stepped in a slightly sideward fashion. This was noted as it was thought to potentially contribute to patterns of greater variability in the outcomes obtained from stair negotiation, particularly stair descent, compared to level walking.

Table 7.1: Frequency of preferential stair turning patterns

	3 months			12 months		
	Towards	Away	No pattern	Towards	Away	No pattern
Standard THR subjects (n)	3	13	0	3	6	0
Resurfacing subjects (n)	11	3	0	7	2	1
TOTAL	14	16	0	10	8	1
% Total Group	46.7	53.3	0.0	52.6	42.1	5.3

Note: Turning directions are expressed as motion relative to the operated limb

Another observation of interest was the variation in the turning direction at the top of the staircase (table 7.1). It was noted that in general, subjects with a hip resurfacing tended to turn towards their operated hip, whereas subjects with a standard THR tended to turn away from their operated hip, in a manner as to avoid pivoting on the prosthetic hip joint. The frequency of this trend is shown graphically in figure 7.1 for both 3 and 12 month data sets.



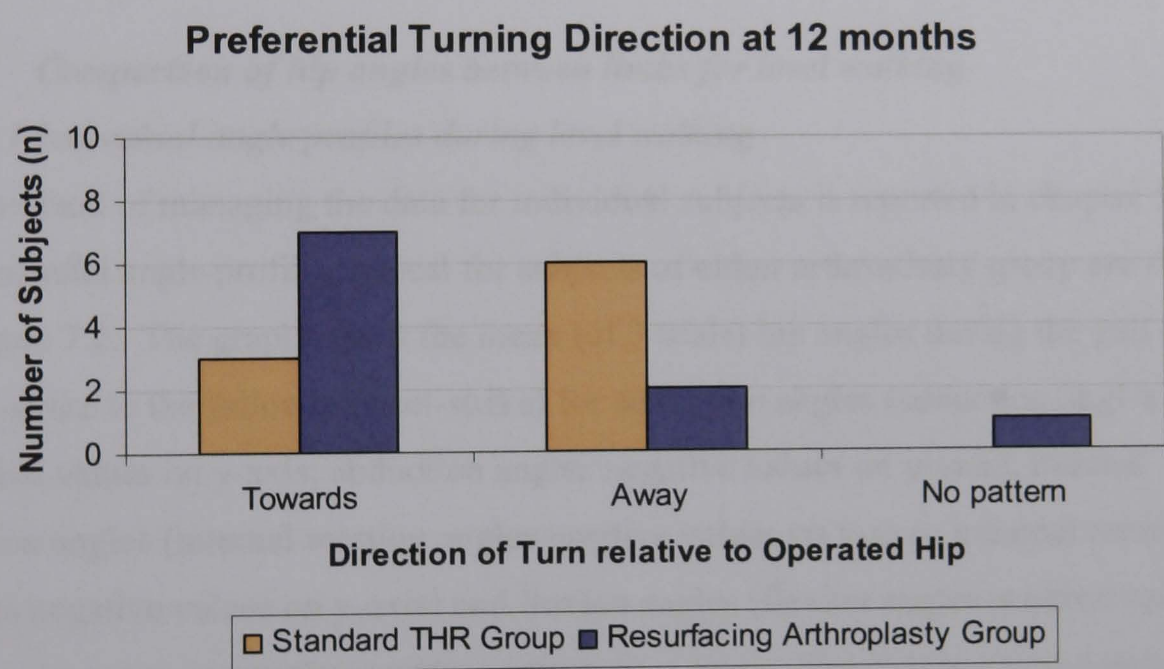


Figure 7.1: Comparison of stair turning direction between arthroplasty groups

The difference in turning direction between the arthroplasty groups was significant at both assessment phases and for repeated measures analysis (3 months, $p=0.021$; 12 months, $p=0.041$; repeated measures, $p=0.05$). At 3 months, the majority of subjects turned away from the operated limb, whereas, by 12 months, the majority of subjects turned towards the operated limb. Therefore, regardless of THR type, there was a trend progression of less protective behaviour as post-operative time increased.

Overall, more variability was observed with the stair negotiation tasks. This was attributed to the greater freedom available to the subjects for performing stair negotiation relative to level gait, which was a more repeatable motor task.

7.2 Results from Level Walking

Stick figure representations of typical motion patterns during level walking are shown in figures 7.3a, 7.4a and 7.5a for sagittal, coronal and transverse plane motion respectively, with pictures every 0.044 seconds from heel strike to heel strike of the right foot.

7.2.1 Comparison of hip angles between limbs for level walking

7.2.1.1 Individual angle profiles during level walking

The method of managing the data for individual subjects is reported in chapter 5. 3-dimensional angle profiles typical for subjects of either arthroplasty group are shown in figure 7.2. The graphs show the mean (of 3 trials) hip angles during the gait cycle (heel-strike to the following heel-strike) for adduction angles (adduction angles positive values on y-axis; abduction angles negative values on y-axis), internal rotation angles (internal rotation angles positive values on y-axis; external rotation angles negative values on y-axis) and flexion angles (flexion angles positive values on y-axis; extension angles negative values on y-axis). The data were randomly extracted from 3 month post-op results although the trends identified from the data were present at 12 months in most cases (section 7.2.1.2).

The adduction angles show that the operated limb followed a similar pattern of angular displacement as the control limb, but the up-shifted position of the line on the graph implies that the operated limb maintained a more adducted position throughout the gait cycle relative to the control limb.

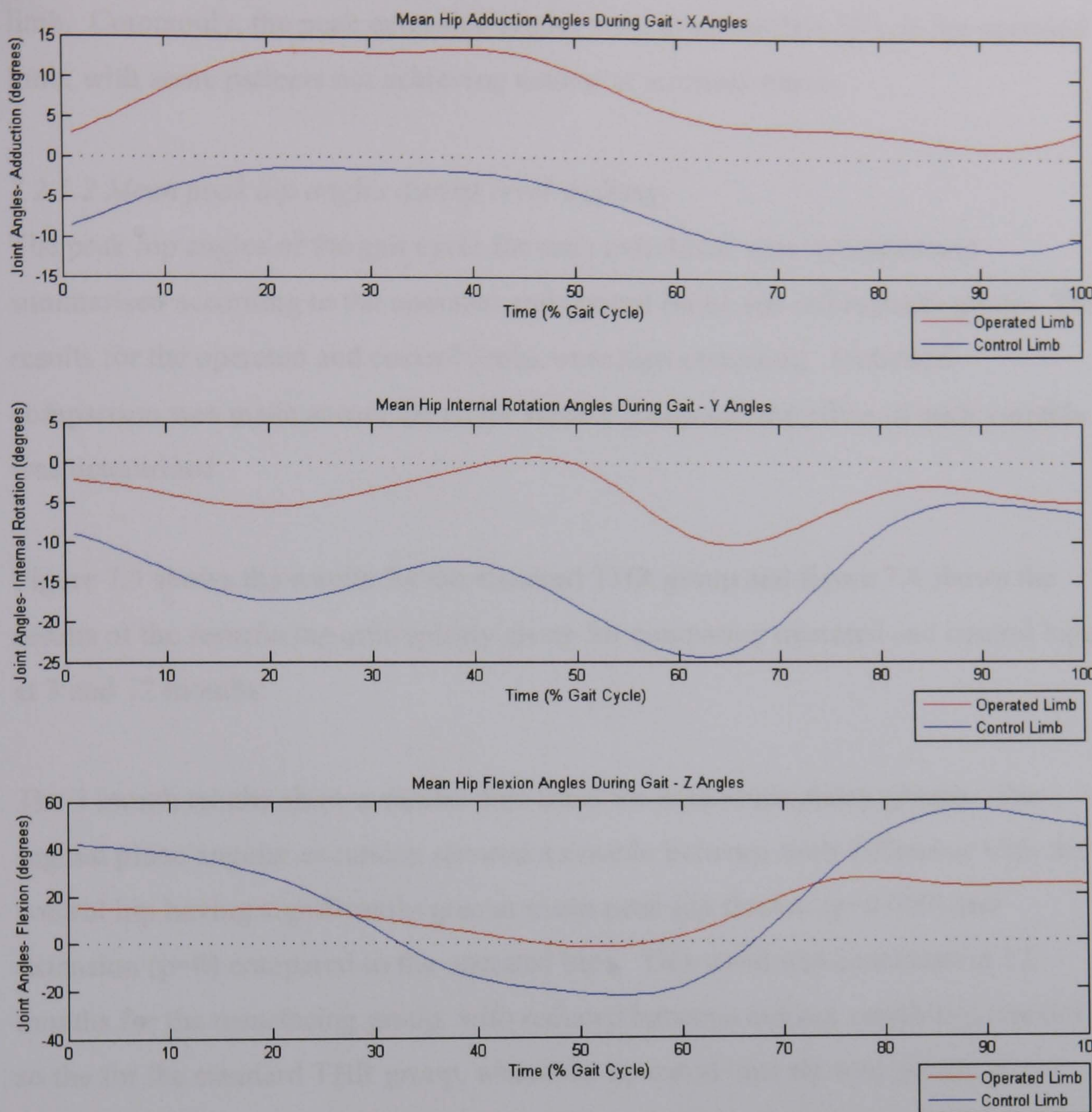


Figure 7.2: Mean 3-D angle profiles typical of operated and control limbs during level walking

The angles of hip rotation show a similar trend where, typically the operated limb follows a similar pattern of angular displacement to the control limb but maintains a more internally rotated position. The control limb showed typically greater angular excursion (range of angular displacement) than the operated limb with greater peak external rotation angles.

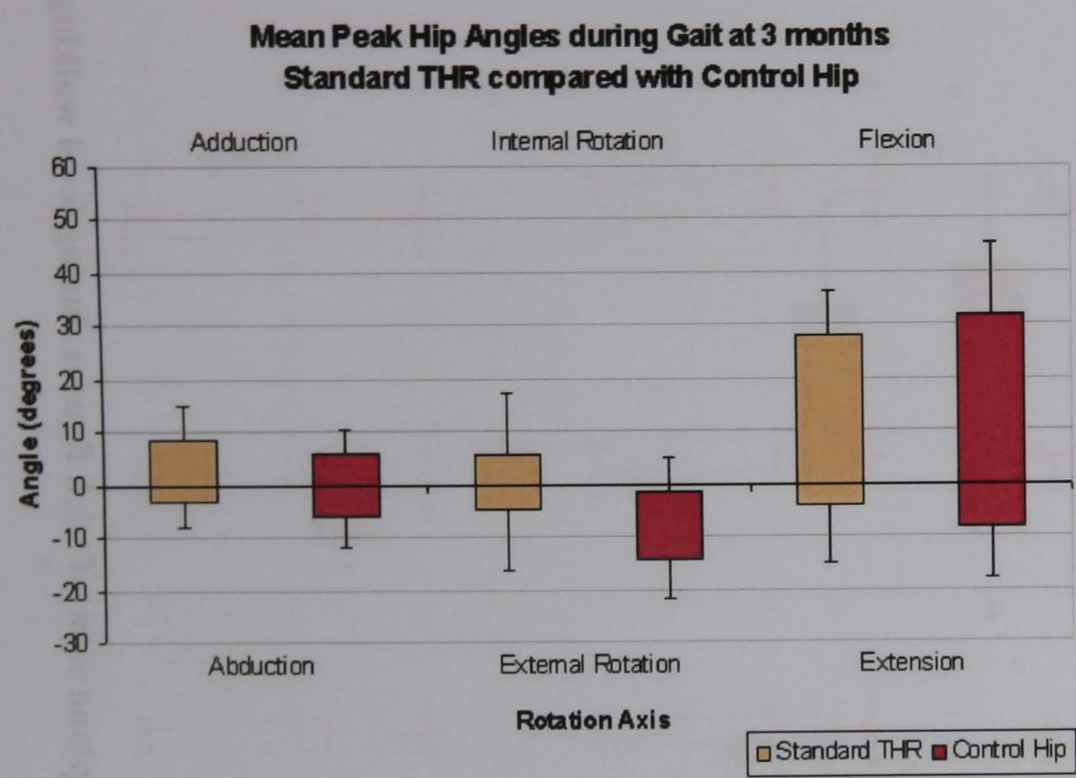
Sagittal plane motion typically showed the greatest differences in angular motion trends between the operated and control limbs. The operated limb achieved less angular excursion and smaller peak flexion and extension angles than the control limb. Commonly, the peak extension angles were about neutral (0°) on the operated limb, with some patients not achieving neutral at terminal stance.

7.2.1.2 Mean peak hip angles during level walking

The peak hip angles of the gait cycle for each individual were averaged and summarised according to the operated and control limbs and arthroplasty group. The results for the operated and control limbs were then compared. Statistical comparison was made across the entire subject group and the effect of each variable was determined.

Figure 7.3 shows the results for the standard THR group and figure 7.4 shows the results of the resurfacing arthroplasty group for comparing operated and control hips at 3 and 12 months.

The 3 month results show a similar data trend for both arthroplasty groups. The sagittal plane angular excursion showed a notable between-limb difference with the control hip having significantly greater mean peak hip flexion ($p=0.009$) and extension ($p=0$) compared to the operated hips. This trend was consistent at 12 months for the resurfacing group, with reduced between-subject variability, but not so the for the standard THR group, where the operated hips showed greater mean peak sagittal plane motion than the control hip. Overall, the between-limb differences for all subjects at 12 months were not significant (flexion $p=0.072$; extension $p=0.968$). The pattern change in peak sagittal angles over time was significantly different between the groups for flexion motion ($p=0.02$) and close to significance for extension motion ($p=0.056$), with an increase in the peak angles achieved on the standard operated hips by 12 months, with an average of nearly 10° more flexion. In contrast, the resurfacing operated hips gained more extension motion (mean 10°) with time but the control hip of the resurfacing group showed a corresponding improvement. Hence, mean peak extension angles of the resurfacing hips improved from 3 to 12 months although they remained sub-optimal compared to the control limb (figure 7.4).



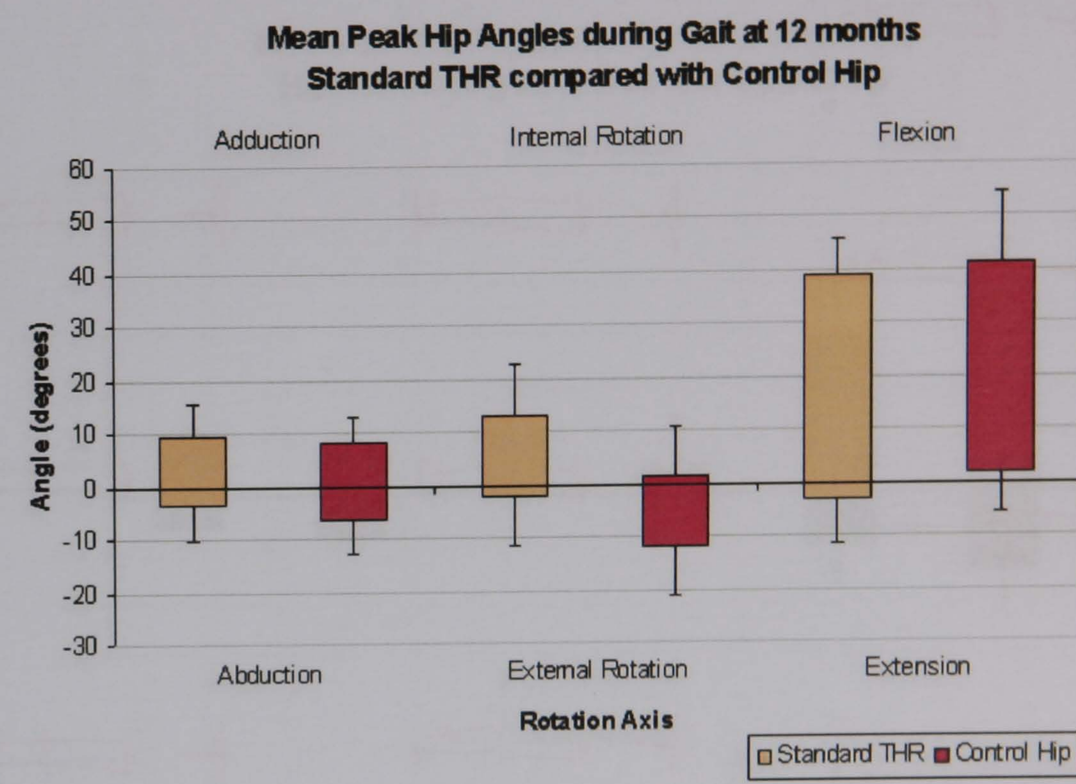


Figure 7.3: Comparison of 3-D hip angles for standard THR group-operated and control limbs during level walking at (a) 3 months and (b) 12 months

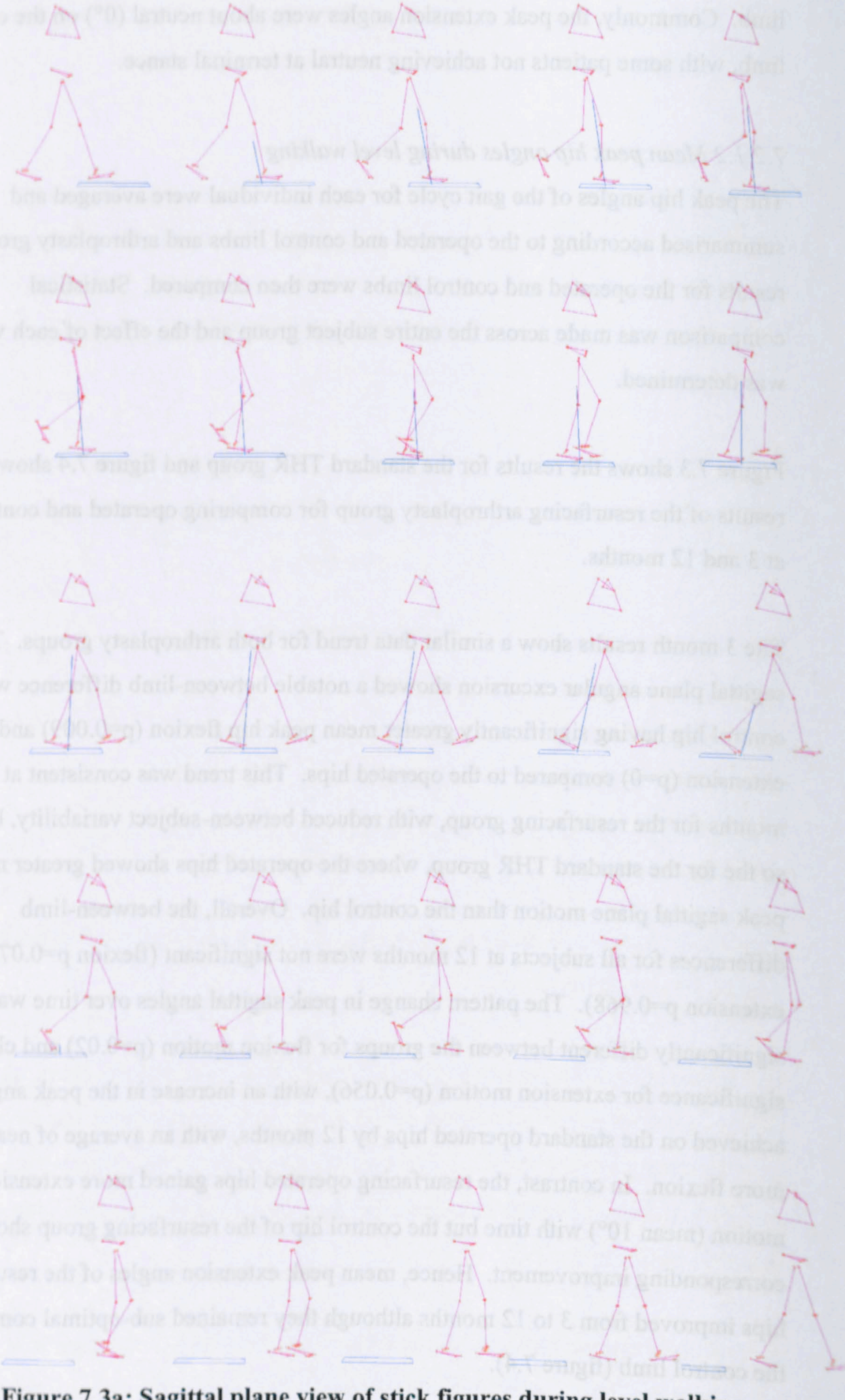
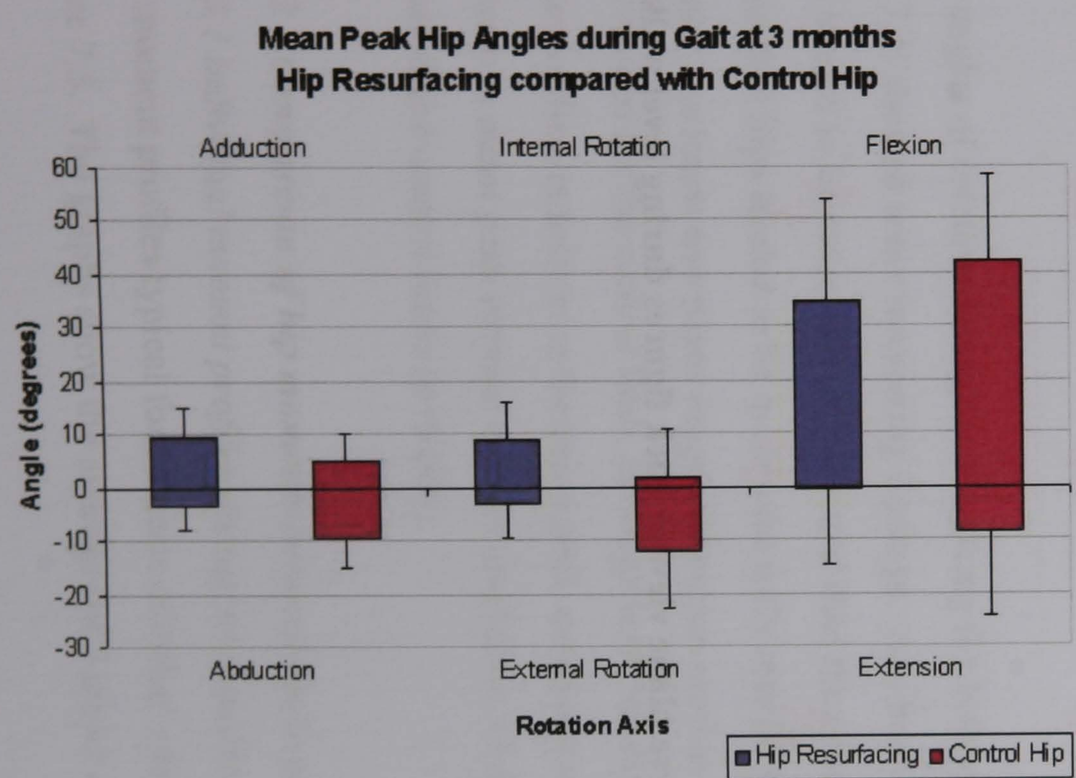


Figure 7.3a: Sagittal plane view of stick figures during level walking



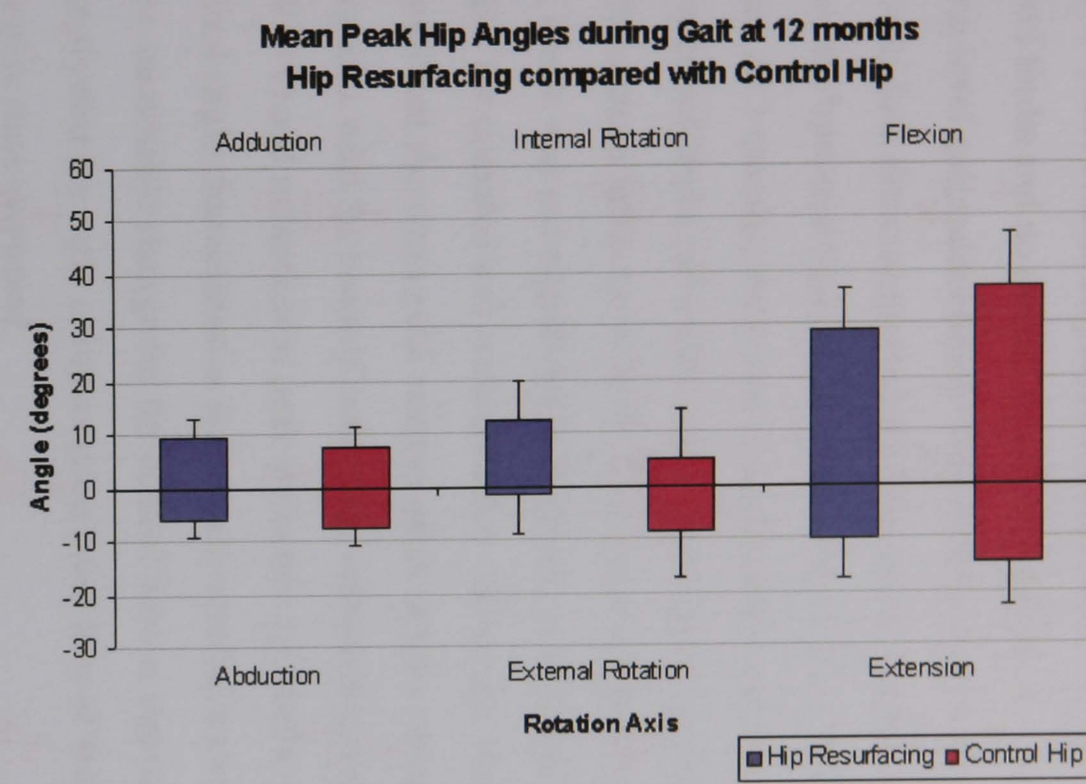


Figure 7.4: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during level walking at (a) 3 months and (b) 12 months

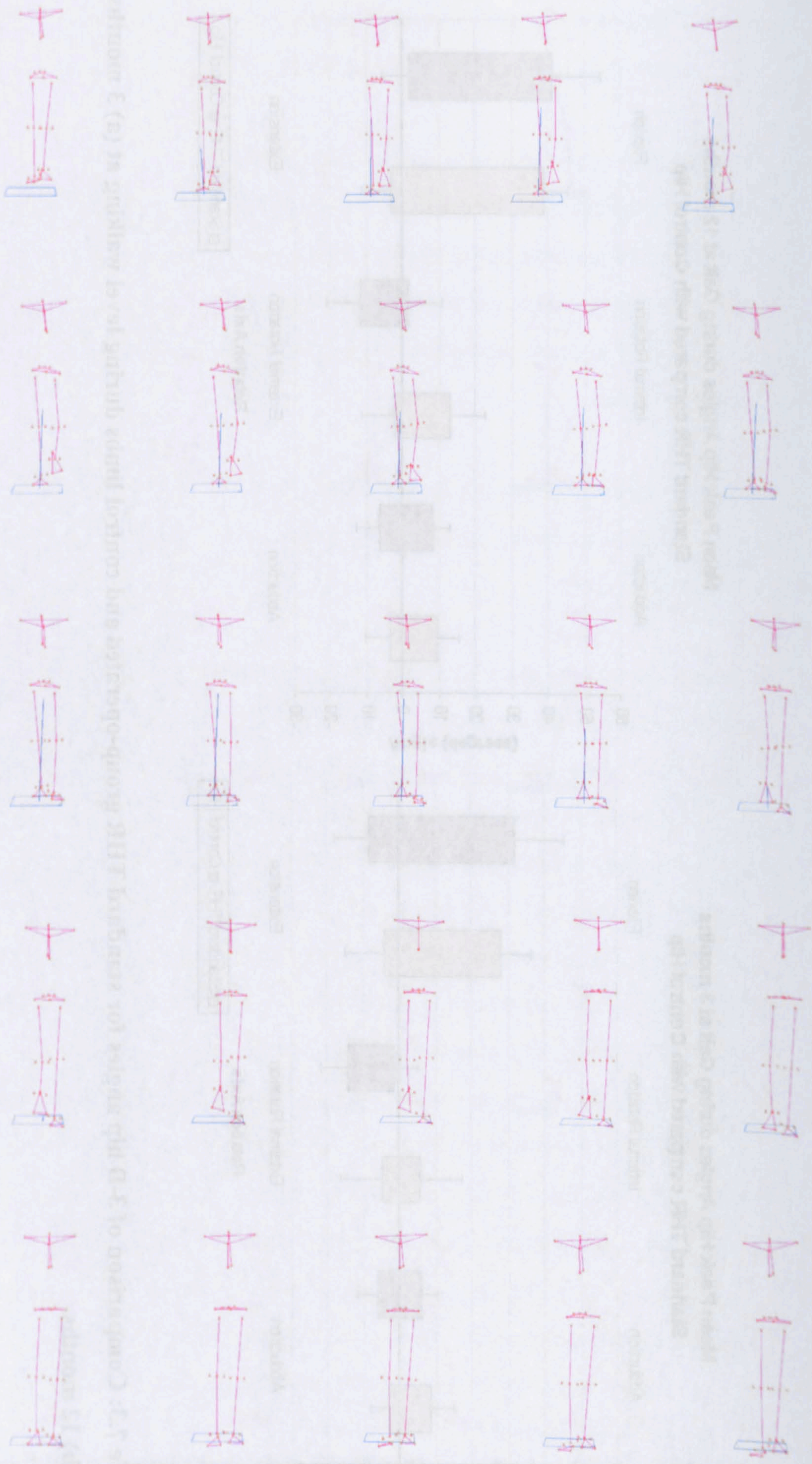


Figure 7.4a: Coronal plane view of stick figures during level walking

Coronal plane results displayed similar values of angular motion for operated and control limbs and no overall significant difference in peak angles between-limbs during level gait (adduction $p=0.166$; abduction $p=0.106$). However, examining the 3 month data alone supported the data trends reported previously from figure 7.2, where the operated limb maintained at more adducted position throughout the gait cycle. At 3 months, there was a significantly greater mean peak adduction angle for the operated limbs ($p=0.044$) and a significantly greater mean peak abduction angle for the control limbs ($p=0.013$). This trend continued at 12 months but weakened and, hence was not significantly different, as the angular motion became more similar for operated and control limbs. Moreover, repeated measures analysis showed that the mean peak coronal plane angles varied significantly between 3 and 12 months with the control limb being responsible for an increase in peak adduction ($p=0.017$) and reduction in peak abduction ($p=0.024$) with time which made the coronal angle characteristics statistically similar between limbs by 12 months. This more remarkable change for the control limb as opposed to the operated limb appears to imply that the early compensation behaviour of the control limb reduced as post-operative time increased.

The angles of rotation during level walking for both arthroplasty groups (figure 7.3 and 7.4) showed some interesting findings. As observed in figure 7.2, the operated hips tended to be more internally rotated than the control hips ($p=0.004$) and, in turn, the control hips tended to be more externally rotated ($p=0.001$). From 3 to 12 months, the transverse plane range of rotation motion increased for the operated limb but reduced for the control limb, although both became more internally rotated with a corresponding reduction in the mean peak external rotation on both limbs. The increase in mean peak internal rotation angles by 12 months was significant for both operated and control limbs ($p=0.006$).

7.2.2 Comparison of hip moments between limbs for level walking

7.2.2.1 Individual moment profiles during level walking

Hip moment profiles typical for subjects of either arthroplasty group are shown in figure 7.5. The graphs show the average (of 3 trials) external hip moments during

Figure 7.5a: Transverse plane view of stick figures during level walking

the stance phase of level gait (heel-strike to toe-off) for adductor moments (adductor moments positive values on y-axis; abductor moments negative values on y-axis), internal rotator moments (internal rotator moments positive values on y-axis; external rotator moments negative values on y-axis) and flexor moments (flexor moments positive values on y-axis; extensor moments negative values on y-axis). As with the typical angle profiles, the data were extracted from 3 month post-op results although the trends identified from the data were present at 12 months in most cases.

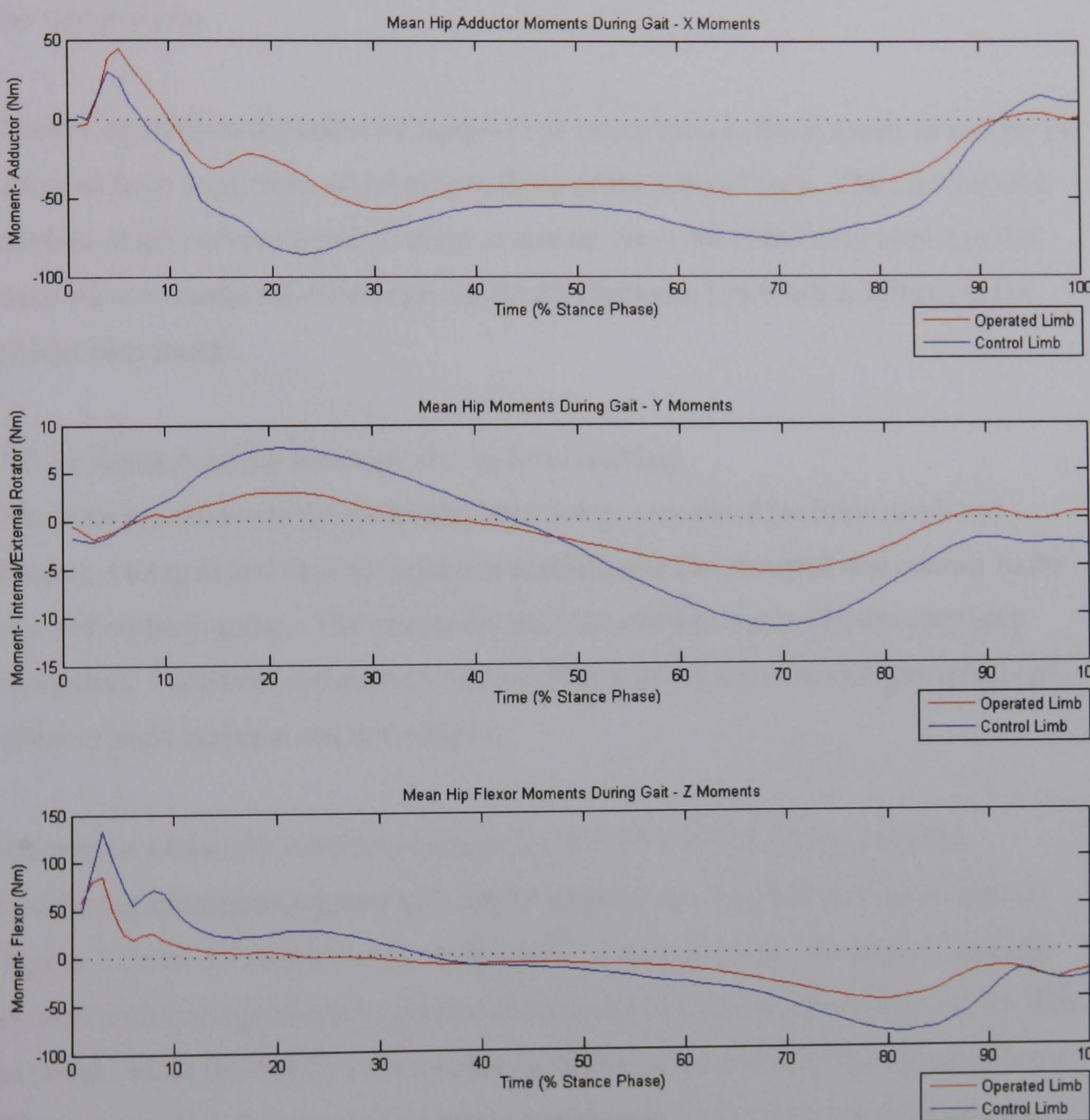


Figure 7.5: Mean 3-D moment profiles typical of operated and control limbs during level walking

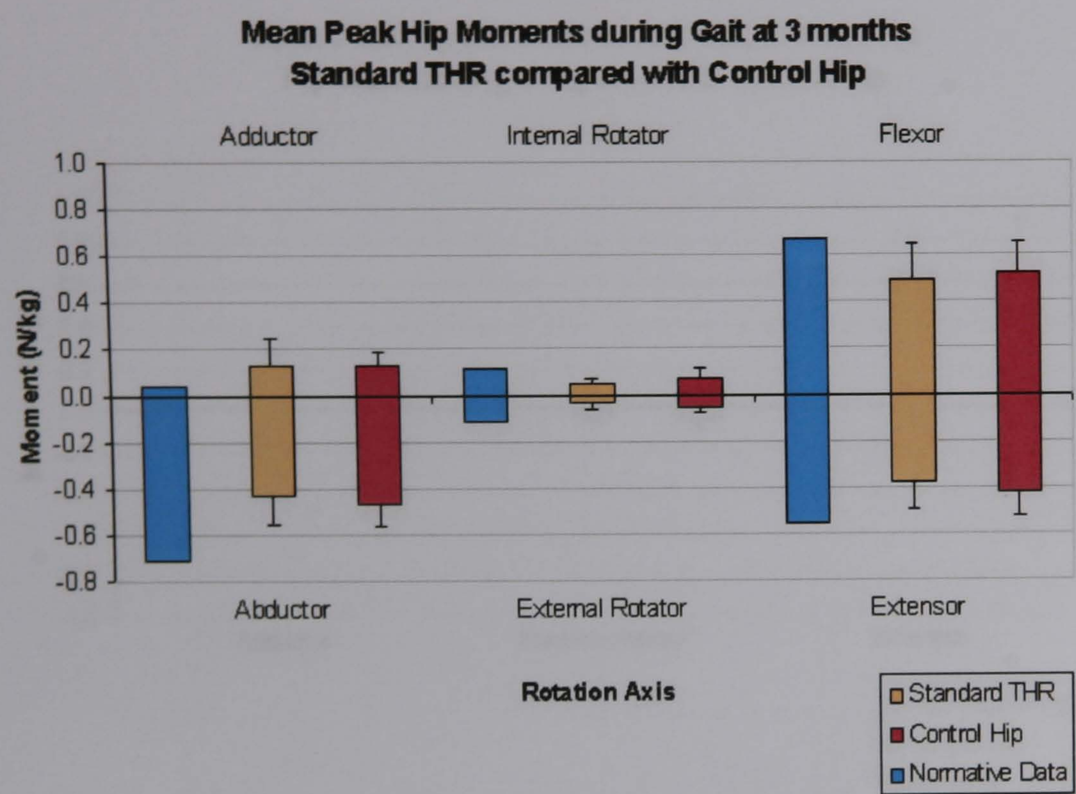
The operated limb tended to have reduced abductor moments compared to the control limb and this pattern was typically maintained throughout the stance phase. The between limb differences for internal rotator moments showed a relative reduction in the peak internal and external rotator moments of the operated limb compared to the control limb. There was however substantial variation in the internal rotator moment profiles between subjects, which will be discussed further in section 7.2.2.2. The hip flexor moment profiles also showed a reduction in the flexor and extensor moments of the operated limb relative to the control limb all throughout the stance phase.

Overall, apart from the adductor moments at initial stance, the moment values for the operated limb were reduced relative to those of the control limb. The maxima and minima of the curves tended to occur at similar times for both limbs implying that there were no temporal differences in the data between limbs when following the typical data trends.

7.2.2.2 Mean peak hip moments during level walking

The peak hip moments for each individual were normalised (to body mass and height), averaged and then summarised according to the operated and control limbs and arthroplasty group. The results for the operated and control limbs were then compared. Statistical comparison was made across the entire subject group and the effect of each variable was determined.

The results of the hip moment outcome for the standard THR group and the resurfacing arthroplasty group at 3 and 12 months are compared between-limb in figures 7.6 and 7.7 respectively. At the early (3 month) stage, for both groups, the control limb had significantly greater abductor ($p=0.032$), internal rotator ($p=0.001$), external rotator ($p=0.003$) and extensor ($p=0.001$) moments than the operated limbs. When accounting for repeated measures (combined 3 and 12 month assessment), statistical analysis showed that the flexor moment was also significantly different between limbs ($p=0.018$) with greater peak flexor moments for the control limb.



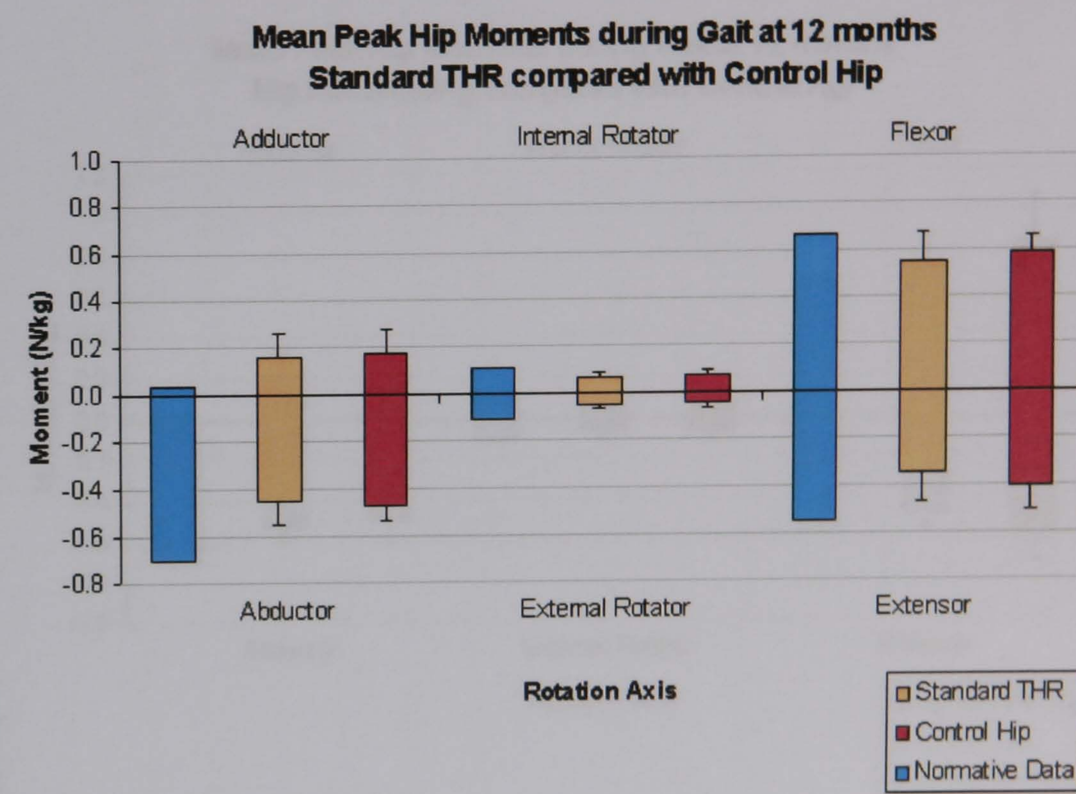


Figure 7.6: Comparison of 3-D hip moments for standard THR group-operated and control limbs during level walking at (a) 3 months and (b) 12 months

(Normative data sourced from Eng and Winter 1995; male subjects (n=9), aged 19-26 (mean 22.2) years, mean gait velocity 1.6m/s)

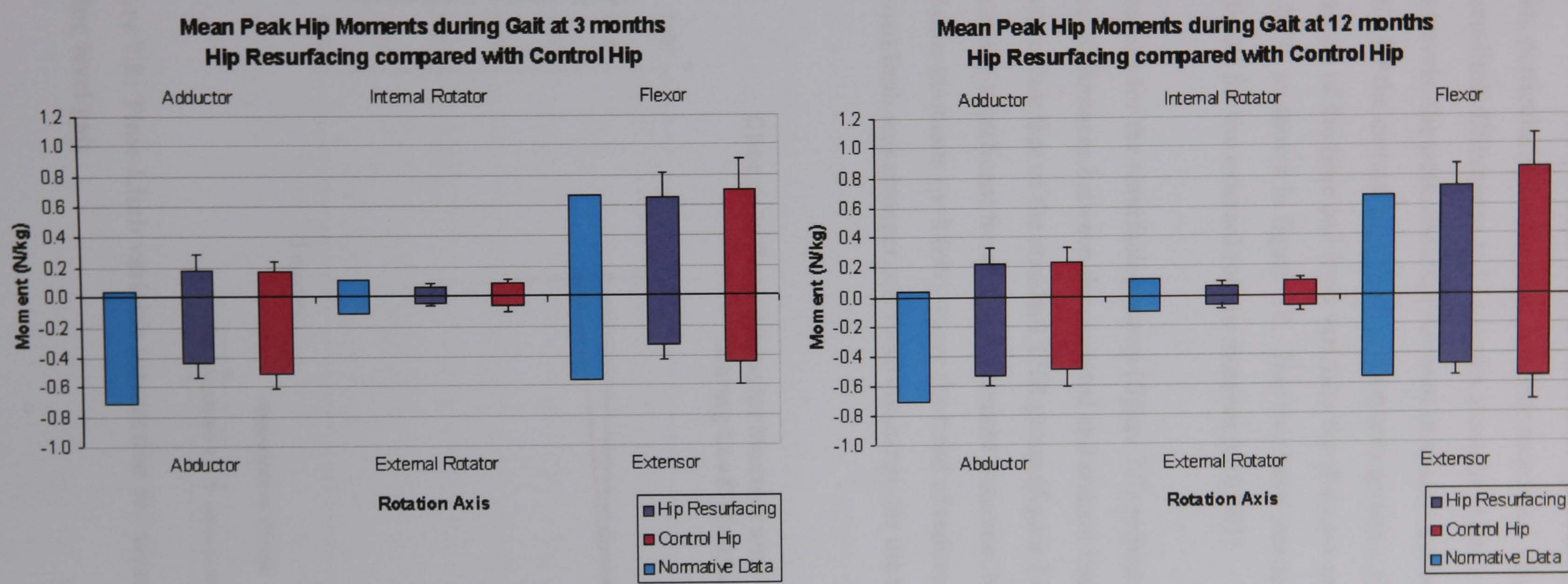


Figure 7.7: Comparison of 3-D hip moments for hip resurfacing group-operated and control limbs during level walking at (a) 3 months and (b) 12 months

(Normative data sourced from Eng and Winter 1995; male subjects (n=9), aged 19-26 (mean 22.2) years, mean gait velocity 1.6m/s)

Between-limb comparisons showed less differences at 12 months compared to 3 months, particularly for the peak abductor moments, which showed no significant between-limb differences at 12 months alone (p=0.642). Specifically, between the test intervals, the peak abductor moment increased for the operated hips and reduced slightly for the control hips (both arthroplasty groups), giving rise to a significant interaction of the time and limb variables for abductor moments during level walking (p=0.039), as shown in figure 7.8. This trend and interaction effect was also significant for the external rotator moment (p<0.001).

The results for the resurfacing group (figure 7.7) appeared to display relatively greater differences between the operated and control limbs for mean peak hip moments than that of the standard THR group (figure 7.6). This observation was not statistically significant for any peak moment outcome, but closest to significance for the flexor moments (p=0.099) where the trend of reduced flexor moments for the operated limb was stronger and more consistent for the resurfacing group.

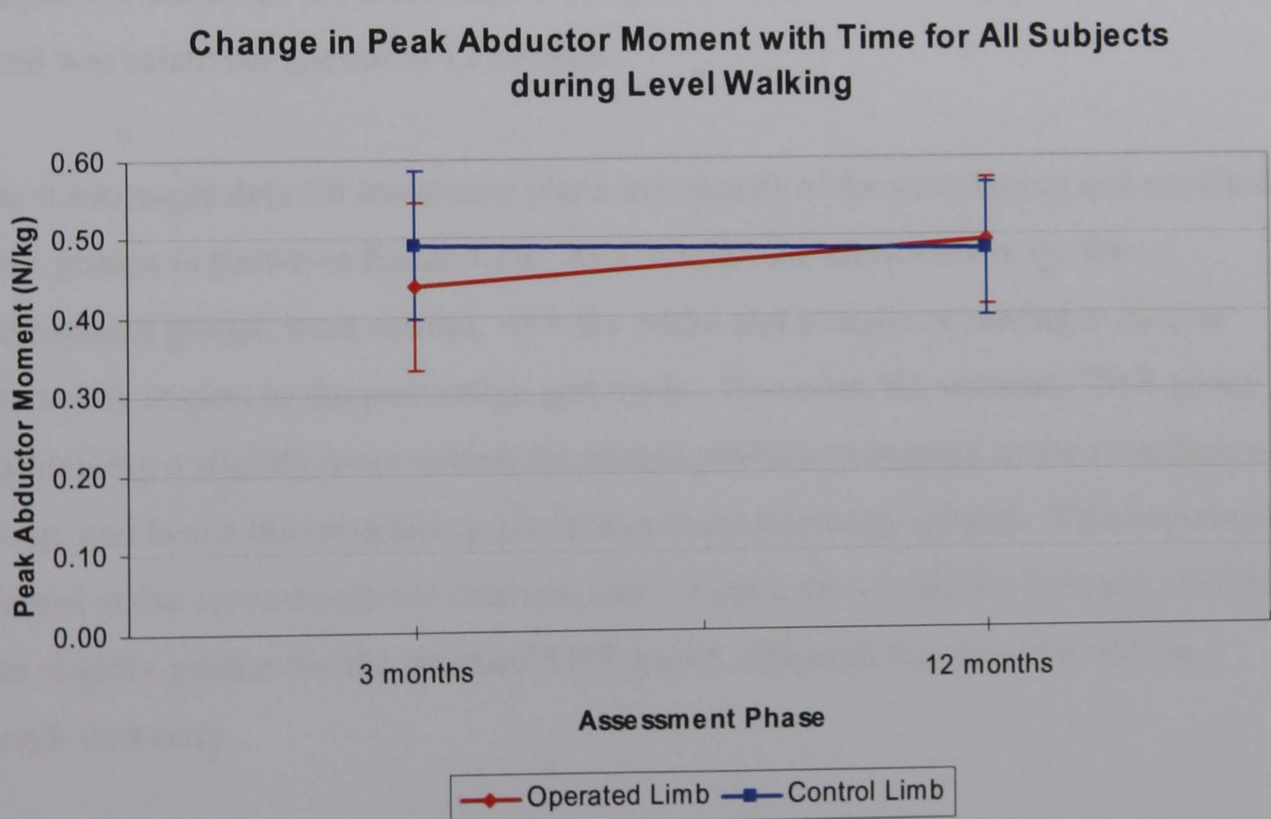


Figure 7.8: Time-Limb variable interaction for mean peak abductor moment during level gait

7.2.3 Comparison of hip angles between arthroplasty groups for level walking

7.2.3.1 Average group angle profiles for level walking

The angle profile data for the level walking gait cycle of the operated hip was summarised according to the arthroplasty type. The mean angle profiles were averaged for each arthroplasty group at 3 and 12 months and the data compared for angles in all 3-dimensions.

Figure 7.9 shows that at 3 months following surgery, the mean angle profiles for movement in the coronal plane were similar for the arthroplasty groups. At 12 months, most of the between group differences occurred after 40% of the gait cycle, where the resurfacing group featured greater abduction angles than the standard THR group, particularly during the swing phase. Overall, the magnitude of the angles throughout the cycle remained similar between 3 and 12 months. The peak adduction angle was achieved slightly quicker for the hip resurfacing group at 3 months, but there were little differences by 12 months. The pattern of variability in adduction angles showed that the standard THR group displayed more between-subject variability of the mean angles compared to the resurfacing group, and this trend was relatively greater at 12 months.

The mean angle data for transverse plane movement of the resurfacing and standard THR groups is shown in figure 7.10. At 3 months the mean curves for the arthroplasty groups were similar, with the peaks and troughs occurring at similar times with respect to the percentage gait cycle. However, the standard THR group maintaining a slightly more externally rotated position compared to the resurfacing group, and hence the resurfacing group was more internally rotated. This was most marked at the second external rotation peak. Again, the variability in angle profiles was slightly greater for the standard THR group, although this was true for the 3 month data only.

The external rotation peak at the start of the stance phase (15-20%) shows interesting developments with time. For the operated limb in both arthroplasty groups, the mean peak external rotation angle reduced by 3°. This was consistent in both groups and may be an important development with post-operative progression. This reduction in the external rotation at early stance was the greatest change for the resurfacing group over time, but the standard THR group showed an overall increase in internal rotation angles during level walking by 12 months.

The hip flexion profiles comparing resurfacing and standard hip arthroplasty (figure 7.11) showed interesting and, albeit inconsistent results. Results from both 3 and 12 months show similar flexion curve profiles for each arthroplasty group, with a similar pattern of angular displacement throughout the gait cycle at both post-operative phases and the peak angles occurring at similar times. However, at 3 months the resurfacing arthroplasty hips had relatively greater flexion angles and at 12 months the opposite was true, with the standard THR group having greater flexion angles. This implies that the resurfacing group had poorer hip extension angles at terminal stance (~50% gait cycle) at 3 months, but featured a greater improvement in hip extension with time compared to the standard THR group (time-THR type interaction, p=0.056). Figure 7.12 shows a visual representation of the change in sagittal angles with time during level walking for each arthroplasty group. Although both groups gained a greater total amplitude of sagittal motion during gait, the mean extension peak of the standard THR hip was maintained at 12 months but the resurfacing group develop a more extended angle profile, as shown by the down-shift of the resurfacing profile, the greater peak extension angles and the reduced flexion angles by 12 months.

The pattern of variability between subjects for flexion angles contrasted with that for adduction and internal rotation angles, with greater variability with the resurfacing group compared to the standard THR group. Overall, the variability reduced from 3 to 12 months, particularly for the resurfacing arthroplasty group.

