Biomechanical Optimization of the Angle and Position for Surgical Implantation of a Straight Short Stem Hip Implant

by

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A thesis submitted in conformity with the requirements
for the degree of Master of Health Science in Clinical Engineering
Institute of Biomaterials and Biomedical Engineering
University of Toronto

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Abstract

Conservative hip implants preserve healthy bone for revision surgeries and improve physiological loading; however, they have few supporting biomechanical data. This thesis determined the 3D orientation of a straight short stem hip implant within the proximal femur that would yield a stress distribution most similar to an intact femur. Synthetic femurs were implanted with a stem in one of seven maximum angles or positions and axially loaded, with resultant strain values used to validate a finite element model. Design of experiments was used to analyze the range of potential implant orientations under three gait cycle loading conditions. A global optimal orientation of 9.14° valgus, 2.49° anteversion, 0.48mm posterior position, and 0.23mm inferior position yielded optimal stress distributions across the range of the gait cycle. Orthopaedic surgeons should therefore implant straight short stem hip implants in valgus up to 10° with an otherwise neutral position and version for maximal physiological loading.

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Table of Contents

Acknowledgments.....	iii
Table of Contents	iv
List of Tables	viii
List of Figures	xi
List of Abbreviations	xvii
CHAPTER 1 Thesis Overview	1
CHAPTER 2 Background.....	3
2.1 Human Femur Bones	3
2.1.1 Composition and Structure	3
2.1.2 Biomechanical Properties	3
2.2 Total Hip Arthroplasty.....	6
2.2.1 Indications for Total Hip Arthroplasty	6
2.2.2 Hip Replacement Demographics	7
2.2.3 Revision Surgery.....	8
2.2.4 Cemented and Cementless Implants	8
2.2.5 Stress Shielding.....	9
2.2.6 Gruen Zones.....	10
2.3 Conservative Hip Implants	11
2.3.1 DePuy Silent™ Hip Implant.....	12
2.3.2 Examination of Orientation.....	14
CHAPTER 3 Thesis Objectives and Hypothesis	15
3.1 Research Objective and Rationale	15
3.2 Research Question	15
3.3 Research Hypothesis.....	15
3.4 Specific Aims.....	15

CHAPTER 4 Relevant Literature.....	17
4.1 Conservative Hip Implant Reviews	17
4.2 Retrospective Reviews.....	20
4.3 Clinical Studies	21
4.4 Biomechanical Studies.....	24
4.4.1 Finite Element Modelling Studies	24
4.4.2 Experimental Studies	26
4.4.3 Implant Orientation Studies	27
4.5 Summary of Findings.....	30
CHAPTER 5 Research Methods.....	31
5.1 General Approach	31
5.2 Mechanical and Material Properties	31
5.2.1 Synthetic Femurs	31
5.2.2 Hip Implant Properties.....	33
5.3 Biomechanical Testing.....	36
5.3.1 Orientation Definition.....	36
5.3.2 Sample Size.....	41
5.3.3 Software-Guided Surgery	42
5.3.4 Silent Stem Implantation.....	46
5.3.5 Orientation Verification	49
5.3.6 Femur Preparation.....	51
5.3.7 Biomechanical Testing.....	54
5.3.8 Strain Gauge Calculations.....	56
5.4 Finite Element Analysis	57
5.4.1 Generation of Solid Models	57
5.4.2 ANSYS Workbench for Finite Element Analysis	59

5.5 Design of Experiments.....	72
5.5.1 OFAT versus Factorial Design	73
5.5.2 Response Surface Methodology Designs.....	74
5.5.3 JMP for Design of Experiments.....	75
5.5.4 Optimal Orientation	81
5.5.5 DOE Validation	83
5.5.6 Global Optimal Orientation Verification	83
CHAPTER 6 Results.....	84
6.1 Verification of Implant Orientation	84
6.2 Validation of FE Model	84
6.3 DOE Results.....	86
6.3.1 Neutral Phase	86
6.3.2 Extension Phase	104
6.3.3 Flexion Phase	105
6.3.4 Global Optimal Orientation	106
6.4 Validation of DOE	109
6.5 Validation of Global Optimal Orientation	111
CHAPTER 7 Discussion.....	112
7.1 Optimal Orientation	112
7.1.1 Common Significant Terms.....	113
7.1.2 Region- and Phase-Specific Significant Terms	114
7.2 Clinical Relevance	116
7.3 Study Limitations and Future Directions.....	118
CHAPTER 8 Conclusions.....	123
References.....	124
Appendix A: DOE Outcome Responses	136

Appendix B: Weighted Averages for Each Gait Phase	148
Appendix C: Complete Neutral Phase DOE Results	151
Appendix D: Complete Extension Phase DOE Results	173
Appendix E: Complete Flexion Phase DOE Results	201

List of Tables

Table 1: Average mechanical and material properties for human femoral bone ^{2, 7}	5
Table 2: Mechanical and material properties for synthetic femur bones ⁶³	32
Table 3: Material properties of the ceramic femoral head and titanium alloy stem ⁶⁸⁻⁷⁰	35
Table 4: Planned angles and positions in the A/P and lateral planes for each of the 6 maximum angles and positions, and neutral	41
Table 5: List of the Young's modulus and Poisson's ratio used for each material in the ANSYS Workbench simulations	60
Table 6: DOE factors and corresponding maximum levels, as achieved in the biomechanical tests	73
Table 7: CCD factor levels with corresponding coded names and values.....	76
Table 8: All of the runs specified by JMP given the chosen inputs, with a pattern indicating the composition of each run: “+” for high setting, “-” for low setting, “A” high axial, “a” low axial, “0” for centre point ¹¹¹	77
Table 9: Overview of each response selection approach and the method used to generate the employed input value.....	80
Table 10: Summary of the reduced Proximal Region stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and <i>p</i> -values. Significant terms have been marked with an asterisk (*) and bolded.....	87
Table 11: Summary of the reduced Proximal Region % of intact femur model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and <i>p</i> -values. Significant terms have been marked with an asterisk (*) and bolded.....	90
Table 12: Summary of the reduced regional stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and <i>p</i> -values. Significant terms have been marked with an asterisk (*) and bolded.....	92