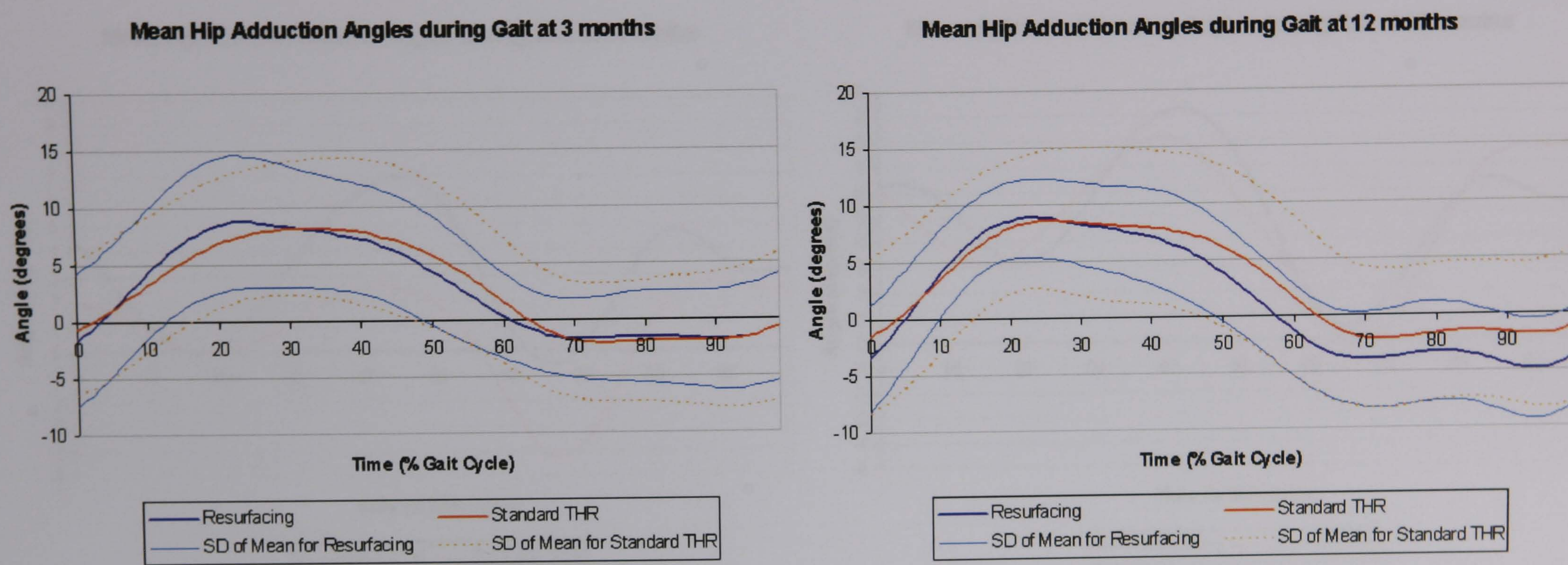


Figure 7.9: Mean hip adduction angles at (a) 3 and (b) 12 months

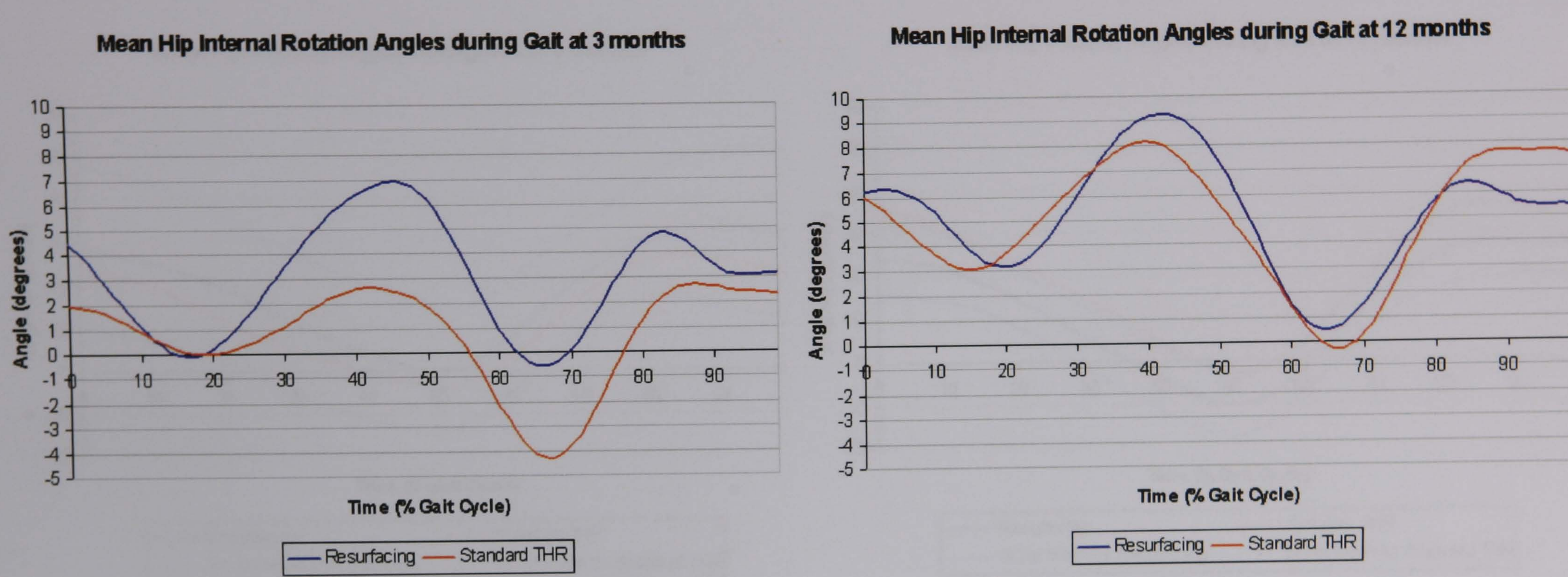
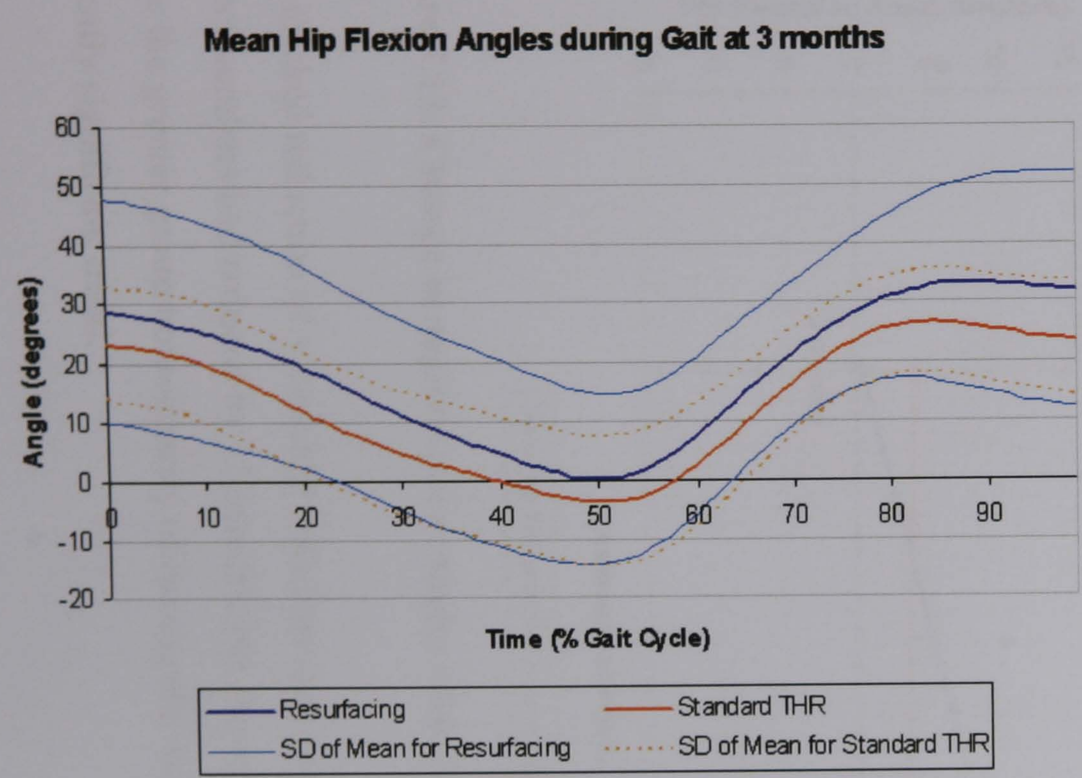


Figure 7.10: Mean hip internal rotation angles at (a) 3 and (b) 12 months



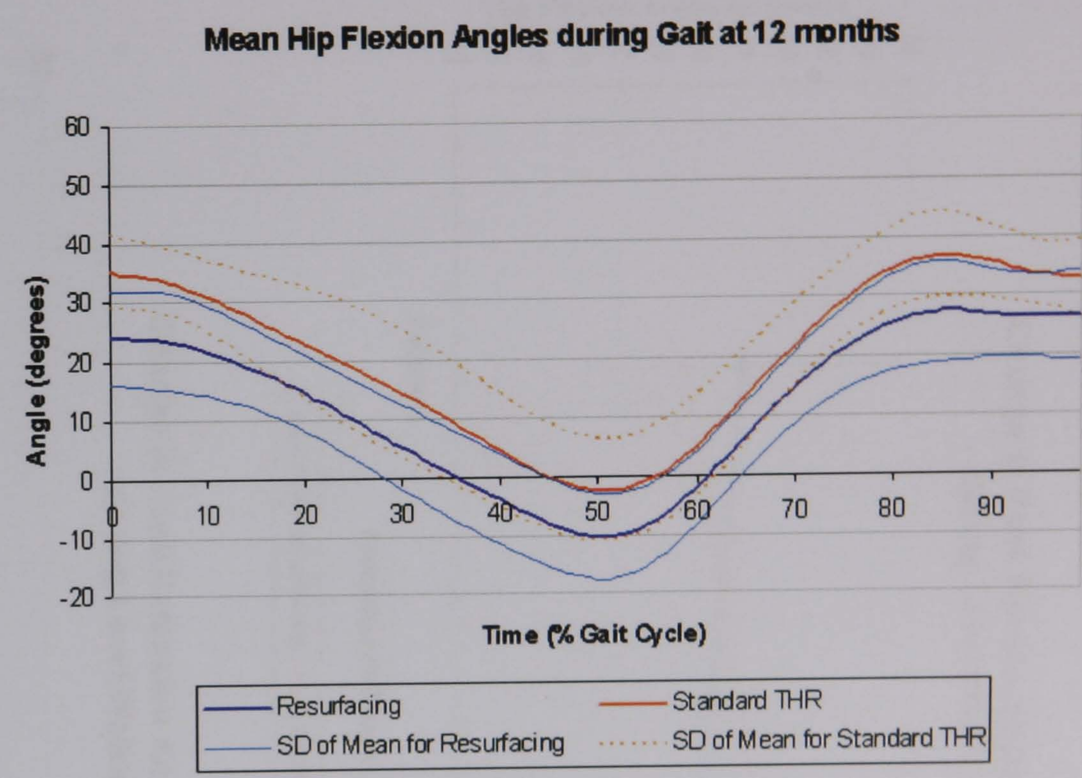
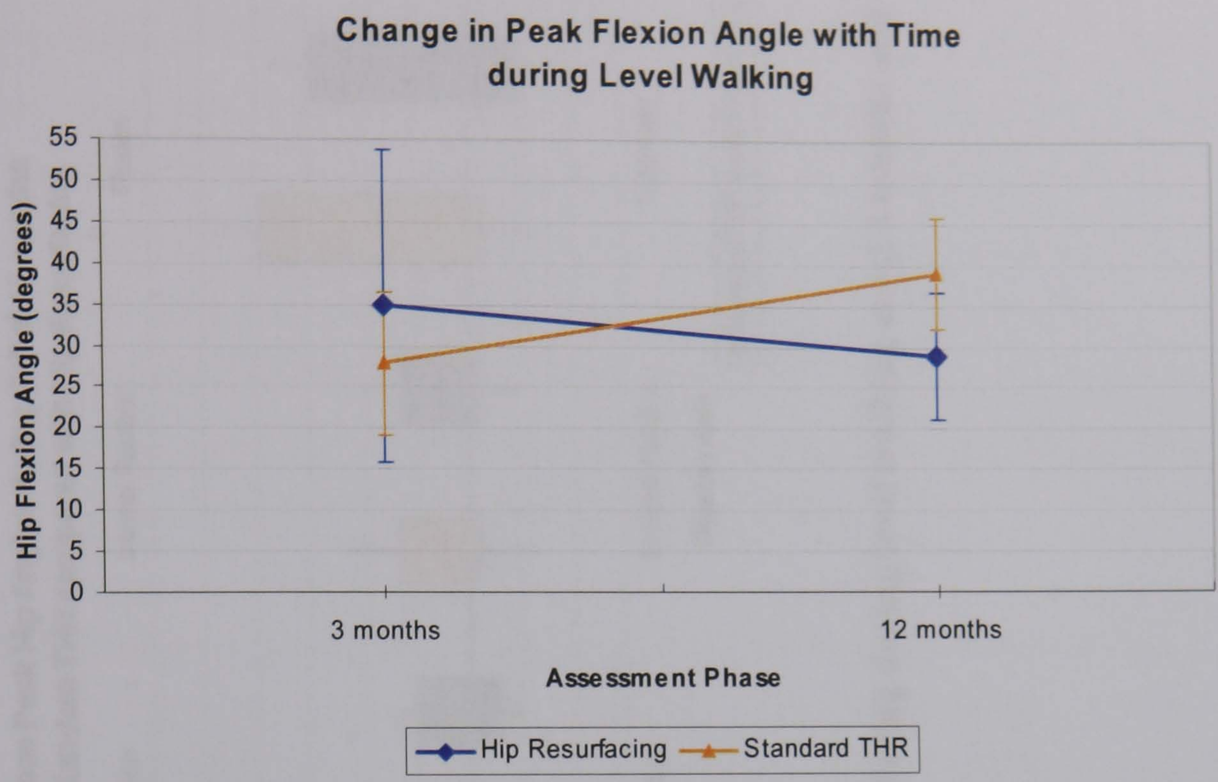


Figure 7.11: Mean hip flexion angles at (a) 3 and (b) 12 months



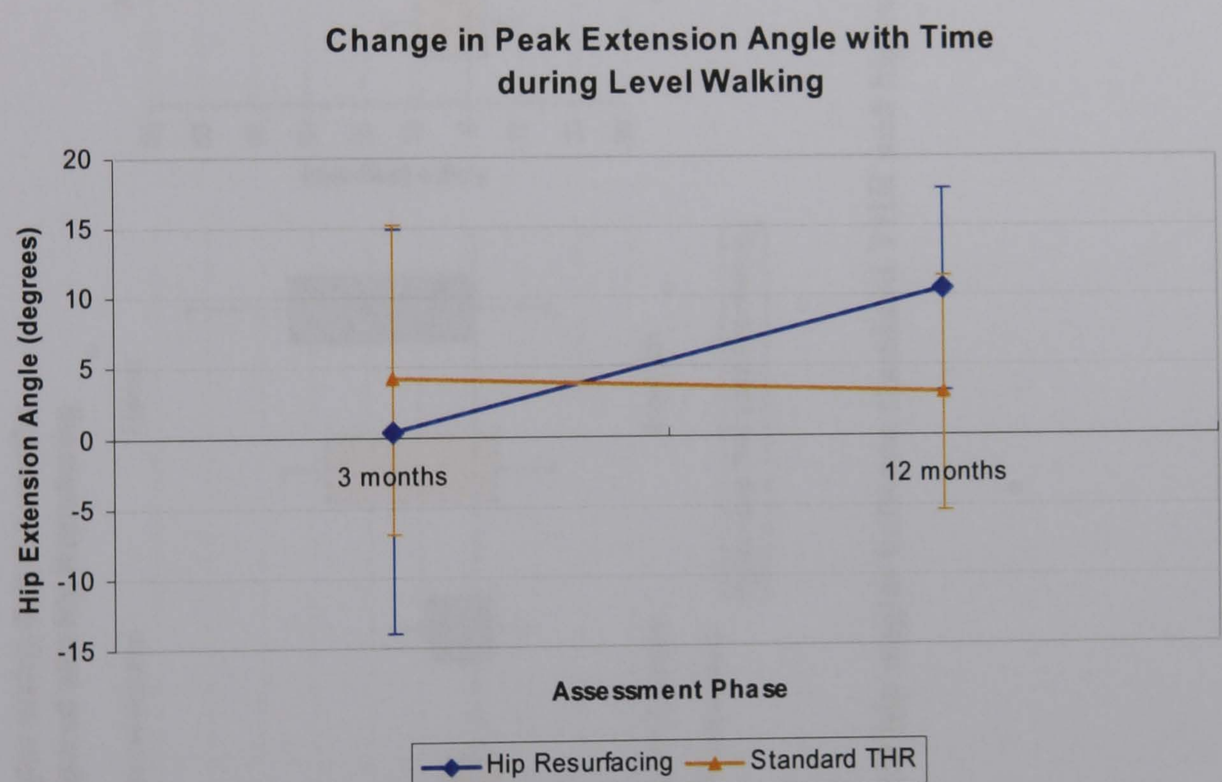
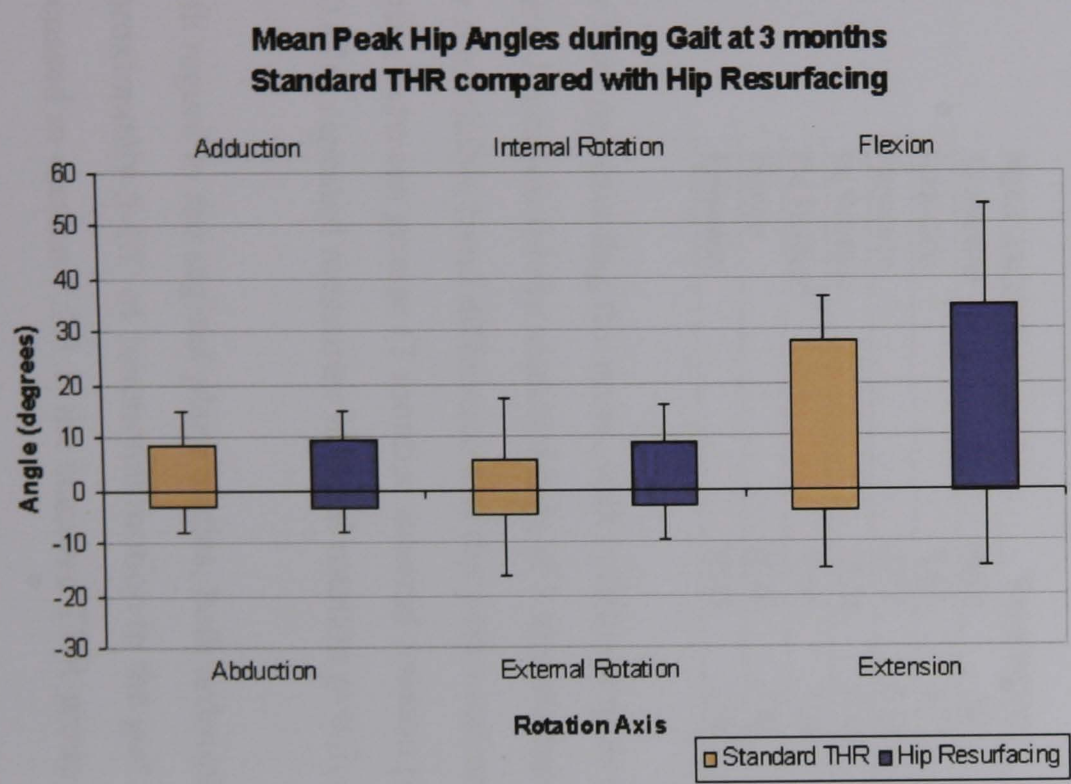


Figure 7.12: Change in sagittal plane angles with Time and THR type

This marked reduction of variability with time was an interesting observation as the subject numbers had reduced by 12 months, but the results became more consistent, hence the greater group homogeneity of the sagittal angle profiles with time may be a clinically significant trend.



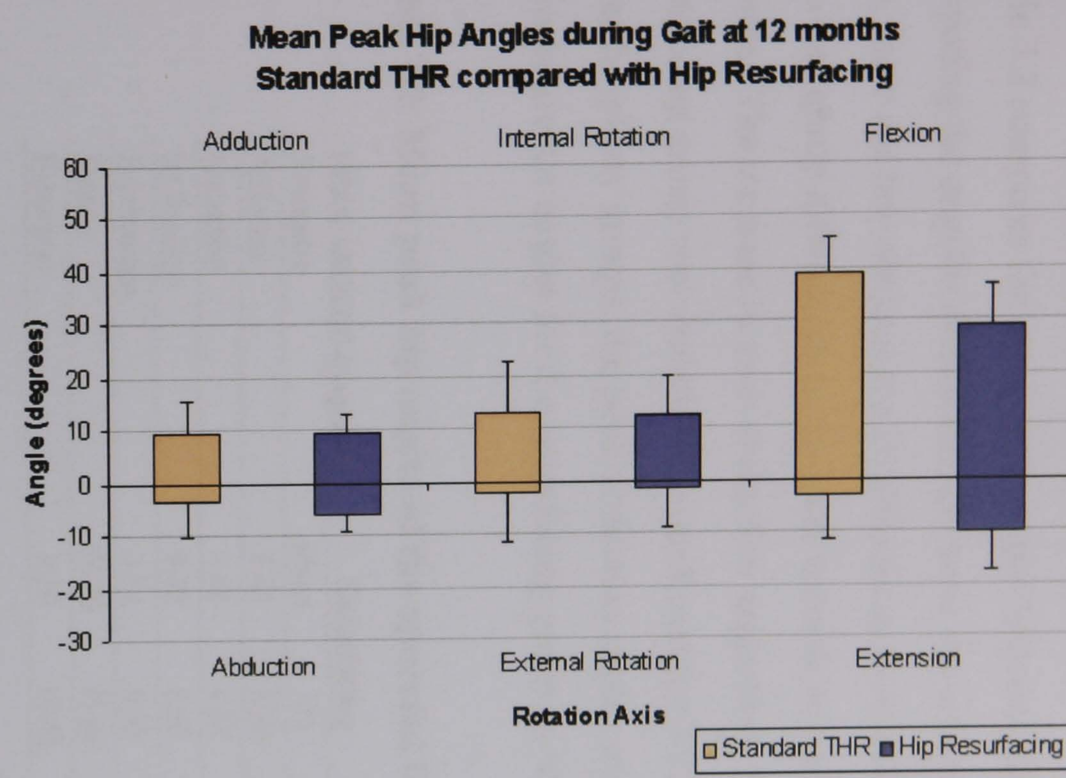


Figure 7.13: Comparison of 3-D hip angles between standard THR and hip resurfacing during level walking at (a) 3 months and (b) 12 months

7.2.3.2 Mean peak hip angles for level walking

Table 7.2 compares the mean peak angles between the arthroplasty groups. When comparing the angular differences between the arthroplasty groups (figure 7.13, table 7.2), there are few obvious trend differences to remark on. There were minimal between-group differences in the peak coronal or transverse plane angles at 3 or 12 months. The increase in peak abduction angle from 3 to 12 months for the resurfacing group was statistically significant (p=0.024). When comparing between the arthroplasty groups, the peak abduction angles showed a strong trend difference towards greater angles for the resurfacing group (p=0.06).

Table 7.2: Mean peak hip angles of the operated limb during level walking

PEAK ANGLES (degrees)	Resurfacing		Standard THR	
3 months	Mean	SD	Mean	SD
Adduction	9.48	5.39	8.62	6.10
Abduction	3.48	4.56	2.92	5.18
Int Rotation	9.08	6.86	5.75	11.72
Ext Rotation	3.12	6.33	4.99	11.44
Flexion	34.79	18.96	27.78	8.67
Extension	0.40	14.45	4.13	11.09

PEAK ANGLES (degrees)	Resurfacing		Standard THR	
12 months	Mean	SD	Mean	SD
Adduction	9.42	3.71	9.51	5.95
Abduction	5.77	3.55	3.40	6.76
Int Rotation	12.28	7.46	13.14	9.81
Ext Rotation	1.44	7.38	2.29	9.03
Flexion	28.86	7.86	38.90	6.77
Extension	10.41	7.26	2.94	8.45

The bars representing the mean peak rotation angles (figure 7.13) showed less internal rotation for the standard hips at 3 months but the large variability negates any remarkable trend difference and the peak rotation angles were statistically similar between groups (3 months: internal rotation p=0.232, external rotation p=0.476; repeated measures: internal rotation p=0.3, external rotation p=0.547).

With regard to the sagittal plane angles, both arthroplasty groups gained approximately 5-10° of functional motion in the gait cycle from 3 to 12 months. As discussed in section 7.2.3.1, the standard THR group display this gain with a greater peak flexion angle whereas the resurfacing group show a relatively greater gain in the peak extension, thus giving rise to the skewed pattern of sagittal plane angles between the groups at 3 and 12 months, as shown in figure 7.13.

7.2.3.3 Average group moment profiles for level walking

The moment profile data for the operated hip during stance phase of level walking was summarised according to the arthroplasty type. The mean moment profiles were averaged for each arthroplasty group at 3 and 12 months and the data compared for moments in all 3-dimensions.

Figure 7.14 shows the mean adductor moment profiles for the standard THR and resurfacing arthroplasty group at 3 and 12 months following surgery. The moment curves were very similar between groups, particularly at 3 months. At 3 months the main differences included the reduced adductor moment immediately after heel strike and the greater abductor moment at mid-stance for the standard THR group. By 12 months, the adductor moment peak at initial contact increased for both groups but remained relatively lower for the standard group. This difference was significant (p=0.049), implying that the resurfacing group had greater peak abductor moments than the standard group for level walking (figure 7.17). The minimum value of the abductor moment at mid-stance was similar between 3 and 12 months for the resurfacing group, whereas that equivalent value for the standard THR group reduced by 12 months, and was similar to that for the resurfacing group. When contrasting the 3 and 12 month moment profiles, it can be seen that the variation in the first abductor moment peak (20-30% stance) was the greatest between-group difference, with the resurfacing group displaying a reduced time to reach the peak and a relatively greater peak abductor moment value.

The variability of the adductor moments showed some small reductions by 12 months but this may be better viewed from figure 7.17 which summarises moments in all 6 degrees of freedom.

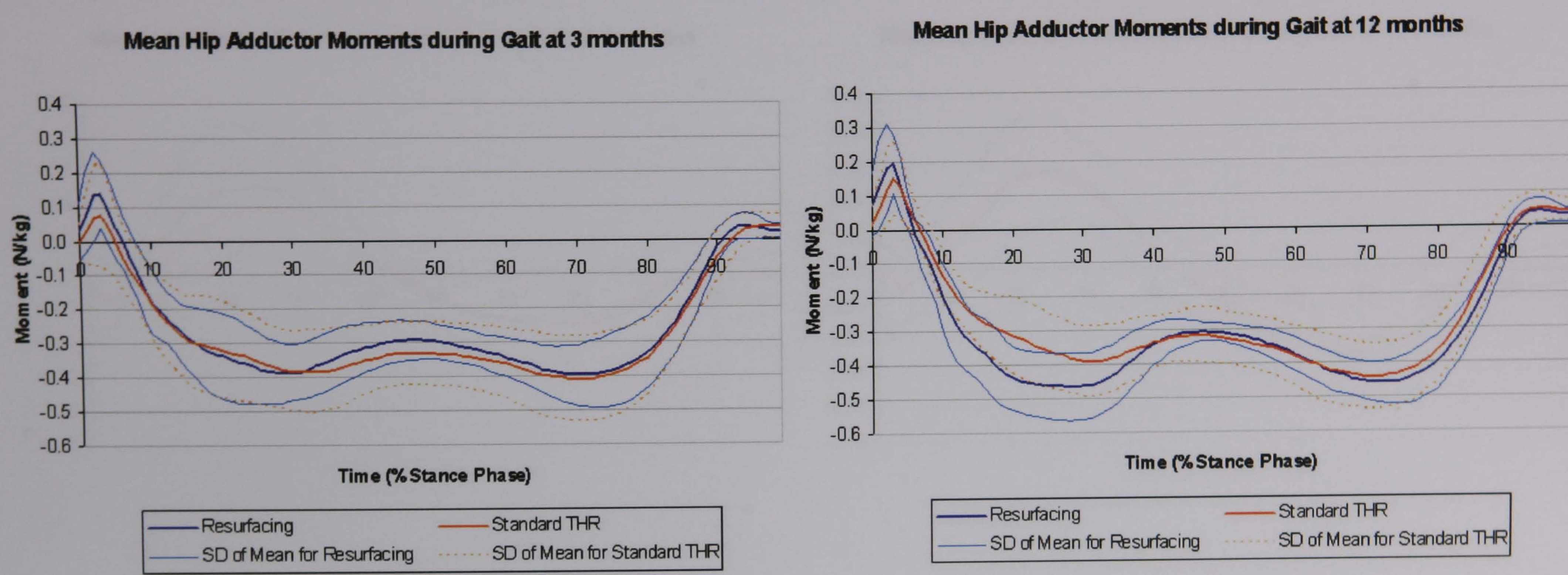
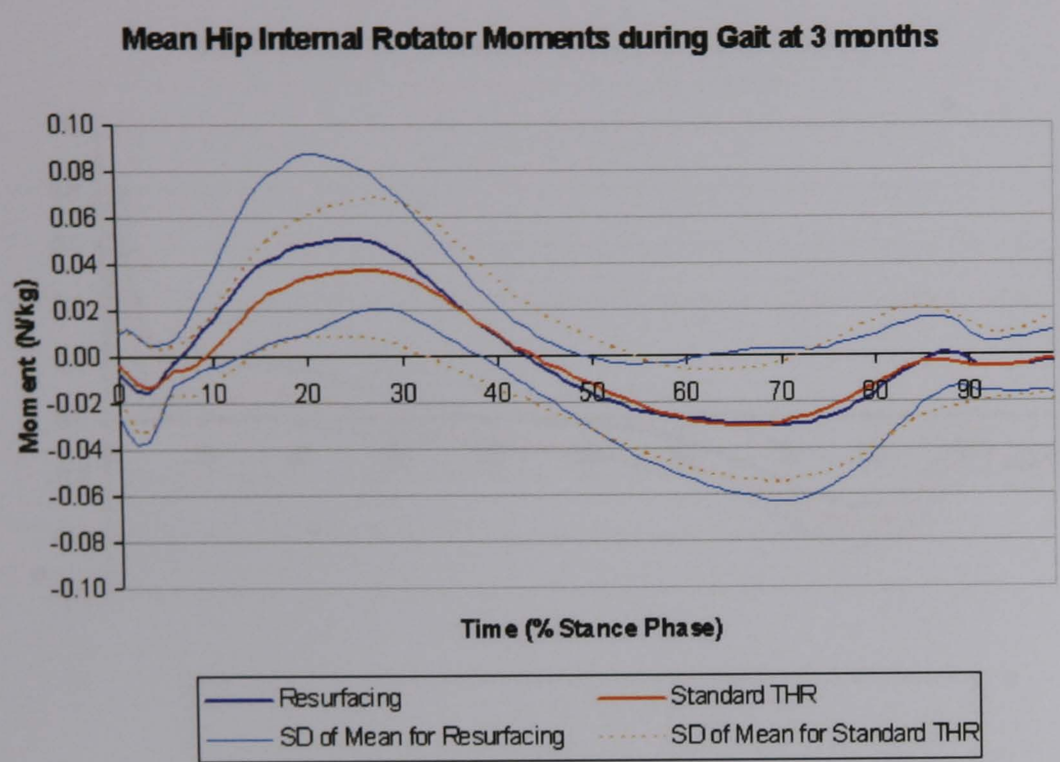


Figure 7.14: Mean hip adductor moments at (a) 3 and (b) 12 months



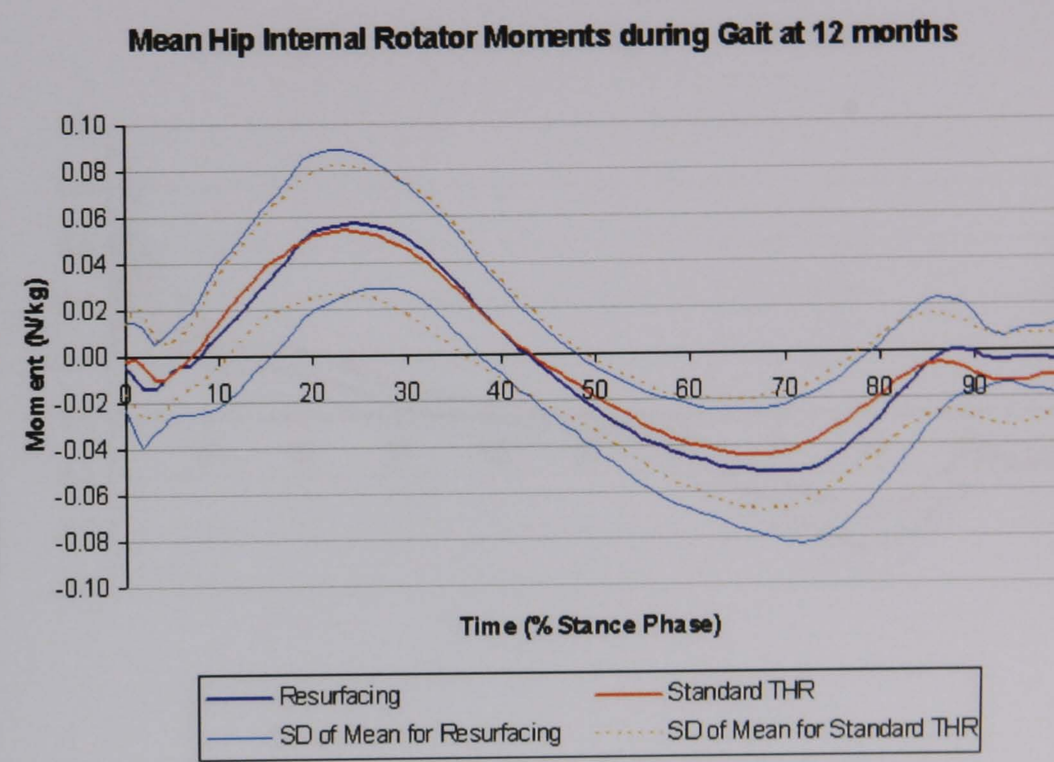


Figure 7.15: Mean hip internal rotator moments at (a) 3 and (b) 12 months

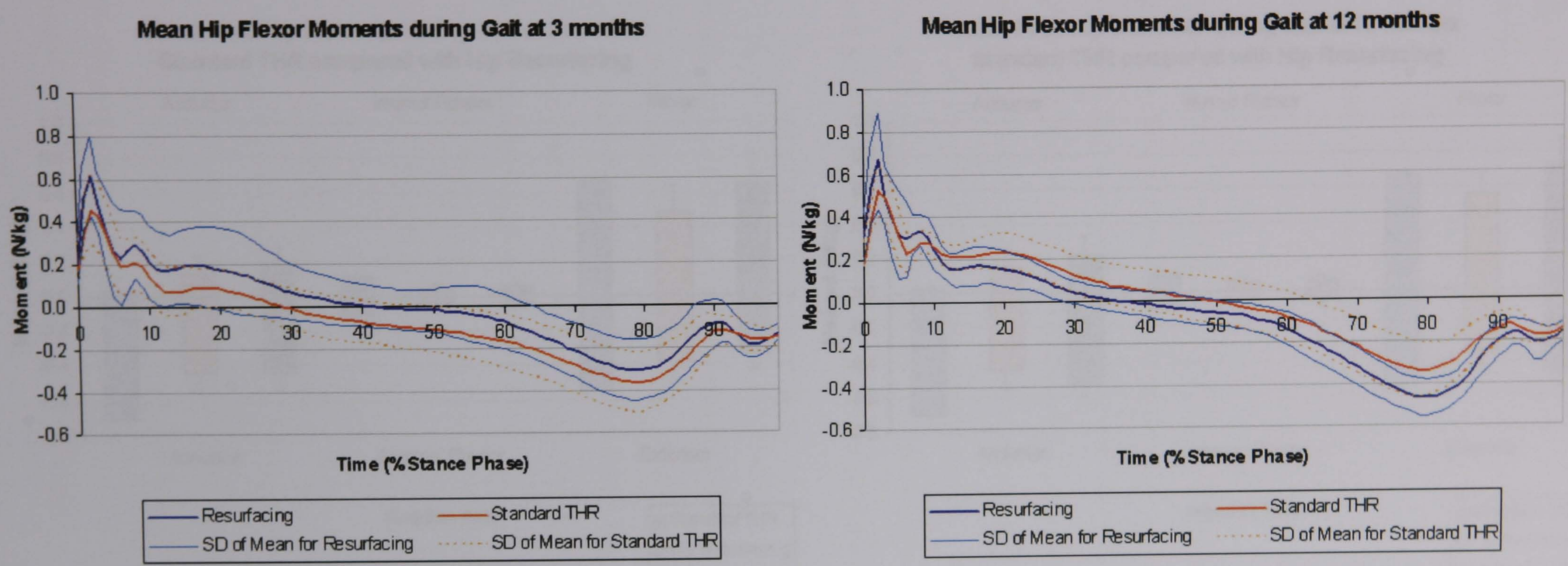


Figure 7.16: Mean hip flexor moments at (a) 3 and (b) 12 months

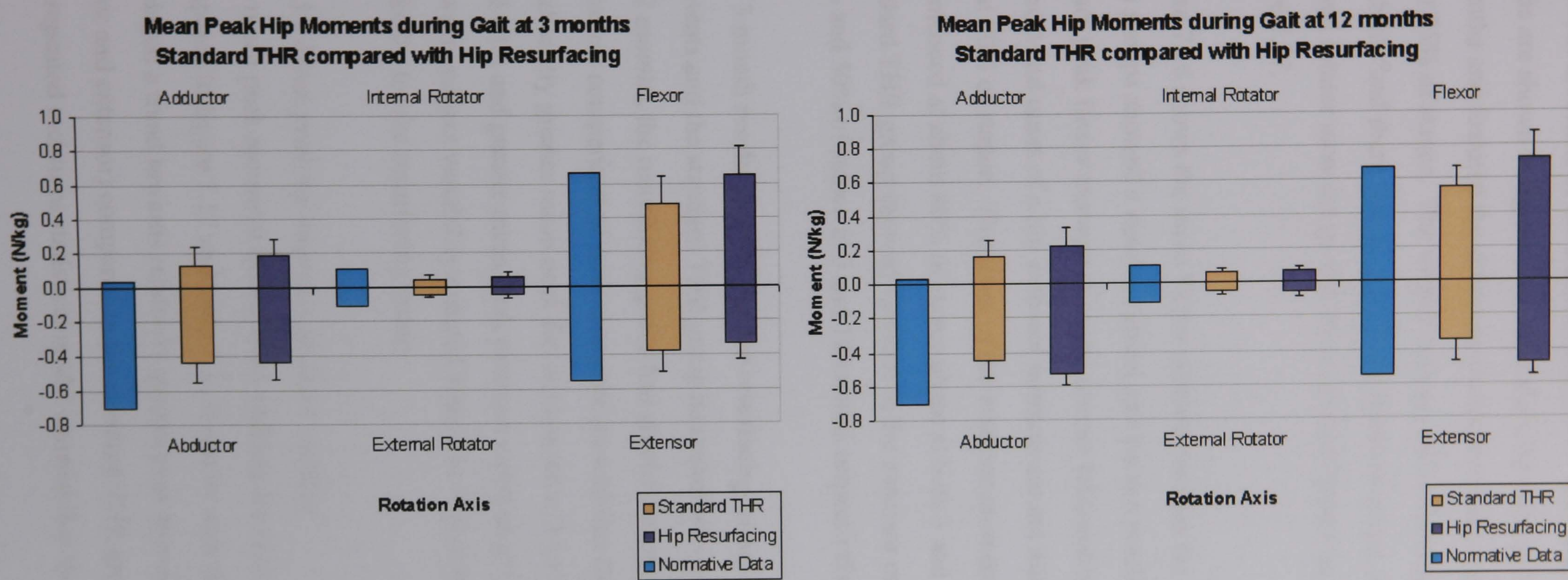


Figure 7.17: Comparison of 3-D hip moments between standard THR and hip resurfacing during level walking at (a) 3 months and (b) 12 months

(Normative data sourced from Eng and Winter 1995; male subjects (n=9), aged 19-26 (mean 22.2) years, mean gait velocity 1.6m/s)

The mean moments of the transverse plane for the resurfacing and standard THR groups are shown in figure 7.15. The main between-group difference was visible at 3 months and featured the difference in the internal rotator peak, which occurred at about 25% of stance. The rotator moment curve featured high between-subject variability and there was no significant between group difference for the peak internal rotator moment (p=0.279) or external rotator moment (p=0.106) as seen in figure 7.17.

Figure 7.16 shows the mean flexor moment profiles for the arthroplasty groups. Both groups showed a similar, typical gait pattern profile of hip flexor moments with a sharp peak flexor moment at initial contact followed by a reduction in the flexor moment and onset of a hip extensor moment around mid-stance, which peaked at about 80% of stance. The onset of the extensor moment in the resurfacing group commenced at about 40% of stance phase at both 3 and 12 months, whereas the standard THR group showed variance in the extensor moment onset, occurring at 30% and 50% of stance at 3 and 12 months respectively.

The 3 month results showed that the resurfacing group had relatively greater flexor moments and the standard THR group had relatively greater extensor moments; and at 12 months, the resurfacing group had greater flexor and extensor moments. Statistical comparison of peak moments showed that the resurfacing group had a significantly greater mean peak flexor moment at 3 (p=0.003) and 12 months (p=0.003) and greater mean peak extensor moment at 12 months (p=0.005). Between-subject variability reduced from 3 to 12 months for both groups, but more markedly for the resurfacing group.

7.2.3.4 Mean peak hip moments for level walking

The mean peak moments during level walking for both arthroplasty groups are compared in figure 7.17 and table 7.3. It can be seen that the resurfacing group displayed a trend towards relatively greater peak hip moments (adductor, abductor, flexor and extensor) compared to the standard THR group, certainly at 12 months. A full repeated measures statistical model showed that the between-group differences in the peak moment outcomes were significant for the adductor (p=0.049), flexor (p=0) and extensor moments (p=0.02). The main difference between 3 and 12 months was the improvement (increase) in the abductor and extensor moments of the resurfacing arthroplasty group.

Table 7.3: Mean peak hip moments of the operated limb during level walking

PEAK MOMENTS (N/kg)	Resurfacing		Standard THR	
3 months	Mean	SD	Mean	SD
Adductor	0.184	0.102	0.130	0.118
Abductor	0.438	0.098	0.439	0.116
Int Rotator	0.059	0.031	0.045	0.025
Ext Rotator	0.045	0.019	0.041	0.018
Flexor	0.649	0.166	0.483	0.157
Extensor	0.331	0.099	0.377	0.126

PEAK MOMENTS (N/kg)	Resurfacing		Standard THR	
12 months	Mean	SD	Mean	SD
Adductor	0.224	0.105	0.163	0.093
Abductor	0.529	0.065	0.456	0.096
Int Rotator	0.063	0.028	0.060	0.026
Ext Rotator	0.057	0.028	0.051	0.021
Flexor	0.721	0.153	0.551	0.117
Extensor	0.477	0.075	0.345	0.122

The greater hip moments of the resurfacing group compared to the standard THR group must be interpreted with care as the resurfacing group tended to walk faster. This will be considered further in section 7.5.

7.2.4 Summary of within- and between-group differences for level walking

Comparing hip joint angles for the operated and control limbs during level walking showed that the operated hip had significantly reduced external rotation and flexion compared to the control hip and tended to maintain a more adducted and significantly more internally rotated position throughout the gait cycle. Sagittal plane angular excursion was reduced on the operated hip and the peak extension angle at terminal stance was significantly reduced at 3 months but improved by 12 months. The control limb adopted a more adducted position by 12 months, showing greater similarities to the operated limb.

When contrasting the finding between the arthroplasty groups, no significant differences for any of the peak angle outcomes were identified. The standard THR group were relatively more externally rotated than the resurfacing group. Aside for the sagittal plane motion at 3 months, the standard THR group displayed more between subject variability in the angle outcomes. With time. the peak internal rotation angles increased (standard THR>resurfacing) and sagittal plane motion increased (standard THR>resurfacing). The resurfacing group had a relatively greater gain in extension and abduction peak angles with time, which was significant for the extension peak.

Moment differences between operated and control limbs during level walking were present and statistically significant for flexor. extensor, internal and external rotator moments and for abductor moments at 3 months alone, where peak moments of the operated hip were sub-optimal. With time the magnitude of difference between-limbs reduced. In particular, the peak abductor moment of the operated limb increased with time, whereas this outcome reduced for the control limb.

Comparing between the arthroplasty groups, the flexor and extensor moments showed the most significant differences. Overall the resurfacing group had greater peak moments, which may have been velocity related. The abductor and extensor moments for the resurfacing group showed a larger trend increase with time.

7.3 Results from Stair Ascent

Stick figure representations during stair ascent are shown in figures 7.18a, 7.19a and 7.20a following the format used for level walking (pictures every 0.051 seconds).

7.3.1 Comparison of hip angles between limbs for stair ascent

7.3.1.1 Individual angle profiles during stair ascent

The typical 3-dimensional angle profiles for stair ascent of the operated and control hips were similar to those observed for level walking (Appendix 9). The angle profiles typically followed the trends observed for level walking, with the operated limb adopting a more adducted and internally rotated position throughout the gait cycle. Sagittal and transverse plane motion of the operated limb typically displayed less angular excursion than the control limb.

7.3.1.2 Mean peak hip angles during stair ascent

The peak hip angles for each individual were averaged and summarised according to each arthroplasty group. The results for the operated and control limbs were then compared. Figure 7.18 and 7.19 show the results for the standard THR group and the resurfacing group at 3 and 12 months respectively. Overall the trends in the data comparisons between-limbs for stair ascent were similar to those observed for gait.

There were significant between-limb differences for the peak adduction and abduction angles during stair ascent, with greater adduction angles for the operated limb (p=0.032) and greater abduction angles for the control limb (p=0.012). This trend was noted for level walking but was not found to be statistically significant. The between-limb differences for peak coronal plane angles appeared to have a stronger trend for the resurfacing group (figure 7.19) but this observation was not significant (adduction p=0.247; abduction p=0.509). For both groups, the control limb had significantly greater external rotation angles than the operated limb (p=0.008). At 3 months, total hip rotation was greater for the control limb but by 12 months the peak external rotation values reduced significantly (p=0.013). The trend towards greater internal rotation of the operated hips continued at 12 months (p=0.407) but the difference in total angular excursion between limbs narrowed. The standard THR group displayed little difference in the mean peak sagittal plane angles between-limbs, whereas the resurfacing group continued to demonstrate reduced sagittal mean peak angles on the operated limb (figures 7.18 and 7.19). Overall, the repeated measures ANOVA for the full sample group showed significantly reduced peak flexion angles of the operated limbs (p=0.047) and a strong trend toward reduced peak extension angles of the operated hip (p=0.058).

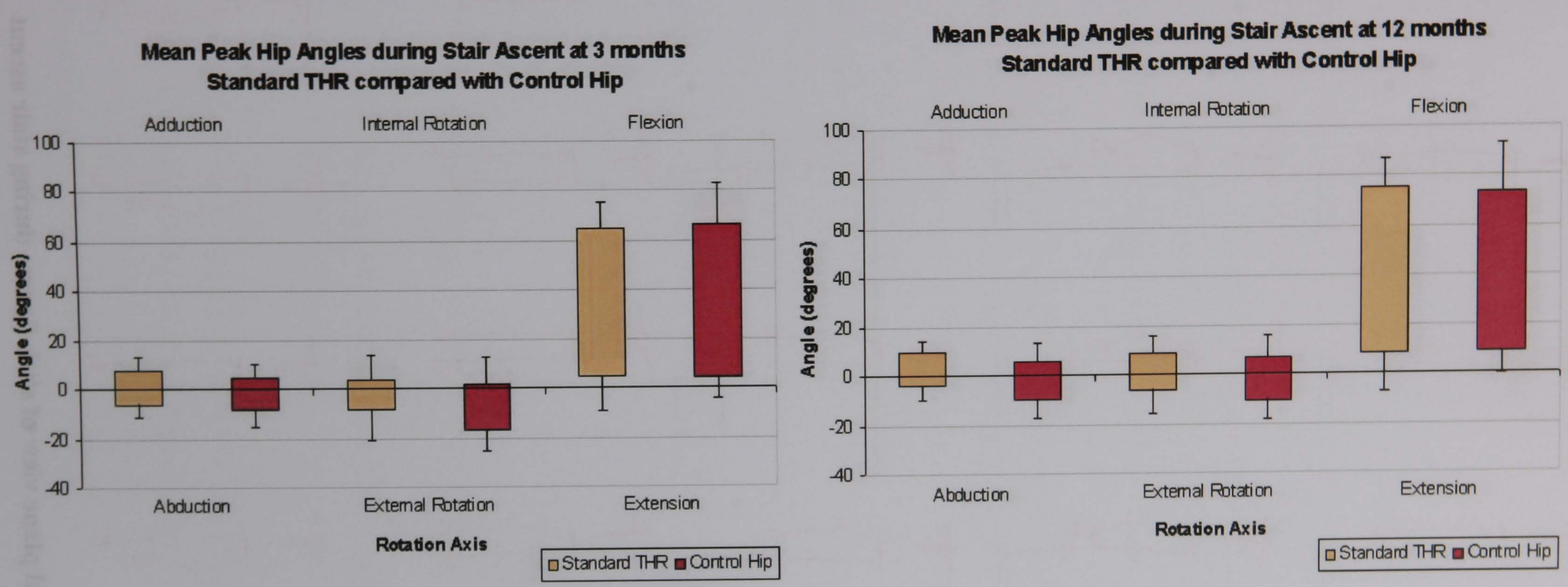


Figure 7.18: Comparison of 3-D hip angles for standard THR group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months

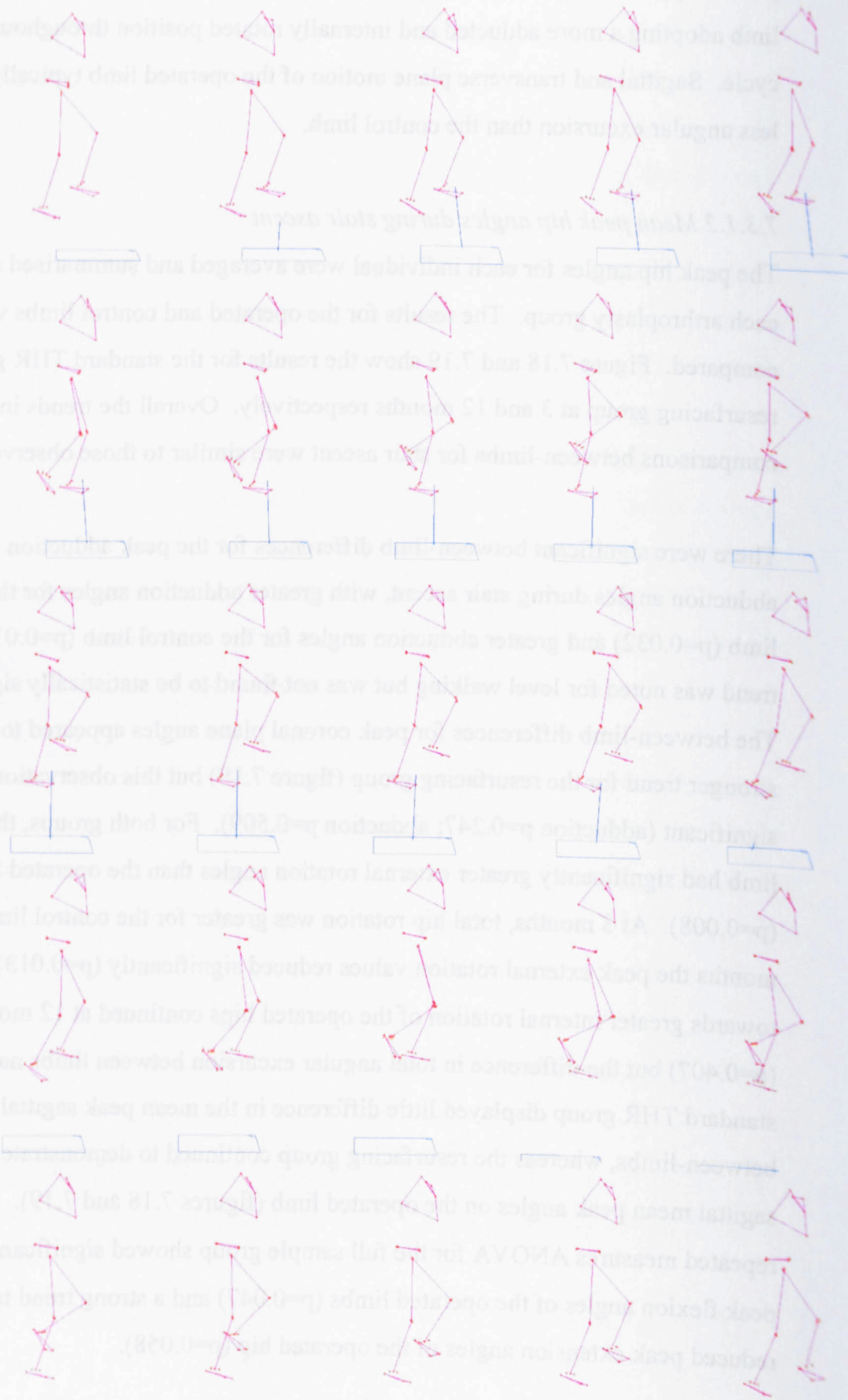


Figure 7.18a: Sagittal plane view of stick figures during stair ascent

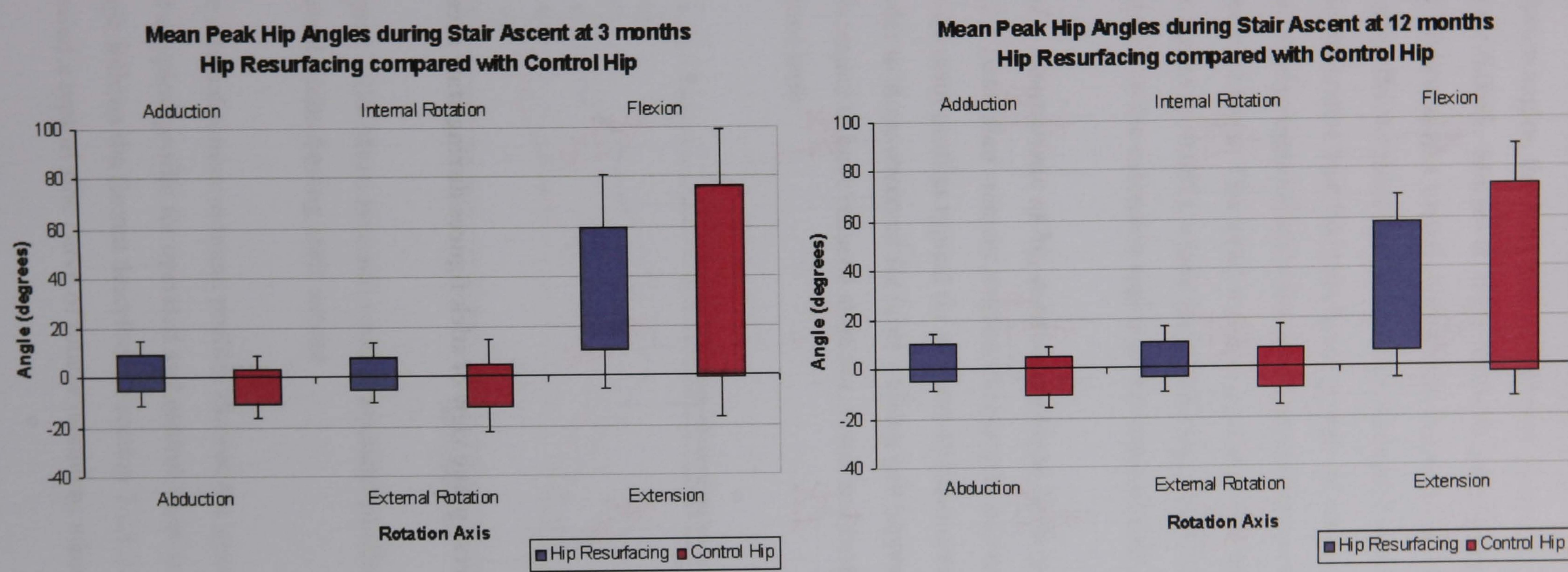


Figure 7.19: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months

Figure 7.19a: Coronal plane view of stick figures during stair ascent

Whilst the 3 month data for level walking showed significantly reduced peak extension angles, the early stair ascent data showed a strong trend lack of extension only (p=0.068). Moreover, the differences between operated and control limbs for peak flexion angles varied significantly between the arthroplasty groups (p=0.036), showing that the standard THR group had significantly less between-limb flexion angle difference than the resurfacing group. In other words, the resurfacing group were largely responsible for the significant reduction in flexion angles compared to the control limb. This trend was also consistent for peak extension angles during stair ascent (p=0.062), where the resurfacing group were responsible for the trend reduction in the extension angles of the operated limb.

7.3.2 *Comparison of hip moments between limbs for stair ascent*

7.3.2.1 Individual moment profiles during stair ascent

Hip moment profiles typical for subjects of either arthroplasty group were also similar to those observed for level walking gait (appendix 9). Typically the operated limb tended to have reduced abductor, flexor and extensor moments compared to the control limb.

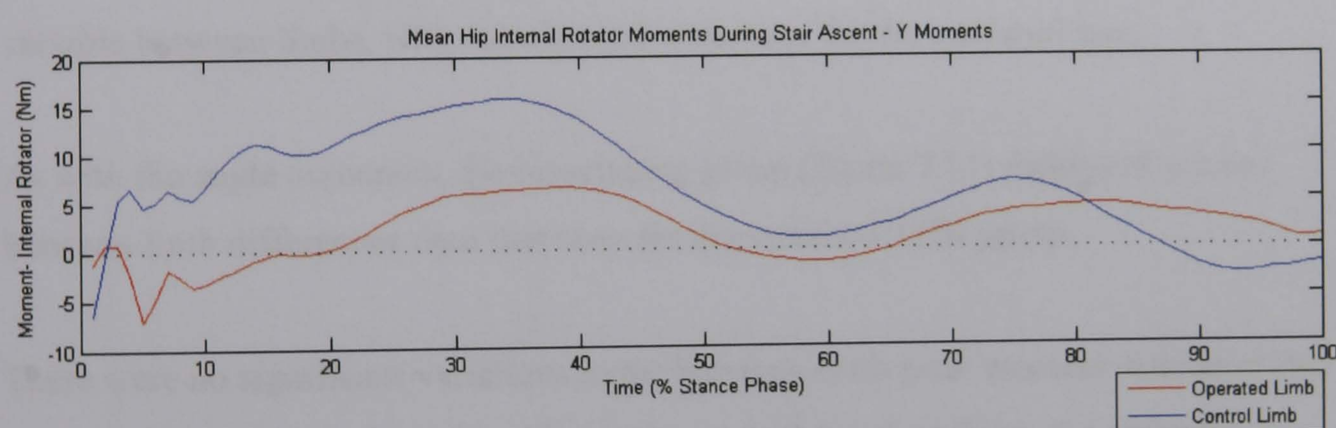


Figure 7.20: Mean internal rotator moment profiles typical of operated and control limbs during stair ascent

The internal rotator moment profiles showed the greatest discrepancy with level gait, and a typical profile for operated and control limbs is shown in figure 7.20. The graph follows the format described in section 7.2.1.2. The curve for level walking showed a typical sine curve of rotator moments, whereas the stair ascent curve was a

Figure 7.20a: Transverse plane view of stick figures during stair ascent

typical two peak internal rotator moment curve with greater internal rotator moments for the control limb throughout the stance phase.

7.3.2.2 Mean peak hip moments during stair ascent

The peak hip moments for each individual were normalised (to body mass and height), averaged and then summarised according to each arthroplasty group. The results for the operated and control limbs were then compared.

The results of the hip moments for the standard THR group are shown in appendix 9 and those for the resurfacing group in figure 7.21.

There were greater variations between-limbs at 3 months than at 12 months. Statistical analysis of the 3 month data alone showed that the operated hips had significantly reduced abductor (p=0.009), flexor (p=0.002) and internal rotator (p=0) moments relative to the control hips. There was also a trend of a reduced peak extensor moment (p=0.059) of the operated limb compared to the control limb. Repeated measures analysis, accounting for both post-operative data sets, showed that flexor (p=0.008) and internal rotator (p=0.019) moments were consistently variable between limbs, with sub-optimal outcomes for the operated hips.

As with the angle outcomes, the resurfacing group (figure 7.21) displayed greater between-limb differences than that seen for the standard THR group.

There were no significant variations in the between-limb peak moment outcomes due to time. Interestingly, from 3 to 12 months, as with level walking, the abductor moments of the operated hips had improved and became slightly greater than the mean peak abductor moment for the control hips, showing a trend interaction of time with the between-limb variable (p=0.058).

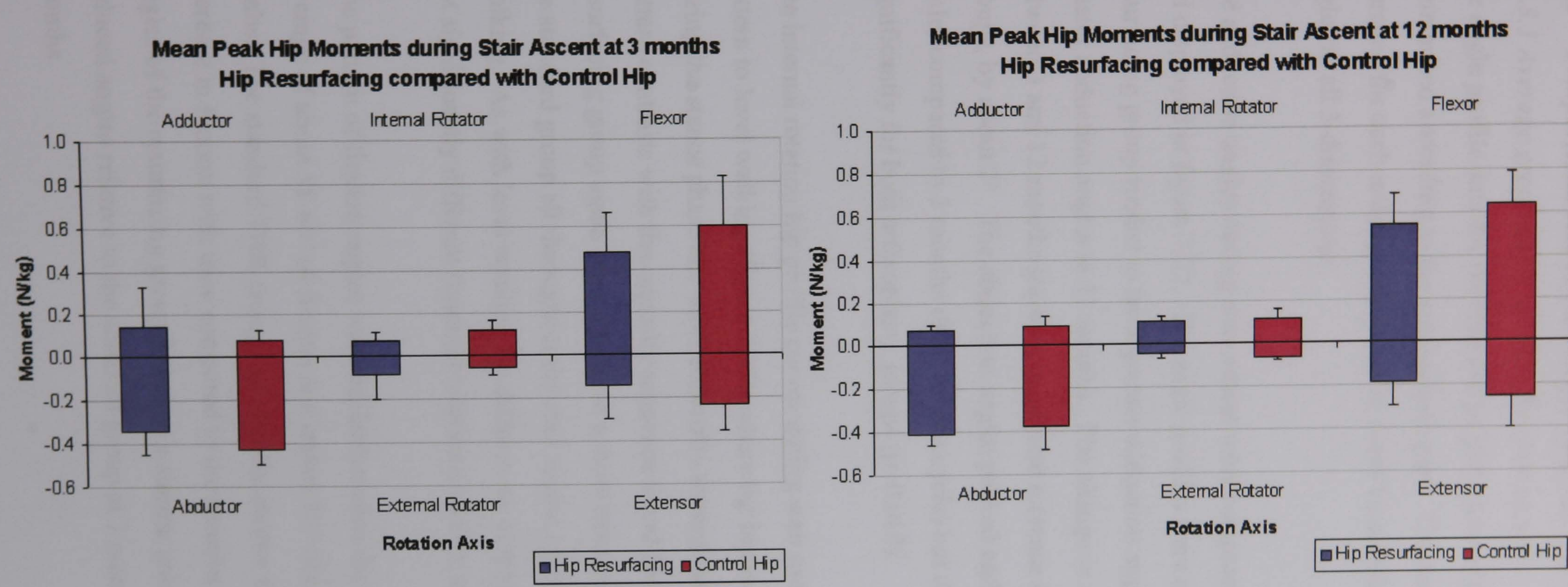


Figure 7.21: Comparison of 3-D hip moments for hip resurfacing group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months

7.3.3 *Comparisons between arthroplasty groups for stair ascent*

7.3.3.1 Average group angle profiles for stair ascent

The angle profile data for the stair ascent gait cycle of the operated hip was summarised according to the arthroplasty type. The mean angle profiles were averaged for each arthroplasty group at 3 and 12 months and the data compared for angles in all 3-dimensions.

The adduction angles during stair ascent were compared between arthroplasty groups and displayed in figure 7.22. The angle profiles were similar for both groups. The resurfacing group tended to have greater adduction angles at 3 and 12 months and greater abduction angles at 12 months. The change in adduction angle curves between 3 and 12 months shows a consistent increase in the adduction peak for both groups, by about 2°. The abduction angles peaked earlier at 12 months (70% of gait cycle) compared to 3 months (90% of gait cycle) but the peak value reduced significantly for both arthroplasty groups ($p=0.019$).

The internal rotation hip profile curves during stair ascent follow a slightly different pattern to level walking (figure 7.23), where the hip was relatively externally rotated during the stance phase and more internally rotated during the swing phase. This pattern contrasts with the opposite sequence found for level walking. The resurfacing group again tended to have a more internally rotated position relative to the standard group all throughout the stair ascent cycle, much like the trends for level walking. As with level walking the differences in rotation between the groups were not significantly different (internal rotation $p=0.333$; external rotation $p=0.234$).

The pattern of flexion angles was similar between the arthroplasty groups with an average of about 55-60° of flexion and minus 10° of extension for the operated limbs. The standard THR group had slightly greater flexion angles and had a greater increase in flexion with time compared to the resurfacing group. The extension angles of the resurfacing group followed a similar pattern to level walking with reduced angles relative to the standard group at 3 months and the opposite trend at 12 months.

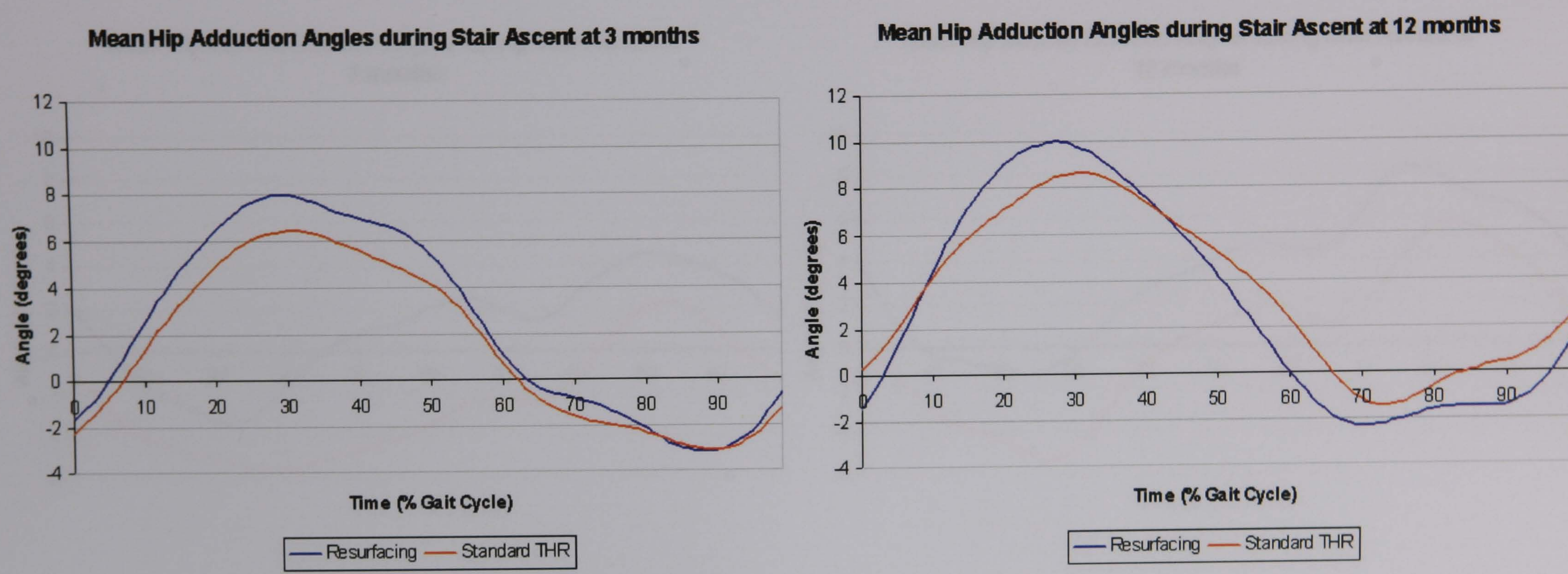


Figure 7.22: Mean hip adduction angles at (a) 3 and (b) 12 months

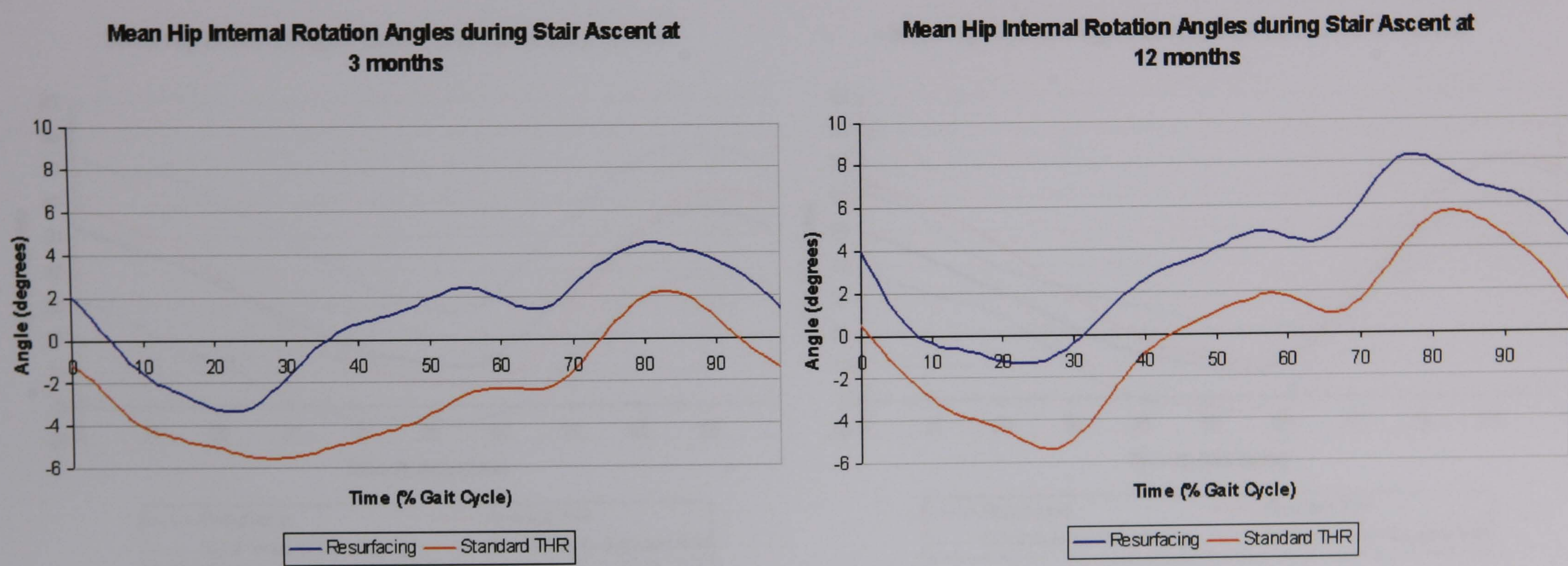


Figure 7.23: Mean hip internal rotation angles at (a) 3 and (b) 12 months

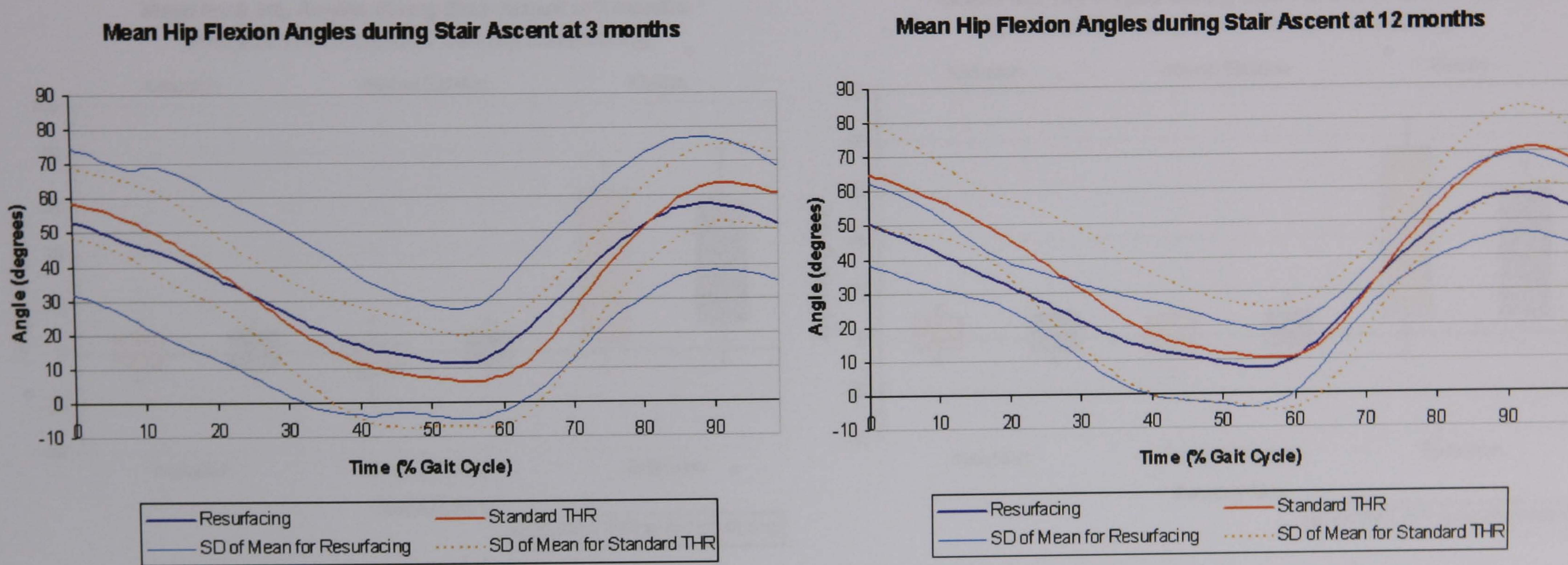
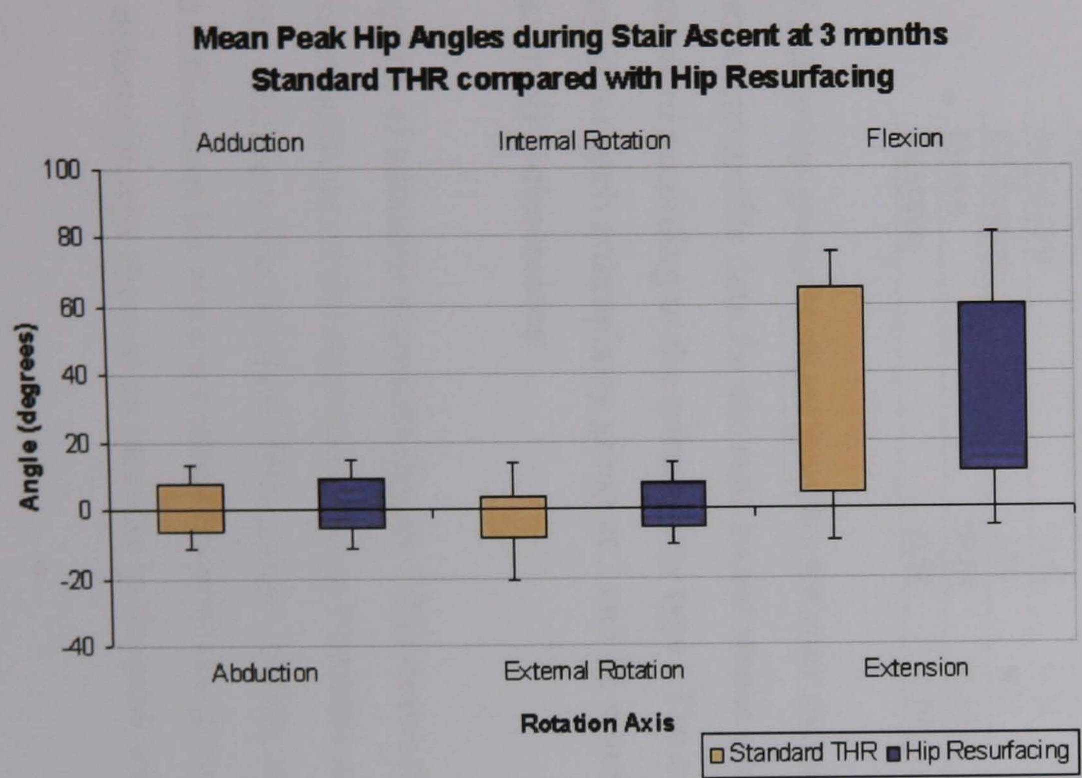


Figure 7.24: Mean hip flexion angles at (a) 3 and (b) 12 months



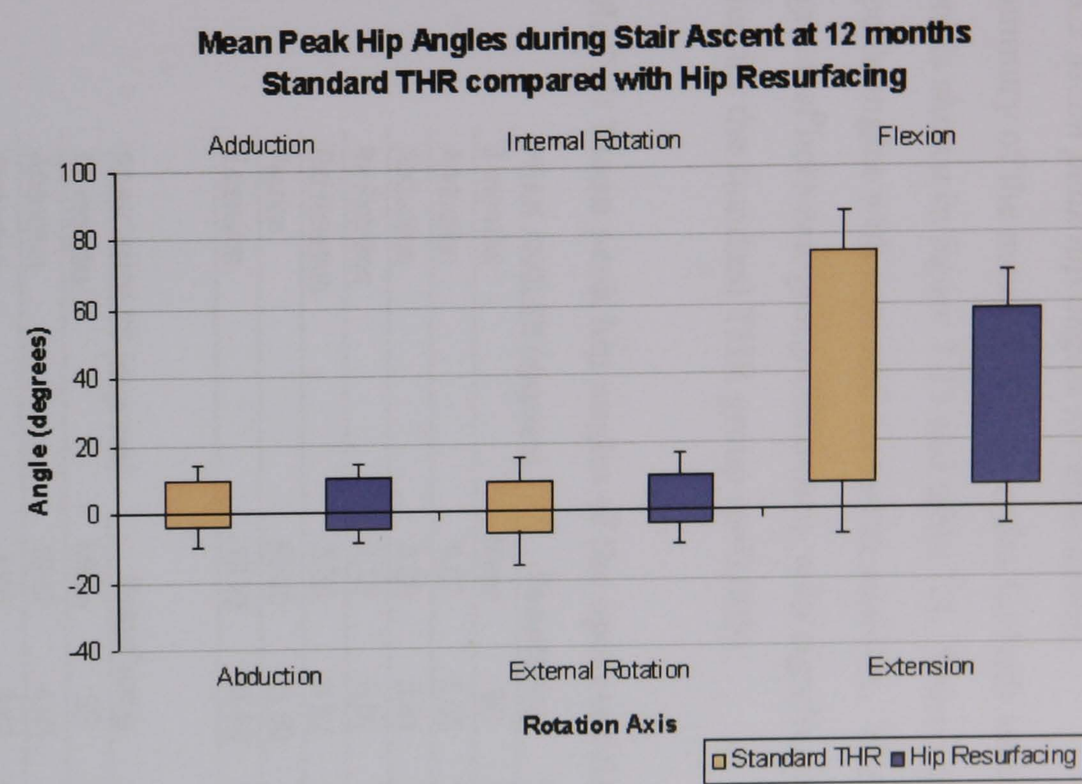


Figure 7.25: Comparison of 3-D hip angles between standard THR and hip resurfacing during stair ascent at (a) 3 months and (b) 12 months

7.3.3.2 Mean peak hip angles for stair ascent

A summary of the mean peak hip angles for both arthroplasty groups during stair ascent is shown in figure 7.25 and table 7.4. There were no significant differences in the peak angles with repeated measures analysis. The 12 month data alone showed the greatest between-group difference, with significantly greater maximum flexion angles for the standard THR group (p=0.005).

Table 7.4: Mean peak hip angles of the operated limb during stair ascent

PEAK ANGLES (degrees)	Resurfacing		Standard THR	
3 months	Mean	SD	Mean	SD
Adduction	9.07	6.08	7.46	5.76
Abduction	5.29	5.84	5.72	5.12
Int Rotation	7.56	6.50	3.76	10.14
Ext Rotation	5.39	4.93	8.50	12.27
Flexion	60.00	20.59	64.73	10.41
Extension	-10.23	15.90	-4.69	14.03

PEAK ANGLES (degrees)	Resurfacing		Standard THR	
12 months	Mean	SD	Mean	SD
Adduction	10.57	4.03	9.86	4.48
Abduction	4.94	3.62	3.63	6.55
Int Rotation	10.40	6.32	8.65	7.71
Ext Rotation	3.86	5.84	6.25	9.29
Flexion	57.84	11.43	75.18	11.74
Extension	-6.54	11.54	-7.87	14.95

7.3.3.3 Average group moment profiles for stair ascent

The moment profile data for the stair ascent stance phase of the operated hip was summarised according to the arthroplasty type. The mean moment profiles were averaged for each arthroplasty group at 3 and 12 months and the data compared for angles in all 3-dimensions.

The graph of adductor moments (figure 7.26) shows that the resurfacing group had poorer magnitude of abductor moments at 3 months and a greater magnitude at 12 months relative to the standard THR group. Hence, stair ascent highlighted greater trend differences for abductor moments between groups. From 3 to 12 months, both arthroplasty groups showed an increase in the peak abductor moments.

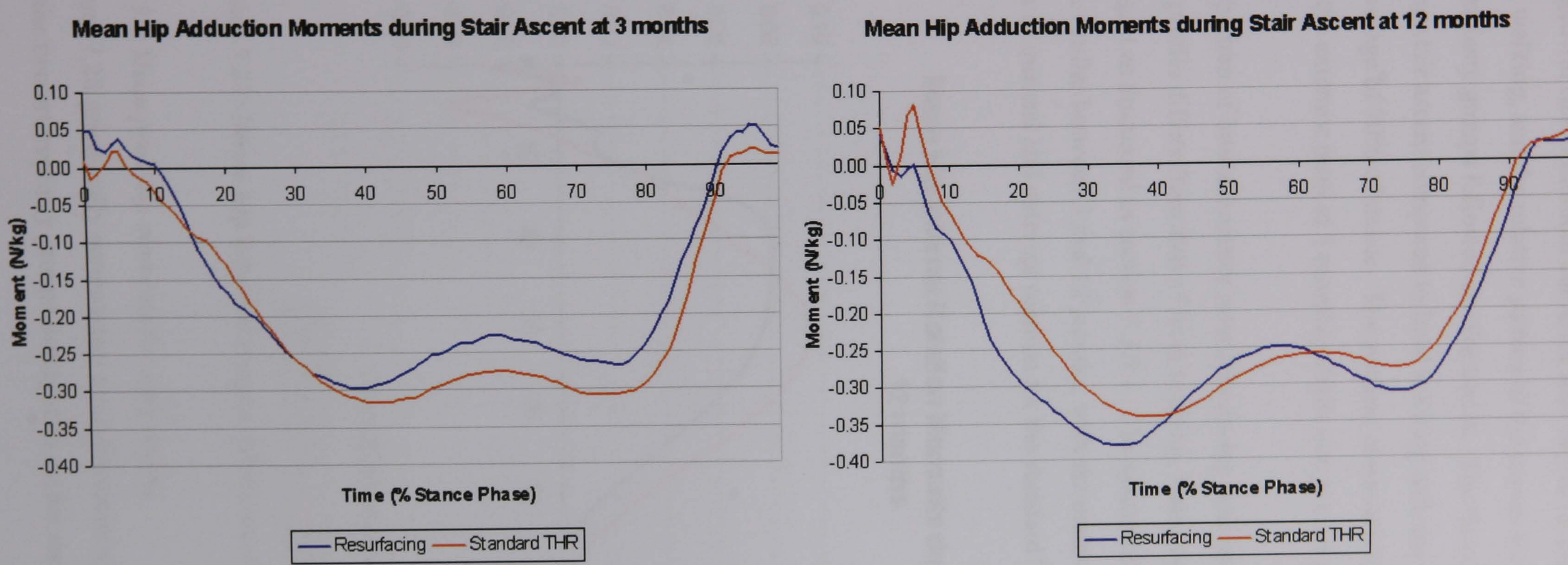


Figure 7.26: Mean hip adductor moments at (a) 3 and (b) 12 months

The general flexor moment curve for stair ascent (appendix 9) was similar to that of level walking, and the relative pattern of the curves for standard and resurfacing arthroplasty groups followed similar trends. The flexor moment was prolonged during stair ascent compared to level walking with the extensor moment on-setting at an average of 70% of stance. The greatest observed difference between the groups was the extensor peak at 3 months, which was greater for the standard THR group.

The pattern of internal rotator moments during stair ascent is shown in figure 7.27. The profile differs from that of level walking, featuring a double peak internal rotator moment as discussed in section 7.3.2.1. The results of the internal rotator moments were similar between 3 and 12 months, with the main differences including a greater peak of internal and external rotation for the standard THR group.

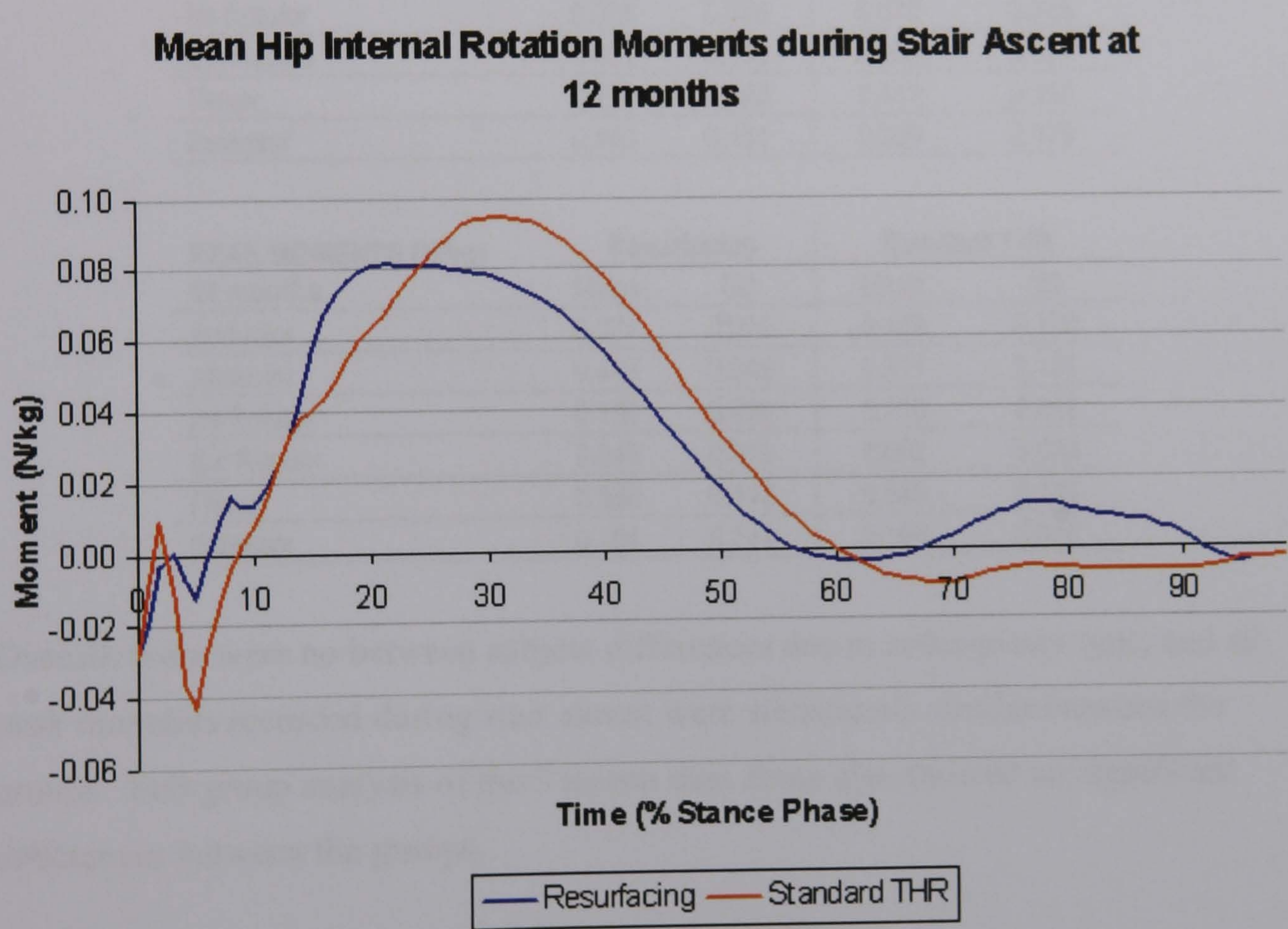


Figure 7.27: Mean hip internal rotator moments at 12 months

7.3.3.4 Mean peak hip moments for stair ascent

Figure 7.28 graphically summarises the differences in the mean peak moment values for the two arthroplasty groups during stair ascent and table 7.5 documents the corresponding peak values. The 3 months results show inconsistent trend differences, much like the level walking trends with skewed trend differences for sagittal and coronal moments. At 3 months the resurfacing hips had slightly poorer abductor moments relative to the standard hips, but greater abductor moments at 12 months. The opposite was true for the adductor moments with the resurfacing hips having poorer adductor moments at 12 months. Similarly, the resurfacing hips had greater flexor moments and poorer extensor moments than the standard hips at 3 months and no difference at 12 months.

Table 7.5: Mean peak hip moments of the operated limb during stair ascent

PEAK MOMENTS (N/kg)	Resurfacing		Standard THR	
3 months	Mean	SD	Mean	SD
Adductor	0.147	0.179	0.069	0.077
Abductor	0.340	0.114	0.359	0.152
Int Rotator	0.074	0.036	0.078	0.026
Ext Rotator	0.081	0.119	0.053	0.031
Flexor	0.476	0.183	0.413	0.150
Extensor	0.143	0.151	0.230	0.179

PEAK MOMENTS (N/kg)	Resurfacing		Standard THR	
12 months	Mean	SD	Mean	SD
Adductor	0.071	0.025	0.129	0.130
Abductor	0.413	0.048	0.371	0.121
Int Rotator	0.106	0.028	0.110	0.042
Ext Rotator	0.045	0.019	0.062	0.023
Flexor	0.550	0.142	0.549	0.136
Extensor	0.184	0.114	0.198	0.078

Overall, there were no between subject differences due to arthroplasty type, and all peak moments recorded during stair ascent were statistically similar between the groups. Sub-group analysis of the 3 month data alone also showed no significant differences between the groups.

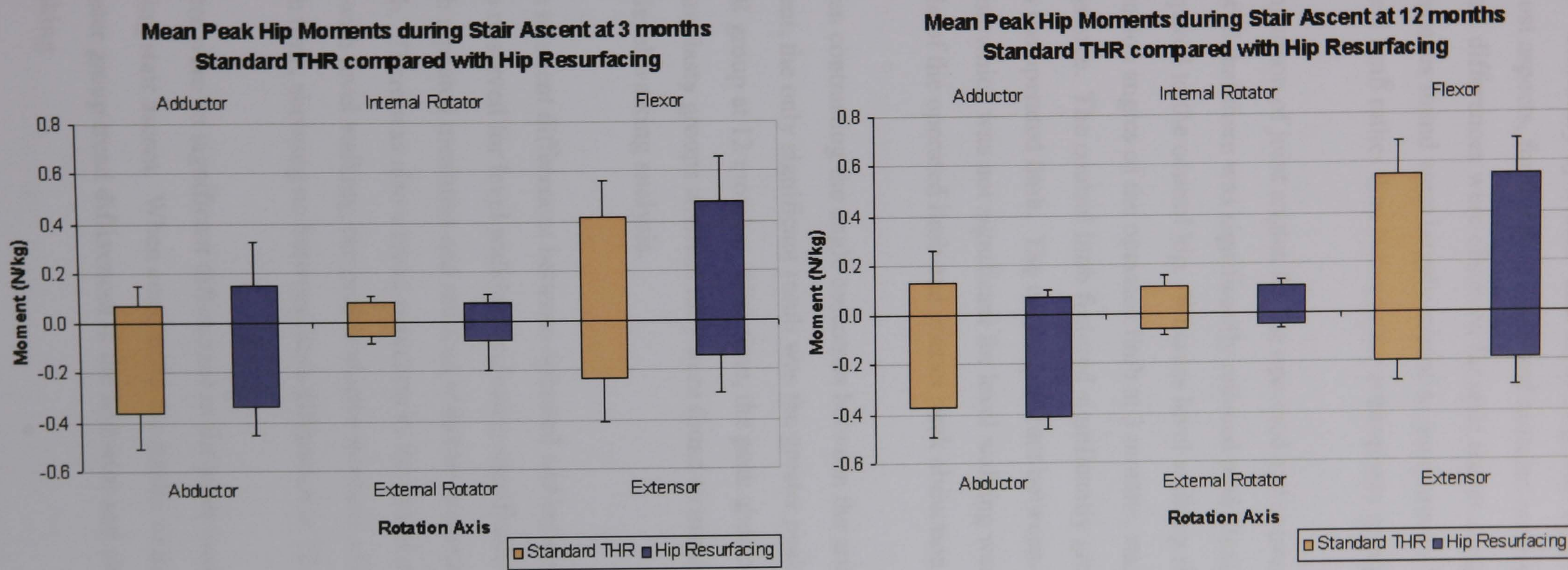


Figure 7.28: Comparison of 3-D hip moments between standard THR and hip resurfacing during stair ascent at (a) 3 months and (b) 12 months

7.3.4 Summary of within- and between-group differences for stair ascent

In most aspects, for the peak angle and moment outcomes, less within- and between-subject differences were observed for stair ascent compared to level walking. The differences found were largely related to comparisons between the operated and control limb rather than between the arthroplasty groups.

Comparison of joint angles for the operated and control limbs during stair ascent showed that there was significantly reduced peak flexion angles on the operated hip compared to the control hip. Whereas level walking showed significantly reduced extension angles of the operated limb at 3 months, stair ascent only showed a trend difference. The control limb featured significantly greater peak external rotation than the operated limb. The only significant between-limb difference found for stair ascent which was not significant for level walking was the greater peak adduction angle of the operated limb and greater peak abduction angle of the control limb.

When contrasting the angle outcomes between the arthroplasty groups for stair ascent, the only significant result was the greater peak flexion angle of the standard THR group at 12 months. With time, the peak abduction angles reduced for both arthroplasty groups whereas they were found to increase between 3 and 12 months for level walking analysis.

The moment differences between operated and control limbs during stair ascent were also observed for level walking, including significantly reduced peak flexor moments with repeated measures and reduced abductor moments at 3 months for the operated limb. There was also a trend reduction in the extensor moment of the operated limb. As with level walking, the peak abductor moment of the operated limb increased with time, showing no between-limb difference at 12 months.

There were no significant differences in the peak moment outcomes between-groups during stair ascent. When comparing the modes of ambulation, stair ascent showed greater group trend differences in the adductor and abductor moments than level walking.

7.4 Results from Stair Descent

7.4.1 Comparison of hip angles between limbs for stair descent

7.4.1.1 Individual angle profiles during stair descent

3-dimensional angle profiles for stair descent which were typical for subjects of either arthroplasty group are shown in figure 7.29. The graphs show the average (of 3 trials) hip angles during the stair descent gait cycle (foot contact to the following foot contact) following the format described in section 7.2.1.1.

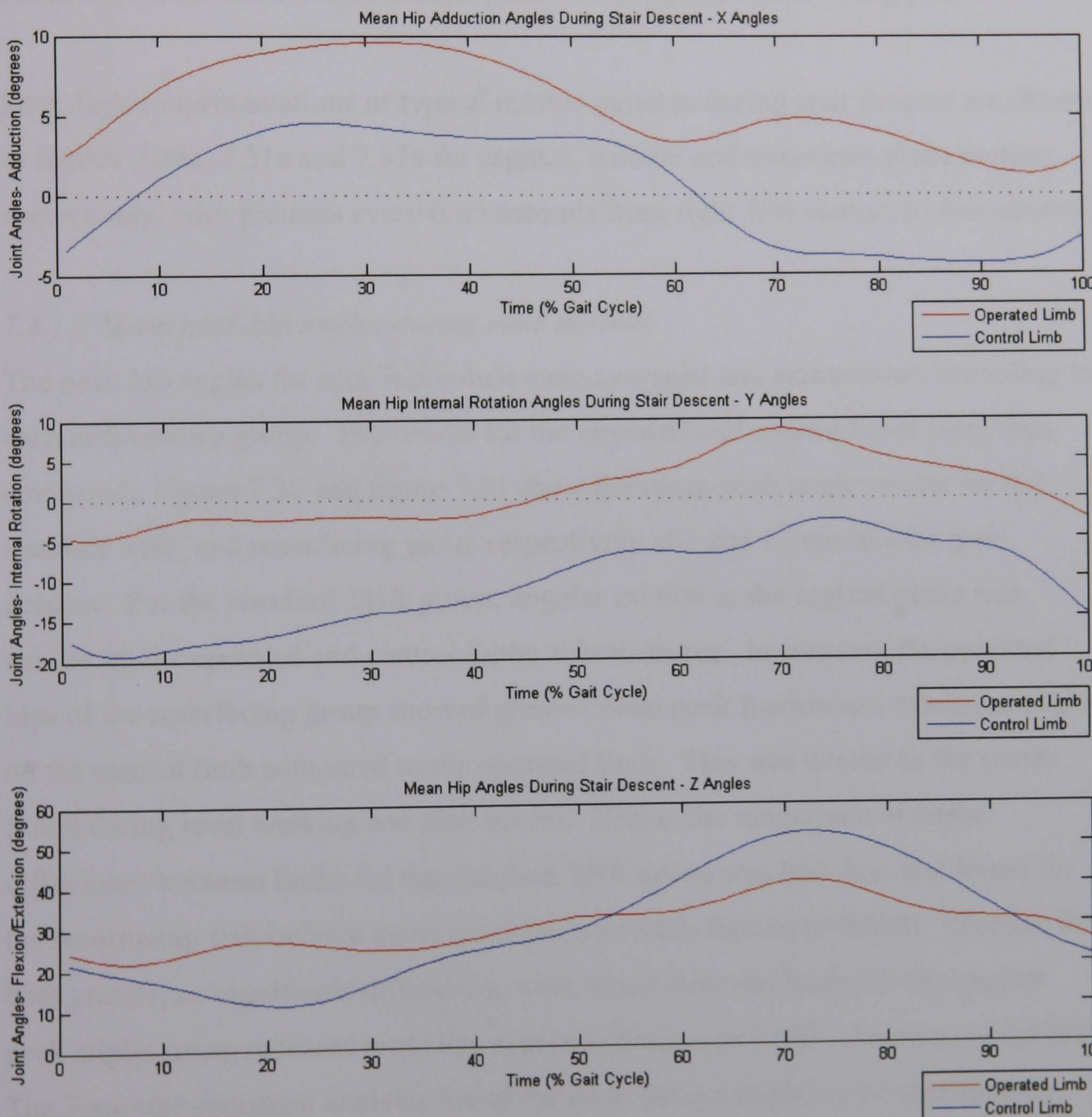


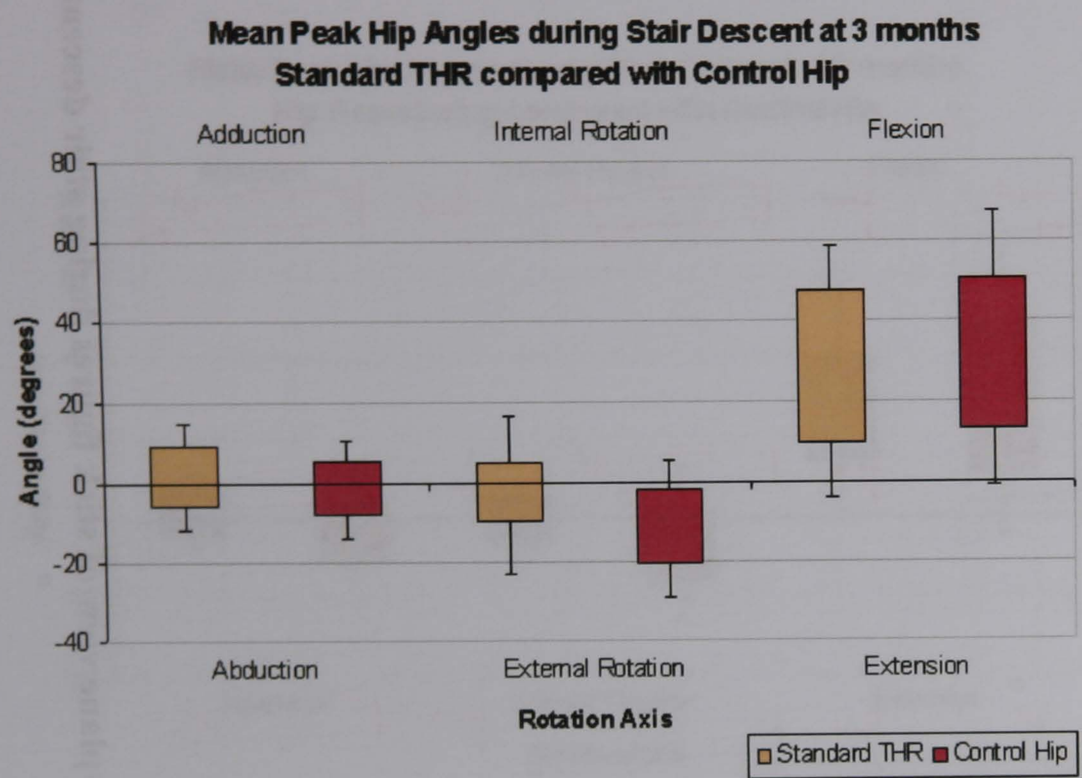
Figure 7.29: Mean 3-D angle profiles typical of operated and control limbs during stair descent

The adduction and internal rotation angles trends typically observed during stair descent were similar to those for level walking and stair ascent, with more adduction and internal rotation of the operated hip compared to the control hip. This pattern was consistent throughout the stair descent cycle. The internal rotation profile followed a similar shape to that of stair ascent. Again, sagittal plane motion of the operated limb typically displayed less angular excursion with smaller peak flexion and extension angles than the control limb. The stair descent flexion curve was also markedly different from the typical curve for level walking and stair ascent. In contrast the curve of the control limb was almost inverted, with the extension peak occurring earlier and a single flexion peak occurring during the swing phase.

Stick figure representations of typical motion patterns during stair descent are shown in figures 7.30a, 7.31a and 7.32a for sagittal, coronal and transverse plane motion respectively, with pictures every 0.05 seconds from right foot contact to foot contact.

7.4.1.2 Mean peak hip angles during stair descent

The peak hip angles for each individual were averaged and summarised according to each arthroplasty group. The results for the operated and control limbs were then compared. Figure 7.30 and figure 7.31 show the mean peak angle results for the standard THR and resurfacing group respectively at 3 and 12 months for stair descent. For the standard THR group, angular motion in the sagittal plane was similar on the operated and control limbs at both stages. In contrast, the operated hips of the resurfacing group showed greater mean peak flexion and extension angles on the control limb compared to the operated limb. This was similar to the trends found during level walking and stair ascent. Hence, the magnitude of angle differences between limbs for the standard THR group was less than that found for the resurfacing arthroplasty group (extension $p=0.04$; flexion $p=0.086$). Overall, for both groups, no significant differences were found between limbs for the sagittal peak angles using repeated measures analysis (flexion $p=0.082$; extension $p=0.973$). The 3 months statistical analysis found the peak flexion angles to be significantly greater on the control limb ($p=0.035$), which was due to the resurfacing group.



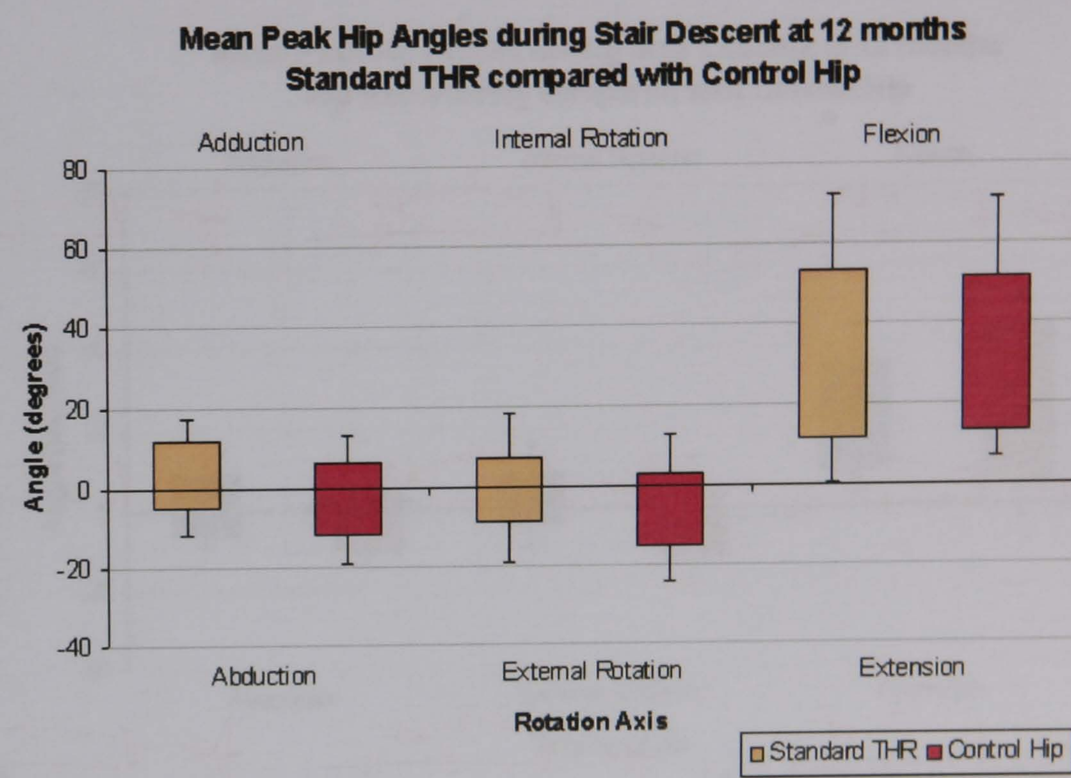
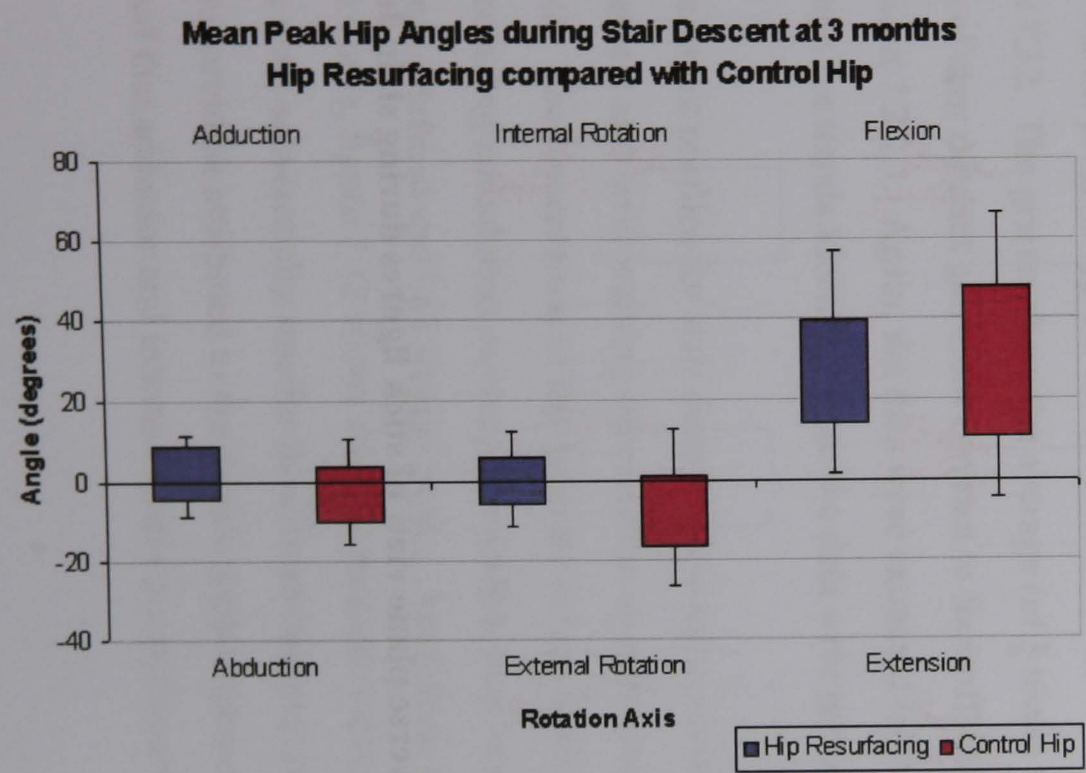


Figure 7.30: Comparison of 3-D hip angles for standard THR group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months

Figure 7.30a: Sagittal plane view of stick figures during stair descent



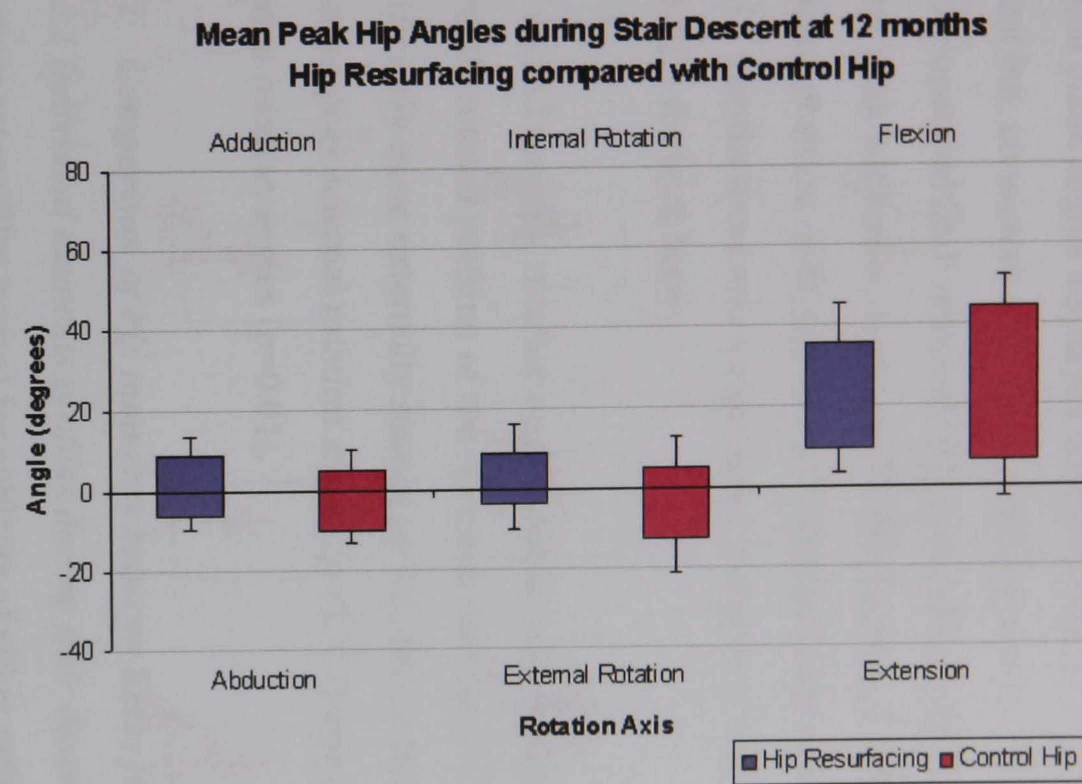


Figure 7.31: Comparison of 3-D hip angles for hip resurfacing group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months

Figure 7.31a: Coronal plane view of stick figures during stair descent

Coronal plane angles displayed similar values of total angular excursion during stair descent but, consistent with the trends from level walking and stair ascent, showed that the operated limb maintained a more adducted position, with significantly greater peak adduction angles (p=0.029) and the control limb maintained a more abducted position, with significantly greater peak abduction angles (p=0.046). From 3 to 12 months there was a significant increase in the mean peak abduction angles (p=0.022) for both limbs.

At 3 and 12 months, rotation angles during stair descent displayed a relatively more internally rotated position of the operated limb (p=0.055) and the control limb was significantly more externally rotated (p=0). By 12 months there was an increase in the mean peak internal rotation angles (p=0.027) and a reduction in the mean peak external rotation angles (p=0.01).