Table 13: Summary of the reduced regional % of intact model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and <i>p</i> -values. Significant terms have been marked with an asterisk (*) and bolded.....	95
Table 14: Summary of the reduced regional 90 th percentile stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and <i>p</i> -values. Significant terms have been marked with an asterisk (*) and bolded.	98
Table 15: Summary of the reduced regional 90 th percentile % of intact stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and <i>p</i> -values. Significant terms have been marked with an asterisk (*) and bolded.	100
Table 16: Summary of common significant neutral terms and their corresponding effects, in addition to the region which was most significantly affected by the term	102
Table 17: Compilation of all optimal orientations generated by each neutral phase response model, specifying the weighted average values for each factor and the factor range.....	102
Table 18: Summary of common significant extension terms and their corresponding effects, in addition to the region which was most significantly affected by the term	104
Table 19: Compilation of all optimal orientations generated by each extension phase response model, specifying the weighted average values for each factor and the factor range.....	105
Table 20: Summary of common significant flexion terms and their corresponding effects, in addition to the region which was most significantly affected by the term	106
Table 21: Compilation of all optimal orientations generated by each flexion phase response model, specifying the weighted average values for each factor and the factor range.....	106
Table 22: Global optimal orientation, including weighted average values for each factor and the total acceptable factor range	107
Table 23: Response outcome measures for each Region of Interest across all gait phase optimal orientations, including the global optimal orientation, for loading under 1000N. Bolded and	

starred terms indicate responses that are more similar to the intact femur value than any other
response outcome measure for that particular DOE 109

List of Figures

Figure 1: Load versus deformation plot for bone, indicating the elastic and yield regions ¹	4
Figure 2: Stress-strain curve for bone, demonstrating the influence of strain rate on behaviour ^{2, 5}	5
Figure 3: a) Anatomy of a healthy hip joint; b) Outcomes of an osteoarthritic hip joint ¹²	7
Figure 4: Radiographs of the right hip: a) before implantation; b) immediately postoperatively; c) 7.5 years after surgery, with severe stress shielding noted ³⁵	10
Figure 5: Diagram depicting the seven Gruen Zones ³⁹	11
Figure 6: A short stem hip implant with a lateral flare for metaphyseal fixation ²⁰	12
Figure 7: Illustration of the DePuy Silent Hip (DePuy, Leeds, UK) ¹⁷	13
Figure 8: Silent Hip implanted in the femoral neck, demonstrating high neck cut ¹⁷	14
Figure 9: a) Short modular femoral stem (Smith & Nephew, Inc., Memphis, TN, USA) ⁵⁷ ; b) Mayo conservative implant double wedge stem with proximal coating (Zimmer, Rochester, MN, USA) ²⁰	18
Figure 10: a) Proxima Hip (DePuy, Leeds, UK) ⁵⁷ ; b) Birmingham mid-head resection (BMHR) (Smith & Nephew, Marl, Germany) ⁵⁸	19
Figure 11: Silent Hip (DePuy, Leeds, UK) ¹⁷	20
Figure 12: Radiograph of an implanted Type A short stem hip implant (right hip) and a Type B ultra-short stem implant (left hip), with a numerical depiction of the 5 modified Gruen zones for short stem implants ⁴³	22
Figure 13: Radiograph of an implanted Gothenburg osseointegrated titanium (GOT) arthroplasty ⁴¹	23
Figure 14: Pre-remodeling and post-remodeling shear stresses in the long and short-stemmed reconstructions, illustrating the implant design and short stem model ⁶¹	24

Figure 15: Depiction of compressive strain and strain shielding in the intact and reconstructed femur models ⁴⁷	26
Figure 16: Depiction of the changes in the absolute value of the major principal strains (in % of the intact femur value) after implantation of the Alloclassic (left), OPTAN (middle), and CUT (right) ³¹	27
Figure 17: Depiction of cortical strain energy in each Gruen zone for all solid models under walking and stair climbing loads. Intact refers to the intact femur model, Ref is the reference implanted model, Med corresponds to the medialised model, and AV is the anteverted model ⁶²	29
Figure 18: Strain distributions for a single femur model in the: a) intact femur, b) resurfaced in varus, c) resurfaced in neutral, d) resurfaced in valgus ⁵⁴	30
Figure 19: Large fourth generation composite femur and femur cross-section (Model #3406, Pacific Research Laboratories, Vashon, Washington, USA). Sizing is as follows: a) 485 mm, b) 52 mm, c) 37 mm, d) 120°, e) 32 mm, f) 93 mm, g) 16 mm canal ⁶³	32
Figure 20: SEM of the Silent Hip stem, depicting the sintered calcium phosphate bead coating at x30 magnification	33
Figure 21: EDX of the stem neck region, illustrating the Ti-Al-V composition of the stem	34
Figure 22: EDX of the stem body region, indicating the composition of the calcium phosphate beaded coating	34
Figure 23: Expanded scale EDX of the ceramic femoral head, indicating the aluminum oxide matrix composition	35
Figure 24: A/P view of the Silent stem neutrally implanted within the femur, where the solid red line indicates the long axis of the femur and the dashed line represents the implant axis at a CCD angle of 120°	37