7.4.2 Comparison of hip moments between limbs for stair descent

7.4.2.1 Individual moments profiles during stair descent

Hip moment profiles typical for subjects of either arthroplasty group are shown in figure 7.32. The graphs show the average (of 3 trials) hip moments during the stance phase of stair descent gait (foot contract to foot-off) following the format described in section 7.2.2.1. Again, the data were extracted from 3 month post-op results although the trends identified from the data were present at 12 months in most cases.

The moment profiles for stair descent were more variable between subjects than the stair ascent and level walking data and the moment curves were occasionally not as smooth. The presentation of the moment curves were different in shape to those for the preceding ambulation modes. The double peak internal rotator moment curve was more defined and had sharper peaks. Apart from the early stance phase on the flexor curve, figure 7.32 shows that the moment values during early and late stance phase were substantially smaller than those between about 15-90% of stance. This finding could be attributed to the double support phase. Generally the trend graphs showed that adductor and internal rotator moment curves were similar for operated

Figure 7.32a: Transverse plane view of stick figures during stair descent

and control limbs and the flexor moment curves showed reduced extensor moments of the operated limb compared to the control limb for the stance phase.

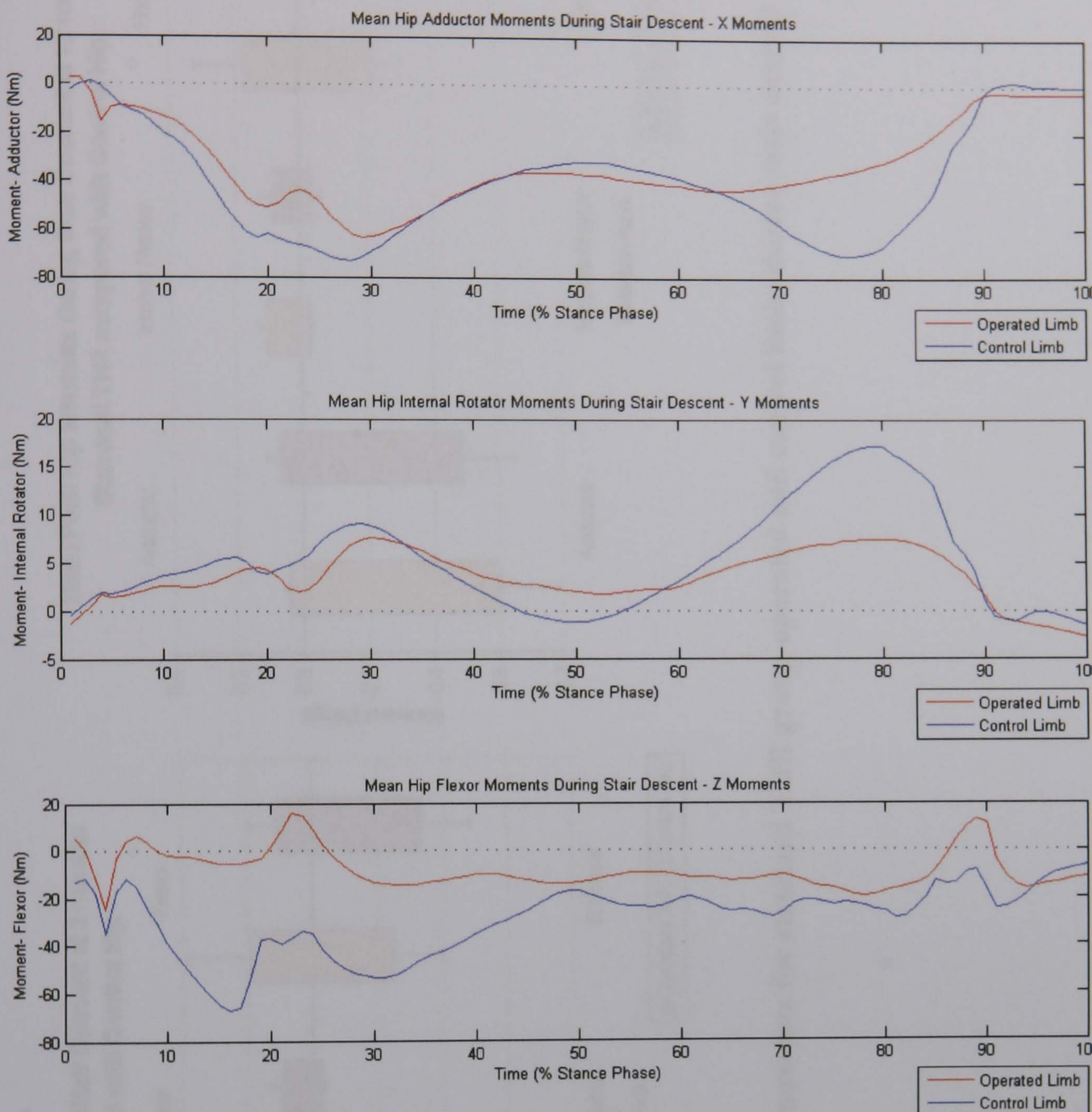
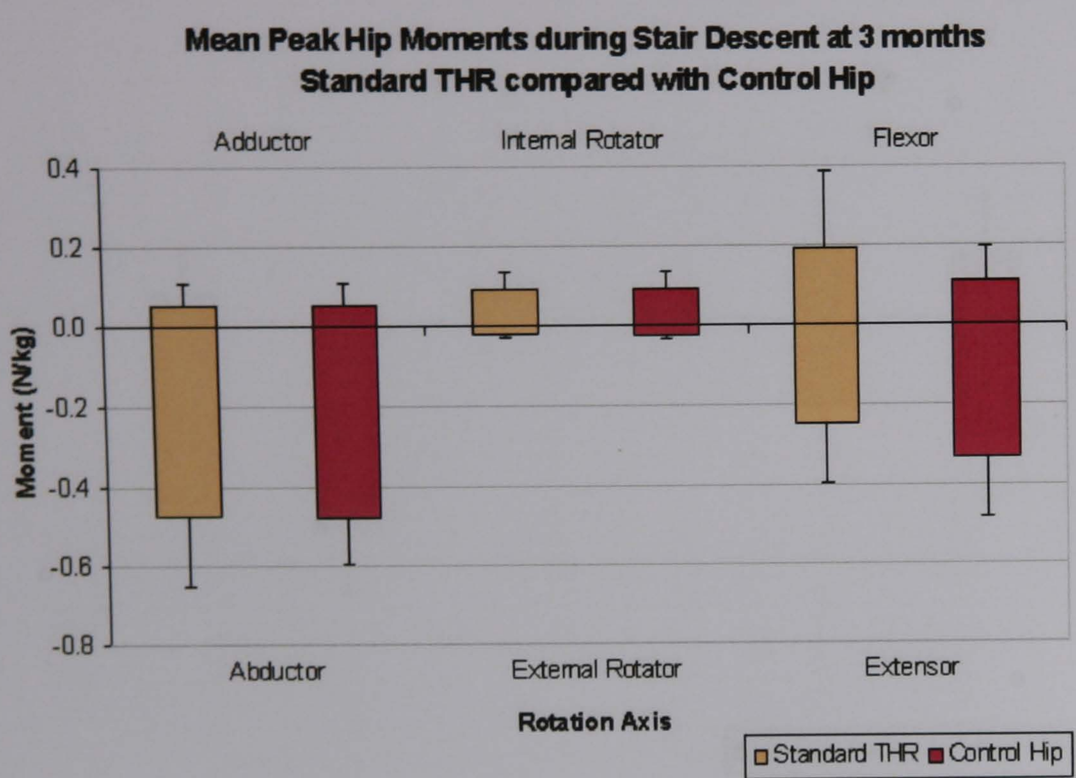


Figure 7.32: Mean 3-D moment profiles typical of operated and control limbs during stair descent

7.4.2.2 Mean peak hip moments during stair descent

The peak hip moments for each individual were normalised (to body mass and height), averaged and then summarised according to each arthroplasty group. The results for the operated and control limbs were then compared.



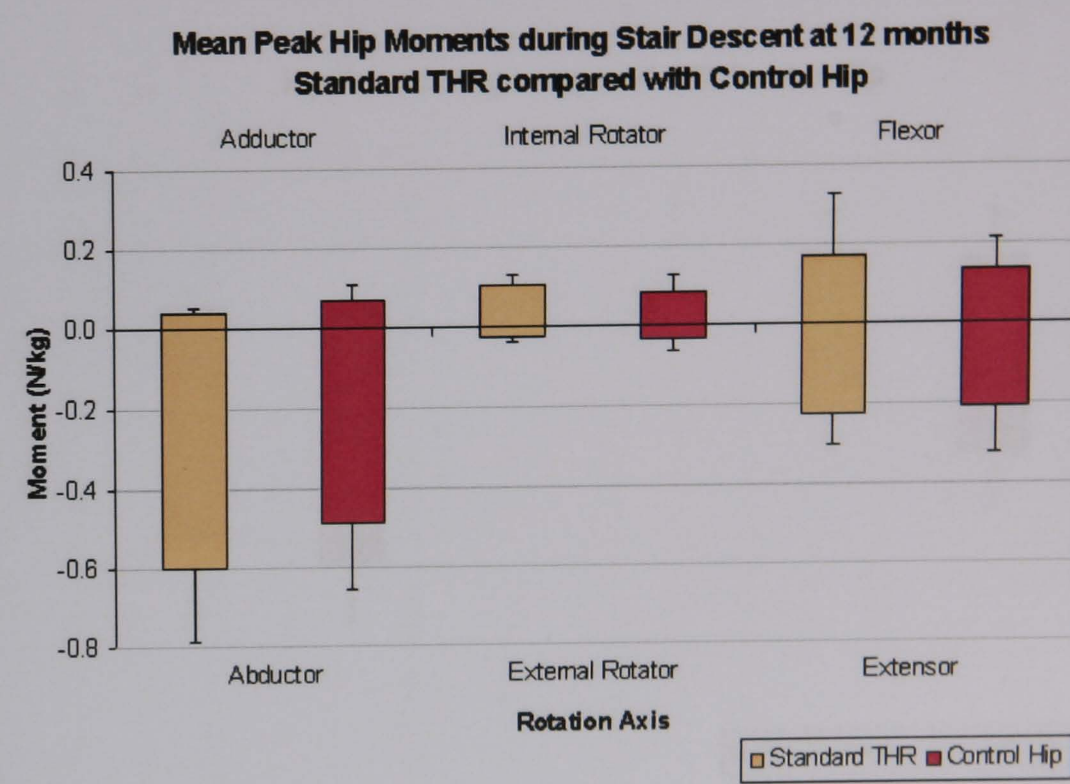
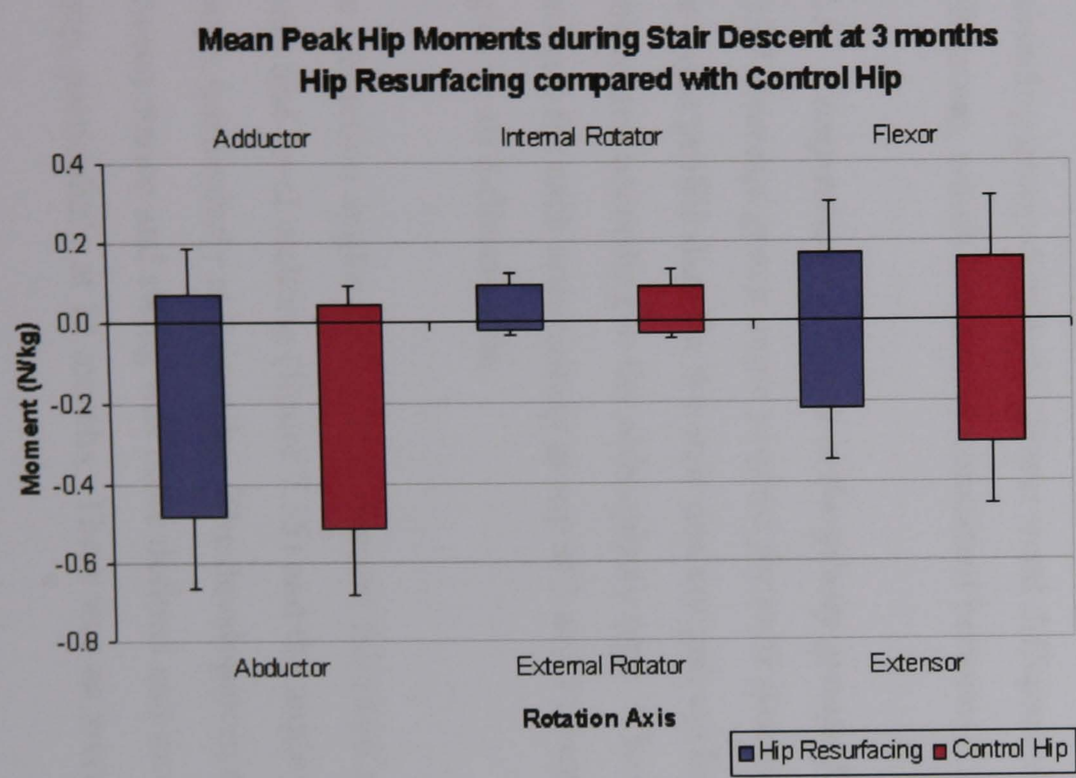


Figure 7.33: Comparison of 3-D hip moments for standard THR group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months



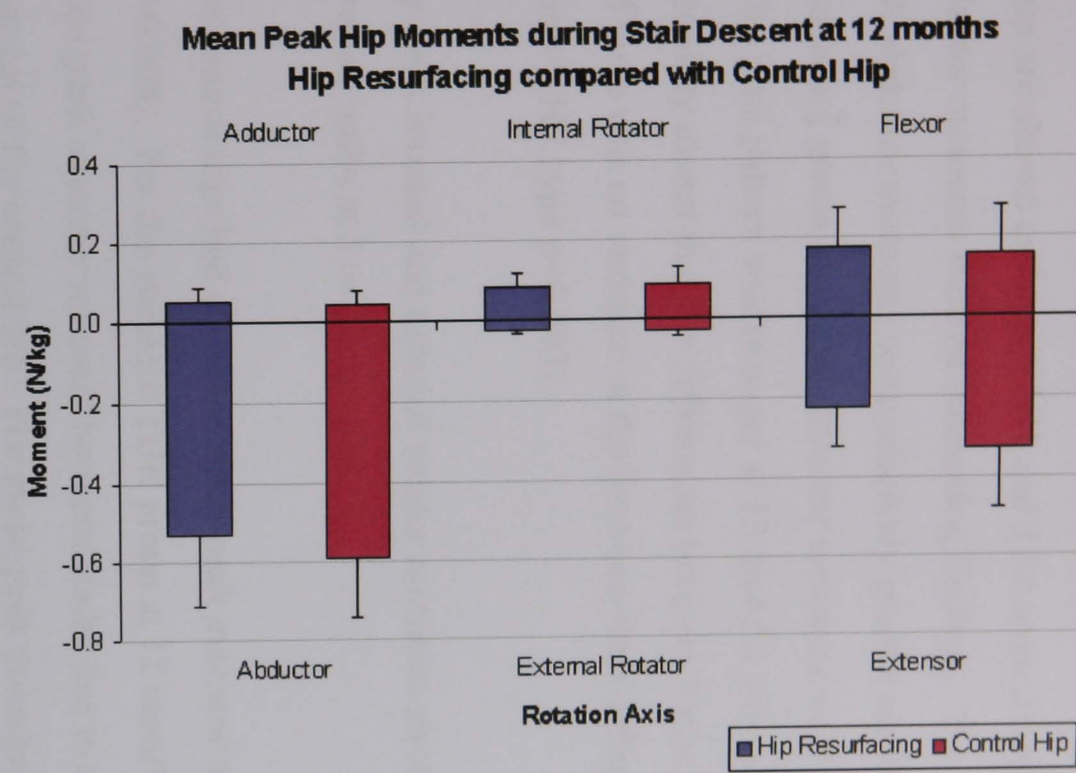


Figure 7.34: Comparison of 3-D hip moments for hip resurfacing group-operated and control limbs during stair descent at (a) 3 months and (b) 12 months

The results of the hip moments for the standard THR group and the resurfacing group are shown in figures 7.33 and 7.34 respectively. The peak abductor and adductor moments showed interesting findings. For the standard THR group, the peak abductor moments were relatively greater on the operated limb: for the resurfacing group, the peak abductor moments were relatively greater on the control limb. This pattern was stronger at 12 months although the high between-subject variability meant that the differences between limbs (p=0.502) were not significant and there was no variance in the between-limb difference due to THR type (Limb*THR type p=0.203).

The peak internal and external rotator moments showed no significant difference between limbs at 3 or 12 months.

The control hips had a greater mean peak extensor moment than the operated hip (p=0.003). For the standard THR group at 12 months, there was no difference in the mean peak extensor moment between-limbs due to the reduction of the extensor moment of the control hip. The mean peak moment values between limbs for the resurfacing group displayed clearer trend differences than that seen for the standard THR group, which were more consistent between 3 and 12 months.

7.4.3 Comparisons between arthroplasty groups for stair descent

7.4.3.1 Average group angle profiles for stair descent

The angle profile data for the stair descent gait cycle of the operated hip was summarised according to the arthroplasty type. The mean angle profiles were averaged for each arthroplasty group at 3 and 12 months and the data compared for angles in all 3-dimensions.

The adduction angles during stair descent followed a similar pattern to that of stair ascent and level walking (figure 7.35) and the angle profiles were similar for both groups, particularly at 3 months. The development of abduction during the transition between stance and swing was more defined and more rapid for the resurfacing group, particularly at 12 months. There was an increase in the adduction peak for the standard THR group with time (p=0.081), whereas the resurfacing group showed an increase in the abduction angle during the swing phase (p=0.022) by the late test session.

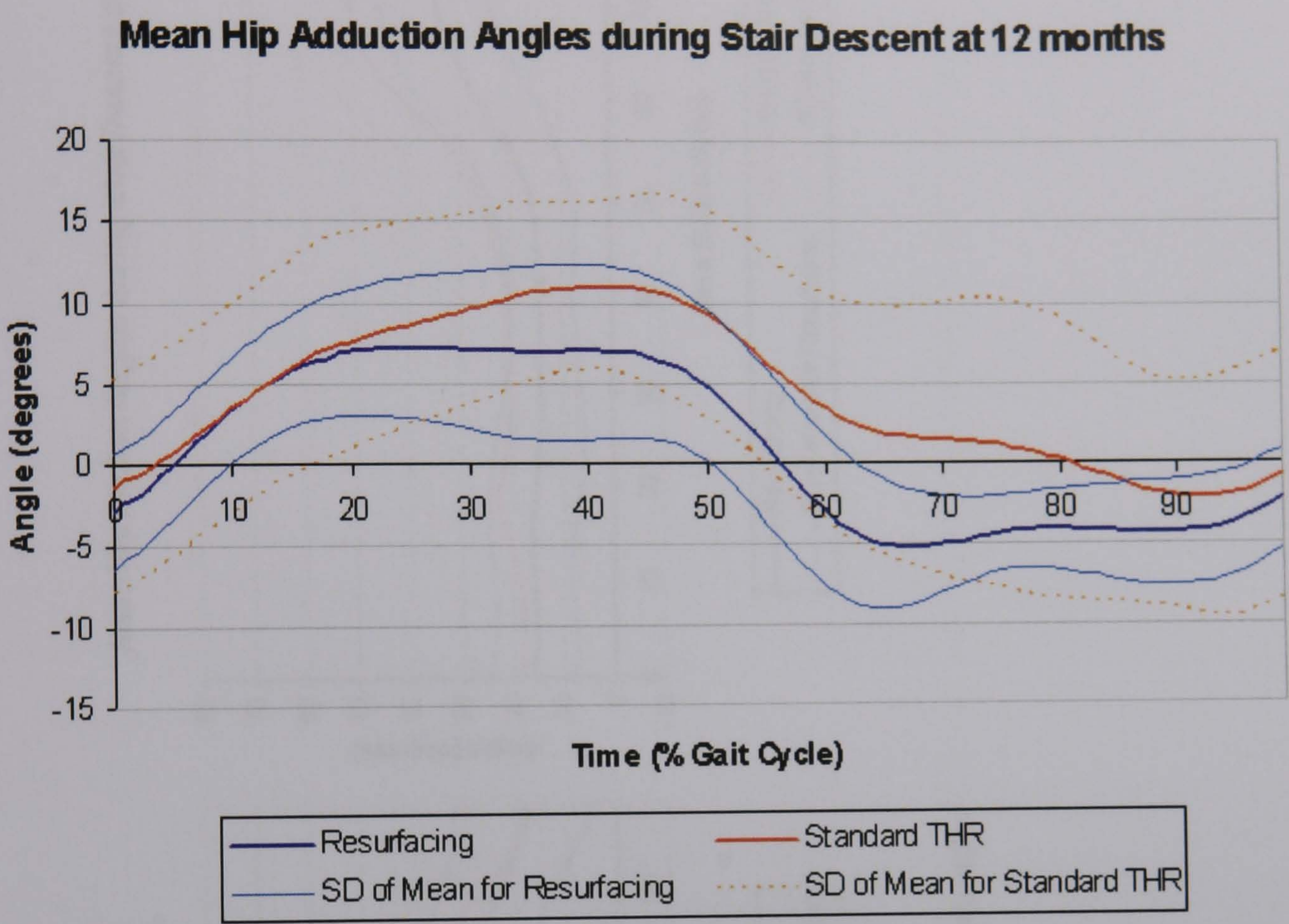


Figure 7.35: Mean hip adduction angles for stair descent at 12 months

The internal rotation hip profile curves during stair descent (appendix 9) followed a very similar pattern to that identified for stair ascent, where the hip was relatively externally rotated during the stance phase and more internally rotated during the swing phase. The relative profile for each arthroplasty group was also similar to that of stair ascent, with the resurfacing group having a more internally rotated position relative to the standard group all throughout the stair descent cycle.

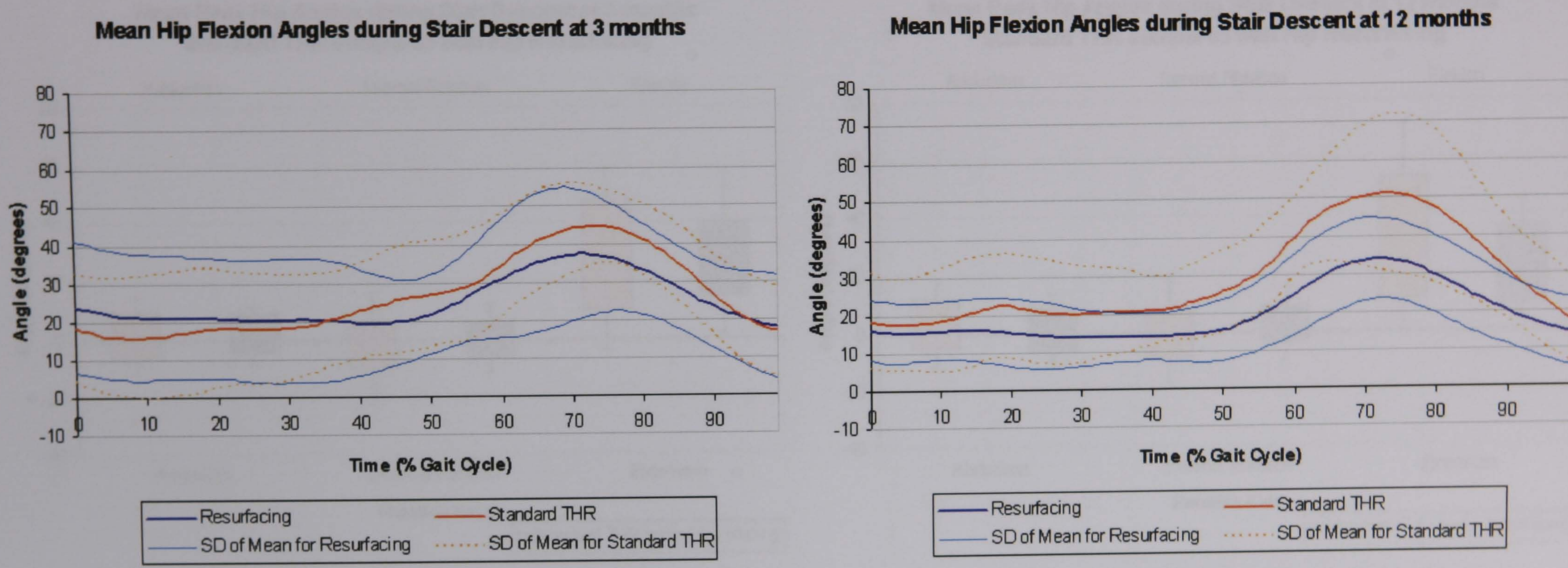
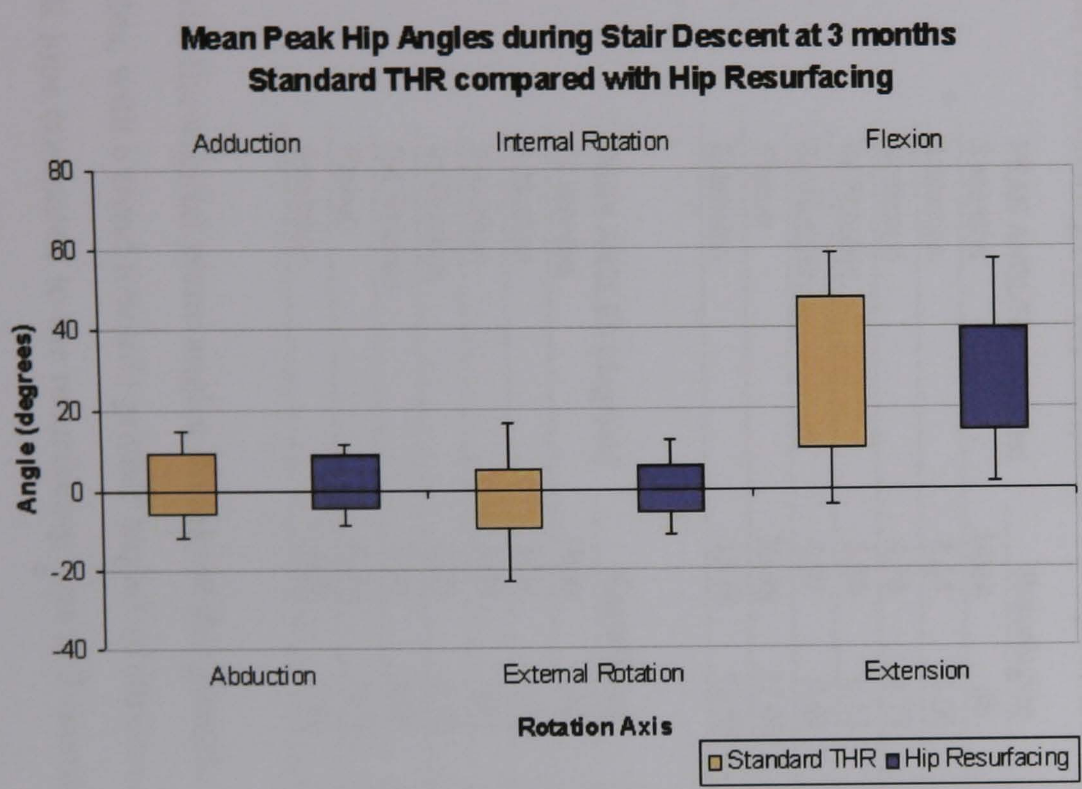


Figure 7.36: Mean hip flexion angles at (a) 3 and (b) 12 months



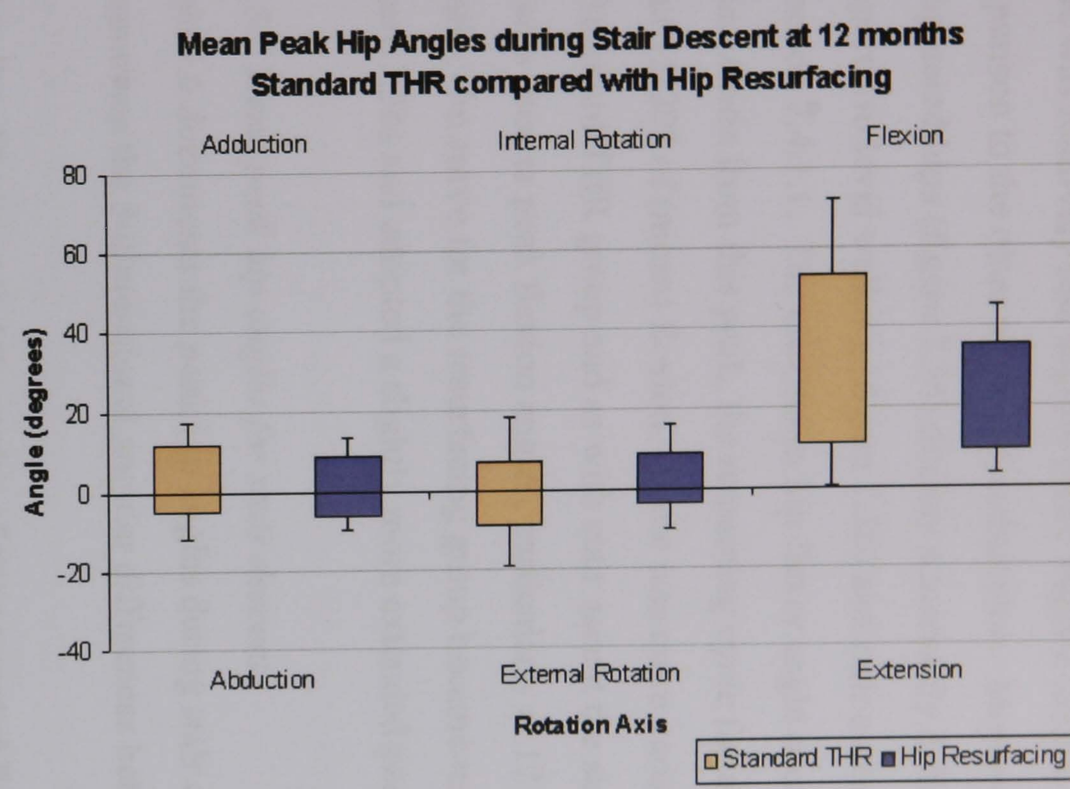


Figure 7.37: Comparison of 3-D hip angles between standard THR and hip resurfacing during stair descent at (a) 3 months and (b) 12 months

There was relatively less sagittal plane angular motion used during stair descent in comparison to the other modes of ambulation. Moreover, the hip flexion profiles of the operated hips (figure 7.36) display a markedly different pattern of sagittal motion compared to level walking (figure 7.10) and stair ascent (figure 7.24), as mentioned in section 7.4.1.1. The maximum hip flexor angle occurred at about 73% of the step cycle. Aside from this peak, the remaining cycle featured a fairly static angle of about 15-20° of (mean) flexion. There was more variability in the mean angles of the standard THR group and as with stair ascent the standard group featured a trend towards greater peak flexion angles, particularly at 12 months. Between 3 and 12 months, the curve for the resurfacing group became more negative on the graph (figure 7.36) and adopted a slightly more extended position throughout the cycle.

7.4.3.2 Mean peak hip angles for stair descent

Table 7.6 documents the peak hip angles during stair descent and figure 7.37 summarises the 3-dimensional angular differences between the arthroplasty groups.

Table 7.6: Mean peak hip angles of the operated limb during stair descent

PEAK ANGLES (degrees)	Resurfacing		Standard THR	
3 months	Mean	SD	Mean	SD
Adduction	8.56	3.06	9.16	5.71
Abduction	4.76	4.18	5.81	6.08
Int Rotation	5.80	6.13	4.89	11.49
Ext Rotation	5.97	5.49	9.75	13.28
Flexion	39.89	17.09	47.23	11.30
Extension	-13.78	12.50	-9.50	13.79

PEAK ANGLES (degrees)	Resurfacing		Standard THR	
12 months	Mean	SD	Mean	SD
Adduction	8.82	4.45	11.98	5.38
Abduction	6.31	3.40	4.89	6.72
Int Rotation	8.39	7.28	6.91	11.41
Ext Rotation	3.39	6.84	9.15	10.08
Flexion	35.44	10.09	53.23	19.20
Extension	-8.87	5.64	-11.00	10.65

Again, the sagittal plane angles displayed the greatest difference in mean peak angles, with a trend towards greater angles of flexion and extension for the standard THR hips compared to the resurfacing hips at 3 months and greater flexion at 12 months. Like the developments seen with stair ascent at 12 months, the peak extension angle of the resurfacing hips increased and became equivalent to the mean peak extension angles of the standard THR group by 12 months. Repeated measures analysis showed that the between group differences in mean peak angles were not significantly different overall (flexion p=0.16; extension p=0.902). However, statistical analysis of the 12 month data alone revealed significantly greater peak flexion angles of the standard THR group relative to the resurfacing group (p=0.02).

There were minimal between group differences in the coronal or transverse plane angles at 3 or 12 months. The mean peak external rotation angle was slightly greater for the standard hips at both assessment phases. There were no significant differences between the groups for coronal or transverse plane angles.

7.4.3.3 Average group moment profiles for stair descent

The moment profile data for the stair descent stance phase of the operated hip was summarised according to the arthroplasty type. The mean moment profiles were averaged for each arthroplasty group at 3 and 12 months and the data compared for angles in all 3-dimensions.

The graph of adductor moments (figure 7.38) shows that the sharp adduction peak present at foot contact during stair ascent was not produced during stair descent ambulation. In fact, stair descent produced an abductor moment almost exclusively, with minimal adductor moments. The abductor moment profile was very similar for both arthroplasty groups at 3 months. The between-subject variability of the mean measures was greater at 3 months, and was greatest for the resurfacing group. As with stair ascent, both groups also featured a significant increase in the magnitude of the abductor moments from 3 to 12 months (p=0.02), although the increase was slightly greater for the standard THR group (p=0.501). This is in contrast with the findings of stair ascent, which showed a development of greater abductor moments for the resurfacing group by 12 months. Hence, stair descent and ascent highlighted the opposite trend differences for abductor moments between-groups.

The internal rotator moment profiles for both arthroplasty groups are shown in figure 7.40). The pattern of the curves for standard and resurfacing arthroplasty groups followed similar trends, with the standard THR group tending to have greater internal rotator moments, particularly at 12 months. Again, there was substantial variability in the between subject measures for each group, particularly for the standard THR group at 3 months.

The flexor moment profiles for stair descent (figure 7.40) were markedly different from those for stair ascent and level walking, which were relatively similar in shape. The magnitudes of the flexor and extensor moments were also lower for stair ascent. There were however fundamental similarities between the flexor curves for stair descent and the other modes of ambulation.

For example, there was a sequence of fluctuating peaks following initial contact. For the preceding ambulation modes, these peaks were flexor moments whereas those for stair descent were particularly variable between subjects and on average, centred around the zero axis following an extensor-flexor-extensor-flexor sequence before continuing on the gradual development of the peak extensor curve for the total stance phase, representative of all three ambulation modes. The final extensor peak was followed by the typical extensor trough preceding foot-off for all ambulation modes.

There were few notable differences between the arthroplasty group flexor moment profiles (figure 7.40). The standard THR group had a slightly greater peak extensor moment which occurred slightly earlier (~70% stance) than the equivalent of the resurfacing group at 3 months. The 12 month profiles showed greater similarities between groups, particularly following mid-stance, where the extensor curves were more reproducible between-groups.

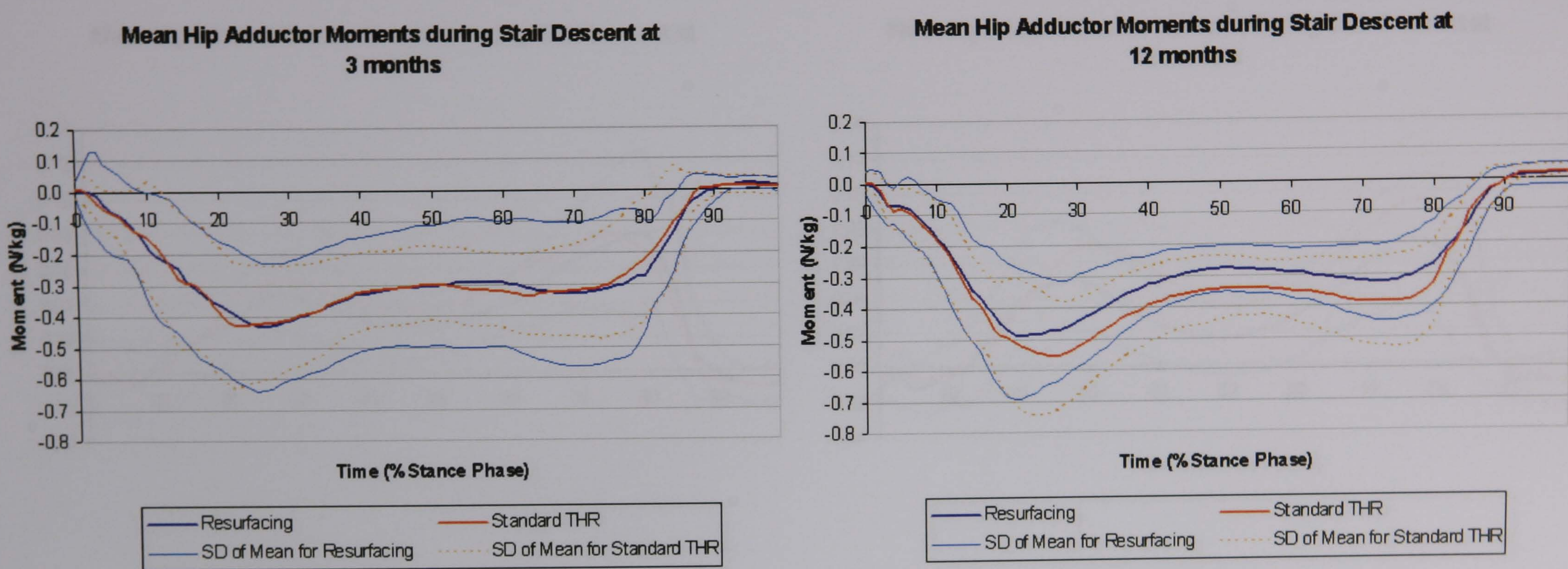
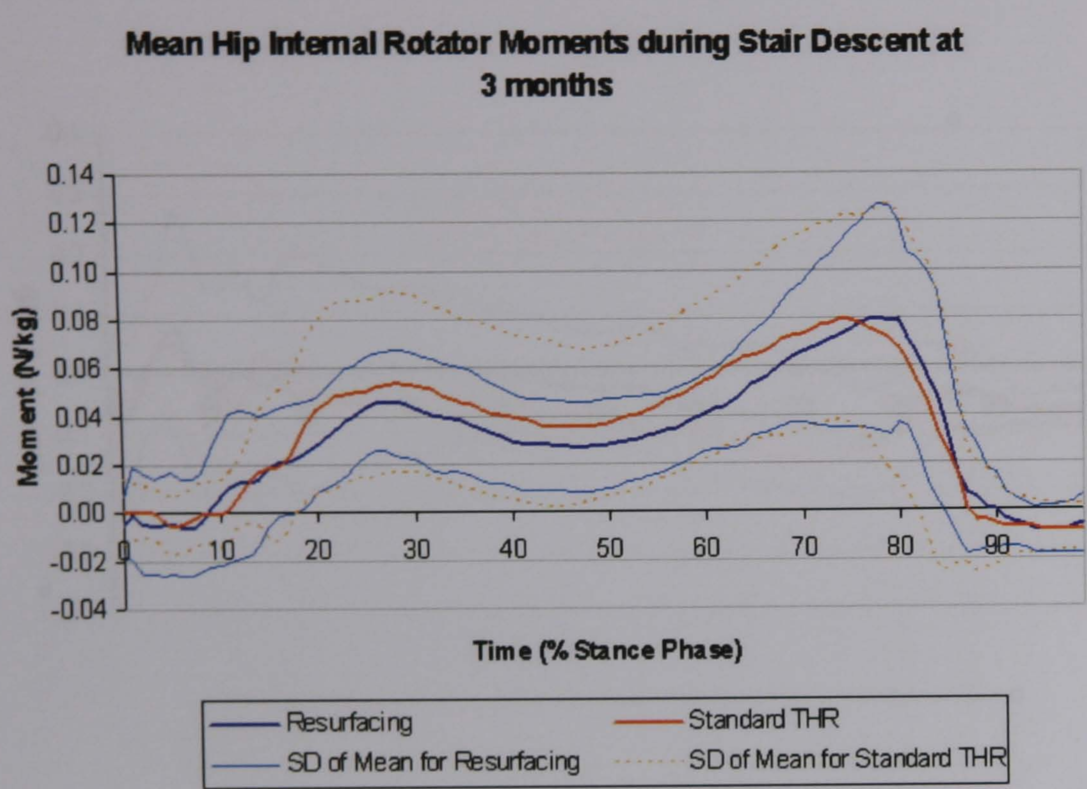


Figure 7.38: Mean hip adductor moments during stair descent at (a) 3 and (b) 12 months



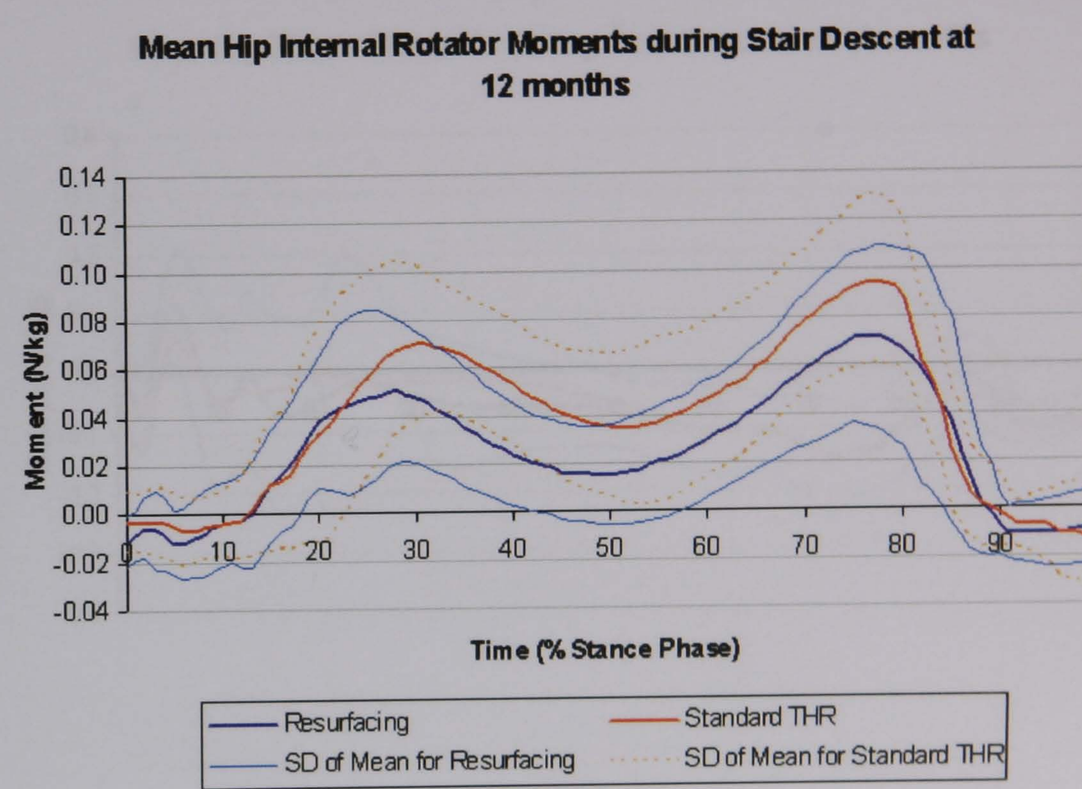
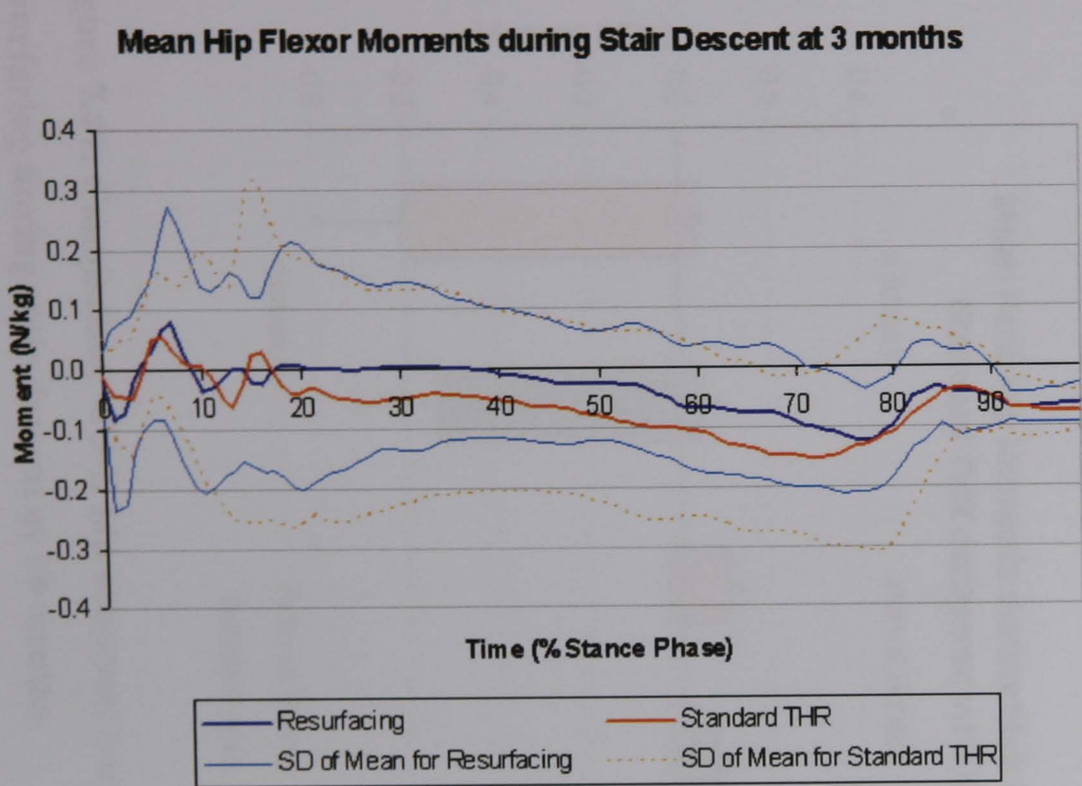


Figure 7.39: Mean hip internal rotator moments during stair descent at (a) 3 and (b) 12 months



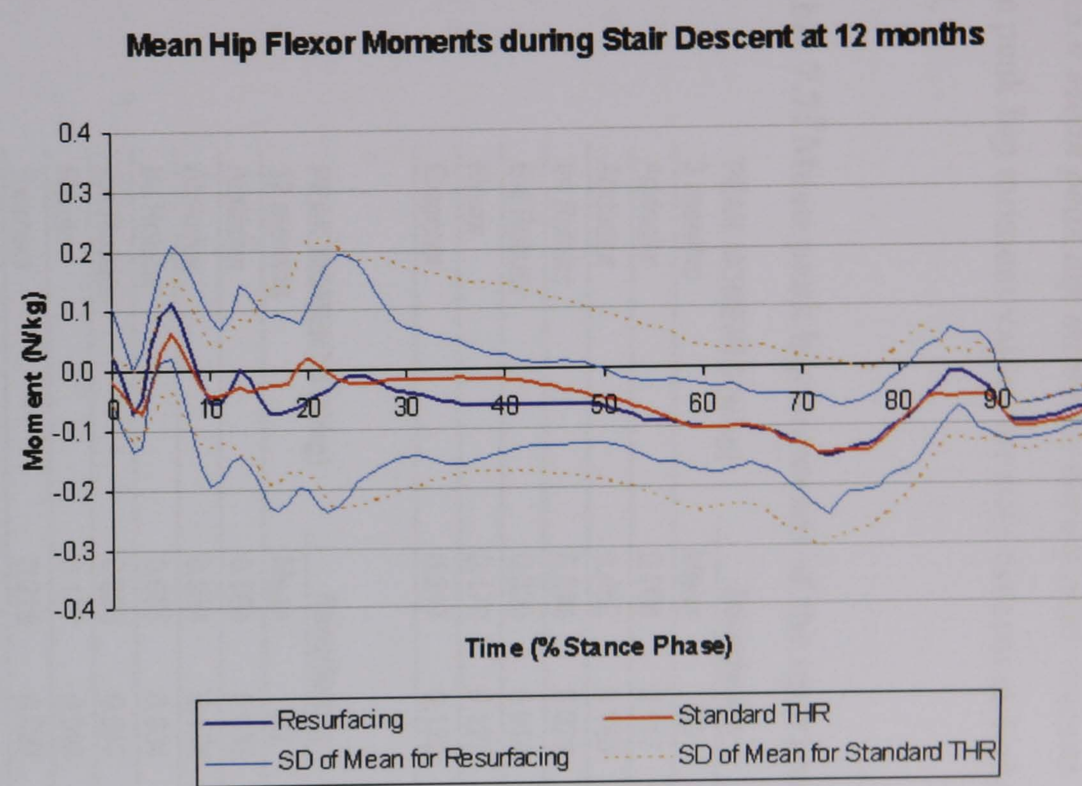


Figure 7.40: Mean hip flexor moments during stair descent at (a) 3 and (b) 12 months

7.4.3.4 Mean peak hip moments during stair descent

The peak hip moment values for stair descent at 3 and 12 months are shown in table 7.7.

Table 7.7: Mean peak hip moments of the operated limb during stair descent

PEAK MOMENTS (N/kg) 3 months	Resurfacing Mean	Resurfacing SD	Standard THR Mean	Standard THR SD
Adductor	0.068	0.117	0.057	0.050
Abductor	0.482	0.183	0.474	0.180
Int Rotator	0.096	0.024	0.095	0.043
Ext Rotator	0.020	0.014	0.020	0.007
Flexor	0.170	0.127	0.193	0.191
Extensor	0.217	0.132	0.246	0.151

PEAK MOMENTS (N/kg) 12 months	Resurfacing Mean	Resurfacing SD	Standard THR Mean	Standard THR SD
Adductor	0.054	0.033	0.039	0.016
Abductor	0.534	0.176	0.601	0.183
Int Rotator	0.085	0.034	0.101	0.032
Ext Rotator	0.023	0.012	0.022	0.016
Flexor	0.177	0.094	0.174	0.152
Extensor	0.229	0.097	0.226	0.081

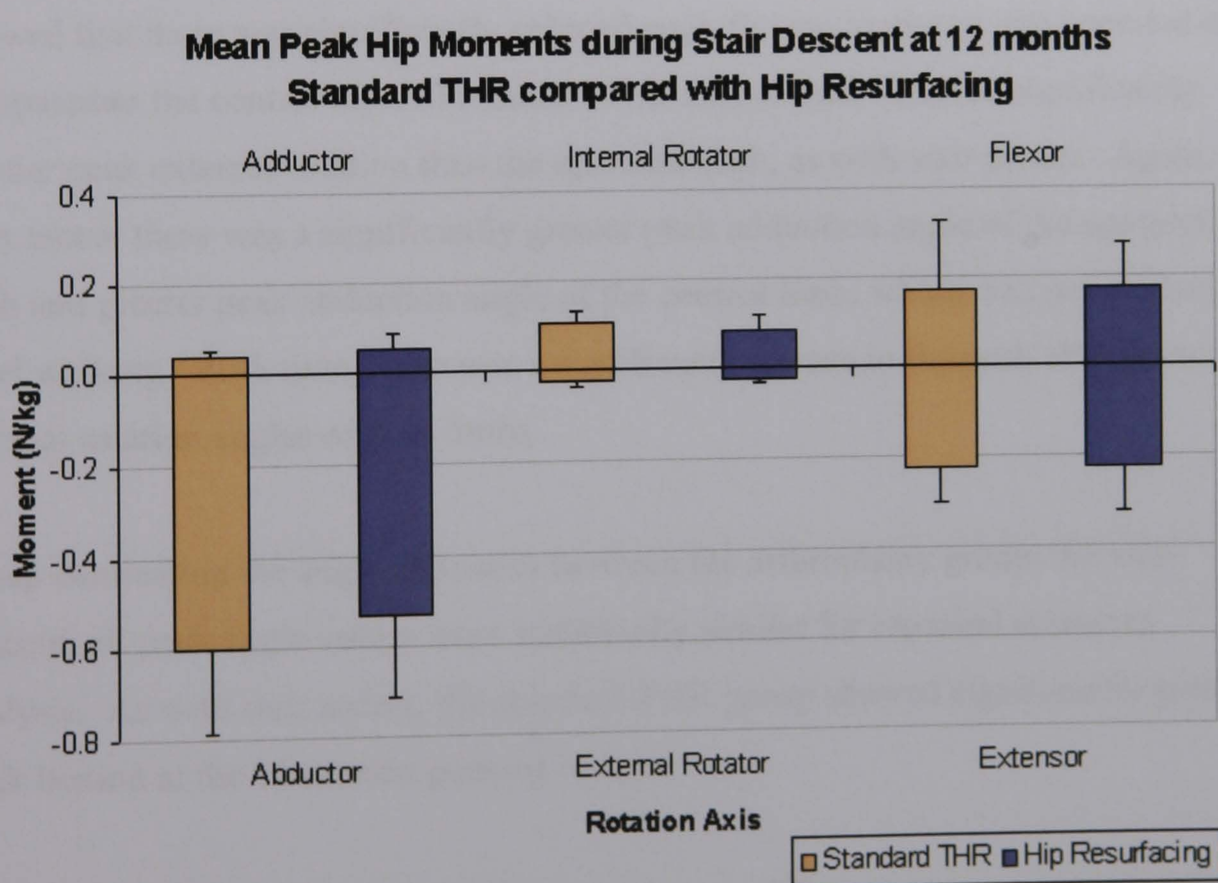


Figure 7.41: Comparison of 3-D hip moments between standard THR and hip resurfacing during stair descent at 12 months

When comparing the moment differences between the arthroplasty groups (figure 7.41) for stair descent, both 3 and 12 month results show minimal between-group difference. There was a slightly greater mean peak flexor and extensor moment for the standard THR group, with a greater range of between-subject measures. More between-group trend differences were present at 12 months with the standard THR group having greater abductor and internal rotator moments. All differences due to arthroplasty type were minimal and not statistically significant for any of the peak moment outcomes.

7.4.4 Summary of within- and between-group differences for stair descent

Stair descent analysis showed less within- and between-subject differences than the other ambulation modes. As with stair ascent, the differences found were largely related to comparisons between the operated and control limb rather than between the arthroplasty groups. Important differentiations were made between the angle and moment profiles for stair descent compared to level walking and stair ascent.

Comparison of joint angles for the operated and control limbs during stair descent showed that there was significantly reduced peak flexion angles on the operated hip compared to the control hip at 3 months. The control limb featured significantly greater peak external rotation than the operated limb, as with stair ascent. Again, like stair ascent there was a significantly greater peak adduction angle of the operated limb and greater peak abduction angle of the control limb, which was not evident for level walking. With time, there was a significant increase in the peak abduction and internal rotation angles of both limbs.

When contrasting the angle outcomes between the arthroplasty groups for stair descent, all peak angle values were statistically similar for repeated measures analysis. As with stair ascent, the standard THR group showed significantly greater peak flexion at the 12 months post-op interval.

Unlike stair ascent and level walking, the moment differences between operated and control limbs showed no significant difference, aside from a greater extensor moment peak for the operated hips of the resurfacing group. As with level walking and stair ascent, the peak abductor moment of the operated limb increased with time.

There were no significant differences in the peak moment outcomes between-groups during stair descent. When comparing the modes of ambulation, stair descent showed the least differences due to THR type.

7.5 Temporospatial Characteristics of the Motion Tasks

7.5.1 Velocity characteristics of level walking

7.5.1.1 Gait velocity of the sample group

The gait velocity was recorded for all subjects at early and late assessment phases, using the methods described in the data analysis chapter (chapter 5). The outcomes were then summarised according to each arthroplasty group. Table 7.8 shows the gait velocity for the level walking motion task.

Table 7.8: Gait velocity during level walking at 3 and 12 months

3 months

VELOCITY (m/s)	STHR	RA
Max	1.38	1.55
Min	0.71	1.00
Mean	1.07	1.27
SD	0.17	0.18

12 months

VELOCITY (m/s)	STHR	RA
Max	1.31	1.60
Min	1.04	1.26
Mean	1.16	1.44
SD	0.08	0.12

STHR: Standard THR
RA: Resurfacing Arthroplasty

Between 3 and 12 months, there was a significant increase in the gait velocity for both resurfacing and standard THR groups ($p=0$). In addition, the between-subject variability of walking speed reduced with time, showing more consistent and faster walking speeds by 12 months.

When comparing between the arthroplasty groups, the results showed that the resurfacing groups walked at a greater mean velocity compared to the standard THR group. The faster walking speed of the resurfacing group was consistent at both test intervals (figure 7.42) and analysis of variance implied that gait velocity varied significantly due to arthroplasty type (3 months p=0.004; 12 months p=0).

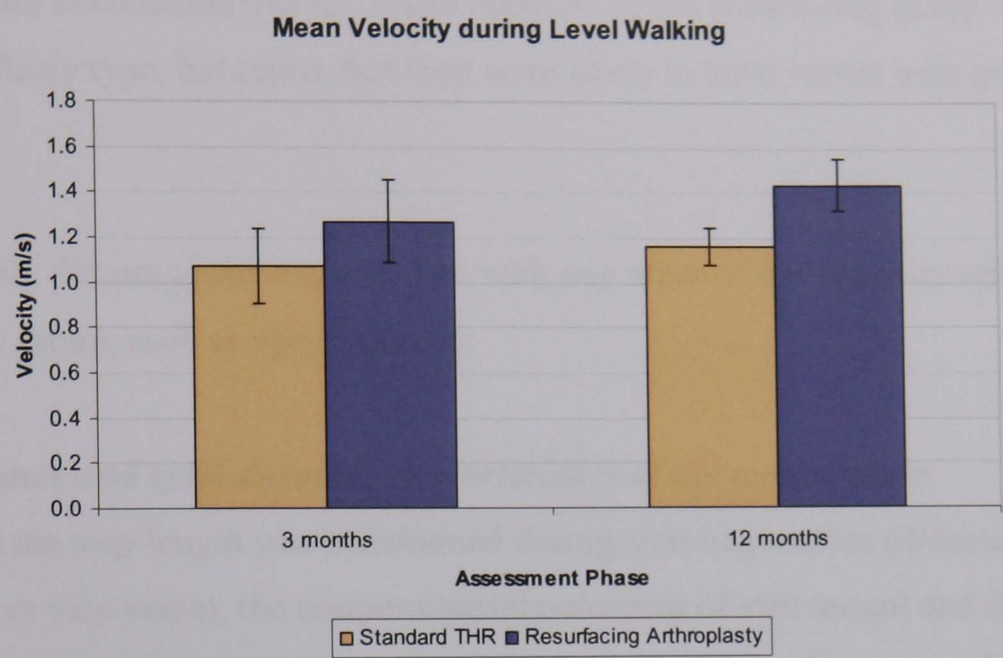


Figure 7.42: Gait velocity during level walking

7.5.1.2 Relationship between gait velocity and moment variables

Gait velocity was an unconstrained variable during data collection and the resultant gait velocity varied significantly between groups. The magnitude of external hip moment values are known to positively correlate with walking velocity (section 2.6.1.2). To rule out a type I error in interpreting the data, the relationship between hip moments and gait velocity during level walking was tested with a multiple variable linear regression model.

At 3 months, the flexor moments correlated significantly with walking speed (operated limb, p=0.003; control limb, p<0.001), and the adductor and extensor moments correlated positively and closely with walking velocity respectively (operated limb: adductor p=0.003, extensor p=0.083; control limb: extensor p=0.086). At 12 months, the linear relationship between adductor and extensor moments and gait speed continued (operated limb: extensor p=0.029; control limb: adductor p=0.007, extensor 0.059). Hence, only the extensor moment at 3 months post-op was not statistically significantly correlated with gait velocity, and may then be positively associated with the hip resurfacing arthroplasty, but this did not continue at 12 months. On the whole, given that the peak adductor, flexor and extensor moment values were significantly greater for the resurfacing group (section 7.2.3.4) and these outcomes had a positive correlation with walking speed, it would be inaccurate to conclude that the better outcome of the resurfacing group was due to the arthroplasty type, but rather that they were likely to have varied with gait velocity.

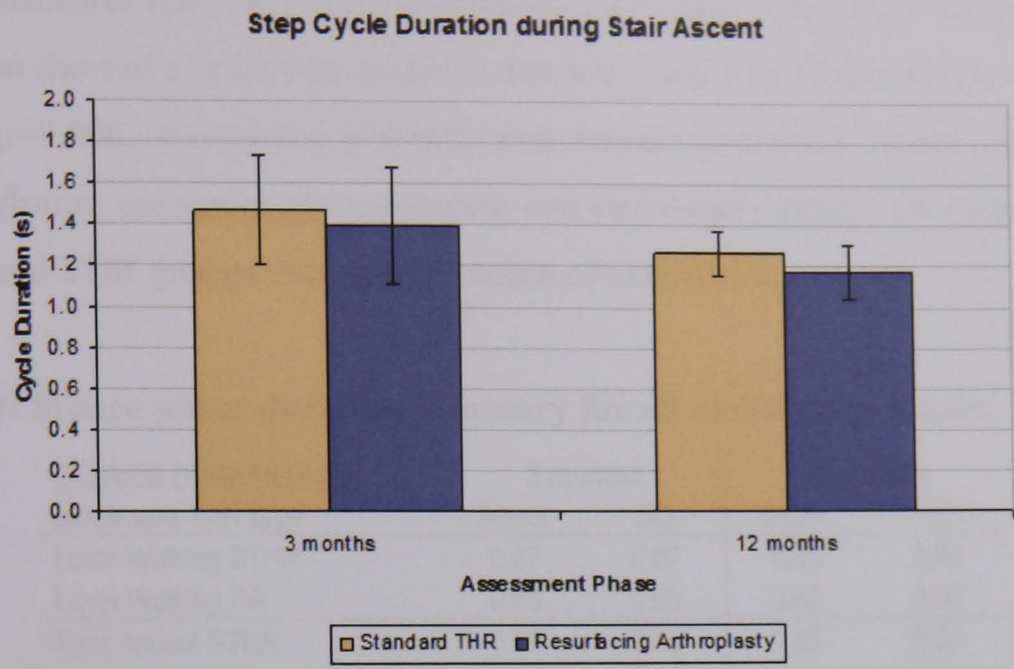
Gait velocity did not positively correlate with any other of the variables relating to the sample group, such as age or gender.

7.5.2 Stance and cycle duration characteristics of the motion tasks

Given that the step length was constrained during stair negotiation (distance from 1st to 3rd step or vice versa), the temporospatial outcomes of stair ascent and descent were quantified by measuring the stance and cycle duration. The stance duration was also assessed for level walking to provide a relative comparison of level and stair gait temporal outcomes.

The stair ascent and descent cycle durations were similar between modes, with an average cycle duration of 1.43s and 1.41s for ascent and descent respectively at 3 months, and 1.22s and 1.20s at 12 months, for the entire sample group. Hence, on average, the ascent cycle was consistently 2 seconds longer. Figure 7.43 contrasts the cycle duration between arthroplasty groups for stair ascent and descent. It can be seen that the standard THR group had consistently greater mean cycle durations than the resurfacing group for both stair negotiation modes at both test intervals. These between group differences were not significant but formed a trend which was consistent with the findings from the level walking gait velocity analysis, where the resurfacing group performed the motion tasks quicker than the standard THR group.

Progression of stair negotiation cycle duration with time was evident. The stair ascent cycle became significantly shorter by 12 months ($p=0.002$) and showed less between-subject variability; the descent cycle showed a trend reduction in cycle duration with time ($p=0.085$) and displayed more variability than stair ascent.



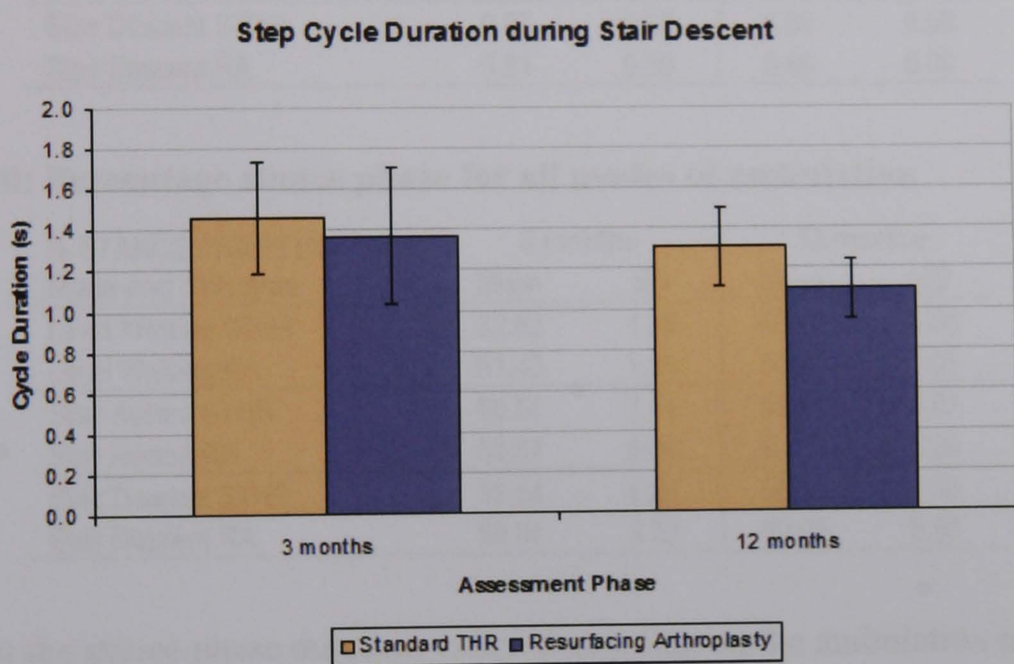


Figure 7.43: Cycle duration for stair negotiation

Statistically, there was symmetry of the cycle duration of operated and control limbs for level walking, stair ascent and descent, with no between-limb effect found with repeated measures ANOVA. Assessment of stance phase duration, on the other hand, showed a significant difference in the stance time of the operated and control limbs during level walking gait at 3 months ($p=0.04$), with the operated limb having a relatively shorter stance phase.

Stance phase duration as an outcome measure showed greater differences with repeated measures than the cycle duration outcome (table 7.9). All 3 modes of ambulation showed a reduction in stance duration from 3 to 12 months (level walking, p=0.006; stair ascent, p=0.005; stair descent, p=0.041). When comparing between groups, the stance phase duration was statistically similar for resurfacing and standard THR groups during all 3 modes of ambulation.

Table 7.9: Stance phase duration summary for all ambulation modes

STANCE DURATION (s)	3 months		12 months	
Mode and THR type	Mean	SD	Mean	SD
Level Walking STHR	0.67	0.07	0.65	0.05
Level Walking RA	0.65	0.08	0.60	0.07
Stair Ascent STHR	0.97	0.15	0.83	0.07
Stair Ascent RA	0.88	0.16	0.73	0.08
Stair Descent STHR	0.87	0.20	0.69	0.08
Stair Descent RA	0.81	0.20	0.66	0.09

Table 7.10: Percentage stance phase for all modes of ambulation

% STANCE PHASE (%)	3 months		12 months	
Mode and THR type	Mean	SD	Mean	SD
Level Walking STHR	62.83	1.98	63.47	1.66
Level Walking RA	61.43	1.29	60.88	1.01
Stair Ascent STHR	66.51	7.41	65.84	4.01
Stair Ascent RA	63.57	5.46	62.18	5.35
Stair Descent STHR	59.54	6.88	55.00	4.26
Stair Descent RA	59.04	3.82	60.85	8.58

Given that the stance phase duration varied depending on the ambulation mode, the outcome of stance as a percentage of the total cycle duration was assessed (table 7.10). This showed that the stance phase of stair ascent was longer than that of level walking, whereas the stance phase of stair descent was shorter relative to level walking. The data showed that the standard THR group had a greater percentage stance phase than the resurfacing group during level walking and stair ascent. This is consistent with the reduced gait velocity and greater stance and cycle time for the standard THR group, equating to overall poorer temporospatial indicators of function. Stair descent, however, showed a different pattern (figure 7.44), with no differences at 3 months and a greater percentage stance phase for the resurfacing group during descent at 12 months.

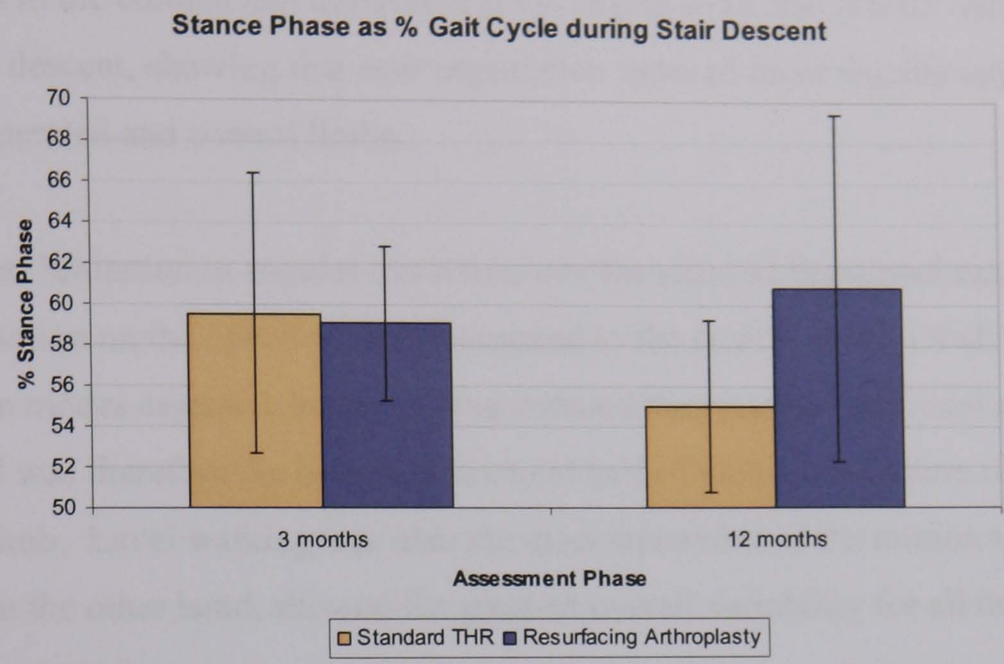


Figure 7.44: Comparison of stance phase duration between groups during stair descent

7.6 Comparative Summary of Functional Tasks

The results gained were contrasted according to the three motion tasks. The peak moment outcome showed more between group differences during the level walking task and very little differences with stair negotiation. Interestingly, the peak abductor moment was greatest for the resurfacing arthroplasty group during ascent and greatest for the standard THR group during descent. Given that there were no significant temporospatial differences between the groups for stair negotiation, this group difference was thought to be induced by the different biomechanical demand of ascent and descent tasks. The greatest between-group contrast for both stair ascent and descent was the greater peak extensor moment of the standard THR group at 3 months, which resolved by 12 months.

In contrast to the moment outcomes, the peak angle outcomes showed least between-group differences for level walking and relatively more differences during stair negotiation. This was particularly notable for the peak flexion angles which were greater for the standard THR group than the resurfacing group. Between-limb differences in the coronal and transverse plane angles were also greater during stair ascent and descent, showing that stair negotiation induced more angular asymmetries between operated and control limbs.

The greatest limitation in angular excursion was the reduced functional extension range of motion on the operated limb compared to the control limb. Of all the ambulation modes assessed, level walking induced the greatest functional extension angles and was therefore the best measurement tool of motion restriction of the operated limb. Level walking was also the most repeatable of the motion tasks. Stair descent, on the other hand, showed the greatest overall variability for all the measures assessed.

Further to the present results from the motion analysis data, the results of the motion detection sense test procedure are recorded in chapter 8. The implications and reasoning behind the results of the motion analysis outcomes will be discussed in chapter 9.

8 Proprioception Results

The preceding results chapters have presented the descriptive characteristics of the sample group, the subjective data and the motion analysis outcomes. In this chapter the results of the proprioception test protocol will be presented, examining the outcome of threshold motion detection sense. MDS results will be reported for the two test variables of abduction and flexion motion conditions of the limb. In addition to this, the results will be further stratified according to subject variables to test the influence of the test limb (operated and control limbs), the surgical category (standard THR and resurfacing) and the post-operative assessment phase (3 and 12 months post-op) on the outcome of hip joint MDS. The results for 3 and 12 months post-operative phases will initially be discussed independently, following which these results will be compared.

8.1 Subject Characteristics

The sample group corresponding to the proprioception test results is identical to that of the preceding results chapters (chapters 6 and 7). Hence the MDS outcome will be presented for 31 subjects (n=17 CoC standard THR; n=14 MoM resurfacing) at the early (3 month) post-operative stage and 20 subjects (n=10 CoC standard THR; n=10 MoM resurfacing) at the late (12 month) post-operative stage. However, there were several omissions to sub-sets of the data which will be highlighted.

Chapter 3 highlighted a small number of potential limitations to the MDS protocol, which were realised during the MDS trials and consequently resulted in omission of data. Firstly, not all subjects were able to complete all of the test conditions within the protocol. For example, some were not comfortable lying on their operated side. This was primarily evident at the 3 month post-operative stage, where several subjects (n=4; 12.9% of total subject group) did not complete the flexion trial of the non-operated control limb, due to discomfort weight-bearing through the operated hip. All subjects in this scenario reported that the main limiting factor was pain from the surgical wound. Of the 4 subjects who did not complete the control limb flexion trials, 3 had a standard THR and one had a hip resurfacing procedure. Further to this, one of those subjects with a standard THR remained unable to lie on their operated hip at the 12 month test session (5% of the total subject group). One other subject was unable to complete all of the four test conditions which lead to no data acquisition for the non-operated limb (abduction and flexion trials of the control limb). The reason for this data omission was twofold, including the onset of mild low back ache due to the positioning during the abduction trial (supine body position) and personal choice to terminate the test session due to fatigue.

Another reason for incomplete data acquisition was due to some trials being missed by subjects due to drowsiness, or the omission of trial data due to the presence of outliers exceeding the mean data values for certain test conditions, which were thought to be due to a momentary reduction in the consciousness of the subjects during the test protocol which they found to be 'very relaxing'. This caused single repetitions of trials from the series of ten to be omitted, and therefore the mean threshold MDS angle was calculated from less than 10 trials for several subjects. The number of trials used to calculate the mean in these circumstances was never less than 8, and these conditions only applied to 3 subjects.

8.2 Early MDS Results - 3 month Data

The early results of MDS data were averaged for each individual. Individual mean results were categorised according to their surgical group and results of each arthroplasty group were averaged producing mean results for the operated and control limbs during both movement conditions (abduction and flexion).

8.2.1 Comparison of MDS between the operated limb and the control limb

The mean threshold detection angles for individuals with standard THR is displayed in figure 8.1. The bars on the graph represent the mean threshold detection angle for subjects within that arthroplasty group and the error bars represent the standard deviation (SD) of the mean threshold angles. The SD therefore corresponds to the between-subject variability of threshold angles.

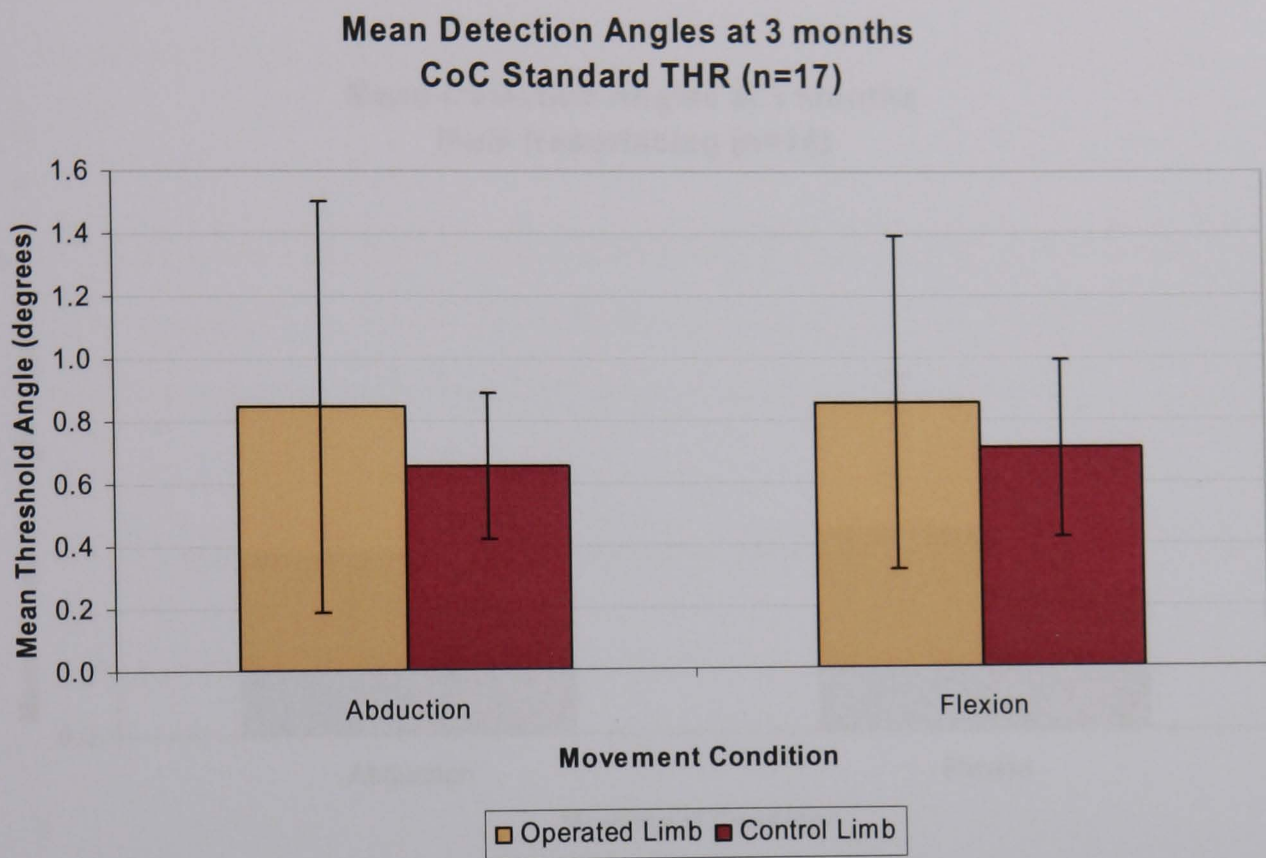


Figure 8.1: Comparison of mean threshold detection angles for operated and control limbs of subjects with standard THR

Figure 8.1 illustrates that the operated limb of the standard THR group had greater mean threshold detection angles than the non-operated control limb. This implies a trend difference in MDS due to the standard THR surgical procedure with greater angular motion required on the operated limb before the movement was detected. In addition, there was also less between-subject variability of MDS measured for the control limb, as shown by the smaller SD. These trends were consistent for both abduction and flexion movement conditions but statistical tests showed that the differences observed between the operated and control limb were not significant for abduction (Mann Whitney: p=0.85) or flexion (2-sample t-test: p=0.34) movement conditions.

When comparing the effect of the movement conditions on MDS of the standard THR group, it was found that there was no difference in the threshold angles resulting from the abduction compared to flexion motion stimuli for either limb (operated limb, Mann Whitney: p=0.89; control limb, paired t-test: p=0.27).

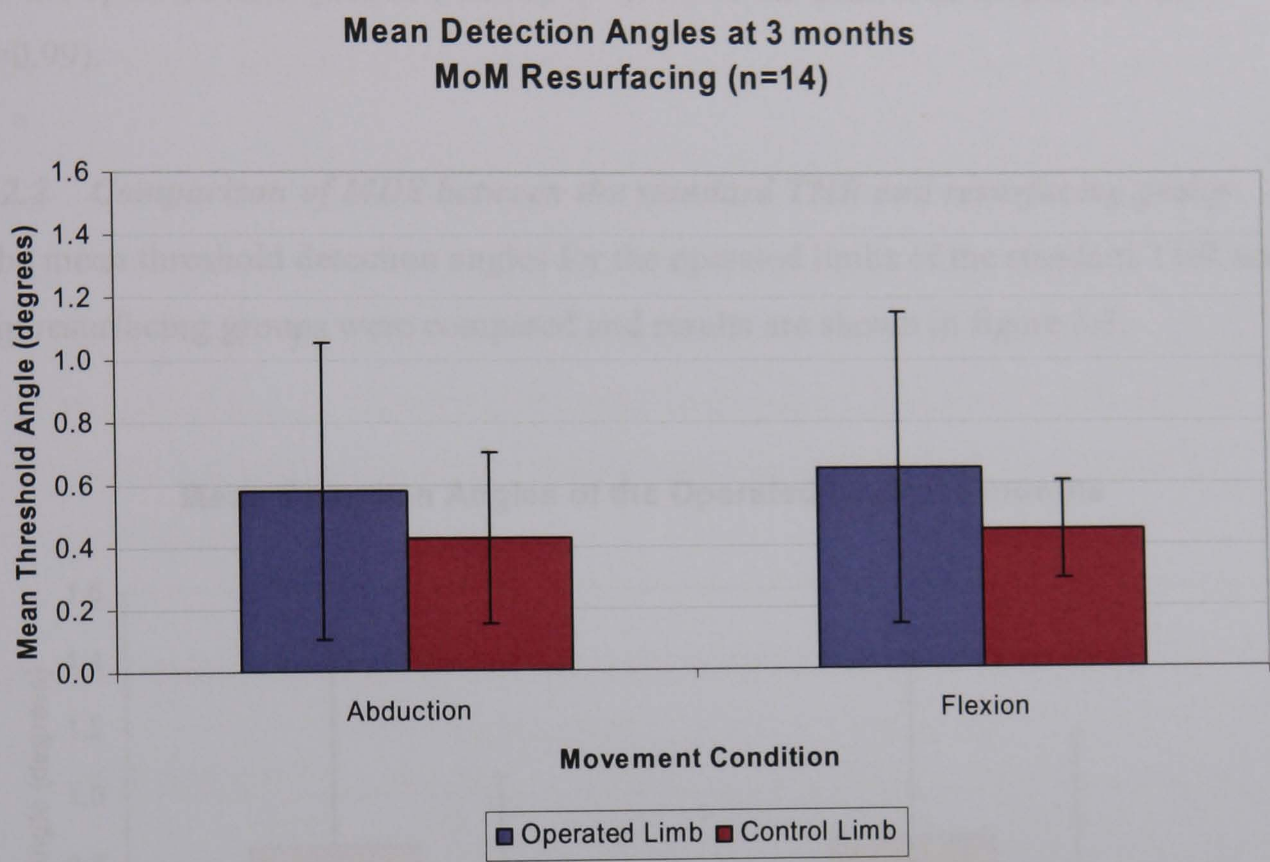


Figure 8.2: Comparison of mean threshold detection angles for operated and control limbs of subjects with MoM resurfacing

The mean threshold detection angles for individuals in the MoM resurfacing group is displayed in figure 8.2. The figure follows the same format as, and is scaled according to figure 8.1. The operated limb of the hip resurfacing group followed a similar trend to that of the standard THR group with greater mean threshold detection angles of the operated limb compared to the non-operated control limb. This implies a trend difference in MDS due to the hip resurfacing procedure with greater angular motion required on the operated limb before the movement was detected, which was consistent for both abduction and flexion movement conditions. However, the differences in threshold angle between limbs were not statistically significant (abduction, Mann Whitney: p=0.48; flexion, Mann Whitney: p=0.59).

Figure 8.2 also showed less between-subject variability of MDS measured for the control limb of the resurfacing group, particularly for the flexion movement condition. MDS of the hip resurfacing group did not appear to vary with the movement condition presented. This was supported by statistical tests which showed no significant difference in the threshold angles during abduction and flexion motion for the operated limb (Mann Whitney: p=0.84) or the control limb (paired t-test: p=0.99).

8.2.2 Comparison of MDS between the standard THR and resurfacing group

The mean threshold detection angles for the operated limbs of the standard THR and hip resurfacing groups were compared and results are shown in figure 8.3.

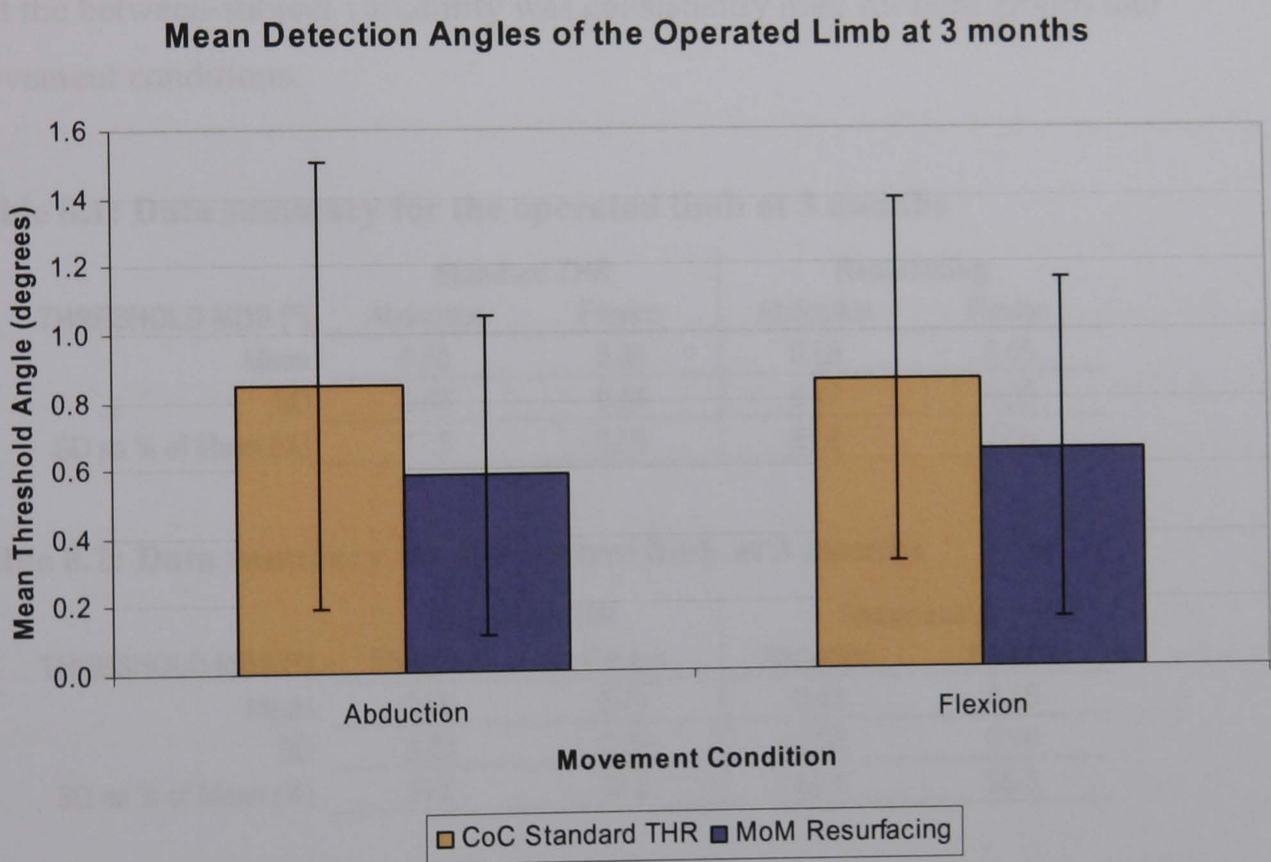


Figure 8.3: Comparison of mean threshold detection angles of the operated limb between arthroplasty groups

Figure 8.3 shows that the operated limb of the standard THR group detected motion at a greater threshold angle compared to the hip resurfacing group. The standard THR group detected abduction motion at a mean angle 0.27° greater than the average threshold angle of the resurfacing group; and detected flexion motion at a mean angle 0.21° greater than the average threshold angle of the resurfacing group. Hence there was a trend towards greater delay in MDS in the standard THR group.

Nevertheless, there was a large amount of variability in the mean angles and the between group differences were not significantly different for either abduction (Mann Whitney: p=0.09) or flexion (Mann Whitney: p=0.24) of the operated limb.

When considering the between-subject variability of MDS measured for the operated limbs, there appeared to be little difference between the standard THR and resurfacing groups, especially for the flexion movement condition. The SD of the mean expressed as a percentage of the mean is displayed in table 8.1. This shows that the between-subject variability was consistently high for both groups and movement conditions.

Table 8.1: Data summary for the operated limb at 3 months

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.85	0.86	0.58	0.65
SD	0.66	0.54	0.47	0.50
SD as % of Mean (%)	77.8	62.9	81.6	77.6

Table 8.2: Data summary for the control limb at 3 months

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.66	0.71	0.43	0.45
SD	0.23	0.29	0.28	0.16
SD as % of Mean (%)	34.8	40.8	65.1	35.6

Differences between the standard THR group and hip resurfacing groups were also analysed for the non-operated limb. These differences are displayed as numerical values in table 8.2 and graphically in figure 8.4.

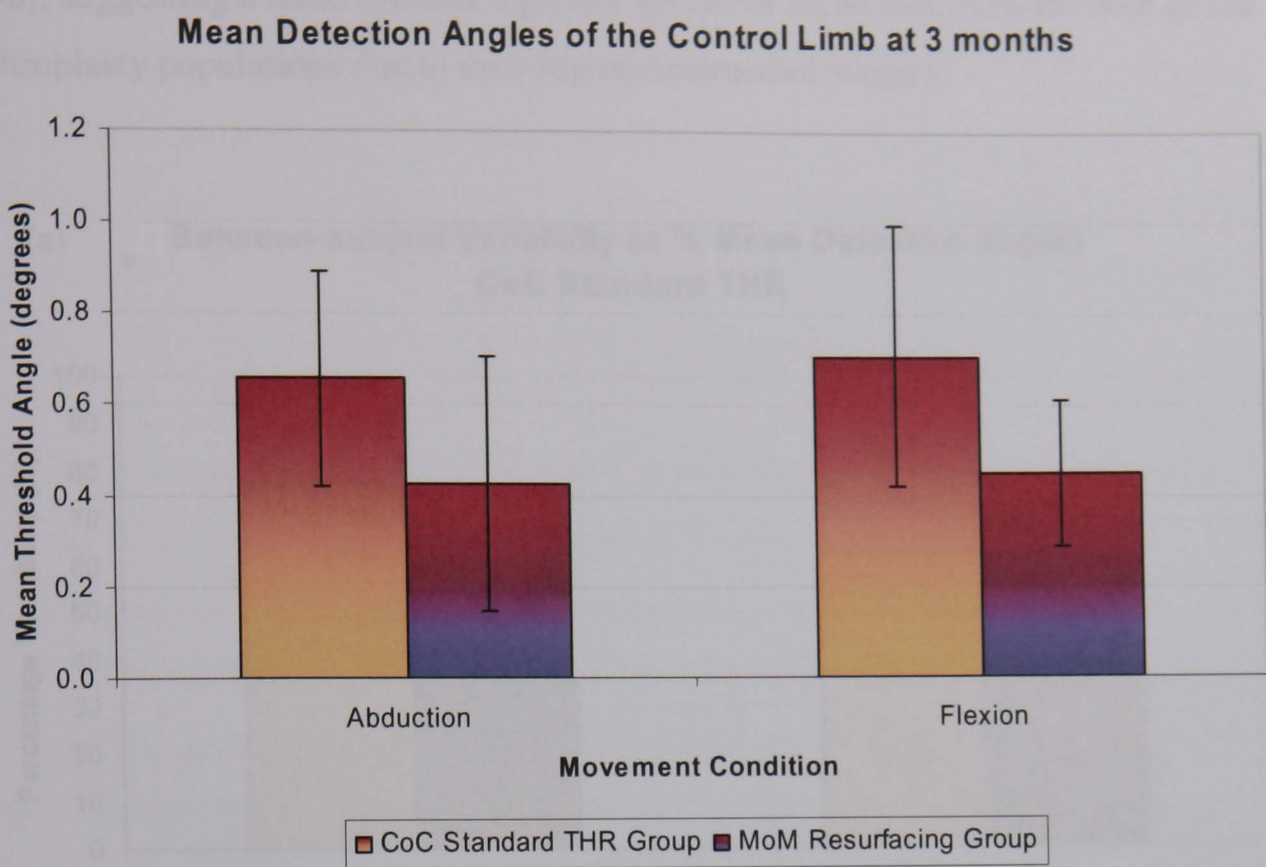


Figure 8.4: Comparison of mean threshold detection angles of the control limb between arthroplasty groups

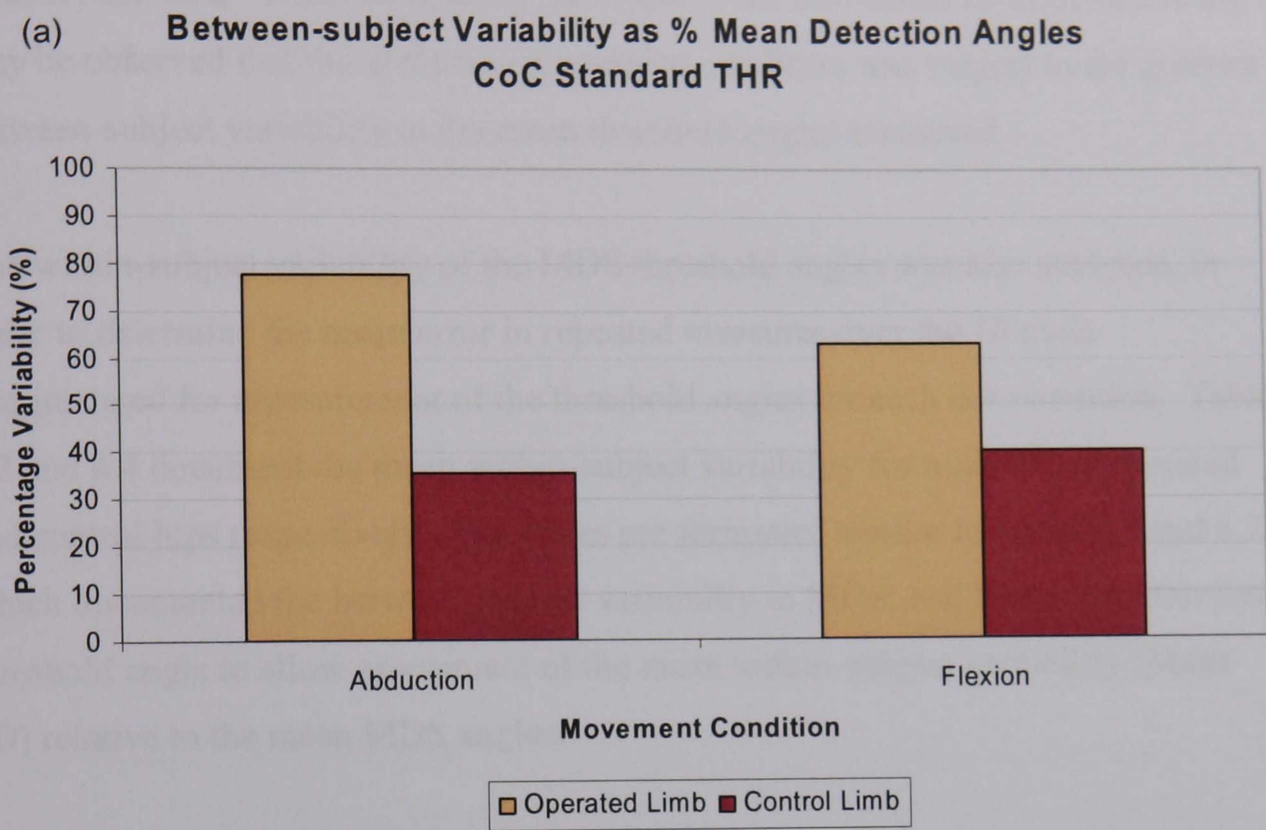
The MDS data of the control limbs shows a similar trend to the data reviewed for the operated limbs in figure 8.3. The standard THR group displayed significantly greater detection angles than the resurfacing group for their 'normal' non-operated control limb (abduction, 2-sample t-test: $p=0.025$; flexion, 2-sample t-test: $p=0.008$). Hence, in contrast to the data from the operated limbs, the control limb data showed statistically significant differences between the arthroplasty groups. In addition, as previously discussed for the control limb data (section 8.3.1), between subject variability was relatively less for the control limb than the operated limb. This observation is supported when the SD is expressed as a percentage of the mean threshold angles (tables 8.1 and 8.2).

8.2.3 Variability of the 3 month post-operative data

The findings at the early post-operative phase showed considerable variability in the comparison between subjects of the same arthroplasty group. This between-subject variability was greater for the operated limb than the control limb (figure 8.5a and

8.5b), suggesting a trend towards a greater spread of mean measures for both of the arthroplasty populations due to their hip reconstructive surgery.

(a)



(b)

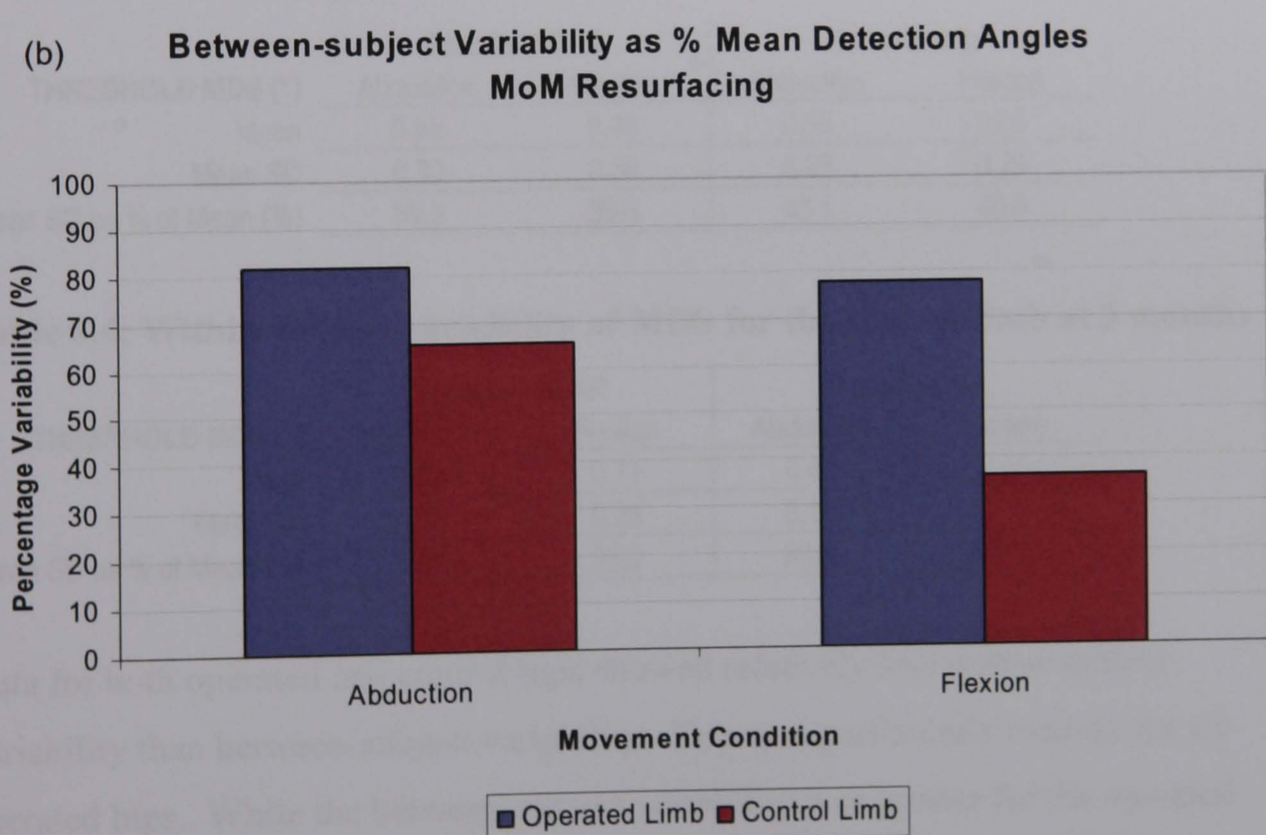


Figure 8.5: Comparison of between-subject variability of the operated and control limb for subjects with (a) CoC standard THR and (b) MoM resurfacing

It can be observed from figure 8.5 that the between-subject variability for the control limb of the resurfacing group during abduction was greater than the remaining control limb data. When comparing the effect of the movement conditions overall, it may be observed that the abduction movement condition was subject to the greatest between-subject variability in the mean threshold angles measured.

The within-subject variability of the MDS threshold angles was also analysed, in order to determine the mean error in repeated measures over the 10 trials administered for measurement of the threshold angles for each test condition. Tables 8.3 and 8.4 document the mean within-subject variability for trials of the operated and control hips respectively. The tables are formatted similar to tables 8.1 and 8.2, which documented the between-subject variability in MDS, and they repeat the mean threshold angle to allow assessment of the mean within-subject variability (Mean SD) relative to the mean MDS angles.

Table 8.3: Within-subject variability of MDS for the operated limb at 3 months

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.85	0.86	0.58	0.65
Mean SD	0.30	0.34	0.25	0.26
Mean SD as % of Mean (%)	35.3	39.5	43.1	40.0

Table 8.4: Within-subject variability of MDS for the control limb at 3 months

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.66	0.71	0.43	0.45
Mean SD	0.23	0.25	0.18	0.17
Mean SD as % of Mean (%)	34.8	35.2	41.9	37.7

Data for both operated and control hips showed relatively less within-subject variability than between-subject variability. This was particularly evident for the operated hips. While the between-subject variability was greater for the operated limb relative to the control limb, the within-subject variability was similar for both limbs, implying that the error in detecting hip motion over repeated trials was similar for both hips of the same subject and did not vary due to having a hip arthroplasty.

The within-subject variability was slightly greater for the resurfacing arthroplasty group, possibly implying less consistency in the detection angles with repeated measures following resurfacing compared to standard THR. Errors in detection of hip motion were similar for motion stimulus in both abduction and flexion planes.

8.3 Late MDS Results - 12 month Data

The late phase results of MDS data were averaged for each individual. Individual mean results were categorised according to their surgical group and results of each surgical group were averaged producing mean results for the operated and control limbs during both movement conditions (abduction and flexion).

8.3.1 Comparison of MDS between the operated limb and the control limb

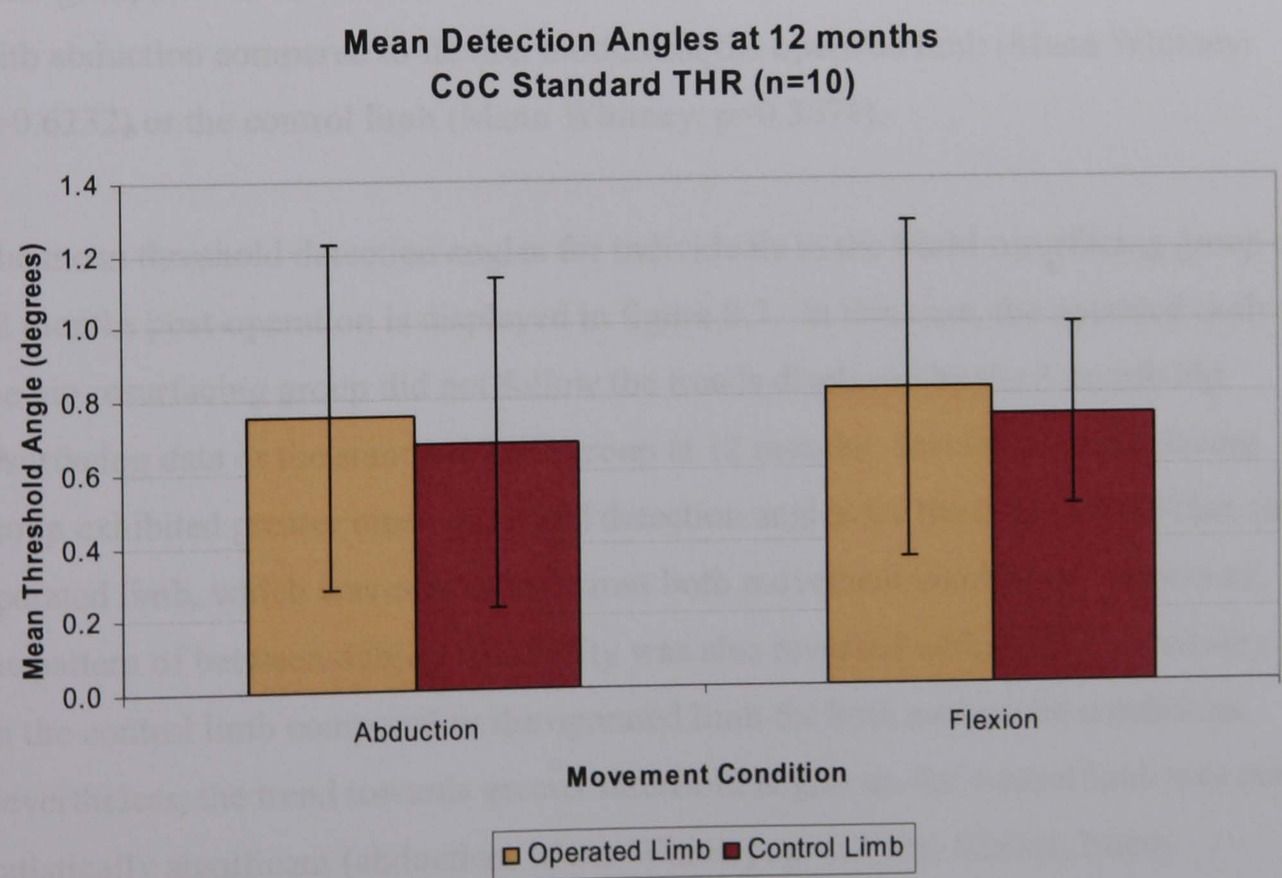


Figure 8.6: Comparison of mean threshold detection angles for operated and control limbs of subjects with CoC standard THR

The mean threshold detection angles at 12 months for individuals with a CoC standard THR are displayed in figure 8.6. Figure 8.6 illustrates that the operated limb of the standard THR group had slightly greater mean threshold detection angles than the non-operated control limb for both abduction and flexion movement conditions. This implies that the trend difference of delayed MDS due to the standard THR surgical procedure persisted at 12 months post-surgery. The early pattern of less between-subject variability of threshold angles measured for the control limb was also consistent at 12 months for the flexion trials. However, there was no observable difference in the variability between the operated and control limbs for the abduction trials at 12 months.

Applying statistical tests to the mean 12 month data sets showed no significant difference between the operated and control limbs for abduction (Mann Whitney: p=0.5708) or flexion (Mann Whitney: p=0.6760) movement conditions.

When comparing the effect of the movement conditions on MDS of the standard THR group, it was found that there was no difference in the threshold angles yielded with abduction compared to flexion motion for the operated limb (Mann Whitney: p=0.6232) or the control limb (Mann Whitney: p=0.3074).

The mean threshold detection angles for individuals in the MoM resurfacing group at 12 months post-operation is displayed in figure 8.7. In this case, the operated limb of the hip resurfacing group did not follow the trends displayed by the 3 month hip resurfacing data or the standard THR group at 12 months. Instead, the resurfacing group exhibited greater mean threshold detection angles for the control limb than the operated limb, which was consistent across both movement conditions. Moreover, the pattern of between-subject variability was also reversed with greater variability on the control limb compared to the operated limb for both movement conditions. Nevertheless, the trend towards greater threshold angles on the control limb was not statistically significant (abduction, Mann Whitney: p=0.8501; flexion, Mann Whitney: p=0.8501).

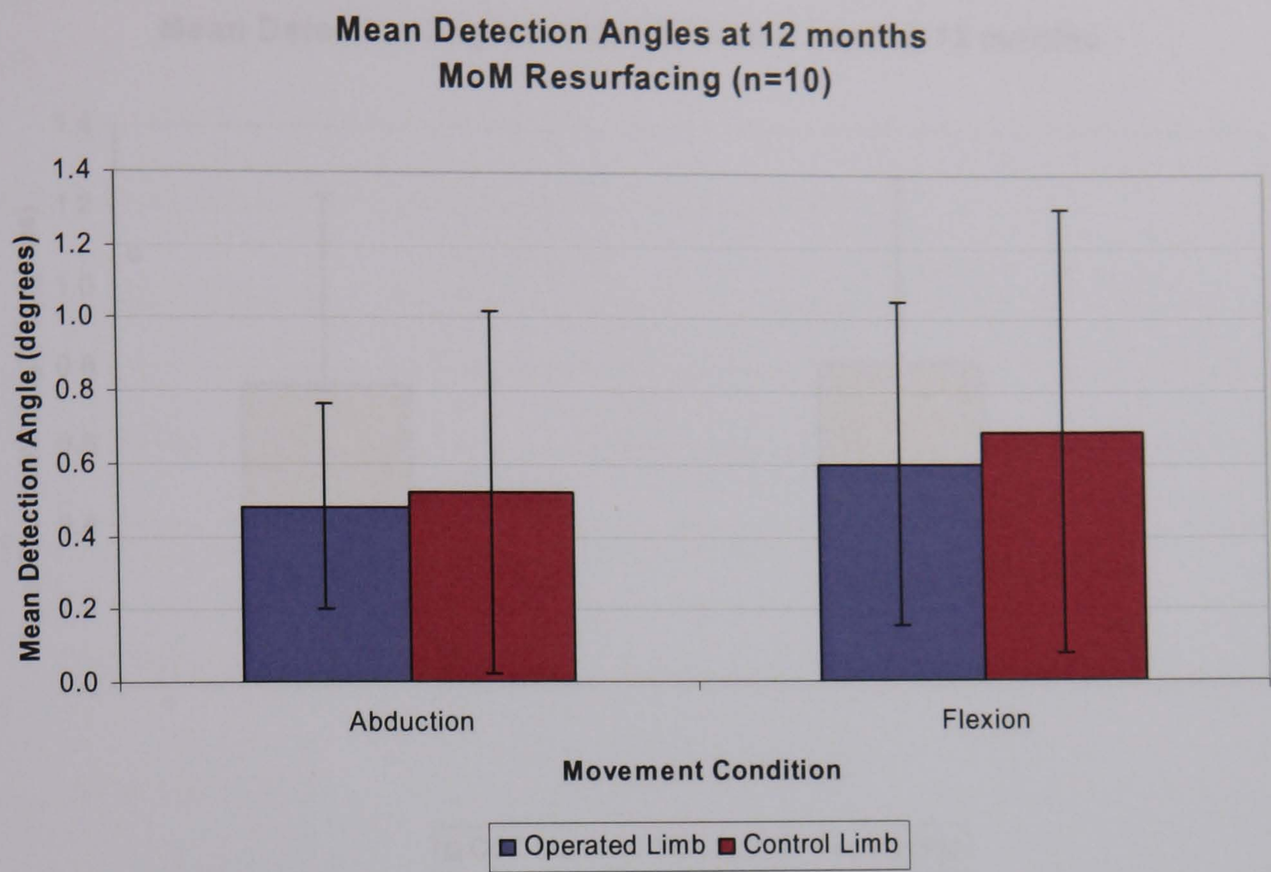


Figure 8.7: Comparison of mean threshold detection angles for operated and control limbs of subjects with MoM resurfacing

These results also showed a trend difference between flexion and abduction movement conditions, with greater threshold detection angles observed for flexion trials. This observation was consistent for both limbs of the resurfacing arthroplasty group and suggests that relatively greater angular motion was required during flexion motion before the movement was detected by the resurfacing group. However, this trend difference was minimal and was not supported by statistical tests which showed no significant difference in the threshold angles for abduction or flexion motion for the operated limb (Mann Whitney: p=0.5708) or the control limb (Mann Whitney: p=0.4274).