Figure 25: Lateral view of the Silent stem neutrally implanted within the femur, where its neutral angle and position are displayed through its central location between lines tangent to the cortical bone.....	37
Figure 26: A/P view of a Silent stem implanted in 130° valgus	38
Figure 27: A/P view of a Silent stem implanted in 110° varus.....	39
Figure 28: Lateral view of a Silent stem implanted in 10° anteversion.....	39
Figure 29: Lateral view of a Silent stem implanted in 2mm anterior	40
Figure 30: Lateral view of a Silent stem implanted in 2mm posterior	40
Figure 31: A/P view of a Silent stem implanted in 4mm inferior.....	41
Figure 32: Depiction of the Birmingham Hip Resurfacing (BHR) implant (Smith & Nephew, Marl, Germany) ⁷⁵	42
Figure 33: Depiction of an optical array of calibration markers attached to the intact femur	43
Figure 34: Screenshot of the surgical planning screen provided by the BrainLAB Hip Essential software.....	44
Figure 35: Image of the registered femur undergoing guidewire placement using a drill and marker-equipped drill guide.....	45
Figure 36: Screenshot of the planned and achieved surgical paths for a neutrally aligned guidewire.....	46
Figure 37: Reamers included in the Silent Hip surgical technique.....	47
Figure 38: Silent stem placement using the implant impactor instrument, indicating the final seating height and the 3mm distal gap	48
Figure 39: Model of the intact femur, where the solid red lines indicate the respective resection lines and the dashed line indicates the longitudinal axis of the femur.....	49

Figure 40: A/P view of a neutrally implanted femur, indicating the neutral 120° CCD angle formed by the long axis of the femur and the implant	50
Figure 41: Lateral view of a neutrally implanted femur, depicting the modified Resubal method for determination of implant neutrality: a) tangent circles drawn on femoral neck; b) line drawn through circle centre-points, and distance to axis of implant measured	51
Figure 42: Synthetic femur located in a steel cube in preparation for potting, indicating the desired 7° of adduction	52
Figure 43: Gauged femur, indicating the location of the medial linear gauges (1 and 2), lateral linear gauges (3, 4, and 5), and the rosette gauge	53
Figure 44: Experimental set-up, depicting the femur locked into a neutral orientation within the Instron machine and attached to the Cronos data acquisition system (DAS)	55
Figure 45: Silent Hip model fitted to the femoral head, indicating location of extra bodies in blue for eventual Boolean operations in ANSYS	58
Figure 46: CAD model of the implanted femur assembly with the Silent stem in neutral	59
Figure 47: Screenshot of ANSYS DesignModeler, depicting the tree outline with each modification step.....	62
Figure 48: Comparison of linear neutral model with load applied directly to femoral head (left) versus a nonlinear neutral model with contact between head and load plate set to frictionless (right), equivalently scaled in von Mises stresses (MPa) for comparison purposes	64
Figure 49: Depiction of a 10-node quadratic tetrahedron ⁹⁵	65
Figure 50: Proximal cross-section of the mesh used for this study, depicting the finer mesh and areas of refinement.....	66
Figure 51: Screenshot of the final mesh used in this study.....	67
Figure 52: Proximal Region of interest selected based on a global coordinate system location, where the pink dots indicate underlying nodes.....	69

Figure 53: Inferior Neck Region, depicting selected nodes.....	70
Figure 54: Superior Neck Region, depicting selected nodes	70
Figure 55: Lateral Region, depicting selected nodes	71
Figure 56: Depiction of a black-box system with variable inputs and corresponding responses ¹⁰⁷	72
Figure 57: Two-level factorial design depicting the contents of each run, and the volume covered by the design ¹⁰⁸	73
Figure 58: Depiction of a two-level CCD design space ¹⁰⁷	74
Figure 59: Comparison of the three types of CCD, indicating factorial ranges (blue) and star points (pink) ¹⁰⁹	75
Figure 60: Colour map depicting the aliasing of effects, where a blue square indicates no correlation and a red square indicates complete aliasing.....	78
Figure 61: Screenshot of the desirability tool, where the lower graphs indicate the desirability response curve for each factor and the corresponding degree of desirability on the y-axis for each of the given factor settings, indicated by the numbers in red	82
Figure 62: Correlation of all measured experimental and corresponding FEM strains including a line of best fit with 95% confidence limits. The equation of fit is $y = 30.52 + 1.18x$, with a correlation coefficient of $R = 0.83$, and $p < 0.0001$	85
Figure 63: Residual plot of measured versus predicted strains.....	85
Figure 64: Bar chart of FEM and average experimental stiffness values for each implant orientation	86
Figure 65: Output of the desirability tool for stress response, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. Starting from the top row and going down, intact femur values for this optimization were 5.13MPa, 3MPa, and 9.19MPa.....	88

Figure 66: Output of the desirability tool for % of intact, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. All intact femur values for this optimization corresponded to 100	91
Figure 67: Output of the desirability tool for mean stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. Starting from the top row and going down, the intact femur values for this optimization were 9.19MPa, 6.11 MPa, 4.06MPa and 5.13MPa.....	93
Figure 68: Output of the desirability tool for percent of intact stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. All intact femur values for this optimization corresponded to 100.....	96
Figure 69: Output of the desirability tool for 90 th percentile stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. Starting from the top row and going down, the intact femur values for this optimization were 9.19MPa, 13.61 MPa, 9.66MPa and 5.88MPa	99
Figure 70: Output of the desirability tool for 90 th percentile % of intact femur stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. All intact femur values for this optimization corresponded to 100	101
Figure 71: FE model of the intact femur loaded at 3000N (top), and the neutral phase optimally implanted femur loaded at 3000N (bottom), equivalently scaled in von Mises stresses (MPa) for comparison purposes.....	103
Figure 72: FE model of the intact femur loaded at 3000N (top), and the global optimally implanted femur loaded at 3000N (bottom), equivalently scaled in von Mises stresses (MPa) for comparison purposes.....	108

List of Abbreviations

ANOVA: Analysis of Variance

A/P: Anteroposterior

BHR: Birmingham Hip Resurfacing

BMD: Bone Mineral Density

BMHR: Birmingham Mid-Head Resection

BW: Body Weight

CAD: Computer-Aided Design

CCD Angle: Caput-Collum-Diaphyseal Angle

CCD: Central Composite Design

CI: Confidence Interval

CT: Computed Tomography

DAS: Data Acquisition System

DEXA: Dual-Energy X-Ray Absorptiometry

DICOM: Digital Imaging and Communications in Medicine

DOE: Design of Experiments

DOF: Degrees of Freedom

EDX: Energy-Dispersive X-Ray Spectroscopy

FE: Finite Element

FEA: Finite Element Analysis

FEM: Finite Element Method

HA: Hydroxyapatite

HHS: Harris Hip Score

IR: Infrared

OA: Osteoarthritis

OFAT: One-Factor-at-a-Time

PMMA: Polymethylmethacrylate

RMSE: Root-Mean-Square Error

RSA: Radiostereometric Analysis

RSD: Relative Standard Deviation

RSM: Response Surface Methodology

SE: Standard Error

SEM: Scanning Electron Microscopy

SNA: Stem-Neck Angle

SE: Standard Error

THA: Total Hip Arthroplasty

CHAPTER 1

Thesis Overview

The following thesis explores the topic of conservative hip implants, in particular biomechanical considerations pertaining to the surgical placement of such implants, and is divided into 8 distinct chapters.

This thesis begins with Chapter 2, which provides background information on the behaviour and composition of human bone, the total hip arthroplasty (THA) procedure, and examines the motivation behind conservative hip implants. The DePuy Silent Hip (DePuy, Leeds, UK) in particular is discussed as a novel straight short stem hip implant with minimal supporting biomechanical evidence, and no research to support one surgical implant position over another, which defines the scope of this thesis. Chapter 3 then expounds upon the objectives of this thesis, discussing the rationale behind the study, the research question and hypothesis, followed by the specific aims which this thesis hopes to achieve.

Chapter 4 presents an overview of the research in this field, reviewing literature pertaining to conservative hip implants currently on the market. A number of review papers are discussed initially, followed by papers evaluating conservative implants in a clinical setting, and those using biomechanical techniques. Finally, the findings are summarized and knowledge gaps are discussed.

Chapter 5 outlines the three phases of this thesis, and the methods used to accomplish each. Initially, relevant mechanical and material properties are discussed, including the material properties of synthetic bone and the chosen conservative hip implant. An overview of the biomechanical testing phase of this study is then provided, wherein 10 synthetic femurs are surgically implanted with the Silent Hip and axially loaded for the purpose of validating a corresponding finite element (FE) model. This is followed by a discussion of the finite element method (FEM) modelling technique, and the approaches used to generate the FE model are then discussed. Finally, design of experiments (DOE) is examined as a method for determining the optimal implant orientation given the full range of potential angles and positions.

The results of these procedures are detailed in Chapter 6, which presents the outcomes for both the biomechanical testing phase and the FEM validation phase, followed by DOE results describing the optimal orientation of the Silent Hip within the proximal femur. Chapter 7 then discusses these findings and elaborates upon their clinical relevance and significance to the field, following which the limitations and potential future directions of this study are assessed. Finally, Chapter 8 comprises a summary section which gives an overview of the study and its subsequent findings.

CHAPTER 2

Background

2.1 Human Femur Bones

The human skeleton composes the framework of the body, and consists of a number of different bones whose functions vary. While some bones provide protection for internal organs, other bones are the main site for hematopoiesis, and still others aid in locomotion. The femur is a long bone within the leg that articulates with the hip and knee joints to allow for human mobility.

2.1.1 Composition and Structure

Bones, in particular long bones such as the human femur, must be both stiff and flexible in order to resist deformation while still enabling energy absorption at impact.¹ Furthermore, bones must be light enough to allow human mobility. This dichotomy is overcome through the material properties and morphology of bone. Bones are composed of a bone matrix with varying proportions of inorganic mineral ions, water, and organic components. Type I collagen is the main organic protein in the bone matrix, while calcium phosphate (primarily hydroxyapatite) is the primary inorganic component.² On a macroscopic level, bones are composed of cortical and cancellous (trabecular) bone. Cortical bone is the minimally porous, hard outer layer of bone, and is composed of lamellae that provide much of the required stiffness. Cancellous bone fills the bony interior and is composed of a woven porous network that makes the bone lighter, allows for blood vessel growth, and provides for some of the energy absorbing capabilities of bone. Bone mass is further reduced through the presence of a medullary cavity, in which the marrow is stored.¹

2.1.2 Biomechanical Properties

When a load is applied to bone it resists deformation, and stress (measured as force/area) is developed in response. The local deformation that occurs due to this applied load is called strain, and refers to the relative change in size or shape.¹ The reaction of bone to an applied load can be defined using either a load-deformation curve or a stress-strain curve. In a load-deformation curve, the amount of force needed to achieve a given deformation is defined, and thus this curve takes into account both the material properties of bone and its shape (Figure 1). Load and

deformation are generally linearly related until the yield point, and this linear portion describes the elastic region of bone during which any deformations will not be permanent. After yield, the bone will begin to undergo permanent deformation until finally it fails. The slope of the elastic region describes the stiffness of the bone.¹

Figure 1: Load versus deformation plot for bone, indicating the elastic and yield regions¹

The stress-strain curve for bone reflects the material behaviour of bone, and contains: an elastic region whose slope is known as the Young's modulus (or elastic modulus) which measures bone rigidity, a yield point with a corresponding yield stress and strain, and an ultimate point following which bone enters the inelastic region and subsequently fails.¹ Bone is a viscoelastic material, and so its strength is influenced by the rate of strain. For example, a higher strain rate increases the apparent stiffness of bone and so causes it to fracture at higher forces, where a strain rate of 0.01s^{-1} to 0.03s^{-1} represents physiological strain rates during rapid walking and running respectively (Figure 2). In general, bone typically fails at a deformation of about 1%.¹⁻⁴