8.3.2 Comparison of MDS between the standard THR and resurfacing group

The mean threshold detection angles for the operated limbs of the standard THR and hip resurfacing groups at 12 months post-surgery were compared. The results are shown in figure 8.8.

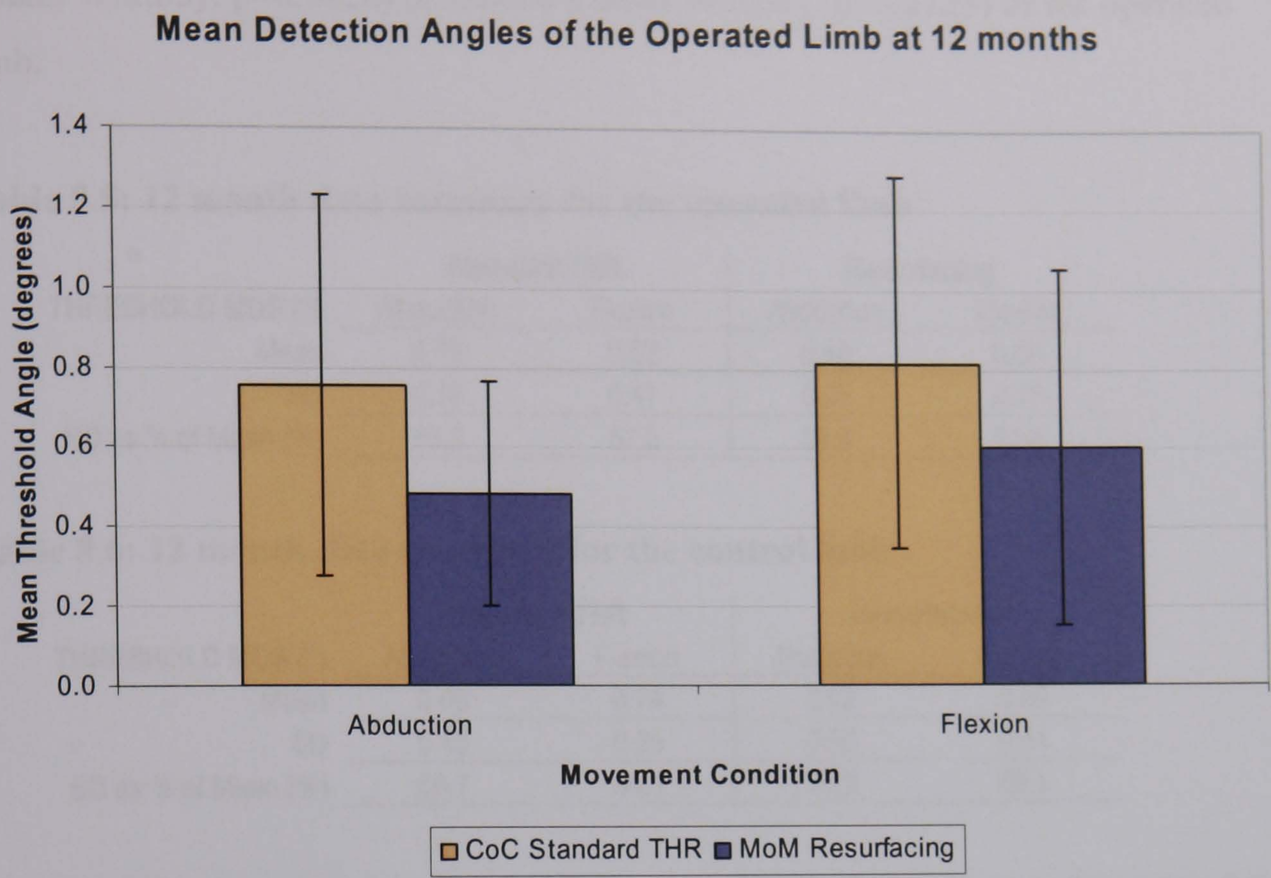


Figure 8.8: Comparison of mean threshold detection angles for subjects with CoC standard THR and MoM resurfacing

The figure shows that the operated limb of the standard THR group detected motion at a greater threshold angle compared to the hip resurfacing group. This 12 month result reflects the trend observed at 3 months post-surgery. At the 12 month phase, the standard THR group detected abduction motion at an angle 0.27° greater than the average threshold angle of the resurfacing group, which exactly matched the mean difference observed for abduction motion at 3 months. The standard THR group detected flexion motion at an angle 0.21° greater than the average threshold angle of the resurfacing group, which again exactly matched the differences observed between the arthroplasty groups at 3 months for flexion motion. Hence the trend towards greater delay in MDS in the standard THR group mirrored that seen at the early post-operative stage.

The large amount of variability in the mean angles was consistent at 12 months. Given this variability in the mean measures, assessment of differences between the arthroplasty groups showed no statistically significant difference for either abduction

(Mann Whitney: p=0.1620) or flexion (Mann Whitney: p=0.2123) of the operated limb.

Table 8.5: 12 month data summary for the operated limb

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.75	0.82	0.48	0.60
SD	0.48	0.47	0.28	0.45
SD as % of Mean (%)	63.3	57.3	58.4	74.8

Table 8.6: 12 month data summary for the control limb

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.68	0.74	0.52	0.69
SD	0.45	0.25	0.50	0.61
SD as % of Mean (%)	66.7	34.1	95.3	89.1

The SD of the mean expressed as a percentage of the mean is displayed in table 8.5. This shows that the between subject variability was consistently high for both groups and movement conditions, but the magnitude of variability was less than that observed at 3 months (table 8.1).

Whereas at 3 months the variability of the control limb was less than that of the operated limb, this pattern did not continue with the 12 month results (table 8.6). Mean differences between the standard THR group and hip resurfacing group at 12 months were analysed for the non-operated control limb and the results are displayed in figure 8.9.

The MDS data of the control limbs (figure 8.9) showed a similar trend to the data reviewed for the operated limbs in figure 8.8. The standard THR group displayed slightly greater detection angles than the resurfacing group for their 'normal' non-operated control limb but, in contrast to the 3 month post-op results, these differences were not statistically significant (abduction, Mann Whitney: p=0.1041; flexion, Mann Whitney: p=0.1530).

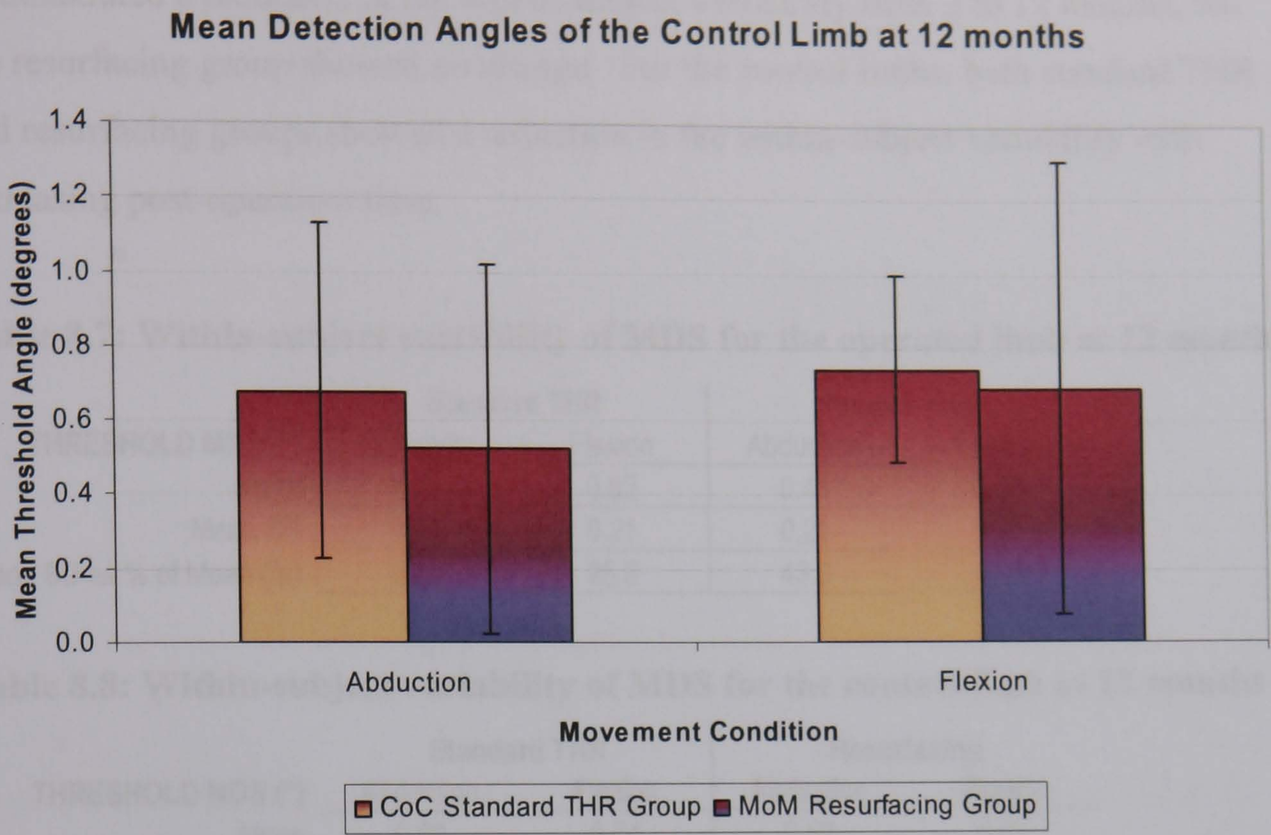


Figure 8.9: Comparison of mean threshold detection angles of the control limb for subjects with CoC standard THR and MoM resurfacing

8.3.3 Variability of the 12 month post-operative data

The pattern of between-subject variability at 12 months post-surgery showed less variability in the mean data values of the operated limb (table 8.5) compared to the 3 month post-op results (table 8.1). This trend was consistent for both arthroplasty groups. The reduction in variability with time suggests some improvement in MDS over the 9 month gap interval in testing. The control limb of the standard THR group reflected this reduction in variability seen for the operated limbs, but the resurfacing group exhibited and increase in variability of the control limb data from 3 months to 12 months (table 8.2 and 8.6).

The within-subject variability of the MDS threshold angles at 12 months is shown in tables 8.7 and 8.8, documenting the mean within-subject variability for trials of the operated and control hips respectively. The format of the tables follows that used for the 3 month data in section 8.2.3. As with the 3 month data, there was relatively less within-subject variability compared to the level of between-subject variability observed for the 12 month results. For the operated limbs, the standard THR group demonstrated a reduction in the within-subject variability from 3 to 12 months, but the resurfacing group showed no change. For the control limbs, both standard THR and resurfacing groups showed a reduction in the within-subject variability with increasing post-operative time.

Table 8.7: Within-subject variability of MDS for the operated limb at 12 months

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.75	0.82	0.48	0.60
Mean SD	0.24	0.21	0.21	0.24
Mean SD as % of Mean (%)	32.0	25.6	43.8	40.0

Table 8.8: Within-subject variability of MDS for the control limb at 12 months

	Standard THR		Resurfacing	
THRESHOLD MDS (°)	Abduction	Flexion	Abduction	Flexion
Mean	0.68	0.74	0.52	0.69
Mean SD	0.20	0.18	0.20	0.22
Mean SD as % of Mean (%)	29.4	24.3	38.5	31.9

At 3 months, the within-subject variability was similar for both limbs. This was also consistent for the 12 month data when accounting for the actual mean error angles between the groups. When accounting for the ratio of the error value, the standard THR group maintained the consistency of variability between limbs but the operated hips of the resurfacing subjects showed less consistency in their threshold MDS angles over repeated measures.

As with the 3 month data, the within-subject variability in the detection of hip motion was similar for abduction and flexion motion trials.

8.4 Comparison of Data at 3 and 12 months post-operation

Figures 8.10 and 8.11 show the mean threshold detection angles at 3 and 12 months post-surgery for the operated limbs. The results for the standard THR group show minimal difference between the two measurement stages, with the greatest difference observed for abduction movement conditions.

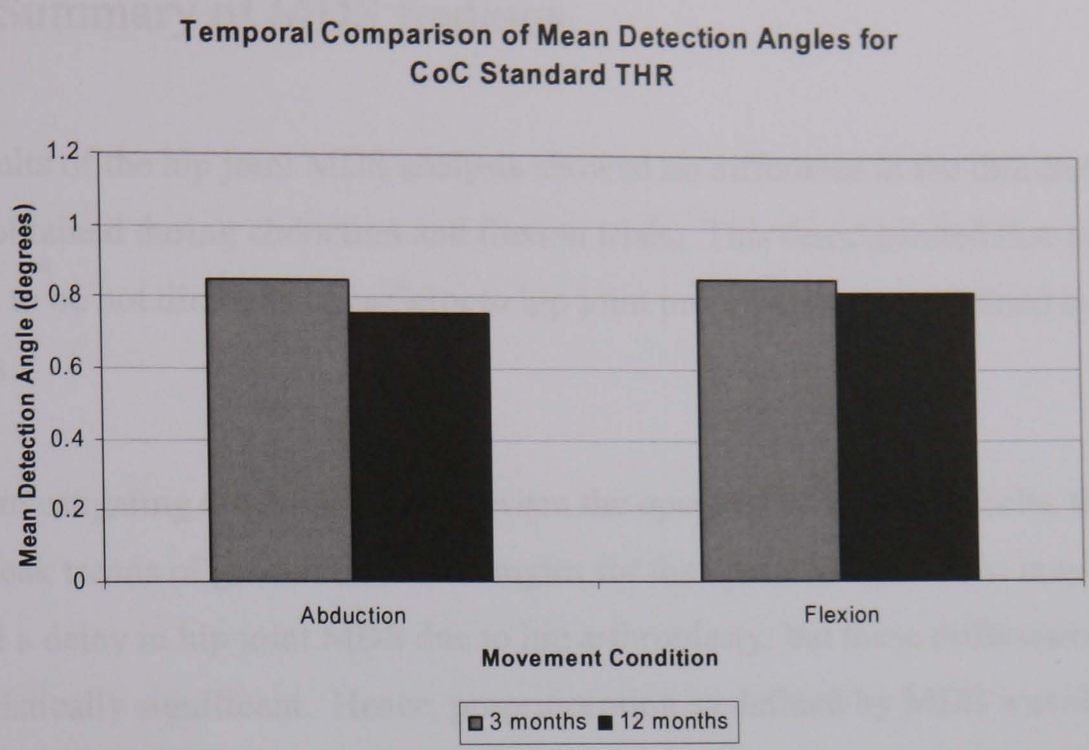


Figure 8.10: Comparison of mean threshold detection angles between 3 and 12 month for hips with CoC standard THR

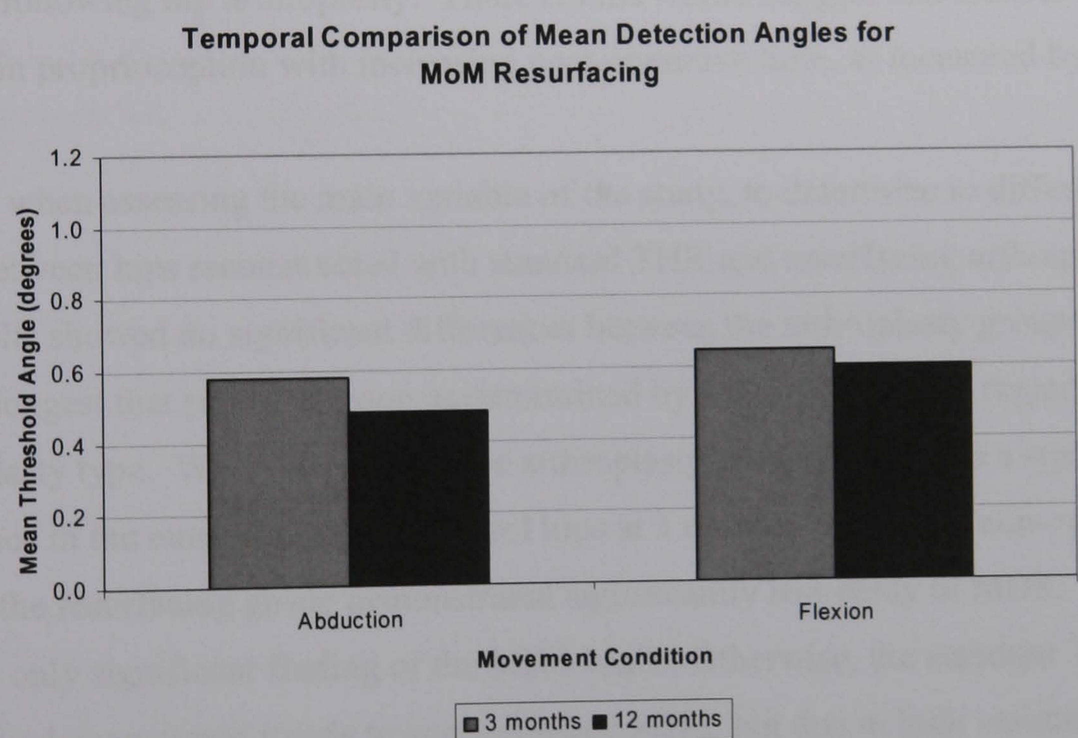


Figure 8.11: Comparison of mean threshold detection angles between 3 and 12 month for hips with MoM resurfacing arthroplasty

Similarly with the resurfacing hips, there was minimal change in the MDS results from 3 to 12 months, and most improvement (reduction) in threshold detection angle was observed for the abduction trials.

8.5 Summary of MDS findings

The results of the hip joint MDS analysis showed no difference in the threshold angles obtained during abduction and flexion trials. This demonstrated that there appears to be not direction specificity to hip joint proprioception as defined by MDS analysis.

When investigating the differences between the operated and control limbs, there were weak trends of greater threshold angles for the operated hips. This may have implied a delay in hip joint MDS due to hip arthroplasty, but these differences were not statistically significant. Hence, proprioception as defined by MDS was similar for operated and non-operated hips.

The threshold angles were statistically similar for measurements recorded at 3 and 12 months following hip arthroplasty. These results would suggest that there is no change in proprioception with increasing post-operative time, as measured by MDS.

Finally, when assessing the main variable of the study, to determine to difference in MDS between hips reconstructed with standard THR and resurfacing arthroplasty, the results showed no significant differences between the arthroplasty groups. This would suggest that proprioception as determined by MDS was similar regardless of arthroplasty type. When contrasting the arthroplasty groups, there was a significant difference in the outcomes for the control hips at 3 months, where the non-operated hips of the resurfacing group demonstrated significantly less delay in MDS. This was the only significant finding of the MDS trials. Otherwise, the standard THR hips only demonstrated trends towards delayed MDS, but due to high variability in the mean threshold angles, these differences due to arthroplasty type were not significant.

The proprioception findings and their implications will be discussed further in the chapter 9.

9 Discussion

This study set out to quantify the differences in functional performance of those having had hip resurfacing arthroplasty compared to those with conventional THR, by way of determining which arthroplasty type optimised post-operative outcomes.

The current chapter will discuss the results of the study, their implications and importance, with specific attention to the core purpose of the thesis; to identify differences between resurfacing and standard THR. The results will be appraised for their usefulness in determining any differences between the functional performance of the standard Trident CoC hybrid bearing THR and the Durom MoM resurfacing hip arthroplasty. The importance of the findings will be judged and the magnitude of the functional differential due to arthroplasty type will be evaluated. In return, the implications of these conclusions will be considered. Impact and inference for the patient population, clinical institutions and industrial companies will be discussed, and recommendations suggested for orthopaedic practice and for further study.

The structure of this discussion has been formatted to firstly address the motion analysis findings, followed by the motion detection sense results. The objective results will be related to the subjective and descriptive data where relevant. The methods will be appraised for their merits and limitations and, where appropriate, adaptations of the protocol will be proposed.

In addition to discussing the main dependent variable of arthroplasty type, the differences in outcomes between operated and non-operated limbs and between early and late post-operative test intervals will also be examined.

9.1 Motion Analysis Findings

The motion analysis study assessed outcomes of hip angles, hip moment and temporospatial parameters of the motion tasks. These outcome parameters will be discussed separately in the sections to follow.

9.1.1 Hip angle data: Differences due to hip arthroplasty

The peak angular values recorded during the present analysis were comparable with existing data (section 2.6 and 2.7). For level gait, the peak flexion values consistently mirrored normative values of previous gait analysis trials (30 to 35°) with the exception of slightly greater flexion peaks for the standard THR group at 12 months (38.9°). Peak extension values were below 'normal' (10 to 15°) for the operated limb and comparable to 'normal' for the control limb, with the exception of the resurfacing operated hips achieving a mean of 10.4° of extension by 12 months post-op. Kadaba et al (1989) published normative coronal and transverse hip angles during gait (figure 2.20). The current subject group showed slightly greater mean peak adduction and abduction angles, but slightly reduced internal and external rotation angles than those reported by Kadaba et al. The smaller range of transverse angles may be due to the greater age as well as the pathology associated with the current subject group (section 2.1.2.1).

Normative hip angle values for stair ascent and descent also compared well with the current findings. Mean peak flexion angles found in literature varied widely from 40° (Andriacchi et al 1980) to 70° (Riener et al 2002). The current data better matched the 60 to 65° range documented by McFadyen and Winter (1988), Nadeau et al (2003) and Protopapadaki et al (2006). The hip extension angles recorded from the current study group also fitted within the range reported within literature (-4.7 to -15°). Only one published study has assessed coronal plane angles during stair ascent (Nadeau et al 2003) with peak adduction of 9.7° and peak abduction of 4.1°. Again the current findings match well with the data previously reported (table 7.4). The angles obtained from the current stair descent data can only be compared with literature data of sagittal plane angles (Andriacchi et al 1980, McFadyen and Winter

1988, Nadeau et al 2003, Protopapadaki et al 2006, Riener et al 2002), but as with stair ascent, these compared well for 'normal' flexion (37 to 45°) and extension (0 to -20°), with the current data falling within these ranges. Hence, the current peak angle values were representative of those found in published literature. Given that the data is comparable with normative functional angles suggests an excellent outcome of functional motion following hip arthroplasty.

Most of the meaningful findings from the angle data highlighted differences between the operated and control limbs as opposed to the differences between the standard THR and resurfacing hips. The difference between angles of operated and control hips were greatest at 3 months, indicating that as time progressed, the magnitude of differences between limbs reduced, or became minimal. The improvement over time may have been due to post-operative healing, increased activity or reduced pain inhibition.

Of the angular differences between-limbs, some may be classified as a restriction and some may be classified as a symmetry imbalance of the operated hip relative to the non-operated control hip.

9.1.1.1 Angular restriction of the operated limb

One of the most significant differences between the operated and non-operated hips was the relatively reduced peak extension angle of the operated hip, which occurred at terminal stance phase. This has been observed in previous studies (Murray et al 1971) and was most likely due to the presence of a fixed hip flexor muscle contracture, limiting hip extension (Hurwitz et al 1997). The reduced extension angle was statistically significant at 3 months post-op during level walking, and showed trend reductions relative to the non-operated hip at 3 months during stair ascent. These early limitations improved by 12 months. The improvements may be attributed to the hip prosthesis allowing more 'normal' movement, the reduction or resolution of pain, the return of more normal activity or a combination of all these factors. Nevertheless, the peak extension angle of the operated hip remained sub-optimal compared to the control hip at 12 months. Functional improvements following hip arthroplasty are reported to plateau by 12 months post-op (Tanaka 1998), hence the relatively poorer extension range of the operated hip at the late assessment phase may imply that a long-term impairment had been established. The persistence of flexion contractures 1 year following surgery has been observed previously (Olsson 1986, cited in Perron 2000), which reinforces the need for management of such soft-tissue restrictions.

There are significant implications associated with the limitation of hip extension range of motion. Although the between-limb difference was not statistically significant at 12 months, the late trends may be clinically significant. Reduced extension angles at terminal stance have the effect of reducing the power generation at push-off, leading to reduced propulsion (Perron et al 2000, Watelain et al 2001). Hence several aspects of the gait cycle would be adversely affected due to this motion restriction. Further reasoning of this chain of biomechanical problems infers that the limp often observed during gait and frequently reported by the subjects (section 6.3.1), particularly at 3 months, may be related to the reduced peak extension angles. Moreover, reporting of limp on the OHS reduced between 3 and 12 months (section 6.3.1), as did the significance of the limitation in extension ROM.

The reduced compliance of the flexor muscle group was unlikely to have varied depending on the arthroplasty group, but may have varied continuously between individuals. Variation in the extent of the contracture may have been influenced by factors such as the severity and duration of the degenerative joint disease, the severity of pain affecting the individual and the activity level of the patients pre-operatively. Quantifying the severity of the pre-operative flexor muscle contracture was possible. This could be done by goniometric measurement of the maximum passive extension range of motion; a measurement outcome used by the clinicians prior to orthopaedic intervention. Unfortunately, this measurement was not consistently reported in pre-operative clinical records. This omission might have been due to some patients having such severe contractures that the extension range was not noted as there was effectively no extension, or negative degrees of extension, qualifying a flexion contracture. Alternatively, it may have been difficult or painful to position the patient in order to carry out the measurement. Most often, if a limitation in extension ROM was present, the triage reports documented a positive Thomas' test. Given that the extent of the extension restriction was not quantified, pre-operatively or during the test sessions, the researcher could not rank this impairment between subjects or attempt to correlate the impairment to the terminal extension angles during gait analysis. Nevertheless, the lack of extension ROM observed was likely to contrast with the extension range measured during surgery, which averaged at 10-15°, as documented in theatre case notes. In cases where flexion contractures (≤0° extension) presented following trial of the arthroplasty components and before surgical closure, the orthopaedic surgeons would surgically release the hip flexor muscles to optimise post-operative ROM. In the current study, no patients required such surgical release, and all achieved a good extension range intra-operatively. This would therefore indicate that the subjects appeared to have lost their optimal range of extension in the early stages following surgery and may imply the need for modified early rehabilitation to encourage maintenance of the hip extension angles achieved in theatre.

In addition to the limitation of hip extension, the operated hips also showed a limitation of flexion motion during all modes of ambulation at 3 months. At 12 months the level walking task showed a trend of poorer flexion angles for the operated limb but the stair negotiation tasks showed no difference between limbs. Hence, again there was an improvement in the outcomes with time, though accumulatively the sagittal plane peak angles showed a reduced total angular excursion for all tasks, particularly at the early stages post-operatively.

There were contrasts to be drawn between the lack of flexion and the lack of extension motion. The maximum extension angles produced during gait were similar to the end range of physiological motion of hip extension (Murray et al 1964), whereas the maximum flexion angles only represented the mid-range of motion of hip flexion, particularly for gait, followed by stair descent and then with stair ascent. The limitation in hip flexion motion during these functional activities was therefore not due to a passive restriction and stretch of soft tissues, but was more likely to be related to a reduction in muscle strength or reduced stride length. The argument of reduced stride length to justify the reduced flexion angles is less valid for stair negotiation compared to level gait, as the step distance and step height were fixed for stair ascent and descent. Hence, the reduced flexion during these activities implies that the operated hips used the minimum required motion in order to clear the step. In addition, or alternatively, other compensatory strategies may have been used; for example circumduction of the hip or increased lateral pelvic tilt and trunk side flexion. Hip circumduction could be ruled out as one might expect the abduction angles of the operated hip to be increased if this were the case; instead, the abduction angles were reduced relative to that of the control hip, and with the operated limb being more adducted, this might represent a pattern of lateral elevation of the pelvis instead.

9.1.1.2 Inverse angle patterns between-limbs

All three functional tasks showed similar results for the coronal plane hip angles. The operated hip was significantly more adducted compared to the control hip and the control hip was significantly more abducted compared to the operated hip, reflective of the frontal plane obliquity characterised by a Trendelenburg-type pattern. This finding supports those of previous studies (Sliwinski et al 2004, Watelain et al 2001) and was an example of a symmetry imbalance of left and right hips due to the unilateral nature of the subjects' condition, relating to both pre-existing hip disease and corresponding arthroplasty.

The pattern of coronal plane angles would imply that there was a lateral shift of the COM over the operated limb, so as to reduce the body weight lever arm and minimise the mechanical work required by the abductor muscles of the operated limb, and consequently reduce the loading of the operated hip (section 2.1.2.2, figure 2.9). Hence this pattern was likely to be a compensatory mechanism and has previously been observed following hip arthroplasty (Sliwinski et al 2004) and secondary to hip OA (Watelain et al 2001). By adopting this behaviour, subjects could favour the operated hip and minimise biomechanical demand (Charles et al 2004, Sariali et al 2008). This would also serve to reduce existing pain by reducing joint loading. On the other hand, it may have been a strategy acquired from the pre-operative phase to alleviate pain and subjects were carrying over this learned behaviour pattern to the post-operative stage. This is an interesting concept and it questions the psychological aspects of the patients' recovery. All patients scored a significant improvement in the OHS, including all pain items, and the VAS pain levels reported at the 3 month assessment phase were minimal. Hence it may be more likely that the angle deviations were due to fear avoidance and learned compensation behaviours rather than pain avoidance.

Alternatively abductor muscle weakness of the operated limb due to the pre-operative joint pathology, pain inhibition, or surgical trauma resulting from the surgical incision may have directly caused the pelvis to drop towards the non-operated limb, resulting in a more adducted position of the operated limb. Muscle weakness allowing high joint reaction forces may have also induced the compensations discussed above. The possibility of muscle dysfunction being responsible for the coronal plane angle pattern will be considered further in section 9.1.2, featuring the discussion of external intersegmental moments of the hip joint.

The relatively greater angles of adduction of the operated hip and abduction of the control hip were present for level walking at 3 months and present for both stair negotiation modes at 3 and 12 months post-op. This suggests that the compensatory behaviour reduced with time, as detected by level walking. Notably, when examining the changes in coronal plane angles between 3 and 12 months for level walking, it was apparent that the control limb was responsible for the return of between-limb symmetry with time. Between 3 and 12 months, the control limb became significantly more adducted. This finding was interesting as it highlighted that the control limb was responsible for the compensatory behaviour observed rather than the operated limb. This is logical; the more able limb was compensating for the impairments of the pathological limb. Therefore, compensatory behaviour, as observed for level walking, diminished by the late assessment phase and there was bilateral symmetry in the coronal plane angles. This progression may have been due to reduced fear avoidance, reduced pain or adoption of more normal motion patterns.

With this reasoning in mind, it must be questioned as to why the bilateral mismatch in coronal angle peaks did not improve with time for assessment of the stair negotiation tasks. At the late assessment phase, as examined from stair ascent and descent trials, the operated limb remained significantly more adducted compared to the control limb and vice versa. With regard to the aforementioned conclusions, the late results of stair negotiation therefore imply that the compensation behaviour did not diminish by 12 months. It is possible that the stair ascent and descent activities placed a greater threshold biomechanical demand on the hip joints compared to level gait and continued to induce the compensatory pattern.

Another finding which highlighted a mismatch of the angle data between limbs was the difference in peak axial rotation values. Similar to the pattern of coronal plane angles, the peak transverse angles were skewed between limbs. The operated hip had significantly greater internal rotation angles than the control hip and the control hip had significantly greater external rotation peaks than the operated hip. This was evident from the analysis of level walking, stair ascent and stair descent. Hence all the ambulation modes highlighted a consistent pattern of a mismatch in functional rotation characteristics, which were due to hip arthroplasty, as shown by statistical analysis.

Perron et al (2000) also documented greater peak internal rotation of the operated hips in their conventional THR group during level walking. When considering the reasons why the operated hip had relatively greater internal rotation angles, several possible explanations were deduced. Intra-operatively the surgeon detached the short external rotator muscles of the hip in order to gain access to the hip joint via the posterior incision. The external rotators were then left unrepaired following standard practice, followed due to the high failure rate and the limited effectiveness associated with such muscle repairs (Stähelin et al 2002). However, leaving the external rotators unrepaired may have the resultant effect of causing a muscle imbalance at the hip, with the internal rotators generating a pull that was not balanced by the external rotators, causing a net internal rotation motion and more internally rotated resting position of the hip joint. Relating this theory to the axial rotation moment patterns found, the profile graphs (chapter 7) showed relatively larger internal rotator moments, particularly during the stair climbing activities. The peak internal rotator moments were larger than the peak external rotator moments during stair climbing, particularly for the standard THR group at 12 months, which was close to being significantly different to the non-operated limb (p=0.053). Hence, although the functional range of hip motion in the transverse plane was similar for operated and control limbs, the operated hip maintained a greater median and peak angle of internal rotation motion and a tendency towards greater internal rotator moments.

Another plausible explanation for the greater relative internal rotation of the hips having undergone arthroplasty was derived from considerations into aspects of the anatomical reconstruction of the hip joint. The posterior approach for THR is associated with higher dislocation rates compared to the other approaches (section 2.4.1), and to compensate for this, the surgeon can alter aspects of the surgical procedure and alignment of implants, within acceptable limits, in order to minimise the risk of dislocation. This may be done by placing the acetabular cup or socket in a more anteverted position, or, in the case of the standard THR, cementing the femoral stem in a rotated position to enhance anteversion of the prosthetic neck. Enhancing anteversion by means of anatomical reconstruction of the hip is known to limit the range of external rotation available to the individual (D'Lima et al 2000), due to early impingement with external rotation motion, but allow internal rotation ROM with reduced risk of posterior hip dislocation. This might explain why the maximum range of external rotation of the operated hip was less than that of the anatomic control hip. Furthermore, it may also explain why the operated hips tended to display a functional rotation range in a more internally rotated median position than the non-operated hips.

Standardised radiographic assessment was not undertaken in the current research protocol so this theory positively associating component anteversion with greater internal rotation angles may not be cross-examined. However, anecdotally, the probability of the reconstructed hips displaying greater angles of anteversion compared to the non-operated control hips has been supported by the surgeons involved in the current study (RI and AS), as these methods are inherent within the intended operative technique using a posterior approach. This concept prompts speculation as to what the functional rotation motion of the hips would have been pre-operatively. Indeed, the likely balance of rotation between limbs pre-operatively was expected to contrast with this post-operative result. Commonly the ROM restrictions due to the degenerative hip pathology result in a loss of internal rotation range. Moreover, there is typically a loss of flexion, abduction and internal rotation associated with the capsular pattern of the hip (Cyriax 1982, cited in Petty and Moore 1998), which is commonly presented on pre-op assessment of this patient group. Hence an externally rotated resting position of the hip joint pre-operatively might be caused by dysfunction of the capsule due to fibrosis or inflammation. Alternatively the externally rotated resting position and lack of internal rotation motion might have been due to impingement of bone. For example, bony spurs or osteophytes, particularly at the anterior rim of the acetabulum or femoral neck, may have caused a restriction in internal rotation or may have caused pain during motion (section 2.2.1). On the whole, the post-operative findings cannot be explained by continuation of the pre-operative rotation characteristics and it is therefore suggested that the hip arthroplasty procedure results in a significant change in the kinematic and kinetic behaviour of rotation at the hip.

Further interpretation of the greater internal rotation of the operated hip was considered with reference to the possible axial rotation of the femur at different geometric positions relative to the pelvis. It was not sufficient to think of the effects of rotation angles (transverse plane) in isolation. Appreciating that functional motion tasks result in a complex combination of hip angles in 3-dimensions, the affect of altering the moment arm of the surrounding muscles on hip joint angles was considered. For the natural or anatomic hip joint, it has been documented that internally rotated gait patterns are often related to anteverted femoral geometry and excessive hip flexion (Arnold and Delp 2001). This study proposed that greater angles of hip flexion during gait acted to optimise the internal rotation moment arm of the gluteus medius and minimus muscles to cause excessive internal rotation of the hip. Relating these findings to the current study could imply that, as well as the greater anteversion resulting from the anatomical reconstruction of the arthroplasty hip, the flexion contractures implicated from the limited extension ROM may cause sufficient hip flexion to mechanically induce greater internal rotation of the operated limb. Assessing the hip rotator moments between the operated and control limb did not fully support this theory as the internal rotator moments of the operated hip during gait were significantly lower than that of the control hip. This may have been because the flexion contractures were not as severe as the subject group that Arnold and Delp (2001) drew their conclusions from. They related their model to children with cerebral palsy, who would likely have had more severe flexion contractures than the young THR population.

9.1.2 *Hip angle data: Differences due to arthroplasty type*

When comparing the angular outcomes between the arthroplasty groups, the standard THR group showed fewer deviations from normal at 3 months but the resurfacing group showed the greatest improvement with time and generally surpassed the outcomes of the standard THR group at 12 months.

9.1.2.1 Differences in sagittal plane angles due to hip arthroplasty type

The results show that the hip resurfacing group had greater peak hip extension angles by 12 months post-op compared to the standard THR group. The opposite was true at 3 months, and although the standard THR group showed no change in hip extension with time, the resurfacing group showed a significant improvement.

The greater improvement and greater peak value of extension angles for the resurfacing group may be due to the relatively greater activity level of the resurfacing group and the high probability that they engaged in more normal patterns of activity compared to the standard THR group. The greater levels of activity and more normal motion patterns associated with the resurfacing group were deduced from the UCLA activity scores and the descriptive motion features described in chapters 6 and 7. A greater frequency, intensity and better quality (more normal and less inhibited) of functional movement would lead to greater dynamic stretch of the flexor muscle group and assist in reversing flexor muscle contractures. In contrast the standard

THR group demonstrated less progression as determined by subjective outcomes and scored a lower level of activity and this was reflected in the static nature of the peak extension angle outcome with time.

The implication that there was some degree of reversal of the flexor muscle contracture and improvement in peak functional extension range due to the return of normal activity is a positive outcome for the resurfacing patient group. This is particularly notable given that there was no provision of exercises to effectively stretch the flexor muscle group post-operatively. The routine hospital exercise programme does not incorporate stretches at the end range of hip extension and there was no progression of exercises following hospital discharge. Therefore the increase in extension range in this study could be attributed to dynamic stretch of the flexors as opposed to formal exercise stretching. However, the results also serve to remind the reader that the peak extension angles of the operated hip remained sub-optimal to the non-operated hip. Therefore the need to promote flexor muscle stretch is implied to achieve a better outcome which matches that of the non-operated limb. Based on the current group analysis, the findings could imply that the less active the individual the less recovery of extension range. Perhaps there are fewer implications of this hypothesis for the current, young active subject group, but when relating this to the less active elderly population, the need to appropriately manage hip flexor contractures is enhanced.

The peak flexion pattern differences between groups also have some interesting implications. While the resurfacing group showed a development of greater extension peak angles, the standard THR group demonstrated a development of greater flexion peak angles with time. These patterns were consistent for all ambulation modes. Given that the standard THR group had an increase in flexion angles without a corresponding increase in extension angles, this might suggest that the group possibly adopted greater anterior pelvic tilt by 12 months. Increased anterior pelvic tilt would result in greater hip flexion angles observed during quiet standing and relatively greater peak flexion angles during the functional activities. In order to verify this hypothesis, kinematic outcomes of pelvic alignment would be required.

It has been mentioned that there was an improvement of peak extension angles with time for the operated limbs, but this was not the case for the control limb of the standard THR group (figure 7.3). As mentioned, of all the ambulatory tasks, level walking challenged extension ROM the most, and repeated measures of peak extension during level gait for the control limb of the standard THR group shows a 10° reduction of the extension angle by 12 months. The flexion angles of both limbs showed a corresponding increase, and although this may have been multifactorial, it seems reasonable to conclude that an increased anterior pelvic tilt was partly responsible, possibly caused by a deterioration in function of the non-operated limb, inducing an increased flexion contracture of the non-operated limb. Lee et al (1997) assessed pelvic tilt and found a strong positive correlation between anterior tilt and limited extension motion during walking. Moreover, given that the resurfacing group showed no deterioration of extension angles of the control limb and showed no corresponding increase in peak flexion angles bilaterally, it seems reasonable to conclude that not only do the lack of extension and increase in flexion appear to be correlated with probable flexion contractures, the resurfacing group appear to have maintained a condition with a unilateral nature. These findings reinforce the significant dynamic implications of hip flexion contractures. Further to this, given the more bilateral nature of hip dysfunction suggested within the standard THR group, their results may have an additional level of complexity which demands care in interpreting the results and could highlight dissimilarities between the groups at 12 months. These arguments do, however, reinforce the importance of using subjects with unilateral THR while still performing bilateral assessment, and highlights the significance of the interaction between limbs, i.e. the kinematics of one limb affects the kinematics of the other (Loizeau et al 1995).

A possible reason for the greater peak flexion angles for the standard THR group, particularly for those produced during stair ascent and descent, may have been related to the differences in the mean height of the groups. The resurfacing group, as reported in chapter 6, were slightly taller than the standard THR group, hence, in relative terms, the standard THR population may have had to step higher than the resurfacing group members in order to clear the step. This would imply that the greater flexion angles were related to the gender and height characteristics of the sample group as opposed to the use of the standard THR. However, statistically, there was no positive correlation between peak flexion angles and height or gender, and hence this trend was associated with arthroplasty type, suggesting a better functional motion outcome for the standard THR group during stair negotiation.

9.1.2.2 Differences in coronal plane angles due to hip arthroplasty type

Contrary to the study hypothesis, the coronal plane angle differences between the arthroplasty groups show fewer differences than the sagittal or transverse plane angles.

There was a trend difference towards greater abduction angles for the resurfacing group during the swing phase of level walking at the 12 month assessment phase. As mentioned in section 9.1.1.2. the operated limb tended to be relatively more adducted compare to the control limb. This might suggest that the resurfacing hips showed greater similarities to the 'normal' hip than the standard THR hips did. However, it was thought that the 'normal' control hips' pattern of greater abduction angles was compensatory (section 9.1.1.2), so the reason for the greater abduction angle for the resurfacing group operated hips is unclear. Perhaps the resurfacing hips had also adopted some kind of mechanical compensatory pattern. The larger abduction angles during swing could suggest that the resurfacing hips used circumduction of the hip, although it is unclear why this would develop with time.

Alternatively, it may be that the abductor muscle strength by the late assessment phase was relatively greater for the resurfacing group and therefore the resurfacing hips achieved greater angles of abduction during swing. An important factor to note regarding the coronal plane differences was the significant increase in the abduction angles of the resurfacing group from 3 to 12 months, which was evident for the stair descent task as well as level walking. This point reinforces the findings that the resurfacing group showed greater changes between the assessment phases.

9.1.2.3 Differences in transverse plane angles due to hip arthroplasty type

The standard THR group tended to have greater angles of external rotation than the resurfacing group. Given that the control limb was significantly more externally rotated than the operated limb (section 9.1.1.2.), this finding would suggest that the hips replaced with a standard stemmed component resulted in better reconstruction of the transverse plane kinematics. It is possible that the standard stemmed THR prosthesis allowed the surgeon greater scope for altering anteversion of the hip joint, by way of compensating for the soft-tissue imbalance generated by detachment of the short external rotators. As the anatomic femur is maintained with hip resurfacing, the methods of altering hip joint anteversion are limited to the acetabular component alignment. This finding is contrary to the hypothesis, as it was thought that the resurfacing procedure would result in more accurate restoration of the subjects' normal anatomy, and hence, the resurfacing prosthesis was thought to better correct, or maintain the rotation axis of the femur. On the other hand, it is possible that although the bony anatomy of the femur following resurfacing best represents the subjects' normal bony geometry, the muscular alterations during surgery have allowed the normal alignment of the femur relative to the pelvis to change (section 9.1.1.2.) and the femur to adopt a more internally rotated position. While the muscular alterations are the same for both arthroplasty procedures, featuring detachment of the short external rotators, the introduction of the prosthetic femoral stem in the standard THR may provide compensation for the imbalance of active structures and limits the excessive internal rotation that results.

The hypothesis that the larger diameter bearing of the resurfacing prosthesis would result in greater functional ROM was not fulfilled. Overall, the kinematic data showed equivalent peak angles outcomes during level walking and stair climbing for both resurfacing and standard arthroplasty groups. It may be that the standard THR components provided a greater head-neck ratio than the resurfacing procedure, thus allowing greater ROM before prosthetic impingement (section 2.1.2.1). Although preservation of the anatomical neck of femur is thought to be beneficial. the anatomic neck diameter is substantially larger than that of the prosthetic neck (Charles et al 2004). Although larger diameter bearings provide greater oscillation angles before impingement contact (Yoshimine and Ginbayashi 2002). the head-neck ratio is also important in determining ROM (Sariali et al 2008). CoC stemmed THR may optimise the head-neck ratio sufficiently in order to accommodate for the relatively smaller diameter bearing used, thus resulting in the largely equivalent angle data observed.

9.1.3 Hip moment data

The current mean peak moment data was compared with existing normative data for gait and stair negotiation. The literature data was mostly reported in alternative units (Nm/kg) compared to the current analysis (N/kg) due to differences in the normalisation procedures employed. Correction of the current values (using a mean study group height of 1.69m) showed that the results were similar to the sagittal moments reported by Riener et al (2002) but lower than those reported by Eng and Winter (1995) for level gait. The mean peak abductor moment reported by Mont et al (2007) for their resurfacing arthroplasty group (0.78Nm/kg) was very similar to that of the current resurfacing group (and standard THR group) at 3 months (0.73Nm/kg) and less than that observed for the current resurfacing group at 12 months post-op (0.88Nm/kg).

As with the angle data, most existing normative data only presents sagittal plane moments during stair climbing tasks (Andriacchi et al 1980, McFadyen and Winter 1988, Nadeau et al 2003, Protopapadaki et al 2006, Riener et al 2002). For stair ascent, the current hip flexor moments were similar to those reported by Protopapadaki et al (2006) (0.76Nm/kg), and were greater than those observed by Riener et al (2002) and Nadeau et al (2003) (0.5Nm/kg and 0.53Nm/kg respectively). The current extensor moments were within the ranges reported within normative literature. For stair descent, the normative sagittal hip moments varied widely as reported in literature, from 0Nm/kg (no flexor moment, extensor moment only) to 0.52Nm/kg for flexor moments to 0.13Nm/kg to 0.55Nm/kg for extensor moments.

When corrected for units, the mean moment values for the current subject group for stair descent fell within these ranges at approximately 0.3Nm/kg and 0.39Nm/kg for flexor and extensor moments respectively. Hence, the current peak moment values for operated and control limbs were representative of those 'normal' values found in published literature, suggesting excellent outcomes and return to function following hip arthroplasty.

As with the angle data, the greatest differences shown by the hip moment outcomes were between the operated and control limbs as opposed to differences between the standard THR and resurfacing hips. The contrast of between-limb differences was greatest for the resurfacing group. In addition, and consistent with hip angle outcomes, the greatest contrasts were shown at 3 months, indicating a positive progression with time for both groups.

9.1.3.1 Differences due to hip arthroplasty

Previous studies examining the kinetics of unilateral hip arthroplasty have demonstrated that the operated hip tends to have reduced peak abductor, flexor and extensor moments compared to the non-operated hip (Perron et al 2000, Sliwinski et al 2004, Tanaka 1993). In contrast, the current research has not fully supported these findings. Indeed the flexor and extensor moments were significantly greater on the non-operated control hip, but the abductor moments were not significantly different between limbs.

The early results alone did show that the operated hip had sub-optimal abductor moments during level gait and stair ascent gait, implying that there was evidence of abductor muscle dysfunction at 3 months, which improved by 12 months, so that the abductor moments were equivalent between limbs, regardless of the presence of a hip arthroplasty. The improvement by the late assessment phase was attributed primarily to soft tissue healing, which was likely to have been complete by approximately 6 months following surgery. The previous studies have shown sub-optimal hip abductor moment peaks at late assessment reviews (6 to 18 months post-operation), but the data presented was from an older, less active subject population, aged 50 to

75 years (Perron et al 2000). The mean age of this population is unknown and the mean assessment phase was not documented. Nevertheless, the current study of young active subjects carries a novelty and suggests important differences in kinetic outcomes of the elderly and those under the age of 60 years. It would appear that the current younger population shows greater improvements in function with time, despite pathology and hip arthroplasty. This accelerated improvement may be due to the greater activity level or greater strength of the younger subjects or be due to other age related variables such as the greater capacity for tissue remodelling of the younger subjects to promote healing.

When considering the flexor moments of the operated and control limbs, the results demonstrated that there was no difference due to hip arthroplasty at 3 months but that a significant difference developed by 12 months. The flexor moment differences therefore followed the opposite temporal trend compared to the abductor moments. As the correlation analysis showed, the peak flexor moments did correlate positively with the gait velocity. Hence the increase in flexor moments from 3 to 12 months was directly related to the increase in walking speed. The velocity dependency of the hip flexor moment has also been documented in previous published works (Perron et al 2000, Riley et al 2001). The extensor moment peaks were consistently greater on the control hip and poorer for the operated hip. This result reflects the findings from published literature and was expected from this study. The result implies that as the ground reaction force vector passed posterior to the hip joint, the external moment tending to extend the hip joint was relatively greater on the non-operated hip, causing a more efficient balance of the sagittal plane forces acting on the hip joint. The extensor moments of the operated limb were expected to be reduced relative to the control limb due to the surgical trauma, particularly given the surgical approach.

Interestingly, the internal and external rotator moments were significantly reduced on the operated limb compared to the control limb, with some exceptions, as already discussed in section 9.1.1.2. This has been largely undocumented in previous studies of hip arthroplasty, which is possibly due to the 2-dimensional and 1-dimensional nature of many previous studies, so this finding is of particular interest.

9.1.3.2 Differences due to arthroplasty type

All the differences identified between the standard and resurfacing arthroplasty groups were detected during level walking. Stair negotiation, contrary to the original hypothesis, showed no differences in moment values between the arthroplasty groups.

The peak abductor and extensor moments of the standard THR deviated less from the control hip values than the resurfacing hips, which was contrary to the study hypotheses, which predicted that hip resurfacing would better reflect the biomechanics of the natural (non-operated) hip. The difference observed was thought to be due to the differences between the femoral offsets in the arthroplasty groups, which has direct influence on the length-tension relationship and activity of the gluteal muscle (section 2.1.2.2). Although resurfacing arthroplasty results in a more anatomic reconstruction of the femoral anatomy (Girard et al 2006, Mont et al 2007), and a statistically similar femoral offset post-operatively (Silva et al 2004), the standard THR was associated with a greater resultant femoral offset (McGrory et al 1995) which is associated with certain biomechanical advantages. McGrory et al (1995) documented the advantages of increased femoral offset on both kinematics and kinetics of the hip joint. In a study examining 86 THR joints, radiological assessment was correlated with physiological ROM measurements (single observer hand held goniometer) and isometric abductor muscle strength (isokinetic dynamometry in supported standing). Greater femoral offsets were found to correlate with a greater magnitude of the length of the abductor lever arm, and correspondingly, greater offsets correlated with greater abductor strength in the assessment of force production of the abductor muscles. In addition, femoral offset was positively correlated with the range of abduction motion.

The greater femoral offset provided by the stemmed THR components in the current study would have the effect of increasing the moment arm of the abductor and extensor muscles, which might explain the relatively greater abductor and extensor moments of the standard hips compared to the resurfacing hips for within-subject contrasts with the non-operated hips. The speculated differences in the femoral offsets of the operated hips between groups cannot be confirmed due to the absence of radiographic assessment within this study. However, literature supporting the offset differentials between standard and resurfacing arthroplasty are numerous and consistent in their findings (Girard et al 2006, Loughead et al 2005, Silva et al 2004), and support the theory proposed. Figure 9.1 demonstrates the differences in femoral offset between resurfacing and standard THR, from sample radiographs of Durom™ (a) and Trident™ (b) prosthesis.

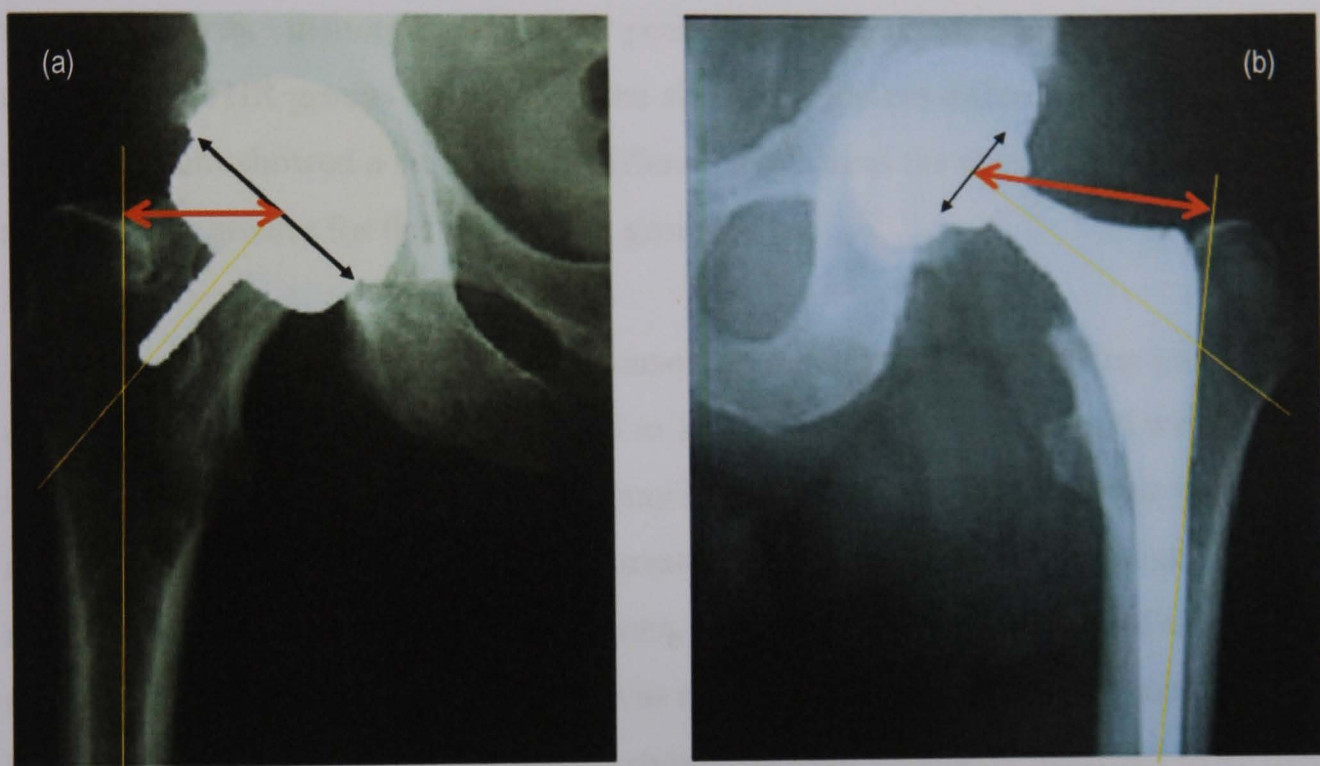


Figure 9.1: Femoral offset comparison between resurfacing (a) and standard THR (b)

The findings are consistent with Charnley's early observations of the importance of soft-tissue balance of the hip, where improvements in biomechanics were achieved by improving abductor muscle function, for which he recommended an increased femoral offset (Charnley 1979, cited in Charles et al 2004). As a result, Charnley predicted that there would be a reduced requirement for abductor muscle force and reduced JRF during gait. Applying these comments to the current data may suggest that the reduced abductor and extensor moments of the resurfacing group could result in a greater prosthetic hip JRF and risk of early wear (Sakalkale et al, cited in Piriou et al 2007) compared to the standard THR. However, given the complex nature of tribology variables (section 2.3.3) and the differences in materials used for the resurfacing (MoM) and standard THR (CoC) bearing components, the relative contribution of potential JRF differences to wear resulting from such functional tasks is largely unknown.

Aside from relative group differences based on within-subject comparison to the control limb, directly comparing the values of the early data for abductor and extensor moments shows that the peak values were equivalent for both groups or slightly poorer for the resurfacing group (extensor moments). By 12 months, the resurfacing group demonstrated greater peak abductor and extensor moments than the standard THR group. Statistically the abductor moment difference was not significant but showed a strong trend difference; whereas the extensor moment was significantly greater for the resurfacing group.

The larger increase in abductor and extensor moments for the resurfacing group compared to the standard group, from 3 to 12 months, may be associated with soft tissue healing. The resurfacing group may have displayed a greater deficit in peak moment values at 3 months due to the greater extent of soft tissue, and in particular, muscular damage from the surgical trauma. More extensive dissection of the gluteus medius and gluteus maximus muscles was required for the resurfacing procedure. For example, most fibres of the gluteus maximus were surgically detached from their insertion into the iliotibial band and the gluteal tuberosity in order to adequately expose the hip joint to perform the hip resurfacing, which would certainly explain the poorer extensor muscle function at 3 months. It may be reasonable to conclude that the potentially greater biomechanical advantages of hip reconstruction with a prosthetic femoral stem compensated for the existing muscle dysfunction from the standard procedure, as the standard operated hips had early outcomes equivalent to those of the control hip and showed minimal improvement in peak abductor and extensor moments with time. Hence, although the introduction of a prosthetic femoral stem changes the offset and the natural biomechanics of the hip joint (Girard et al 2006), this may be advantageous in compensating for muscle weakness secondary to the hip pathology as well as the surgical trauma related to the orthopaedic procedure.

Following healing of the surgical wound and recovery of tensile muscle strength by 12 months, the resurfacing hips appeared to surpass the function of the standard hips in terms of the peak extensor moment outcomes. However, correlation analysis of walking velocity with the moment outcomes revealed that the greater extensor moments of the resurfacing group at 12 months were in fact related to the faster walking speed of this group. The abductor moment values for the resurfacing group, which were close to being significantly greater than those of the standard THR group, were not correlated with walking speed, so it may be reasonable to suggest that the more clinically significant finding between arthroplasty groups was the trend of superior abductor moments for the resurfacing group at the late assessment phase.

In a study of 'normal' individuals up to 40 years old, Riley et al (2001) found that gait velocity correlated positively with extensor moments of the hip, but did not correlate with abductor moment outcomes. Their findings are reflected in the current results. Tanaka (1998) found that walking velocity reduced in individuals with hip extension limitation. As extension range was limited more so in the standard THR group, at the late assessment stage (section 9.1.2.1), this may have been a contributory reason for the slower walking velocity and reduced extensor moment peaks in this group. Concluding that the resurfacing group had superior abductor moments during level gait would appear to be contradictory following the earlier argument which related the probable increased femoral offset of the standard THR group with greater abductor moments. Although this argument concerning the greater femoral offsets may be substantiated, it may be that the greater activity levels of the resurfacing group could be overriding this mechanical advantage shared by the standard THR group. Both the greater activity levels and bias towards greater numbers of males in the resurfacing group could have amounted to an increased strength and muscle control of the resurfacing group. McGrory et al (1995) concluded that the length of the abductor arm and gender were of the most important factors influencing abductor muscle strength after hip arthroplasty, with males having significantly greater abductor strength. Considering that there was a gender bias in the current study, with proportionally more males in the resurfacing group, this might also account for the greater peak abductor moments associated with hip resurfacing compared to standard THR.

Repeated measures assessment of level walking also showed that the resurfacing group had significantly greater peak adductor and flexor moments than the standard THR group. As with the extensor moment at 12 months, these moment outcomes were correlated with walking speed. Specifically, there was a strong positive correlation between gait speed and flexor moments at 3 months and with the adductor moments at both test sessions. Therefore the significance of these findings, which would appear to merit the resurfacing group with superior function, have limited clinical significance. However, the flexor moments did not correlate with gait speed at 12 months, so the result of greater flexor moments of the resurfacing group at 12 months compared to the standard THR group would appear to remain sound. The possible reason for the greater flexor moments demonstrated by the resurfacing group may have been related to differences is stride length between the groups. If the stride length was greater for the resurfacing group this may have caused a greater flexor moment peak at the initial stance phase, following foot contact.

9.1.4 Temporal aspects of the motion tasks

One of the main significant findings observed from this study was the significantly greater walking velocity of the resurfacing arthroplasty group. The mean gait velocity of the current resurfacing population at 3 months (1.27m/s) was similar to that observed by Mont et al (2007) for their resurfacing group at a mean follow-up time of 6 to 15 months post-op (1.26m/s). In contrast, the current 12 month post-op data showed a substantially greater mean velocity of 1.44m/s. This observation shows the value of assessing subjects at defined post-operative assessment intervals and the improvements of gait velocity with time. Comparison of the standard THR group between the current study and that of Mont et al (2007) showed substantial differences. Mont et al found their standard THR group (0.96m/s) to be comparable with the OA population for gait velocity (0.94m/s) whereas the current group walked at 1.07m/s at 3 months and 1.16m/s at 12 months, which became close to that observed for a younger 'normal' population (18-40 years, 1.19m/s) by Riley et al (2001).

Attempts to control walking speed were not employed in the current analysis, as this was thought to impose restrictions on the motion tasks and limit or misconstrue the information gathered. Since the gait velocity was not controlled between subjects and the resultant gait velocity varied significantly between groups, it was questioned whether the greater moment values of the resurfacing arthroplasty group were truly related to arthroplasty type or whether they were greater due to the faster walking speed of the hip resurfacing subjects (section 7.5.1.2). In the latter case, associating the resurfacing group with optimal function due to their greater moment outcomes (section 7.2.3.4) would be a type I error in the interpretation of the data. Statistical analysis showed that this was the case and hence the greater moment values for the resurfacing group could not be related to optimal function provided by the resurfacing arthroplasty, as discussed in section 9.1.3.1.

9.1.5 Comparison of ambulation modes

Specifically in relation to joint pathology and joint arthroplasty, Hunter and Felson (2006) documented that a mobility disability is defined by an individual "needing help walking or climbing stairs" (page 235). Hence the use of gait and stair negotiation for the analysis of motion in the current study is highly relevant to the practical and clinical aspects of mobility assessment.

Stair negotiation was a more complex and demanding task compared to level walking. The stance phase, like level walking required control of the pelvis in the coronal plane, but unlike level walking, required additional dynamic control of the pelvis as the foot of the contralateral limb was lifted up or lowered to the next step. This required greater concentric activity during ascent and eccentric activity during descent for the abductor muscle group. However, stair negotiation showed no differences in moment outcomes between the two groups so did not induce differences in the outcomes assessed.

Although the stair negotiation task did not show any significant differences between the types of prosthetic hips, there were trend differences which have interesting implications. The late assessment phase showed that the resurfacing hips had greater abductor moments during stair ascent but the standard THR group had greater abductor moments during stair descent. These differences were not significant due to the high between-subject variability but it could suggest a subtle difference in the muscle activity between the groups, which may have been ascertained with a larger and more homogeneous population. Given that ascent and descent require more concentric and eccentric muscle control respectively, the ascent findings could imply that the resurfacing group had better concentric muscle activity of the abductors; whereas the descent findings could imply that the standard THR group had greater eccentric abductor muscle control than the resurfacing group.

Overall, level walking analysis showed greater differences between the arthroplasty groups. This conflicted with the expected outcome, where stair negotiation was predicted to highlight most differences due to the greater expected biomechanical demand imposed. Given that level gait analysis provided an effective comparative model, the use of gait analysis may be recommended and is a more practical and simple tool with which to perform objective comparisons of human ambulation.

9.2 Proprioception Findings

The argument as to whether hip joint proprioception is altered following THR not only aims to establish the effect of anatomical reconstruction of joint propriocepsis, but also seeks to classify which structures may be responsible for proprioception of the synovial joint. The THR procedure maintains the cutaneous and musculo-tendinous soft tissue structures, albeit disrupted temporarily to some extent by the surgical trauma. The capsular and ligamentous structures, on the other hand, undergo greater structural changes, with circumferential detachment of the capsular ligaments and the capsule. Hence, given the soft tissue alterations due to THR, it would be logical to assume that there would be differences between the proprioceptive function of an arthroplasty joint and the 'normal' anatomic hip joint. In addition to the soft tissue alterations, the bony modifications and the introduction of foreign materials with dissimilar stress and strain properties might have altered the load receptor function of the joint (section 2.8.1).

Traditionally it was thought that the receptors governing proprioception of a joint were located within the joint (Jerosch and Prymka 1996, Grigg et al 1973, Karanjia and Ferguson 1983); passive soft-tissue structures such as the capsule or the ligaments surrounding the joint. More recent literature has dismissed these theories in favour of evidence suggesting that extra-capsular structures are responsible for joint proprioception, such as muscles. Hence a study of hip joint proprioception in the unilateral THR population presented an appropriate model to test which structures might be responsible for joint proprioception. If proprioception of the arthroplasty hip was similar to proprioception of the non-operated hip, then it was unlikely that capsular mechanisms were responsible for joint proprioception due to the significant disruption of these structures following capsulotomy. If proprioception differed between the operated and non-operated hip, then the difference could be attributed to the main structural variable between the hips, which was the presence or absence of intra-capsular structures, in which case, it would be reasonable to conclude that these structures were responsible for proprioception of the joint.

When describing the proprioception results, there will be conclusions regarding the influence of THR on proprioceptive function of the hip joint, as determined by motion detection sense testing. In addition to this, the differences in MDS due to the type of arthroplasty prosthesis will be discussed. Finally, the findings as a whole will be discussed in the context of the effectiveness of the MDS procedure used.

9.2.1 Influence of hip arthroplasty on MDS of the joint

The current results were similar to those of previous threshold MDS studies. Mean values obtained for the operated and control limbs fell within the corresponding ranges found by Grigg et al (1973). The slightly lower mean threshold angles observed in the current study may have been due to the difference in the test protocol or the relatively younger age of the current study group (section 2.8.2.3; chapter 8).

The current hip joint proprioception results showed that there were no significant differences in the threshold MDS between the operated and non-operated control limbs. This implied that the hip motion detection sense was similar bilaterally and there was no difference in proprioception of the hip joint due to replacement arthroplasty, as measured by motion detection sense.

The mean threshold detection angles of the operated hip did show a trend towards greater values than the control hip, but the values of both limbs were subject to great variability. This was consistent for both 3 and 12 month results for the standard THR group and for 3 month results for the resurfacing group. The trends would imply that on average, a slightly greater magnitude of angular motion was required before the operated hips detected the motion. Hence there may have been some delay in detection of motion due to the presence of the hip arthroplasty, but the inconsistent and variable nature of the data negated the statistical significance of this trend.

Given that proprioception, as measured by MDS, was not altered due to hip arthroplasty, it can be concluded that extracapsular structures are responsible for MDS of the hip joint, such as receptors within the surrounding muscle and tendons. Intracapsular receptors located within the joint capsule or the capsular ligaments of the hip were unlikely to have been the prime structures governing MDS function as these structures were detached and therefore could not generate tension within their soft tissues for the receptors to function. The current findings therefore agree with those of Grigg et al (1973) and Karanjia and Ferguson (1983), and support the evidence that extracapsular structures appear to provide kinaesthetic sensibility of the hip joint.

The observation that there was no difference in MDS results between the operated and non-operated hip at 3 or 12 months implies that potential healing of capsule and ligament occurring between the assessment phases had no significant effect on altering proprioception of the hip joint, as there was no significant difference in the MDS threshold values between the 3 and 12 month assessment phases. This finding reinforces the conclusion that the hip joint capsule and capsular ligaments do not appear to contribute to MDS.

It was observed that there was greater variability of MDS of the operated hip. Therefore, the non-operated hip demonstrated slightly greater consistency in the detection of motion over repeated measures and between subjects, which could imply that MDS was more variable following THR. A reason for the greater variability is possibly that there was a process of motor relearning occurring at the operated hip in response to the altered sensory feedback. Altered proprioceptive feedback following the orthopaedic surgery may have been related to the anatomical changes to the hip joint and the resultant biomechanical status. For example, change in the femoral offset could have altered the length-tension relationship of the surrounding muscles, which may in turn, have altered the response of stretch receptors and golgi tendon organs within the muscle. Greater tension in the muscle due to a greater resting length may have lead to early firing of receptors and vice versa. Alternatively, the presence of pain around the joint may have dulled the response of the appropriate receptors responsible for MDS. Similarly, swelling around the joint might have had a similar effect. Intermittently, the subject may have experienced overriding fear avoidance and had difficulty allowing their limb to be moved passively by the test equipment. Greater active muscle tension in response to this would possibly alter their ability to detect the passive motion stimulus given to the limb. It is possible that any one or any combination of these factors may have lead to greater variability of the subjects MDS over repeated trials of motion stimuli.

Following on from these points, arguing the reasons for the greater variability of MDS on the operated limb, it would be reasonable to predict that the variability of measures would reduce with time as motor relearning improved and neural processes adapted to the presence of the hip prosthesis. Indeed, the within- and between-subject variation did reduce with time for all trials for the standard THR group, showing that there was greater consistency in the responses of the subjects and post-operative healing increased. In contrast, the resurfacing group did not share this expected response. Apart from the flexion trial of the operated limb, which followed the response of the standard THR group, the resurfacing group showed an increase in the between-subject variability at 12 months. This would imply a greater range of the mean MDS responses of the resurfacing population at the late post-operative stage. However, this was also true for the control hip in the resurfacing group, hence variation between subjects of this arthroplasty group may have occurred due to the wider spread of activity variables observed for this group (chapter 6).

Finally, when comparing the motion directions, it was observed that the similarities in MDS for the operated and control hips were consistent for both physiological motion trials (abduction and flexion). This finding was consistent for all subjects, across both arthroplasty groups. Hence, regardless of the motion plane, the amount of motion required at the hip joint in order to detect the motion was similar. This is a novel finding as no known study has compared both sagittal and coronal movement planes to determine whether there is a direction specific element to MDS. By same subject comparison with the same controlled protocol, the current findings indicate that MDS is consistent despite the direction of the motion stimuli.

9.2.2 Differences in MDS due to arthroplasty type

The results showed that the standard THR group tended to have slightly greater threshold MDS angles than the resurfacing group, implying that there was greater delay in detecting motion following standard hip arthroplasty, and that resurfacing arthroplasty resulted in superior hip joint MDS. The standard THR hips therefore required a greater magnitude of sensory stimulation than the resurfacing group before gaining awareness of the stimulus. Nevertheless these differences were not significant between groups, so it cannot be concluded that the resurfacing hips had superior proprioceptive function. Despite this, the reasons for the small but consistent group differences were considered. Although the greater threshold MDS angles of the standard THR group were not above the threshold of statistical significance, the difference due to arthroplasty type may have been clinically significant. A possible reason for the relative delay in MDS following standard THR may have been due to the relatively poorer activity levels exhibited by the standard THR group. As mentioned in chapter 2 (section 2.8.2.1), activity variables have an influence on joint proprioception, with greater proprioceptive acuity associated with individuals who are more active. Subjective reporting of activity of the current subject population (chapter 6) showed that there was a significant difference between the arthroplasty groups, with the resurfacing subjects showing higher activity levels, including participation in sport activities. It would therefore be reasonable to assume that the slightly sharper motion detection abilities of the resurfacing group may have been derived from their more sophisticated proprioception resulting from their more active lifestyle.

This theory was reinforced by the observation that the same between-limb trend was present for the non-operated hip. Assessment of the control limbs of the resurfacing and standard THR groups showed the same trend of greater threshold MDS angles for the control hips of the standard THR group relative to the resurfacing group. If the activity variables were responsible for the slight between-group difference in MDS angles, then it would be expected that this would be observed bilaterally within-subjects. Given that there was consistency between the findings of the operated and control limbs, it would be appropriate to suggest that activity variables were likely to be responsible for the slight proprioceptive difference between groups, rather than the type of arthroplasty prosthesis used.

Interestingly, this finding of reduced threshold MDS of the resurfacing group was consistent at both 3 and 12 month post-op assessments, and showed the same differential of MDS angles for both abduction (standard THR group delayed by 0.27°) and flexion (standard THR group delayed by 0.21°) motion trials at both early and late stages. This would suggest that the difference due to arthroplasty type was consistent with time. It also confirms that the MDS of the operated hips did not alter in response to post-operative healing (section 9.2.1). Since muscle mechanisms are thought to be responsible for MDS, it was expected that improvements would occur with post-operative healing. This could then imply that despite the trauma to the muscles due to the surgical incision, the stretch receptors had maintained their capability to detect motion. This appears logical given that the stretch receptors of the incised muscles were aligned with the fibres of the muscle and parallel with the line of the incision, and would therefore have been subjected to minimal disruption.

Interestingly, the resurfacing group showed some change in MDS with time, but this was relevant to the control hip. The control limb of the resurfacing group at 12 months featured a greater delay in the detection of motion than the operated hip. This difference was not significantly different due to the vast variability of the mean measurement, but shows an interesting interruption to the typical pattern otherwise observed. The observations may suggest that there was a deterioration in MDS of the control limb with time, which may have been due to a number of factors such as an onset or deterioration in joint pathology or pain in the non-operated joint.

The possible deterioration of the non-operated limb in the resurfacing group conflicts with the motion analysis findings. The motion analysis findings suggest that if there was deterioration in the functional outcomes of the non-operated hips, then that was more likely to be represented by the standard THR group, as they displayed an onset of limitation in the extension angles for example. Hence there seems to be conflict in the findings of the two objective assessment protocols. Nevertheless, the current argument relates to a small, statistically insignificant point, and only limited conclusions should be made regarding this.

9.3 Relating Findings to Subjective & Descriptive Data

9.3.1 Use of subjective outcome measures to predict function

The OHS seemed to have a ceiling effect and was best applied to quantifying the difference between the pre- and post-operative outcomes. The scale did not seem to have enough range to quantify changes in post-operative intervals and as a result the subjects appeared to be fairly generic in their report of function. In contrast, the conversational aspects of function between subjects seemed very different to the researcher, so it would appear that the OHS was not sensitive enough to identify post-operative variations in function between the subjects.

As mentioned in section 9.1, the extension limitation may be related to the limp reported by subjects within the OHS. This item of the questionnaire may be very important. However, often subjects completing the questionnaire tended to report that they did not personally notice the limp but their family and friends would point it out. Similarly, patients would feel that their limp had improved very quickly following THR and be unaware of their limp at the early test session; however, the limp was still observed by the researcher and the patient's kin at this stage. Therefore, although there may be links between the objective findings and these subjective reports, the subjective data may often be unreliable and skewed by other subjective reports rather than just that of the patient. Objective data collection is therefore considered invaluable.

Of all the cases where there was progression of outcomes with time, the possible reasons are listed below. Progression may have been due to a single reason or a combination of the reasons.

- Reduced pain and pain behaviour
- Reduced fear avoidance
- Reduced compensation strategies
- Increased confidence
- Improved motor relearning
- Use of more normal motion patterns
- Increased frequency of more normal motion tasks

9.3.2 Variability in activity levels of subject group

In many aspects, the resurfacing group tended to show relatively greater improvements with time, which may have been related to the greater activity levels used by this group, as discussed in chapter 6 and measured retrospectively using the UCLA activity scale. The greater activity of individuals with hip resurfacing has also been reported by Amstutz et al (1984) post-operatively and by Gore et al (1985) both pre- and post-operatively. As pre-operative measures of activity and function were not recorded in the current study, it is difficult to predict with certainty whether the greater levels of activity and better outcomes of the resurfacing group were due to their pre-existing high activity lifestyle or due to the influence of their hip resurfacing arthroplasty. Appraising all the evidence, it would be fair to conclude that the resurfacing group were typically more active and that the greater activity levels observed during the study were not directly related to their hip resurfacing.

9.3.3 Influence of psychological factors

Psychological factors appear to have had minimal effect and no group differences were recorded using the satisfaction questionnaire. Anecdotally, subjects occasionally reported that they felt there was limited education of the expected progression required once discharged from the hospital.

9.4 Merits and Limitations of the Motion Analysis Study

9.4.1 Merits of the motion analysis protocol

The motion analysis protocol included the use of sophisticated and validated equipment to collect highly objective motion data, from which the hip moment and angle data was derived. The use of the data collection to calculate 3-dimensional characteristics of motion was very useful, particularly as there is a large gap in literature regarding the transverse plane analysis. Since the reconstruction of the hip causes some degree of manipulation of the rotation features of the hip, this 3-dimensional analysis was particularly relevant and useful in the current study. The results highlighted that the analysis of angles in the transverse plane was of particular value (section 9.1.1.1). This emphasises that the standard to current motion analysis studies should maximise their methodologies to include full 3-D analysis.

9.4.2 Limitations of the motion analysis protocol

A potential limitation of the motion analysis protocol was the use of the predictive method in locating the hip joint centre (HJC). Although the functional approach is classified as more subject-specific, there were difficulties in the formulation of the functional hip joint centre location algorithm. Pendular motion of the thigh segment was recorded during kinematic data capture, but there was a wide range of maximum oscillation angles achieved by the current subject group, which were generally small in magnitude compared to those required for functional HJC estimation (section 2.5). Several subjects could not complete the limb oscillation task due to severe difficulties achieving active motion of the test limb or prolonged standing on the support limb. As the errors in determining the functional HJC position increase as the performed ROM decreases (Seidel et al 1995), the functional method may have produced large errors and a wide variation in errors between individuals or over repeated measures at 3 and 12 month test intervals. Moreover, the predictive method used has been shown to produce lower errors than that of the functional method (table 2.2) providing greater accuracy of predicting the true HJC position (Bell et al 1990).

9.5 Merits and Limitations of the Proprioception Study

9.5.1 Merits of the MDS protocol

The MDS protocol was valid and effectively measured meaningful threshold angles, which were similar to the values reported in previous literature (Grigg et al 1973, Karanjia and Ferguson 1983). The reliability of the results with repeat testing has also been supported using subjects without hip arthroplasty. Also, the consistency of the findings between 3 and 12 months highlights the sound repeatability of the MDS protocol with the current subject group.

The MDS protocol had several merits in contrast to the pervious protocols of Grigg et al (1973) and Karanjia and Ferguson (1983). Firstly, the measurement methods of the current study were more accurate, with the angular data being collected from an electrogoniometer as opposed to a hand held goniometer or a potentiometer with an analogue numerical indicator scale (Grigg et al 1973, Karanjia and Ferguson 1983), and measured directly from the hip joint as opposed to the centre of rotation of the beam (Grigg et al 1973). These aspects improved the validity and the reliability of the findings, by eliminating errors in reading the start and detection angles and improving the accuracy of the angular measurement. Secondly, the use of a silent running mechanical rig allowed there to be efficient, smooth movement of the limb at a controllable velocity, while still minimising the additional sensory stimuli. Thirdly, the current protocol, in some respects, merged the two previous experimental works by combining the motion directions tested and testing both of them under the one consistent protocol. This allowed the difference between two planar motions to be examined for the effects on motion detection sense.

9.5.2 Limitations of the MDS protocol

The main limitation of the protocol was the usability of the rig. Skill was required in setting up the subject within the rig to the accurate alignment the limb while maintaining their comfort. Nevertheless, the rig had to have substantial adaptability to allow testing of subjects with all different body shapes, so had to feature all the fixtures and attachments that it did. However, this did mean that the protocol would be difficult to apply in a clinical setting with multiple users, due to the time and skill required in accurately collecting MDS data in a repeatable nature. Specialist design adaptations would be required before this could be done.

There was some level of subjectivity that may have skewed the data collection between subjects and possibly added to the between-subject variability. Perhaps in following the standardised instruction to the test (appendix 7), some subjects adhered to these explicitly, and for some, as the test conditions progressed, they became either less particular or became sleepy, and the physical reaction of pressing the switch became a little sluggish. It was believed that these cases were in the minority and that they had minimal effect of the outcomes. The effect of such errors was also limited by randomising the order to the test conditions between subjects and between test sessions, and by the researcher reiterating the instructions between conditions to emphasise adherence to the instructions and making the subject more alert before continuation of the next test.

Another aspect of the subjectivity of the test which was beyond the researchers control was the variation in the reaction time of the subjects. A slower reaction time would add extra time before pressing the switch and the resultant measure of the threshold angle would appear to be larger than an individual with a fast reaction time. Moreover, the use of a lever switch as opposed to a push-button switch was disadvantageous and perhaps added to a small delay in the recording of the detection and possibly some additional between-subject variability. Some subjects had intermittent difficulty in 'flicking' the switch perhaps due to a momentary lapse in remembering which position the lever was in. In such cases, the researcher discarded these trials. Variation between subjects regarding their dexterity and coordination may have lead to small differences in the time taken to operate the switch, and hence contributed to further variation between detection times of the subjects in addition to the reaction time variable. Such patterns were not observed within the data and are therefore believed to be minimal.

In order to account for the reaction time variable, perhaps it would have been appropriate to formulate an accurate and relevant test of reaction time to a sensory stimulus. If the subjects had performed that test on the same day as MDS testing, then the MDS threshold angles could have been analysed as an angular value normalised to reaction time. This proposed modification to the protocol would have to be piloted and validated, as would the reaction time test, but may lead to a more rigorous adaptation of the current methodology to account for, or test the significance of the reaction time variable on the threshold angle outcome.

9.5.3 Effectiveness of MDS as an outcome measure

The researcher has highlighted in chapter 2 that the outcome of MDS represents one facet of proprioception of a joint and that joint proprioception can also be measured by joint reposition sense. The MDS protocol was favoured due to the literature support (section 2.8.3) and as it was credited with less subjectivity compared to the joint reposition tests. Nevertheless, having tested a wide range of subjects with the current MDS protocol, it is fair to conclude that the MDS protocol was also exposed to some level of subjectivity, as described in section 9.5.2.

Bearing in mind that muscular mechanisms were believed to be responsible for MDS, and that if this were the case, the stretch receptors of the muscles were the structures facilitating the proprioceptive feedback, the named muscles triggering the detection of motion can be predicted. During the abduction motion, the abductor muscles were shortening and the adductors lengthening. During the flexion motion the flexor muscle group was shortening and the extensor group lengthening. Hence with the motions concerned, the antagonist was the muscle that was likely to have been responsible for the detection of motion through the stretch of the muscle, be it the adductor or extensor groups. The implication of this is that the previous argument concerning the receptors of the abductor muscles, and the effect of the surgical incision on these receptors, would therefore be void, as the adductors would be more likely to determine the motion detection sense through the activation of the stretch receptors. Nevertheless, studies of the knee joint have failed to demonstrate direction specific differences in threshold MDS (Hall et al 1995), with similar results for flexion and extension. The reason for studying flexion and abduction motion was primarily due to the inclusion of these measures in previous published literature. In addition to this, it was thought that these were two of the prime muscle groups involved in functional motion and control of the hip joint. In light of the former discussion of the muscles thought to be triggering MDS, it might be a more appropriate suggestion to test adduction motion and extension motion using the current protocol. Testing adduction is proposed as this motion would stretch the abductors and activate the stretch receptors within this muscle, thus presenting a more challenging test for the surgically impaired abductor muscle group. Testing extension would stretch the flexors, and as previously mentioned, the flexors may be tight, and hence this too might be a more challenging use of the MDS protocol which would highlight difference between subjects.

9.6 Implications of the Study

The findings of this study, identifying the differences in functional outcomes between the standard THR and the hip resurfacing procedures show that there were few clinically significant differences and that the hip arthroplasty types are functionally equivalent. The main differences between subjects with a standard and resurfacing hip arthroplasty were their activity characteristics. The resurfacing group showed evidence that they were more active than the standard THR group. Perhaps the results imply that the groups were not functionally homogeneous on entering the study, which reinforces the notion that these different types of hip arthroplasty appear to be suited to slightly different patient groups and supports the evidence that there is difficulty randomising a patient population to these arthroplasty groups. Randomising subjects to a resurfacing arthroplasty group is considered ethically questionable due to the greater risks and restrictions of the procedure, and the importance in suitable patient selection criteria for the success of a resurfacing procedure has long been recognised (Amstutz and LeDuff 2006). If the resurfacing arthroplasty and standard THR are suitable for distinct patient groups, then the question may be raised as to whether resurfacing arthroplasty is a preliminary operation and a "conservative time-buying procedure" (Amstutz et al 1998, page 172), although this is generally refuted by surgeons, who argue that both are primary hip arthroplasty procedures in their own right. Nevertheless, the need for greater understanding of the differences between resurfacing and standard THR still stands (Vale et al 2002), despite the inability to perform a randomised control study. In response to this, the current research has provided a comprehensive record that minimal functional differences exist between these arthroplasty options.

The resurfacing arthroplasty is associated with more surgical and post-surgical risks. It is a highly specialised procedure that requires a large learning curve and great skill, but offers a potentially bone sparing arthroplasty and is supported by claims of several perceived benefits. The perceived benefits of resurfacing over the conventional stemmed THR with low wear bearings, relate to the more anatomic reconstruction of the hip which was thought to benefit hip joint biomechanics, the potential for greater ROM relating to the large diameter bearings, and the future considerations of greater ease of revision. However, the ease of the acetabular revision is questioned, given the lack of bone sparing methods required to accommodate a large diameter femoral component.

Moreover, the current study has shown that there are minimal biomechanical advantages relating to hip resurfacing, as measured by level gait and stair climbing activities. Aside from a trend towards greater flexor moments at 1 year, the optimal biomechanical parameters observed for the resurfacing group were related to the greater gait velocity adopted by this group. Hence the expectation of relatively superior kinetic outcomes following resurfacing was not supported in this study. It is thought that the greater offset provided by the prosthetic femoral stem may benefit hip biomechanics to an extent which demonstrates equivalent functional biomechanical outcomes between these arthroplasty options.

In addition, there was no significant difference in the peak hip angles between the arthroplasty groups. This implies that the large diameter bearings do not provide greater functional ROM as predicted from in vitro measures. It could be that the lower head-neck ratio of the typical resurfacing arthroplasty hip limits ROM due to early impingement. Conversely, the relatively greater head-neck ratio of the Trident™ system with Exeter stem may have provided more ROM before impingement, allowing a functional ROM equivalent to that of the Durom™ resurfacing.

The current results will allow orthopaedic surgeons to achieve a greater level of informed consent with their patients, to better advise the pre-operative planning and choice of implant selection. The evidence goes some way to balancing or dispelling the current information supporting hip resurfacing which is largely theoretical and derived from radiographic studies. Based on the current study, surgeons and associated health practitioners can advise patients that there are no apparent differences in functional and proprioceptive outcomes resulting from either arthroplasty type, and no apparent functional gains from the selection of one over the other. From an understanding of the functional equivalence of resurfacing and standard THR, orthopaedics surgeons may base their advice on patient-specific aspects such as the appropriateness of the bearing materials (metallurgy), bone stock and bone viability, the likelihood of revision and their perspective on revising resurfacing prostheses and so on.

Given the growing concerns about MoM bearings, with the emergence of ALVAL syndromes for example, the current evidence may encourage the clinical environment to pursue use to more biologically inert hard-on-hard bearings such as CoC. As CoC is currently only available with stemmed THR designs, this study may ease decision making towards CoC stemmed THR given that it provides functionally equivalent outcomes to resurfacing arthroplasty post-operatively.

Orthopaedic practice may use the current research to balance the theoretical advantages presented by industrial and commercial companies which advocate the use to their product. Dissemination of the findings from this study will promote a more clinical approach with a firm scientific basis, to direct the choice of the arthroplasty prosthesis, which will carry more strength than the marketing policies promoted by orthopaedic companies.

The main findings from the study will be summarised in Chapter 10, by way of concluding the content of this thesis. The current research has provided a valuable and topical perspective on hip arthroplasty for the young and active population, which appears to have significant implications for the ongoing selection process of hip arthroplasty and the expectation of functional outcomes achieved. Given these implications, the sections to follow will direct the key recommendations which may be derived from this work, and will close by outlining the possible recommendations for future work within the current subject area.

10 Conclusion and Further Study

10.1 Conclusions of the Study

The current study has investigated the differences in functional performance of patients having had hip resurfacing arthroplasty compared to those with conventional THR, by assessing biomechanical aspects of their level gait and stair climbing activities and by characterising joint proprioception as measured by joint motion detection sense.

The more anatomic reconstruction of the hip resulting from resurfacing arthroplasty was thought to result in improved abductor muscle function in particular, due to the maintenance of the natural femoral offset. In addition, the larger diameter head of the resurfacing prosthesis was thought to provide greater physiological and functional ROM.

The findings of the study have shown that there were no differences between individuals with resurfacing or standard THR for any of the parameters assessed. Therefore, the function of individuals following either arthroplasty may be classed as equivalent.

In terms of the motion tasks, the angle and moment data revealed no significant differences between the arthroplasty groups. It was observed that the resurfacing group tended to walk quicker and displayed more confident behaviour, particularly during the stair negotiation tasks. However it is thought that these characteristics were inherent to the group and were not related to the subjects having had a resurfacing prosthesis as opposed to a standard THR. Despite the faster gait and more confident motion, hip moment and angle parameters remained similar to those of the standard THR group, and hence, the hip resurfacing did not appear to result in greater functional ROM or improved abductor muscle function as hypothesised. Instead, the prosthetic neck of the standard THR may have increased the femoral offset and optimised the length tension relationship of the abductors, resulting in the equivalent moment outcomes observed in this study. Also, despite the larger diameter femoral head of the resurfacing prosthesis, the bony neck of the anatomic femur would have meant that the neck had a significantly greater diameter than the prosthetic necks used. Therefore, the resultant head-neck ratio following the hip resurfacing would have been lower than that of the standard THR, which may explain why the resurfacing hips did not exceed the angular motion of the standard THR hips.

The MDS findings illustrate that proprioception of the hip joint is not altered following hip arthroplasty and does not vary due to arthroplasty type. Hence, differences in proprioception between the groups cannot explain the more confident motion patterns and greater activity levels of the resurfacing group, and these are believed to be characteristics which were inherent to the group before the deterioration in function due to their hip pathology.

The main conclusions of the study are listed below:

- Subjects with hip resurfacing walked faster and were more active, but this was attributed to the characteristics of the subject group derived from the selection bias induced by the current non-randomised study design
- Activity levels featured as one of the main variables between individuals with stemmed THR and hip resurfacing, and warrants use of activity measurement or monitoring in future studies
- Hip resurfacing does not appear to provide greater functional hip ROM
- Preservation of the femoral neck with hip resurfacing does not appear to improve muscle function
- There was no difference in the MDS of subjects with standard stemmed THR or hip resurfacing and the greater level of activity of resurfacing subjects cannot be attributed to greater kinaesthetic sense
- Even with the current young subject group, residual deficits such as restricted extension ROM and asymmetrical movement and muscle function persist following hip arthroplasty

The recent increased use of resurfacing arthroplasty in orthopaedics has been favoured due to the potential for superior hip function and improved options for revision surgery. It may be too early to comment on the ease of converting a hip resurfacing to a standard THR, but potential difficulties in the revision of the acetabular component have been recognised. Additional problems with hip resurfacing include the difficulty of carrying out the procedure and the steep learning curve for training surgeons which is associated with a high number of failures before proficiency is achieved. In addition to the higher risk of early failure due to femoral neck fracture, MoM hip resurfacing is also associated with poorly understood phenomena such as femoral head bone resorption, ALVAL, metal sensitivity syndromes and cancer risk. With wider consideration to the numerous disadvantages and potential risks of MoM hip resurfacing, the findings of the current study may have significant implications, as the potential for greater functional performance provided by the resurfacing prosthesis has in fact, not been supported.

The current study has shown no biomechanical or proprioceptive evidence to support the use of hip resurfacing over standard THR. The functional performance following standard THR was equivalent to hip resurfacing, but the procedure carries less risks and is well established within orthopaedics. Given the risks and potential disadvantages associated with resurfacing arthroplasty, together with the evidence from the current study, there appears to be little functional advantage to having a resurfacing over a standard THR. It is therefore recommended that the standard THR, with low wear bearings such as CoC, is adopted as the preferred hip arthroplasty option for the 40-60 year old active population.

10.2 Recommendations for Further Study

More studies are required to look at the effects of extension ROM on gait. Future work should assess how flexor contractures affect the biomechanics of gait and categorise the extent of the biomechanical impairments related to the magnitude of the contracture. Recommendations would be to compare 'normal' subjects with those with limited extension ROM and positive Thomas' tests, to eliminate the issues associated with the current within-subject comparisons (section 9.1.1.1). Furthermore, randomised controlled trials should be undertaken to assess the effects of incorporating flexor muscle stretching in an early rehabilitation programme for patients following hip arthroplasty, by means of optimising extension ROM and maintaining the hip extension angles observed intra-operatively.

The proprioception protocol would benefit from several adaptations, as highlighted in section 9.5. To summarise, these could include the use of a push-button motion detection switch, the introduction of a normalisation procedure to account for the reaction time variable and possibly a repeat of the study using biplanar assessment of the operated limb, allowing testing of adduction and extension motion in addition to abduction and flexion trial.

To assist in more accurately assessing the characteristics of the sample group, a UCLA activity scale or another activity monitor should be used prospectively to quantify activity of the subjects. As the variability in activity between the arthroplasty groups was believed to result in a type II error in interpreting the motion analysis data outcomes, it is imperative to quantify activity so it can be related to the results and more accurate conclusions can be drawn with greater certainty.

Finally, the current study may have benefited from radiographic assessment of the hip joint. Firstly, radiographic data would have allowed the hip joint centre position to be accurately located with geometric reference to the ASIS to allow a patient-specific predictive algorithm in the BodyBuilder code. This would assist in minimising the errors associated with hip joint centre location and provide a more accurate understanding of the change of hip joint centre position due to hip arthroplasty. Secondly, reference to accurate radiographs would have been helpful in drawing conclusions regarding the motion analysis data. The hypothesis drawn from the angle and moment results and the association of these findings with the probable anatomical reconstruction of the hip could be verified. This, in turn, would allow more confident recommendations to be given to the orthopaedic clinicians.

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Appendix 1

Influence of Component Design and Placement

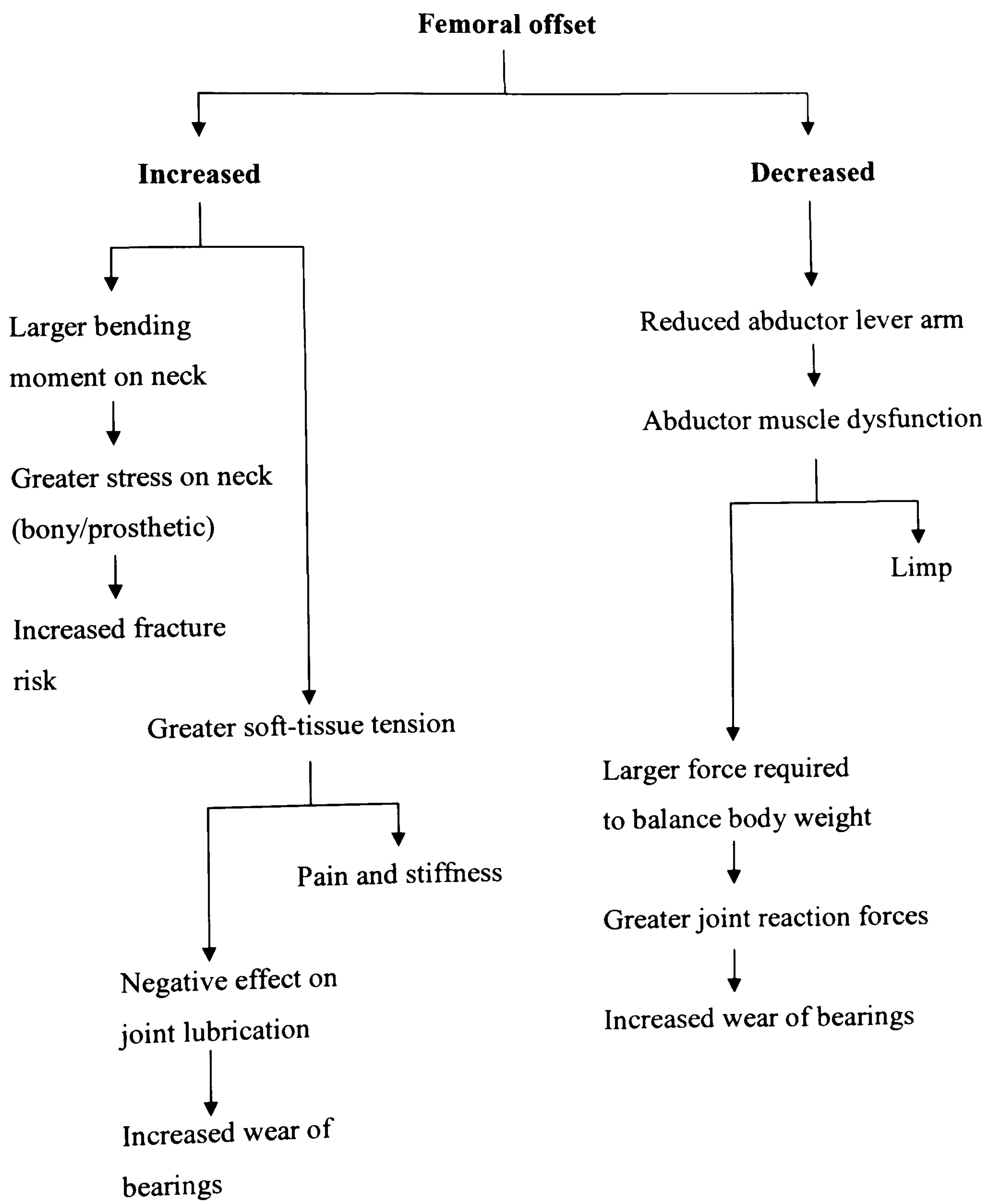
The variables concerned with the acetabular and femoral component design and placement are summarised in this appendix, with reference to the effects on function.

Aspects of Anatomical Reconstruction	Recommendations and Guidelines	Influence and Justification
Acetabular Inclination	**>40°** (Coventry, cited in D'Lima et al 2000) **45-55°** (D'Lima et al 2000) **30°** (Harris, cited in D'Lima et al 2000) **45°**; 60° causes dislocation without impingement (Kluess et al 2007) **35-45°** (Kummer et al 1999) **45-50°** (Sariali et al 2008)	Greater angles of acetabular abduction increase the flexion, extension and abduction ROM available before impingement, but reduce adduction and rotation (D'Lima et al 2000); Greater angles of acetabular abduction have also been found to reduce the extension ROM before impingement (Kluess et al 2007).
Acetabular Anteversion	**>15°** (Coventry, cited in D'Lima et al 2000) **20°** (Harris, cited in D'Lima et al 2000) **15-30°**; Significant impingement at >30 (Kluess et al 2007) **0-10°** (Kummer et al 1999) Place cup parallel with the transverse acetabular ligament (Murray, cited in Sariali et al 2008)	Greater angles of acetabular anteversion increase the flexion ROM but reduce the extension, abduction and external rotation ROM before impingement; Combining acetabular and femoral anteversion has an additive effect on increasing the flexion ROM (D'Lima et al 2000); Acetabular anteversion has more influence on ROM than stem anteversion (Kluess et al 2007).
Femoral Anteversion	Combined anteversion >40° and <60° (Jolles et al, cited in Yoshimine 2006) **0-10°** (D'Lima et al 2000)	Greater angles of acetabular anteversion increase the flexion ROM but reduce the extension, abduction and external rotation ROM before impingement; Combining acetabular and femoral anteversion has an additive effect on increasing the flexion ROM (D'Lima et al 2000).
Head-neck Ratio	Greater head-neck ratios recommended (D'Lima et al 2000, Kluess et al 2007, Kummer et al 1999, Yoshimine and Ginbayashi 2002)	Greater head-neck ratios increase the ROM before impingement; head size influences wear, joint stability and dislocation rates (D'Lima et al 2000).
Femoral Offset/ Neck-shaft Angle	Aim to restore or slightly increase natural offset (Charles et al 2004). See Appendix 2. Slightly reduced neck-shaft angle (varus) to enhance offset with stemmed THR; increased (valgus) with hip resurfacing (McGrory et al 1995, Girard et al 2006).	Restores and balances leg length, soft-tissue tension and resultant joint stability; optimised joint reaction forces and abductor muscle function; reduces wear; increases abduction ROM (Asayama et al 2005, Charles et al 2004, McGrory et al 1995); Neck-shaft angle influences offset: varus angles increase offset for the above desired effects; valgus angles with resurfacing reduce the risk of femoral neck fracture by enhancing compressive stresses and stability (Girard et al 2006).

Appendix 2

Modification of Femoral Offset

The adverse effects of manipulating the femoral offset are documented in this appendix.



Information sourced from Charles et al 2004, McGrory et al 1995, Sariali et al 2008

Appendix 3

Definition of Severity of Pain

Given the subjectivity of grading the severity of pain, existing pain of the contralateral, non-operated hip was graded according to numerical scoring of pain from visual analogue scale of 0-10 units, where:

Mild pain:	scored 1-4
Moderate pain:	scored >4-6
Severe pain:	scored >6-10

Such methods were originally validated for malignant pain and have been supported for hip OA subject groups.

Kapstad H., Hanestad B.R., Langeland N., Rustøen T., Stavem K. (2008): Cutpoints for mild, moderate and severe pain in patients with osteoarthritis of the hip or knee ready for joint replacement surgery. BMC musculoskeletal disorders http://journal.medscape.com/viewarticle/574958

Appendix 4

Oxford Hip Score

supported by stryker

Orthowave Hip Form Oxford Score (OHa1)

Last Name ______________________

First Name ______________________

Patient ID __________ Study ID __________

Reviewer ______________________

Patient's Bar Code ID

Data Entry on: ________ by: ________

Reviewed by: ☐ Operator ☐ Other MD ☐ CRA ☐ Phone ☐ Questionnaire

Side ☐ Right ☐ Left Follow-Up: ________ ☐ Mth ☐ Yr Charnley Type ☐ A ☐ B ☐ C

Date of Surgery (dd/mm/yy): ___/___/___ Date of Evaluation (dd/mm/yy): ___/___/___

During the past four weeks

1. How would you describe the pain you usually had from your hip?

☐ None ☐ Very Mild ☐ Mild ☐ Moderate ☐ Severe

2. Have you had any trouble washing and drying yourself (all over) because of your hip?

☐ No trouble at all ☐ Very little trouble ☐ Moderate trouble ☐ Extreme difficulty ☐ Impossible to do

3. Have you had any trouble getting in and out of a car or using public transport because of your hip?
(whichever you tend to use)

☐ No trouble at all ☐ Very little trouble ☐ Moderate trouble ☐ Extreme difficulty ☐ Impossible to do

4. Have you been able to put on a pair of socks, stockings, or tights?

☐ Yes, easily ☐ With little difficulty ☐ With moderate difficulty ☐ With extreme difficulty ☐ No, impossible

5. Could you do the household shopping on your own?

☐ Yes, easily ☐ With little difficulty ☐ With moderate difficulty ☐ With extreme difficulty ☐ No, impossible

6. For how long have you been able to walk before the pain from your hip became severe? (with or without a stick)

☐ No pain/more than 30 mins ☐ 16 to 30 mins ☐ 5 to 15 mins

☐ Around the house only ☐ Not at all

7. Have you been able to climb a flight of stairs?

☐ Yes, easily ☐ With little difficulty ☐ With moderate difficulty ☐ With extreme difficulty ☐ No, impossible

8. After a meal (sat at a table), how painful has it been for you to stand up from a chair because of your hip?

☐ Not at all painful ☐ Slightly painful ☐ Moderately painful ☐ Very painful ☐ Unbearable

9. Have you been limping when walking because of your hip?

☐ Rarely/Never ☐ Sometimes or just at first ☐ Often, not just at first ☐ Most of the time

☐ All of the time ☐ None of the time

10. Have you had any sudden, severe pain, shooting or spasm, from the affected hip?

☐ No days ☐ Only 1 or 2 days ☐ Some days ☐ Most days ☐ Every day

11. How much has pain from your hip interfered with your usual work (including housework)?

☐ Not at all ☐ A little bit ☐ Moderately ☐ Greatly ☐ Totally

12. Have you been troubled by pain from your hip in bed at night?

☐ No nights ☐ Only 1 or 2 nights ☐ Some nights ☐ Most nights ☐ Every night

Appendix 5

Satisfaction Questionnaire

supported by **stryker**

Orthowave Patient Satisfaction Questionnaire (PSa1)

Page 1 of 1

Patient Name ____________________

Patient ID ____________ Study ID ____________

Reviewer ____________________

Patient's Bar Code ID

Data Entry on: ________ by: ________

Side ☐ Right ☐ Left

Date of Evaluation (dd/mm/yy): ______/______/______

How well did the surgery on your joint:

Relieve pain in your affected joint?

☐ Excellent ☐ Very good ☐ Good ☐ Fair ☐ Poor

Increase your ability to perform regular activities?

☐ Excellent ☐ Very good ☐ Good ☐ Fair ☐ Poor

Allow you to perform heavy work or sport activities? (if applicable or allowed by doctor)

☐ Excellent ☐ Very good ☐ Good ☐ Fair ☐ Poor

Meet your expectations?

☐ Excellent ☐ Very good ☐ Good ☐ Fair ☐ Poor

Would you have this operation again if required on another joint?

☐ Definitely yes ☐ Possibly yes ☐ Probably not ☐ Certainly not

Notes

Appendix 6

BodyBuilder Code

Model file:

```
{*Start of macro section*}
{*====================*}
macro SUBSTITUTE4(p1,p2,p3,p4)
{*Replaces any point missing from set of four fixed in a segment*}
s234 = [p3,p2-p3,p3-p4]
p1V = Average(p1/s234)*s234
s341 = [p4,p3-p4,p4-p1]
p2V = Average(p2/s341)*s341
s412 = [p1,p4-p1,p1-p2]
p3V = Average(p3/s412)*s412
s123 = [p2,p1-p2,p2-p3]
p4V = Average(p4/s123)*s123

p1 = p1 ? p1V
p2 = p2 ? p2V
p3 = p3 ? p3V
p4 = p4 ? p4V
endmacro

macro POINTER(Anatomy,Segment)
{*Calculates the position of the end of the pointer for calibration in the technical frame it belongs to*}
{*1st determine the "point" in the Global system and outputs it as point#Calib. Then converts the point into*}
{*the appropriate technical reference frame and stores it as parameter $%#point#Calib*}

unitPointer=((Pointer1-Pointer2)/DIST(Pointer1,Pointer2))
Anatomy#Calib=Pointer1+123*unitPointer
OUTPUT(Anatomy#Calib)
PARAM(Anatomy#Calib)
%#Anatomy#Calib=Anatomy#Calib/Segment
PARAM(%#Anatomy#Calib)
endmacro
```

```
Macro DYNPOINTER(AnatPoint,Segment)
AnatPoint=%#AnatPoint#Calib*Segment
OUTPUT(AnatPoint)
PARAM(AnatPoint)
EndMacro

macro SEGVIS(Segment)
{*outputs a visual representaion of the segment to be viewed in the Workspace*}
{*0(Segment) is the origin of the segment*}
ORIGIN#Segment=0(Segment)
XAXIS#Segment=0(Segment)+(1(Segment)*100)
YAXIS#Segment=0(Segment)+(2(Segment)*100)
ZAXIS#Segment=0(Segment)+(3(Segment)*100)
OUTPUT(ORIGIN#Segment,XAXIS#Segment,YAXIS#Segment,ZAXIS#Segment)
endmacro

macro ColeJCS(seg1,seg2,joint)
{* Procedure to calculate the rotations about defined embedded axes using the joint
co-ordinate system.
```

References: Cole,G.K. et al (1993). Application of the Joint Co-ordinate System to Three-dimensional Joint Attitude and Movement Representation : A Standardization Proposal. Journal of Biomechanical Engineering. November 1993 : Vol 115 : pp 344-349

aEone,aEtwo,aEthree =unit vector describing the attitude of the 1st,2nd and 3rd axis of the joint co-ordinate system between the reference segment (seg1) and the target segment (seg2), relative to an inertial reference system.

If the axes of a body segment co-ordinate system are identified as an axis of Flexion, a Longitudinal axis and a Third axis, then Fone, Lone, Tone are unit vectors that describe the attitude of the Flexion, Longitudinal and Third axes respectively, in an inertial reference system.

Input: 'seg1', 'seg2' describing the axes of the co-ordinate systems embedded in each segment.
Fone, Lone, Tone describe the flexion, longitudinal and third co-ordinate axes of the proximal segment.
Ftwo, Ltwo, Ttwo describe the flexion, longitudinal and third co-ordinate axes of the distal segment.

'joint' is the name given to the joint at which the specified segments interact.