Figure 2: Stress-strain curve for bone, demonstrating the influence of strain rate on behaviour^{2, 5}

Due to the presence of directional constituents, such as lamellae, haversian canals and collagen, bone is anisotropic. This means that it responds differently to mechanical stimuli depending on the direction of load application.¹ In the femur, anisotropy makes the bone strongest in its primary loading direction, namely axial compression (Table 1). Density also plays a role in bone strength, with increasing density generally being associated with higher compressive strengths.² Bone is also an inhomogeneous material, both at the macroscopic and microscopic levels. As such, the elastic modulus for bone varies depending on both the large-scale anatomic location and the particular microscopic bone lamella.⁶

Table 1: Average mechanical and material properties for human femoral bone^{2, 7}

Bone	Loading Mode	Density (g/cm ³)	Ultimate Strength (MPa)	Young's Modulus (GPa)
Cortical	Longitudinal Tension	1.6 - 2.1	133	17
	Longitudinal Compression	1.6 - 2.1	195	17
	Longitudinal Shear	1.6 - 2.1	68	3.3
	Transverse Tension	1.6 - 2.1	51	11.5
	Transverse Compression	1.6 - 2.1	33	11.5
Cancellous	Compression	0.1 - 1.0	6	0.76

2.2 Total Hip Arthroplasty

Attempts at hip replacement have been recorded as early as 1891 in Germany, in which ivory was used to replace the femoral heads of patients whose joints had degenerated due to tuberculosis.⁸ The total hip arthroplasty (THA) procedure as it is known today was first established by Sir John Charnley in the early 1960's, and is one of the most successful orthopaedic procedures of the 20th century.⁸ Presently, THA is an elective surgical procedure in which the surgeon removes damaged sections of the hip joint and replaces them with prosthetics made from metal alloys, ceramic, or plastic.

2.2.1 Indications for Total Hip Arthroplasty

In a healthy hip joint, the femoral head articulates smoothly with the acetabulum due to the lubricating effect of the synovial fluid and the smooth articular cartilage, which also acts to cushion the hip joint during load bearing. However, with age and repeated overuse, this cartilage may begin to degenerate, causing the bones of the femoral head and acetabulum to grate against one another. This can lead to the formation of bony spurs called osteophytes, swelling in the joint, and ultimately pain and reduced flexibility (Figure 3).^{9, 10} This type of degenerative joint disease is called osteoarthritis (OA), and is the most common reason for joint replacement surgery in Canada, accounting for 82.1% of primary hip replacements.¹¹

Other indications for THA include acute hip fractures, rheumatoid arthritis, osteonecrosis, and childhood hip diseases; however, all of these diagnoses combined are responsible for only 13.9% of hip replacements in Canada.¹¹

Figure 3: a) Anatomy of a healthy hip joint; b) Outcomes of an osteoarthritic hip joint¹²

2.2.2 Hip Replacement Demographics

Presently, orthopaedic surgeons make recommendations for hip replacement surgery based on the patient's pain and mobility levels, in addition to their overall health.¹² While certain health conditions may contraindicate hip replacement surgery, an operable patient who is both healthy and active will be more likely to have a successful outcome.¹³ Thus while THA used to be reserved for those aged 60 and over, it is increasingly being successfully performed in the younger patient population in order to maintain quality of life.^{13, 14} Furthermore, the prevalence of arthritis in the younger population is predicted to increase by half in the coming years, suggesting that THA in this population will become even more widespread.¹⁵

However, most hip implants have a functional lifespan of only 10 – 20 years before a revision surgery is required, due to factors that include aseptic loosening, osteolysis and decreased implant function.¹⁶ Thus, undergoing hip replacement surgery at a younger age introduces the possibility that a revision surgery will be required within the patient's lifetime.¹⁷

2.2.3 Revision Surgery

Revision surgery is a complex procedure that requires removal of the failed component, reconstruction of the remaining bone and soft-tissue, and then determination of enough good bone stock to ensure fixation of the new implant.¹⁶ Secure fixation of the revision component is crucial to ensuring long-term survivorship of the implant.¹⁸ Traditional hip implants extend into the diaphysis of the femur and are cemented in place or undergo osseointegration along their length, making them difficult to remove.¹⁶ This often makes stable fixation challenging to achieve proximally, as this bone stock tends to be deficient and weak from the previous implant.¹⁸

Manufacturers have therefore begun designing conservative hip implants that preserve a greater portion of healthy proximal bone stock initially to facilitate future revision surgeries if necessary. These hip implants include short stem or ultra-short stem designs that are usually uncemented and designed to be surgically implanted within the metaphysis of the proximal femur.^{19, 20}

2.2.4 Cemented and Cementless Implants

Hip implants can be affixed within the femur through either cemented or cementless means. In cemented THA, an acrylic polymer called polymethylmethacrylate (PMMA) is used to mechanically interact with the implant and the bone, providing stable fixation for the implant. Cementless THA utilizes a textured implant surface topography, often with a hydroxyapatite (HA) coating, that encourages direct bony ongrowth and ingrowth. There is much debate as to the efficacy of one fixation method over another, with the main difference in survivorship arising from the age of the patients at the time of surgery. Revision rates for cemented stems were found to be higher in the younger patient population when compared to the older population, typically due to aseptic loosening and osteolysis.²¹⁻²³ The 10-year survivorship of cementless THA was found to be significantly better than cemented THA, particularly in the younger population.^{21, 24} Use of cementless fixation has been steadily increasing in North America, and in Canada, cementless THA is currently the most common fixation method.¹¹

2.2.5 Stress Shielding

Hip implants have begun to diverge from designs based on the Koch model of the hip.²⁵ According to this model, the axial forces experienced during normal walking act as compressive and tensile loads on the medial and lateral proximal femur respectively.²⁶ It is now known that the iliotibial band, which runs concurrent to the lateral edge of the femur, acts as a tension band to turn lateral tensile forces into a compressive load.²⁶ This is important when considering implant design, as a good implant should therefore have the capacity to laterally distribute load in the proximal femur in order to align with physiological conditions.²⁵

Physiological loading is defined as load transfer that produces a stress distribution equivalent to that present in a healthy hip joint under normal loading conditions. In THA, achievement of physiological loading imparts biomechanical stability to the implanted femur, and so is thought to prevent a cascade of events initiated by “stress shielding,” in which the host bone is understressed due to the stiff femoral component supporting the majority of the load. According to Wolff’s law, bone is formed or resorbed in accordance with the magnitude and direction of the applied forces.²⁷ Inclusion of a femoral hip component would therefore cause the loads felt by the bone to decrease, eliciting bone adaptations that include decreased mass and increased porosity (Figure 4).²⁸ This can lead to the development of osteopenia, bone resorption around the implant, loosening of the implant with a concurrent decrease in biomechanical stability, and ultimately failure, which can necessitate a revision surgery in compromised bone stock.²⁹⁻³¹

Stress shielding can be difficult to define clinically, as a bone mineral density (BMD) loss of approximately 30% is required before changes can be seen radiographically.³² For that reason, a qualitative four degree system of bone remodelling is typically utilized when evaluating the severity of stress shielding.³³ In this system, degrees 1 and 2 represent low levels of stress shielding represented by decreased cortical bone density in only the proximal femur. More severe stress shielding is defined by degrees 3 and 4, which characterize bone resorption extending into the diaphysis.³⁴

Figure 4: Radiographs of the right hip: a) before implantation; b) immediately postoperatively; c) 7.5 years after surgery, with severe stress shielding noted ³⁵

Many implants have been designed with a lateral flare in order to engage the metaphysis and maintain proximal loading.²⁹ Furthermore, studies have found that short stem hip implants, which do not extend as far into the femoral diaphysis as traditional hip implants, exhibit more physiological loading proximally with less BMD loss.^{31, 36, 37}

2.2.6 Gruen Zones

In 1979, Gruen et al. developed a system to standardize radiographic assessment of loosening around the femoral component interface by dividing it into 7 zones (Figure 5).³⁸ This method was initially used to determine the location and type of loosening present in cemented femoral hip components.³⁸ Currently, these zones are used as a standardized method to compare spatial and temporal BMD measurements across varying implants types. They are therefore useful when evaluating stress shielding, as it is possible to explore the progression of BMD loss or gain within consistent areas of the proximal femur.

Figure 5: Diagram depicting the seven Gruen Zones ³⁹

2.3 Conservative Hip Implants

Short stem hip implants can vary significantly in geometry; however, they are all similar in that they extend only minimally into the femoral diaphysis, and are intended for metaphyseal fixation (Figure 6).^{36, 40-44} These stems are typically designed for a minimally invasive surgical implantation technique, and a high femoral neck cut to allow for optimal bone stock preservation.^{25, 45} Maintaining the neck also ensures that the trabecular system of the metaphyseal cancellous bone remains intact, thereby allowing for more physiological stress propagation through the femur.⁴⁶ Mid-term results for conservative implants have been favourable, illustrating that these implants have good initial stability, a tendency to provide more physiological conditions with less bone resorption, and a good implant survival rate.^{25, 36, 37, 41, 46-48} However, there is a dearth of data supporting short stem implants, and hence there is still a need for more biomechanical and clinical analyses.⁴⁴

Figure 6: A short stem hip implant with a lateral flare for metaphyseal fixation²⁰

2.3.1 DePuy Silent™ Hip Implant

The DePuy Silent™ Hip (DePuy, Leeds, UK) is an example of a modular, cementless short stem hip implant with limited supporting clinical or biomechanical data. It features a straight short stem with a collarless tapered profile for implantation in the femoral neck (Figure 7). It is composed of a Ti 6Al-4V alloy, and utilizes the DuoFix™ fixation system, which consists of a layer of hydroxyapatite coated beads to promote bony ingrowth. A 12/14 taper enables the Silent Hip to be combined with either a ceramic or metal head, for optimal choice of friction couple.¹⁷

Figure 7: Illustration of the DePuy Silent Hip (DePuy, Leeds, UK) ¹⁷

This implant was designed specifically for younger, active patients with primary or secondary arthritis, for which hip resurfacing may not be suitable.⁴⁹ It facilitates minimally invasive surgical implantation techniques, enabling preservation of soft tissues and the potential for earlier recovery and rehabilitation. A very high neck cut preserves maximal bone stock, and the tapered stem engages the proximal femur to distribute loads in a physiological manner (Figure 8).¹⁷ Contraindications for this hip implant include poor bone quality or gross deformity in the femoral neck, morbid obesity, and males over the age of 65 years or females over the age of 60 years.⁴⁹

A pilot clinical study used radiostereometric analysis (RSA) to evaluate implant stability, finding acceptable levels of femoral stem migration at two years (<1 mm in any direction), and no evidence of continuing migration at five years to suggest implant loosening.⁵⁰ The Kaplan-Meier survivorship at three years based on stem revision was 97% (95% CI of 94-100%).⁵¹ While these are promising results, further long-term clinical evidence and biomechanical research would still be beneficial in fully understanding how this implant functions within the femur.