Output: Angles of rotation about axes aEone,aEtwo,aEthree, flexion, abduction and rotation respectively. Counterclockwise rotations are chosen as positive*}

```
Fone=3(seg1)
Lone=2(seg1)
Tone=1(seg1)
Ftwo=3(seg2)
Ltwo=2(seg2)
Ttwo=1(seg2)

{*Defines e1 and e3*}
aEone=Fone
aEthree=Ltwo

{*Calculate the Vector or Cross Product between the Vectors*}
Va={2(aEthree)*3(aEone)-3(aEthree)*2(aEone),3(aEthree)*1(aEone)-1(aEthree)*3(aEone),1(aEthree)*2(aEone)-
2(aEthree)*1(aEone)}
Vb=DIST({2(aEone)*3(aEthree)-3(aEone)*2(aEthree),3(aEone)*1(aEthree)-
1(aEone)*3(aEthree),1(aEone)*2(aEthree)-2(aEone)*1(aEthree)},{0,0,0})
Vc={2(Va)*3(aEthree)-3(Va)*2(aEthree),3(Va)*1(aEthree)-1(Va)*3(aEthree),1(Va)*2(aEthree)-2(Va)*1(aEthree)}

{*Calculate the Scalar or Dot Product between the Vectors*}
DPone=(1(Va)*1(Ttwo))+(2(Va)*2(Ttwo))+(3(Va)*3(Ttwo))
DPtwo=(1(Vc)*1(Ftwo))+(2(Vc)*2(Ftwo))+(3(Vc)*3(Ftwo))

{*Calculates A (AA) and then e2*}
IF DPone < 0 AND DPtwo > 0 THEN AA=-1 ELSE AA=1 ENDIF
aEtwo=(Va/Vb)*AA

{*Calculate the value of r.*}
Rone={2(Fone)*3(aEtwo)-3(Fone)*2(aEtwo),3(Fone)*1(aEtwo)-1(Fone)*3(aEtwo),1(Fone)*2(aEtwo)-
2(Fone)*1(aEtwo)}
Rtwo=DIST(Rone,{0,0,0})
r=Rone/Rtwo

{*Calculate the Scalar or Dot Product between the Vectors.*}
aEtwoTonedp=(1(aEtwo)*1(Tone))+(2(aEtwo)*2(Tone))+(3(aEtwo)*3(Tone))
```

```
aEtwoLonedp=(1(aEtwo)*1(Lone))+(2(aEtwo)*2(Lone))+(3(aEtwo)*3(Lone))
rLtwodp=(1(r)*1(Ltwo))+(2(r)*2(Ltwo))+(3(r)*3(Ltwo))
FoneLtwodp=(1(Fone)*1(Ltwo))+(2(Fone)*2(Ltwo))+(3(Fone)*3(Ltwo))
aEtwoTtwodp=(1(aEtwo)*1(Ttwo))+(2(aEtwo)*2(Ttwo))+(3(aEtwo)*3(Ttwo))
aEtwoFtwodp=(1(aEtwo)*1(Ftwo))+(2(aEtwo)*2(Ftwo))+(3(aEtwo)*3(Ftwo))

IF aEtwoLonedp >= 0 THEN aEtwoLonesign=1 ENDIF
IF aEtwoLonedp < 0 THEN aEtwoLonesign=-1 ENDIF
IF FoneLtwodp >= 0 THEN FoneLtwosign=1 ENDIF
IF FoneLtwodp < 0 THEN FoneLtwosign=-1 ENDIF
IF aEtwoFtwodp >= 0 THEN aEtwoFtwosign=1 ENDIF
IF aEtwoFtwodp < 0 THEN aEtwoFtwosign=-1 ENDIF

joint#Flex=(acos(aEtwoTonedp))*(aEtwoLonesign)
joint#Abd=(acos(rLtwodp))*(FoneLtwosign)
joint#Rot=(acos(aEtwoTtwodp))*(aEtwoFtwosign)
joint#angles=<joint#Flex,joint#Abd,joint#Rot>

OUTPUT(joint#angles)

joint#JCS=[joint,aEtwo,aEone,xyz]

ORIGIN#joint#jcs=0(joint#jcs)
XAXIS#joint#jcs=0(joint#jcs)+(1(joint#jcs)*100)
YAXIS#joint#jcs=0(joint#jcs)+(2(joint#jcs)*100)
ZAXIS#joint#jcs=0(joint#jcs)+(3(joint#jcs)*100)
OUTPUT(ORIGIN#joint#jcs,XAXIS#joint#jcs,YAXIS#joint#jcs,ZAXIS#joint#jcs)
ENDMACRO

macro FORCEVECTOR(FP)
{*This defines the quantities of force(F), moment(M) and Centre(C) from the reaction (FP)*}
{*P_#FP is the centre of pressure and is set at the forceplate centre if load is below 10N*}

If ExistAtAll( FP )
        F_#FP = FP(1)
        M_#FP = FP(2)
        C_#FP = FP(3)
        if ( ABS ( F_#FP ) > 10 )
                P_#FP = C_#FP + { -M_#FP(2)/F_#FP(3), M_#FP(1)/F_#FP(3), -C_#FP(3) }
```

```
        else
                P_#FP = C_#FP
        endif
        F_#FP = F_#FP + P_#FP
        OUTPUT ( P_#FP, F_#FP )
EndIf
endmacro

{*Macro for Dot Product*}
MACRO DotProduct (One,Two,DotProd)
        DotProd = (1(One)*1(Two)+2(One)*2(Two)+3(One)*3(Two))
ENDMACRO

{* Macro to do a cross product *}
MACRO CrossProduct ( First, Second, Result )
        Result = { First(2)*Second(3)-First(3)*Second(2),
        First(3)*Second(1)-First(1)*Second(3),
        First(1)*Second(2)-First(2)*Second(1)}
ENDMACRO

macro LINVELACC(Point,Segment)
{*When called, this macro calculates the linear velocity in m/s and the linear acceleration in m/s^2 of a
point, using numerical differentiation. For numerical differentiation, reference one of the following:
Hildebrand, F.B. (1974). Introduction to Numerical Analysis, 2nd Edition, pp.111
Kreyszig, Erwin (1983). Advanced Engineering Mathematics, 5th Edition, pp.793
Yakowitz, Sydney and Szidarovsky, Ferenc (1989). An Introduction to Numerical Computations, 2nd Edition,
pp.185*}

$SamplingRate= 120
$FrameTimeLength=1/$SamplingRate
LVel#Point=((Point[-2]-(8*Point[-1])+(8*Point[1])-Point[2])/(12*$FrameTimeLength))/1000
LAccel#Point=((LVel#Point[-2]-(8*LVel#Point[-1])+(8*LVel#Point[1])-LVel#Point[2])/(12*$FrameTimeLength))

{*%LVel#Point=LVel#Point/Segment
%LAccel#Point=LAccel#Point/Segment*}
output(LVel#Point,LAccel#Point)
param($FrameTimeLength)
endmacro
```

```
macro ANGVELACC(child,parent,Joint)
{*When called, this macro calculates the angular velocity in rad/s and the angular acceleration in rad/s^2
at a joint, using numerical differentiation. For numerical differentiation, reference one of the following:
Hildebrand, F.B. (1974). Introduction to Numerical Analysis, 2nd Edition, pp.111
Kreyszig, Erwin (1983). Advanced Engineering Mathematics, 5th Edition, pp.793
Yakowitz, Sydney and Szidarovsky, Ferenc (1989). An Introduction to Numerical Computations, 2nd Edition,
pp.185*}

$SamplingRate= 120
$FrameTimeLength=1/$SamplingRate
pi=3.1415927
Joint#Angle= joint#angles
Joint={Joint#Angle(1),Joint#Angle(2),Joint#Angle(3)}
Rad#Joint=Joint*pi/180
AVel#Joint=((Rad#Joint[-2]-(8*Rad#Joint[-1])+(8*Rad#Joint[1])-Rad#Joint[2])/(12*$FrameTimeLength))
AAccel#Joint=((AVel#Joint[-2]-(8*AVel#Joint[-1])+(8*AVel#Joint[1])-AVel#Joint[2])/(12*$FrameTimeLength))

output(AVel#Joint,AAccel#Joint)
param($FrameTimeLength)
endmacro


{*End of macro section*}
{* Anthropometric Data: From DA Winter, Biomechanics and Motor Control of Human Movement *}
AnthropometricData
DefaultPelvis 0.142 0.865 0.5 0.3
DefaultFemur 0.1 0.567 0.323 0
DefaultTibia 0.0465 0.567 0.302 0
DefaultFoot 0.0195 0.5 0.475 0
EndAnthropometricData

{*Optional points are points which may not be present in every trial*}
OptionalPoints(Pointer1,Pointer2)
OptionalPoints(RASIS,LASIS,RPSIS,LPSIS,PELF)
OptionalPoints(WAIST1,WAIST2,WAIST3,WAIST4)
OptionalPoints(RTH1,RTH2,RTH3,RTH4)
OptionalPoints(LTH1,LTH2,LTH3,LTH4)
OptionalPoints(RSH1,RSH2,RSH3,RSH4)
OptionalPoints(LSH1,LSH2,LSH3,LSH4)
```

```
OptionalPoints(RMET1,RMET5,RHEEL,LMET1,LMET5,LHEEL)
OptionalPoints(RHJC,LHJC,RKJC,LKJC,RAJC,LAJC)
OptionalPoints(CalRLEPI,CalRMEPI,CalLLEPI,CalLMEPI)
OptionalPoints(RLMAL,RMMAL,LLMAL,LMMAL)
OptionalPoints(CalRASIS,CalLASIS,CalRPSIS,CalLPSIS)

{*Substitutes missing markers based on clusters of 4 markers*}
SUBSTITUTE4(WAIST1,WAIST2,WAIST3,WAIST4)
SUBSTITUTE4(RTH1,RTH2,RTH3,RTH4)
SUBSTITUTE4(LTH1,LTH2,LTH3,LTH4)
SUBSTITUTE4(RSH1,RSH2,RSH3,RSH4)
SUBSTITUTE4(LSH1,LSH2,LSH3,LSH4)

{*Marker cluster axis definitions.....CHECK DIRECTIONS WITH RESPECT TO MARKERS*}
WaistSeg=[WAIST2,WAIST2-WAIST3,WAIST3-WAIST4,xyz]
RThighSeg=[RTH1,RTH1-RTH3,RTH3-RTH2,zyx]
RShinSeg=[RSH1,RSH1-RSH4,RSH2-RSH4,zxy]
LThighSeg=[LTH1,LTH1-LTH3,LTH3-LTH2,zyx]
LShinSeg=[LSH1,LSH1-LSH4,LSH2-LSH4,zxy]

{*STATIC CALIBRATIONS*}
If $Static==1

If EXIST(CalRLEPI)
Pointer (RLEPI,RThighSeg)
EndIf

If EXIST(CalRMEPI)
Pointer (RMEPI,RThighSeg)
EndIf

If EXIST(CalLLEPI)
Pointer (LLEPI,LThighSeg)
EndIf

If EXIST(CalLMEPI)
Pointer (LMEPI,LThighSeg)
EndIf
```

```
If EXIST(CalRASIS)
Pointer (RASIS,WaistSeg)
EndIf

If EXIST(CalLASIS)
Pointer (LASIS,WaistSeg)
EndIf

If EXIST(CalRPSIS)
Pointer (RPSIS,WaistSeg)
EndIf

If EXIST(CalLPSIS)
Pointer (LPSIS,WaistSeg)
EndIf

%RMMAL=RMMAL/RShinSeg
%LMMAL=LMMAL/LShinSeg
%RLMAL=RLMAL/RShinSeg
%LLMAL=LLMAL/LShinSeg

PARAM (%LMMAL,%LLMAL,%RLMAL,%RMMAL)
EndIf

{*Dynamic Trials*}

If $Static==0

{*Anatomical frame definition*}
RMMAL=%RMMAL*RShinSeg
RLMAL=%RLMAL*RShinSeg
LMMAL=%LMMAL*LShinSeg
LLMAL=%LLMAL*LShinSeg
OUTPUT (RMMAL,RLMAL,LMMAL,LLMAL)

DYNPOINTER (RLEPI, RThighSeg)
DYNPOINTER (RMEPI, RThighSeg)
DYNPOINTER (LLEPI, LThighSeg)
```

```
DYNPOINTER (LMEPI, LThighSeg)
DYNPOINTER (RPSIS, WaistSeg)
DYNPOINTER (LPSIS, WaistSeg)
DYNPOINTER (RASIS, WaistSeg)
DYNPOINTER (LASIS, WaistSeg)

{*Pelvis Segment...Using "Bell et al. 1990"..Hip Offset*}
{*Hip joint centre is 14%, 30% and 19% From interAsis Distance*}
{*0.36 represents 50% from the ASIS less the 14%*}

SACR=(LPSIS+RPSIS)/2
OUTPUT(SACR)
PARAM(SACR)

PELF=(LASIS+RASIS)/2
OUTPUT(PELF)
PARAM(PELF)
Pelvis=[PELF, RASIS-LASIS, SACR-PELF, zyx]
{*Pelvis=ROT(Pelvis,3(Pelvis),$PelvisTilt)*}
{*Pelvis=ROT(Pelvis,1(Pelvis),$PelvisObliquity)*}
SEGVIS(Pelvis)

%RHipOffsetFactor={-0.19,-0.3,0.36}
%LHipOffsetFactor={-0.19,-0.3,-0.36}

InterASISDist=DIST(LASIS, RASIS)

RHJC= (InterASISDist*%RHipOffsetFactor)*Pelvis
LHJC= (InterASISDist*%LHipOffsetFactor)*Pelvis

OUTPUT (RHJC, LHJC)
PARAM (RHJC, LHJC)

RHipSeg=[RHJC, RASIS-LASIS, SACR-PELF, zyx]
LHipSeg=[LHJC, RASIS-LASIS, SACR-PELF, zyx]
SEGVIS(RHipSeg)
SEGVIS(LHipSeg)

{*Right Thigh Segment*}
```

```
RKJC=(RLEPI+RMEPI)/2
OUTPUT(RKJC)
PARAM(RKJC)

RFemur=[RKJC, RHJC-RKJC, RMEPI-RLEPI, yxz]
SEGVIS(RFemur)

{*Left Thigh Segment*}
LKJC=(LLEPI+LMEPI)/2
OUTPUT(LKJC)
PARAM(LKJC)
LFemur=[LKJC, LHJC-LKJC, LLEPI-LMEPI, yxz]
SEGVIS(LFemur)

{*Right Shin System*}
RAJC=(RMMAL+RLMAL)/2
OUTPUT(RAJC)
PARAM(RAJC)
RTibia=[RAJC, RKJC-RAJC, RMMAL-RLMAL, yxz]
SEGVIS(RTibia)

{*Left Shin System*}
LAJC=(LMMAL+LLMAL)/2
OUTPUT(LAJC)
PARAM(LAJC)
LTibia=[LAJC, LKJC-LAJC, LLMAL-LMMAL, yxz]
SEGVIS(LTibia)

{*Foot System*}
{*Considered to represent a shoe rather than a foot. The markers are put on so that they lie in a plane perpendicular to the floor in a neutral position. This can be considered as ankle joint neutral*}

{*Right Foot System*}
RmidFOOT=(RMET1+RMET5)/2
RFootSeg=[RHEEL, RHEEL-RmidFOOT, RMET1-RMET5, yxz]
OUTPUT(RmidFOOT)
SEGVIS(RFootSeg)

{*Left Foot System*}
```

```
LmidFOOT=(LMET1+LMET5)/2
LFootSeg=[LHEEL, LHEEL-LmidFOOT, LMET5-LMET1, yxz]
OUTPUT(LmidFOOT)
SEGVIS(LFootSeg)

{*The joint names are given values to allow the creation of dummy JCS*}
LHip=LHJC
RHip=RHJC
LKnee=LKJC
RKnee=RKJC
LAnkle=LAJC
RAnkle=RAJC

ColeJCS(LHipSeg,LFemur,LHip)
SEGVIS(LHipJCS)
ColeJCS(RHipSeg,RFemur,RHip)
SEGVIS(RHipJCS)
ColeJCS(LFemur,LTibia,LKnee)
SEGVIS(LKneeJCS)
ColeJCS(RFemur,RTibia,RKnee)
SEGVIS(RKneeJCS)
ColeJCS(LTibia,LFootSeg,LAnkle)
SEGVIS(LAnkleJCS)
ColeJCS(RTibia,RFootSeg,RAnkle)
SEGVIS(RAnkleJCS)

{*corrects so that flexion, adduction and internal rotation are positive*}
{*Order of angles is flexion, add, IR*}
RHipangles=<1(RHipangles),2(RHipangles),3(RHipangles)>
LHipangles=<1(LHipangles),-2(LHipangles),-3(LHipangles)>
RKneeangles=<-1(RKneeangles),2(RKneeangles),3(RKneeangles)>
LKneeangles=<-1(LKneeangles),-2(LKneeangles),-3(LKneeangles)>
RAnkleangles=<(1(RAnkleangles)-90),2(RAnkleangles*-1),3(RAnkleangles)>
LAnkleangles=<(1(LAnkleangles)-90),2(LAnkleangles),-3(LAnkleangles)>

Output(RHipangles,LHipangles,LKneeangles,RKneeangles,LAnkleangles,RAnkleangles)

EndIF
```

```
{*KINETIC CALCULATIONS*}
{*===================*}
{*Hierarchy*}
RFemur=[RFemur,Pelvis,RHJC, DefaultFemur]
LFemur=[LFemur,Pelvis,LHJC, DefaultFemur]
RTibia=[RTibia,RFemur,RKJC, DefaultTibia]
LTibia=[LTibia,LFemur,LKJC, DefaultTibia]
RFootSeg=[RFootSeg,RTibia,RAJC, DefaultFoot]
LFootSeg=[LFootSeg,LTibia,LAJC, DefaultFoot]
{*Force Vectors*}
{*=============*}
OptionalReactions(ForcePlate1, ForcePlate2, ForcePlate3, ForcePlate4)
ForceVector(ForcePlate1)
ForceVector(ForcePlate2)
ForceVector(ForcePlate3)
ForceVector(ForcePlate4)
{* Forces and Moments *}
{*===================*}
{* These moments are external moments*}
{* Not Normalised to body mass (NN)*}

{*NN=$BODYMASS*}

{*LOWER LIMB*}
RAnkleForce = 1(REACTION(RFootSeg))
RAnkleMoment = 2(REACTION(RFootSeg))
RAnkleMoment = RAnkleMoment/(1000)

RKneeForce = 1(REACTION(RTibia))
RKneeMoment = 2(REACTION(RTibia))
RKneeMoment = RKneeMoment/(1000)

RHipForce = 1 (REACTION(RFemur))
RHipMoment = 2(REACTION(RFemur))
RHipMoment = RHipMoment/(1000)

LAnkleForce = 1(REACTION(LFootSeg))
LAnkleMoment = 2(REACTION(LFootSeg))
LAnkleMoment = LAnkleMoment/(1000)
```

```
LKneeForce = 1(REACTION(LTibia))
LKneeMoment = 2(REACTION(LTibia))
LKneeMoment = LKneeMoment/(1000)

LHipForce = 1(REACTION(LFemur))
LHipMoment = 2(REACTION(LFemur))
LHipMoment = LHipMoment/(1000)

{*Corrects for inverse sign for right side of body in frontal and transverse plane*}

RHipMoment = {1(RHipMoment), 2(RHipMoment), -3(RHipMoment)}
RKneeMoment = {1(RKneeMoment), 2(RKneeMoment), 3(RKneeMoment)}
RAnkleMoment = {1(RAnkleMoment), 2(RAnkleMoment), -3(RAnkleMoment)}
LHipMoment = {-1(LHipMoment), -2(LHipMoment), -3(LHipMoment)}
LKneeMoment = {-1(LKneeMoment), -2(LKneeMoment), 3(LKneeMoment)}
LAnkleMoment = {-1(LAnkleMoment), -2(LAnkleMoment), -3(LAnkleMoment)}


OUTPUT(LHipForce, LKneeForce, LAnkleForce, RHipForce, RKneeForce, RAnkleForce,
RHipMoment,RKneeMoment,RAnkleMoment,LHipMoment,LKneeMoment,LAnkleMoment)
```

Parameter file:

```
{*VICON BodyLanguage (tm)*}
{*copyright 1995,1996,1997 Oxford Metrics Ltd*}
{*parameters for use with JulieAx.MKR, JulieAx.MOD*}
{*Static Set*}
{*==========*}
        $Static   = 1
{*Marker diameter and joint widths*}
{*================================*}
        $MarkerDiameter = 14
        $Height = 158.5
        $BODYMASS = 60.5

DistanceThreshold = 500
```

{*Output from file*}

LLEPICalib = {64.5615,-174.064,501.852}

%LLEPICalib = {-10.1408,-56.6719,-189.464}

LMEPICalib = {149.097,-57.3717,504.346}

%LMEPICalib = {96.3524,-59.1591,-198.507}

RASISCalib = {-56.7836,109.71,964.629}

%RASISCalib = {77.7152,41.9145,-165.755}

LASISCalib = {-388.708,-229.228,1068.68}

%LASISCalib = {-175.389,-129.198,-138.177}

%LMMAL = {52.6549,145.872,-134.297}

%LLMAL = {13.0151,80.7849,-177.434}

%RLMAL = {-20.4881,58.0341,-192.619}

%RMMAL = {-54.9299,131.369,-153.704}

RPSISCalib = {95.0867,6.78671,977.996}

%RPSISCalib = {-17.404,18.7006,-27.3744}

LPSISCalib = {77.2636,-77.2583,986.185}

%LPSISCalib = {-98.4398,17.978,-35.4216}

RMEPICalib = {46.7817,-75.6781,467.858}

%RMEPICalib = {31.0072,101.614,-171.09}

RLEPICalib = {46.1912,120.872,479.673}

%RLEPICalib = {-53.1335,53.8463,-145.83}

Marker file:

!MKR#2

[Autolabel]

Pointer1

Pointer2

RASIS

LASIS

LPSIS

RPSIS

SACR

RACJ

LACJ

JUG

XYPH

C7

T8

WAIST1

WAIST2

WAIST3

WAIST4

RTH1

RTH2

RTH3

RTH4

RSH1
RSH2
RSH3
RSH4

RMET1
RMET5
RHEEL

LTH1
LTH2
LTH3
LTH4

LSH1
LSH2
LSH3
LSH4

LMET1
LMET5
LHEEL

LMMAL
LLMAL
RMMAL
RLMAL

Pointer1,Pointer2

RACJ,LACJ

JUG,XYPH
XYPH,T8
T8,C7
C7,JUG

WAIST1,WAIST2
WAIST2,WAIST3
WAIST3,WAIST4
WAIST4,WAIST1

RASIS,LASIS
LASIS,SACR
SACR,RASIS

RTH1,RTH2
RTH2,RTH3
RTH3,RTH4
RTH4,RTH1

LTH1,LTH2
LTH2,LTH3
LTH3,LTH4
LTH4,LTH1

RSH1,RSH2
RSH2,RSH3
RSH3,RSH4
RSH4,RSH1

LSH1,LSH2
LSH2,LSH3
LSH3,LSH4
LSH4,LSH1

RMET1,RMET5
RMET5,RHEEL
RHEEL,RMET1

LMET1,LMET5
LMET5,LHEEL
LHEEL,LMET1

FrontalTrunkSeg
SagittalTrunkSeg

WAISTSeg

Pelvis

RThighSeg
RShinSeg
RFootSeg

LThighSeg
LShinSeg
LFootSeg

Pelvis,RThighSeg
RThighSeg,RShinSeg
RShinSeg,RFootSeg

Pelvis,LThighSeg
LThighSeg,LShinSeg
LShinSeg,LFootSeg

FrontalTrunkSeg=RACJ,JUG,LACJ
SagittalTrunkSeg=JUG,XYPH,T8,C7

WaistSeg=WAIST1,WAIST2,WAIST3,WAIST4
Pelvis=RASIS,LASIS,LPSIS,RPSIS

RThighSeg=RTH1,RTH2,RTH3,RTH4
RShinSeg=RSH1,RSH2,RSH3,RSH4
RFootSeg=RMET1,RMET5,RHEEL

LThighSeg=LTH1,LTH2,LTH3,LTH4
LShinSeg=LSH1,LSH2,LSH3,LSH4
LFootSeg=LMET1,LMET5,LHEEL

[Virtual Points]

RACJ
LACJ

JUG
XYPH
T8
C7

RASIS
LASIS
LPSIS
RPSIS
SACR
PELF

RMET1
RMET5
RHEEL

LMET1
LMET5
LHEEL

RLEPI
RMEPI
LMEPI
LLEPI

RLMAL
RMMAL
LLMAL
LMMAL

RHJC
LHJC

RKJC
LKJC

RAJC
LAJC

RACJ,JUG
JUG,LACJ
LACJ,RACJ

JUG,C7
C7,T8
T8,XYPH
XYPH,JUG

RASIS,LASIS
LASIS,SACR
SACR,RASIS

RHJC,RKJC
RKJC,RAJC

LHJC,LKJC
LKJC,LAJC

RMET1,RMET5
RMET5,RHEEL
RHEEL,RMET1

LMET1,LMET5
LMET5,LHEEL
LHEEL,LMET1

[Calib points]

CalRLEPI
CalRMEPI

CalLMEPI
CalLLEPI

CalRASIS
CalLASIS
CalLPSIS
CalRPSIS

[Kinematics]

LHipangles
RHipangles
RKneeangles
LKneeangles
LAnkleangles
RAnkleangles

[Force Vectors]

P_ForcePlate1	Base of Plate1 Vector
F_ForcePlate1	Tip of Plate1 Vector
P_ForcePlate2	Base of Plate2 Vector
F_ForcePlate2	Tip of Plate2 Vector

RAnkleForce
RKneeForce
RHipForce
LAnkleForce
LKneeForce
LHipForce

P_ForcePlate1, F_ForcePlate1
P_ForcePlate2, F_ForcePlate2

[Moments from model]
RHipMoment
RKneeMoment
RAnkleMoment
LHipMoment
LKneeMoment
LAnkleMoment

Appendix 7

Standardised Instructions for MDS protocol

The instructions given to all subjects in preparation for conducting the MDS protocol are outlined:

"During this part of the experiment, you will lie down on this bed and your leg will be suspended so it hangs freely, in a comfortable position without touching the bed. Once your leg is suspended comfortably, your leg will move. When you feel your leg moving, I would like you to press this switch [demonstration of the switch] to let me know the moment when you feel your leg starting to move. We will repeat this several times. There are two positions what you will lie in; on your back, where your leg will be moving out to the side and in [demonstrated by the operator] and on your side, where your leg will be moving forwards and back [demonstrated by the operator]. We will do this for both legs, so in total that's four different positions. Any questions so far?"

[Subject invited to mount the rig and assume the first position]. "I will explain everything as we go but please ask any questions that you have. You need to be comfortable throughout these tests to give you the best chance of being able to detect movement of your leg, so please let me know if you feel uncomfortable at any stage or need to alter your position. Are you comfortable at the moment? [Subject comfort verified by the operator. Proximal bridge of the rig secured and the beam lifted into place and secured]. Your leg will be suspended from this beam. When the movement takes place, we want to make sure that the movement is happening at your hip, and not from your knee. To do this I will put this brace on your leg. [PROM brace applied]. It is padded so it will also add to your comfort. To make sure there is no movement at your waist, these pads will be secured against your pelvis. [Padded clamps secured]. Now your leg will be suspended and we'll use these cradles under your thigh and your calf to support your leg. [Operator suspends the limb using cradles, aiming to achieve equal pressure for both cradles].

"Now close your eyes. Can you tell me whether you are aware of pressure from one cradle touching more than the other, or is one 'digging in' more than the other? [Feedback from the subject and subsequent adjustment of the length of the straps attached to the cradles].

"Your leg will move from this position and I want you to press the switch as soon as you become aware of the movement. However, I need to be able to measure the movement, and in order for me to do so, I'm going to attach a measurement device over your hip and another over the beam. This device measures angles electronically so these wires go to the computer but you won't feel anything from them as they are for measurement only. So that the legs of the measurement device do not move on the surface of your shorts, I am going to stick these plastic bases between your shorts and the device. [Plastic panels and electrogoniometers applied].

"Now everything is set up and we are ready to do the movement trials. Are you still comfortable? [Subject feedback]. During this test, your leg will more and I want you to press the switch when you *start* to feel your leg move. You do not have to be concerned about the end of the movement, only that start. When we do the tests I would like you to lie as still and relaxed as possible. This will give you the best chance of detecting when your leg starts to move. Your leg will either move towards this leg or away from this leg [operator indicates by brushing the contralateral limb]. With each new movement, I would like you to flick the switch as soon as you become aware of the movement, then tell me which direction your leg moved in by using the words 'towards' or 'away'. After you have told me the direction, flick the switch back again, ready for the next movement. We will do this about 20 times. Each time, the length of the movement and the speed of the movement will be different, but there will be long enough breaks between the movements for you to know when a new movement begins. Any questions? We will do a few practise trials before we begin to make sure you are comfortable with what you're going to do and to make sure there are no areas of pressure on your leg. Keep your leg as heavy and as relaxed as possible at all times, to give you the best chance at sensing the start of the movement and please keep the switch close to your upper body so that your leg does not jump as you press the switch. Make sure that you only press the switch when you are sure the movement has began, and you are sure that you are not mistaking things like muscle twitches and vibrations for instead of the movement. [Practise trials and subject feedback. Alteration of the viscous damper to achieve the correct motion velocity].

"Now we are ready to begin. The trials that will follow will be silent and I will not ask for feedback; however you are free to stop me at any time and can ask questions at any time. Remember to keep your leg as heavy and as relaxed as possible at all times."

Appendix 8

UCLA Activity Score

UCLA Activity Score	Hip ID:
	Study Hip: ☐ Left ☐ Right
	Examination Date (MM/DD/YY): / /
	Subject Initials: |___|___|___|
	Medical Record Number:

Interval: ____________

Check one box that best describes current activity level.

☐ 1: Wholly inactive, dependent on others, and can not leave residence

☐ 2: Mostly Inactive or restricted to minimum activities of daily living

☐ 3: Sometimes participates in mild activities, such as walking, limited housework and limited shopping

☐ 4: Regularly Participates in mild activities

☐ 5: Sometimes participates in moderate activities such as swimming or could do unlimited housework or shopping

☐ 6: Regularly participates in moderate activities

☐ 7: Regularly participates in active events such as bicycling

☐ 8: Regularly participates in active events, such as golf or bowling

☐ 9: Sometimes participates in impact sports such as jogging, tennis, skiing, acrobatics, ballet, heavy labor or backpacking

☐ 10: Regularly participates in impact sports

Appendix 9

Results Figures

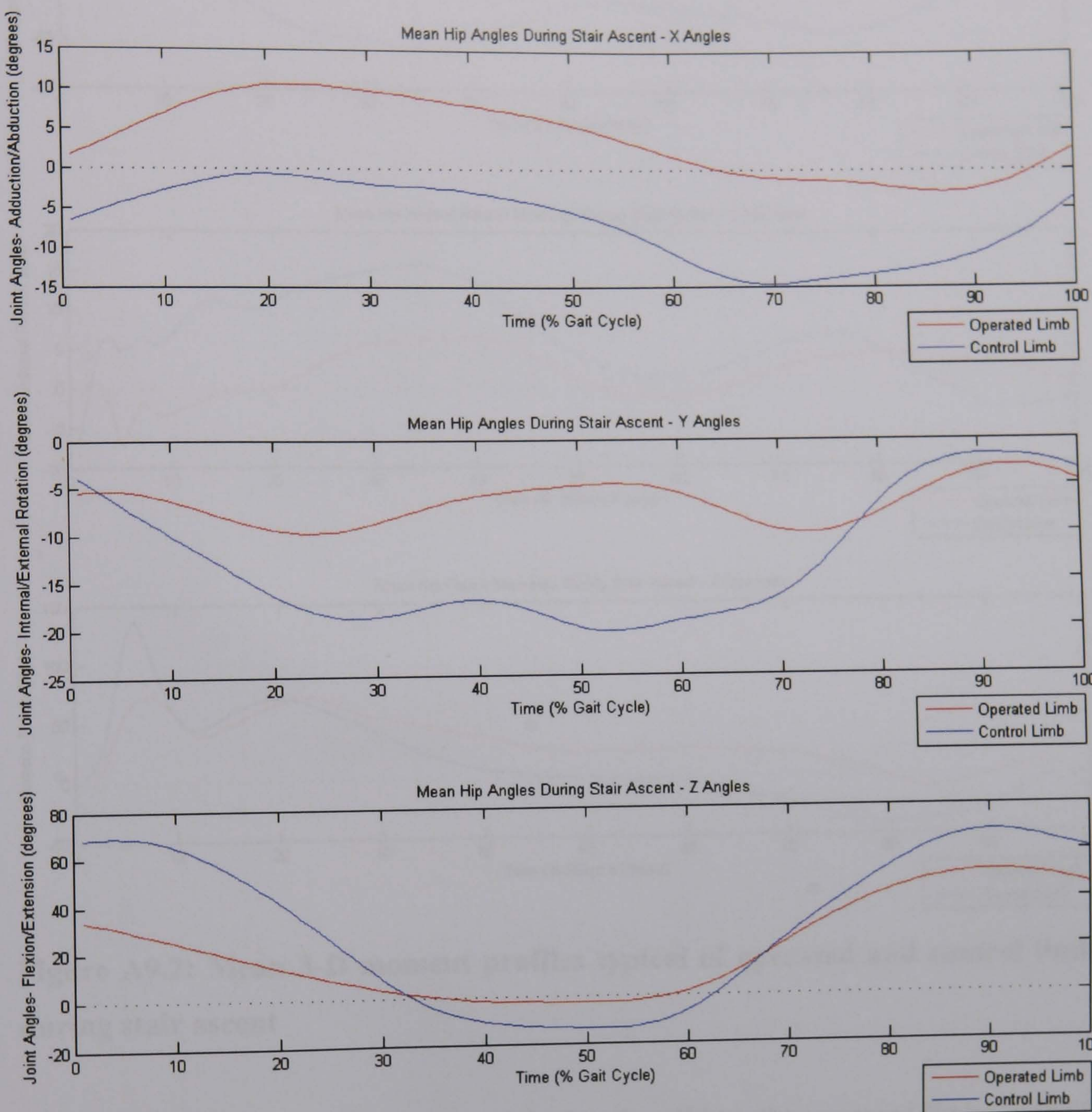


Figure A9.1: Mean 3-D angle profiles typical of operated and control limbs during stair ascent

The graphs show the average (of 3 trials) hip angles during the stair ascent gait cycle (foot contact to the following foot contact), following the format described in section 7.2.1.1. The data was extracted from 3 month post-op results although the trends identified from the data were present at 12 months in most cases.

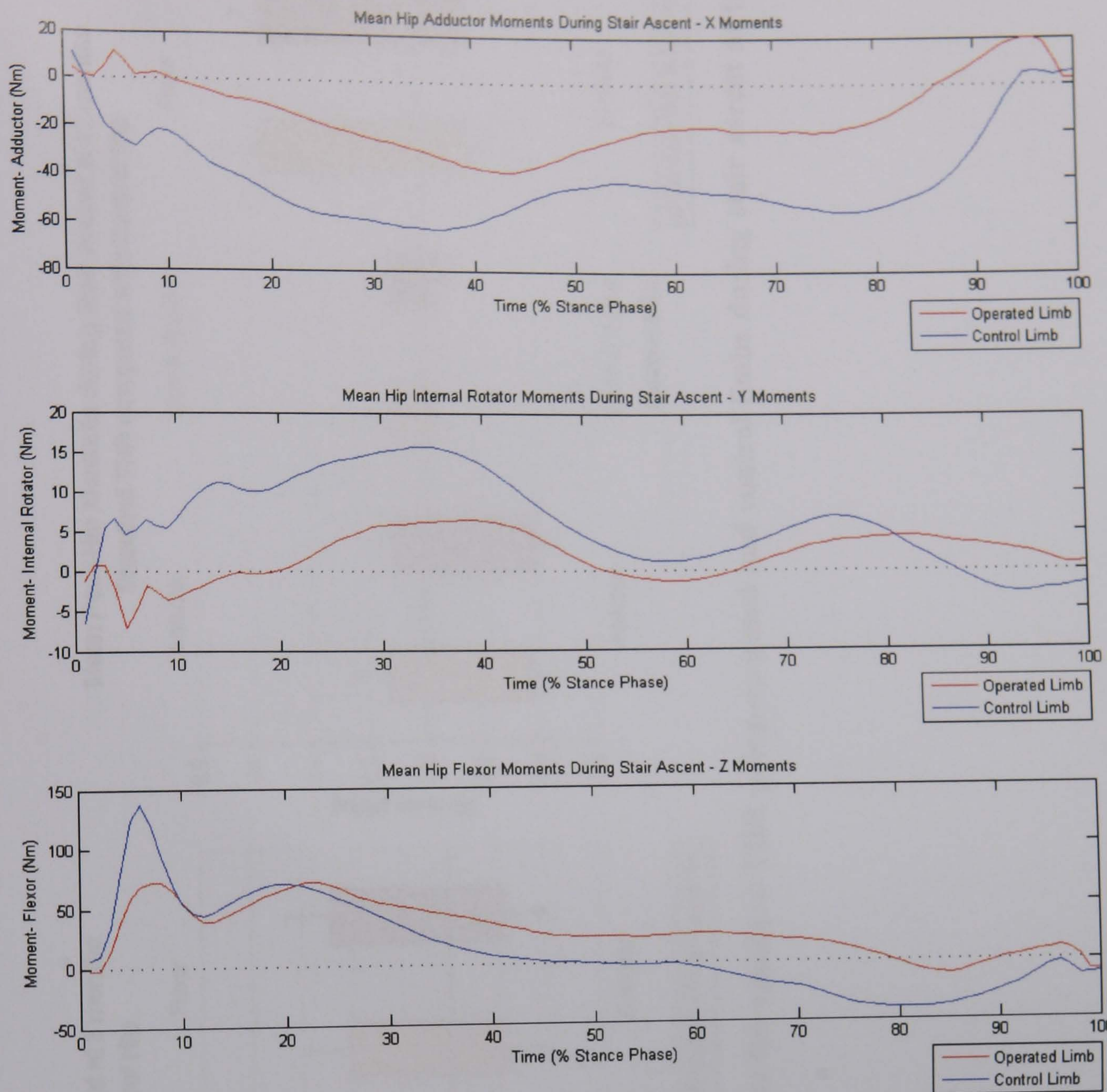
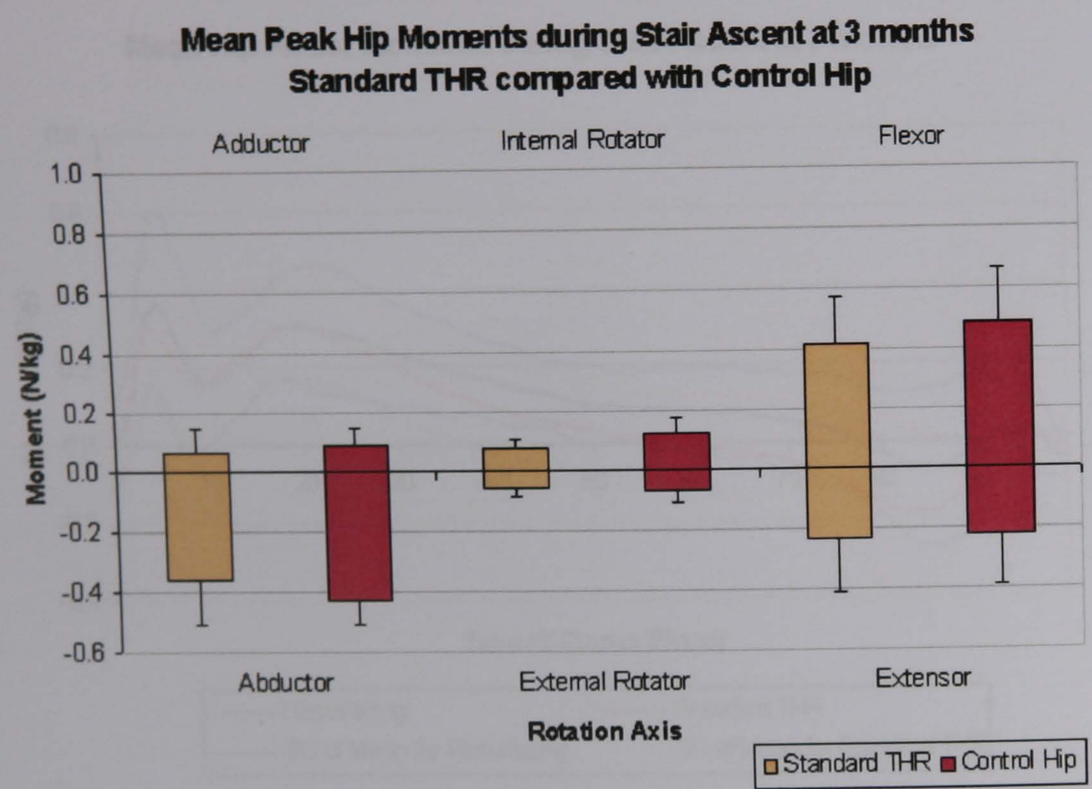


Figure A9.2: Mean 3-D moment profiles typical of operated and control limbs during stair ascent

Similar to the typical trends observed for level walking, the operated limb tended to have reduced abductor moments compared to the control limb. The internal rotator moments show a relative reduction in the peak internal rotator moment of the operated limb compared to the control limb. The hip flexor moment profiles also showed a reduction in the flexor and extensor moments of the operated limb relative to the control limb all throughout the stance phase of stair ascent.



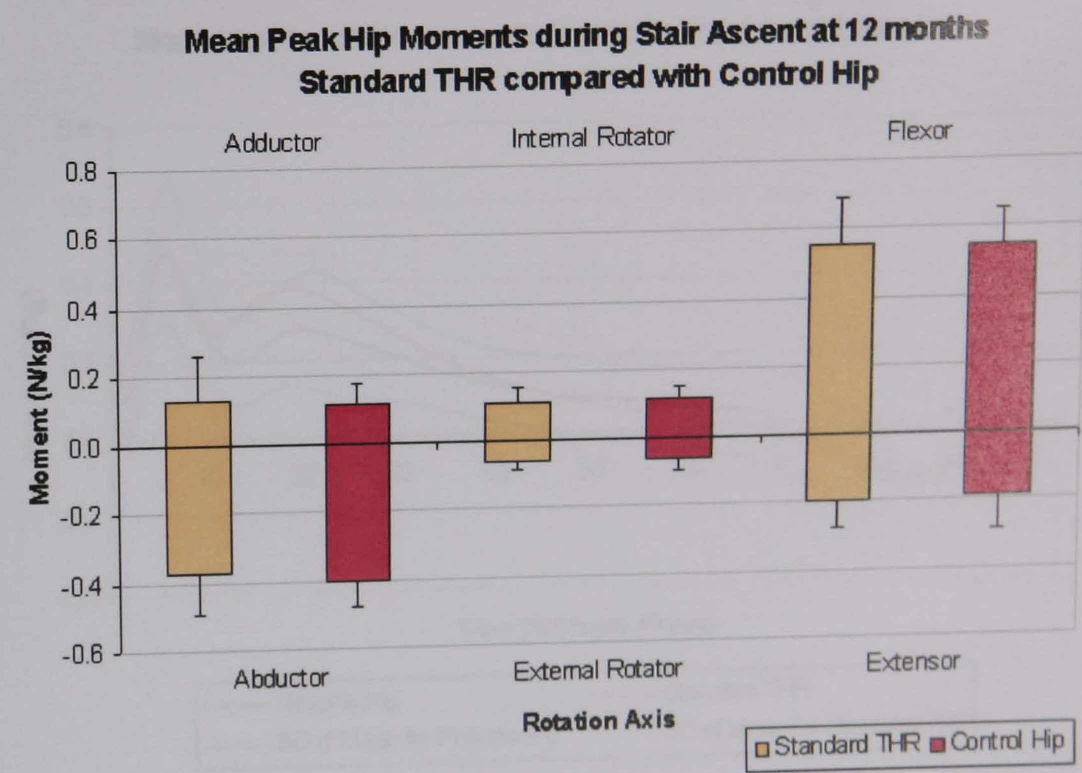


Figure A9.3: Comparison of 3-D hip moments for standard THR group-operated and control limbs during stair ascent at (a) 3 months and (b) 12 months

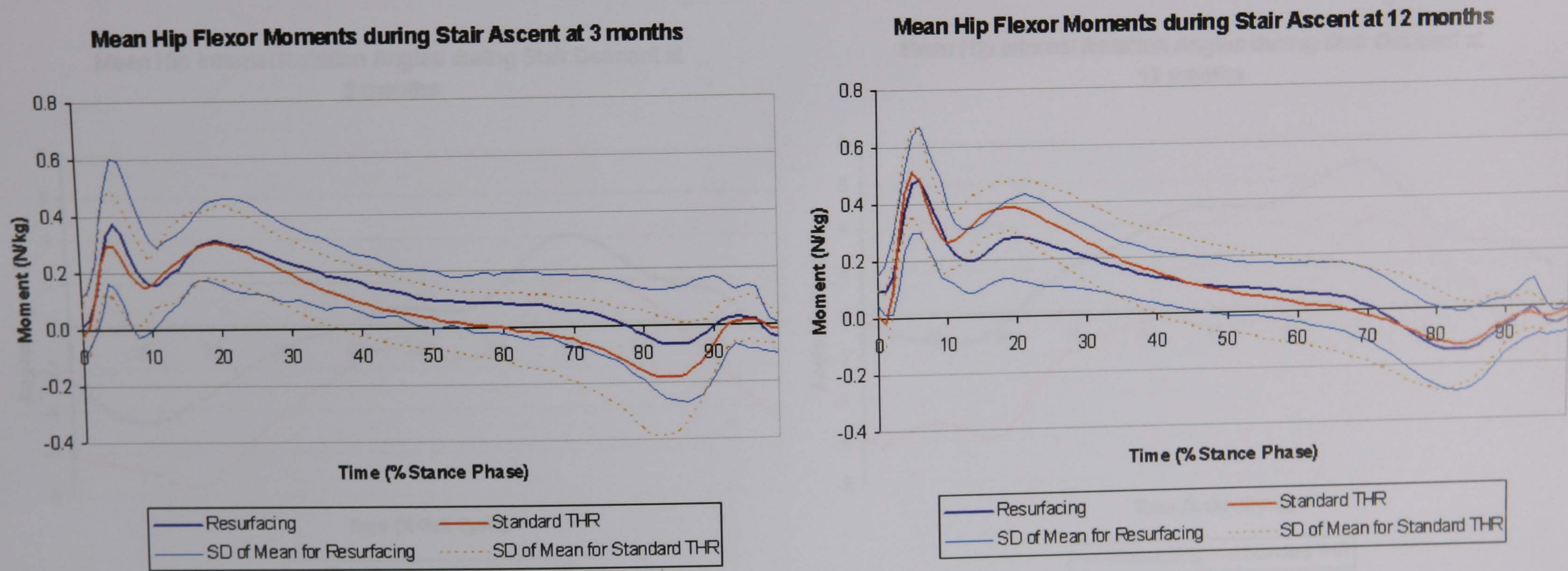


Figure A9.4: Mean hip flexor moments at (a) 3 and (b) 12 months

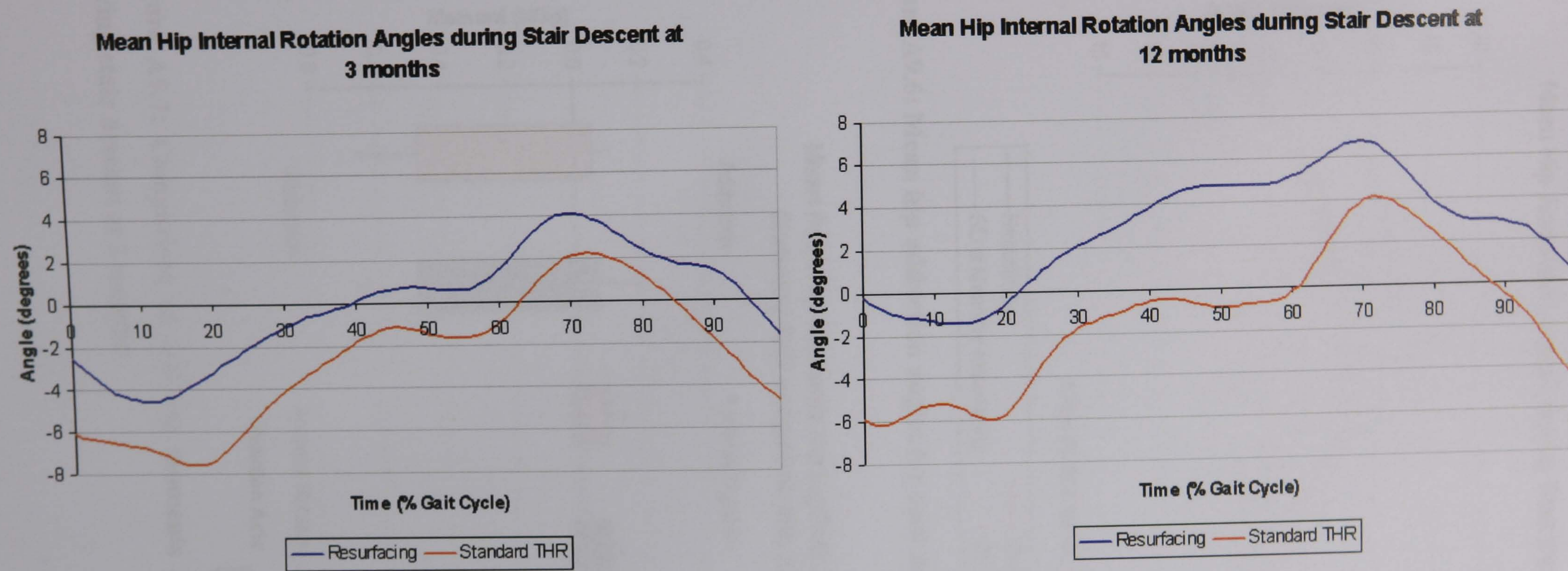


Figure A9.5: Mean hip internal rotation angles for stair descent at (a) 3 and (b) 12 months

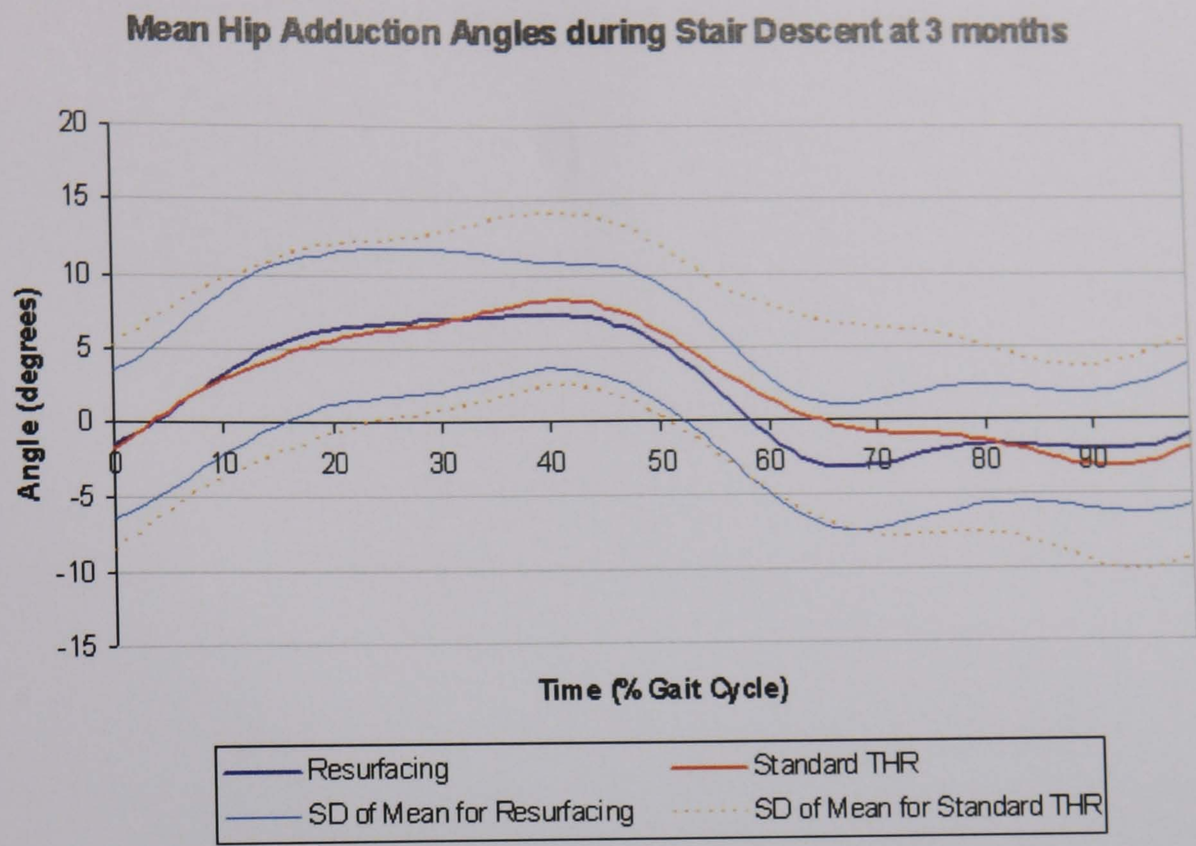


Figure A9.6: Mean hip adduction angles for stair descent at 3 months

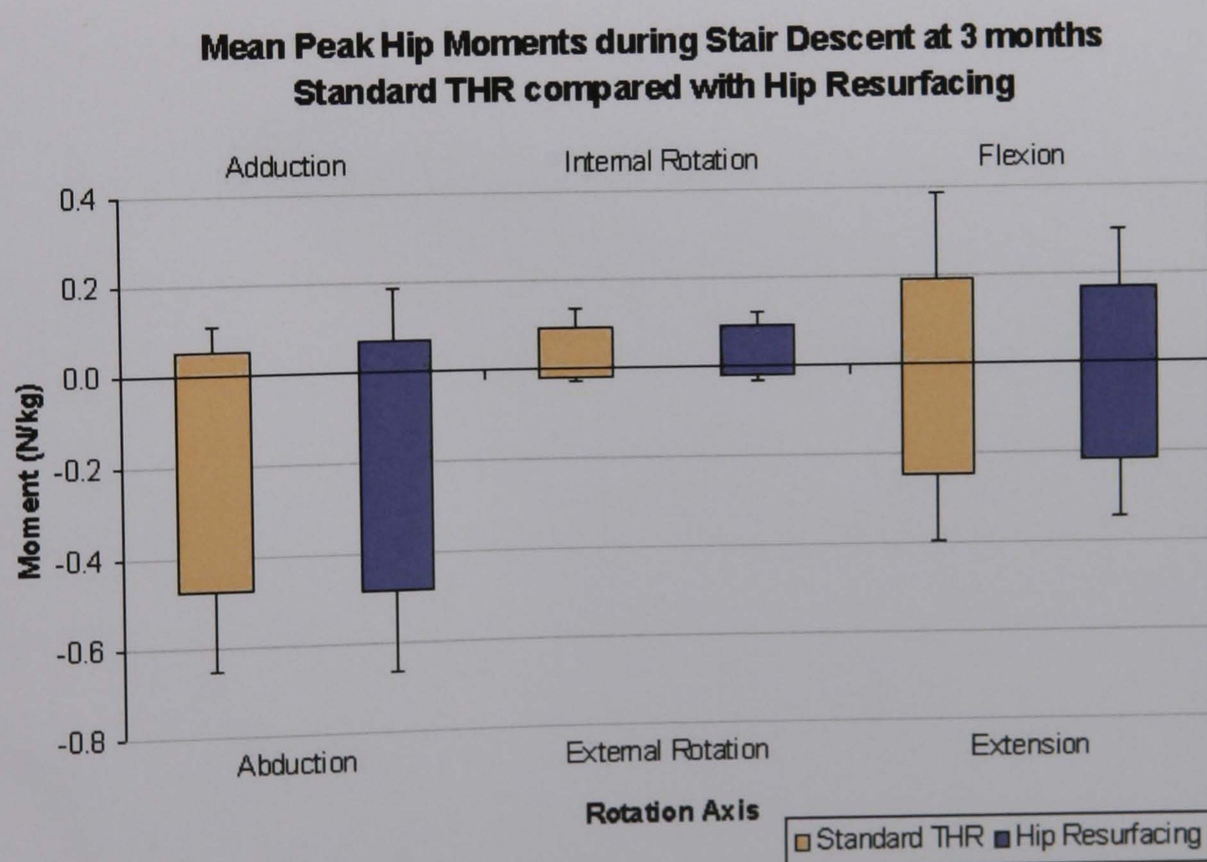


Figure A9.7: Comparison of 3-D hip moments between arthroplasty groups during stair descent at 3 months