Figure 8: Silent Hip implanted in the femoral neck, demonstrating high neck cut ¹⁷

2.3.2 Examination of Orientation

One area of research on conservative hip implants for which there is minimal data is the optimal orientations of these stems within the femur. Biomechanical studies examining the varus/valgus placement of the femoral component of hip resurfacing arthroplasties have shown a valgus orientation to be optimal. It was found that a valgus placement increased the proximal femoral strength of the femur, and so resulted in a higher femoral fracture load.^{52, 53} Valgus placement was also found to induce strain patterns more similar to an intact femur, suggesting physiological loading.⁵⁴ These results were further supported in a study that found valgus to be the optimal coronal alignment of the Birmingham Mid-Head Resection (BMHR) (Smith & Nephew, Marl, Germany) to increase proximal femur strength and provide protection in the case of superior femoral notching.⁵⁵

Another study assessed the effect of femoral head lag screw positioning on proximal femur stiffness and load-to-failure. It was determined that a central position in the lateral plane coupled with an inferior orientation in the anteroposterior plane significantly maximized both axial stiffness and load-to-failure.⁵⁶ However, no prior studies have examined the ideal 3D orientation of conservative short stem hip implants to optimize biomechanical stability.

CHAPTER 3

Thesis Objectives and Hypothesis

3.1 Research Objective and Rationale

There are numerous conservative hip implants on the market that optimize lateral flare and are designed for femoral metaphyseal fixation; however, there is no corresponding biomechanical research on the ideal 3D orientation of these devices in order to enhance long-term mechanical stability of the femur-implant interface and extend the implant's service life. The goal of this study was therefore to determine the optimal 3D angle and position of a straight, short stem hip implant designed for metaphyseal fixation to achieve maximal biomechanical stability.

3.2 Research Question

What is the 3D orientation of a straight short stem hip implant within the proximal femur that will yield a stress distribution most similar to that of an intact femur, thereby providing optimal physiological loading?

3.3 Research Hypothesis

Given that the stem of a straight, short stem total hip replacement, such as the Silent Hip, has a similar geometric structure to hip resurfacing implants, the BMHR, and lag screws, it is postulated that the Silent Hip will be biomechanically similar. It was therefore hypothesized that a valgus and inferior coronal plane orientation of the stem combined with a neutral lateral placement would result in an optimal stress distribution and superior biomechanical stability.

3.4 Specific Aims

This research endeavoured to assess the optimal orientation of a straight, short stem hip implant within the proximal femur through:

A) Biomechanical Testing

- a. Biomechanical testing was completed on implanted femurs that corresponded to six maximum achievable angles and positions of the hip implant, in addition to neutral.

B) Finite Element Analysis (FEA)

- a. Biomechanical tests were used to validate a corresponding finite element (FE) model of the implant-femur construct.
- b. The validated model was then parameterized such that the range of implant angles and positions could be attained.

C) Design of Experiments (DOE)

- a. DOE was used to optimize the parameterized FE model by determining the orientation which yielded the minimum difference from the intact femur in average proximal cortical bone stress.

This study will have clinical implications for the optimal surgical placement of straight, short stem hip implants within the proximal femur. It will also be one of the few existing biomechanical studies on any type of short stem hip implant, and the first study to optimize the 3D orientation for superior biomechanical stability.

CHAPTER 4

Relevant Literature

4.1 Conservative Hip Implant Reviews

There are a few studies which have examined available conservative hip implants, with the purpose of appraising their biomechanical characteristics and short- to mid-term clinical results. In their review, Valverde-Mordt et al.²⁰ determined that currently available prostheses ranged from shortened classic stems (e.g. short modular femoral stem) to metaphyseal wedged stems (e.g. Mayo) (Figure 9). A newer line of implants included ultra-short stem devices with a high neck cut (e.g. Proxima) and prostheses which are designed to be used when a hip resurfacing procedure has failed or is not indicated (e.g. BMHR) (Figure 10). Shortened classic stems were determined to have too few available published results to conclusively establish their primary stability, while metaphyseal wedged stems were found to have a good survival rate with less bone resorption than traditional long-stem prostheses. However, hip implants such as the Mayo stem (Zimmer, Rochester, MN, USA) had reports of perforation of the external cortical bone, intraoperative fractures, and post-operative complications.

Figure 9: a) Short modular femoral stem (Smith & Nephew, Inc., Memphis, TN, USA) ⁵⁷; b) Mayo conservative implant double wedge stem with proximal coating (Zimmer, Rochester, MN, USA) ²⁰

The Proxima (DePuy, Leeds, UK) was found to have early primary stability, greater bone sparing effects, and decreased fracture risk in young patients with good bone quality, as its lateral flare and short stem allowed a conservative high neck cut with metaphyseal fixation. The Birmingham mid-head resection (BMHR) (Smith & Nephew, Marl, Germany) was defined as an intermediate prosthesis designed for osteonecrotic patients and as a secondary implant when a hip resurfacing fails. It consists of a solid metal hemisphere with a porous-coated stem for fixation in the femoral neck, and a metal-on-metal couple with the acetabular component.

Figure 10: a) Proxima Hip (DePuy, Leeds, UK) ⁵⁷; b) Birmingham mid-head resection (BMHR) (Smith & Nephew, Marl, Germany) ⁵⁸

A review completed by Kluge¹⁹ found that implants which have a stem extending into the femoral diaphysis exhibited stress shielding and bone density loss, while implants with proximal load transfer and no distal stem fixation required less intraoperative bone removal, preserved proximal bone stock in the long term, and had better physiological loading. However, it was noted that hip implants relying solely on metaphyseal fixation required sufficiently good cancellous bone quality to ensure primary stability.

Both Kluge and Valverde-Mordt et al. briefly mentioned the Silent Hip, which is a new hip implant similar in design to the BMHR (Figure 11). According to Kluge, this tapered press-fit implant, designed for fixation in the femoral neck and proximal metaphysis, allowed for nearly physiological loading.¹⁹ Valverde-Mordt et al. stated that the Silent Hip is still in the clinical trial phase and therefore no conclusive evidence can be drawn; however, it could prove to be a very versatile implant given its 12/14 taper that allows the surgeon a choice of acetabulum and friction couple.²⁰

Figure 11: Silent Hip (DePuy, Leeds, UK) ¹⁷

4.2 Retrospective Reviews

Oldenrijk et al.⁴⁴ completed a systematic review of the revision rates reported in 49 clinical short stem hip arthroplasty studies. There were 19 different types of stems used overall, which they classified into three stem categories: “collum” (straight neck-sparing stems), “partial collum” (partial femoral neck-sparing), and “trochanter-sparing” (shortened tapered stem but not neck-sparing). Of these stem types, only 3 fell under the collum classification. As such, the clinical evidence for these stems was limited to a small number of studies with mid-term results, evaluating primarily the CUT (ESKA Implants GmbH & Co, Luebeck, Germany) stem. Overall, it was found that partial collum and trochanter-sparing stems displayed adequate mid-term survival rates (above 90% survival at 10 years of follow-up), while collum stems did not.

When looking at some of the individual conservative stem reviews available, the results are slightly more favourable. Choy et al.⁵⁹ completed a retrospective review in which data from the Australian Orthopaedic Association National Joint Replacement Registry was used to compare the survivorship of short Exeter stems with the standard-length Exeter implants (Stryker, Kalamazoo, Michigan, USA). It was found that there was no significant difference in the revision rates between standard and short stems, and short stems had the advantage of being able to fit smaller patient sizes, for whom oversizing and subsequent component loosening and failure is a common concern.

Xing et al.⁶⁰ looked at the effect of stem length on reoperation and complication rates for patients with proximal femur metastases. They retrospectively reviewed 203 patients with proximal femur metastases who had been treated with hip implants ranging in size from short stem (12 – 14 cm stem length) to long stem (25 – 35 cm). It was discovered that short stem implants had a slightly higher reoperation rate than long stems (8.6% as compared to 1.4%) due to aseptic loosening and periprosthetic fracture, which was attributable to poor proximal bone quality. However, long stems were associated with a much higher risk of cardiovascular complications and a correspondingly increased postoperative complication rate. This implies that long stems might not always be indicated depending on the quality of the proximal bone stock.

4.3 Clinical Studies

There are a number of clinical studies that have evaluated the performance of individual conservative hip implants, including custom-made implants. Santori et al.⁴³ published mid-term results for a clinical trial in which 129 custom-made short stem hip implants were inserted into 109 patients. There were two types of hip replacements initially implanted: Type A consisted of implants with a short stem extending no more than 3 cm past the lesser trochanter, while Type B were ultra-short stem implants with a lateral flare shape (Figure 12). The patient criterion for the clinical trial included an age of less than 60 years and good bone stock, and the follow-up period was 11.8 years for Type A implants and 6.9 years for Type B. Radiographs of the hip replacement site were taken immediately post-operatively, and at 3 months, 6 months, 1 year and then annually.

Figure 12: Radiograph of an implanted Type A short stem hip implant (right hip) and a Type B ultra-short stem implant (left hip), with a numerical depiction of the 5 modified Gruen zones for short stem implants⁴³

Santori et al.⁴³ then consolidated patient data and measured the mid-term implant alignment, endosteal spot welds, and proximal bone resorption among other features.⁴³ Stress-shielding only occurred in Type A implants, which also exhibited a decreased bone mineral density in all modified Gruen zones. Conversely, Type B implants showed spot welds in Gruen zone 2 and an increased bone density in zones 2 and 4, indicating that stress was distributed to the lateral proximal femur. These conclusions are supported in a previous study by Santori et al.,²⁵ which determined that the bone mineral density of ultra-short stem implants 2 years post-operatively was nearly identical to the non-operated contralateral hip. This implies that for young patients with good metaphyseal spongy bone, it is possible to rely on this trabecular bone for complete lateral load transfer.

In a prospective cohort study of the collum femoris preserving (CFP) prosthesis (Waldemar LINK GmbH & Co, Hamburg, Germany), Hutt et al.⁴⁸ assessed the Harris Hip Scores (HHS), radiographic lucency, and implant stability of 67 implanted stems at 10 years. They found a statistically significant improvement in HHS from those reported preoperatively (mean of 50 preoperatively and 91 postoperatively), and no radiographic evidence of loosening or osteolysis. Overall, the stems were found to be stable with good bony ingrowth.

Similar measures were evaluated in a clinical study by Carlsson et al.,⁴¹ in which the Gothenburg osseointegrated titanium (GOT) (Figure 13) hip arthroplasty was implanted in 20 patients, following which the HHS, implant migration, and radiographic lucencies at 2 years postoperatively were determined. HHS were high (>90) at 2 years, and radiographic analysis showed only minor resorption under the collar of the femoral component. Radiostereometric analysis (RSA) was used to evaluate implant migration and rotation, and only minimal movement was noted (within the threshold of 1.2mm). These results indicate that the implant has achieved osseointegration in the metaphysis, with good predictions for long-term stability.

Figure 13: Radiograph of an implanted Gothenburg osseointegrated titanium (GOT) arthroplasty⁴¹

Finally, Tóth et al.⁴⁰ presented a study examining the short-term outcomes of the Proxima™ (DePuy, Leeds, UK) short stem hip implant. There were 35 patients included in the clinical trial with a total of 41 implanted hips, and HHS were recorded pre- and postoperatively. Mean 2 year HHS were 88 and radiographic analysis showed no signs of loosening or migration. However, it was noted that 2 stems were implanted in severe varus and 8 in varus orientation, as a result of the implant shape and “round the corner” broaching method.

4.4 Biomechanical Studies

4.4.1 Finite Element Modelling Studies

While many suggest that conservative metaphyseal implants are the optimal solution for bone conservation and proximal load transfer, not all investigations are in agreement. In a study by Van Rietbergen et al.,⁶¹ finite element analysis (FEA) was used to determine whether shortening of the distal implant stem would reduce stress shielding and improve proximal load transfer. A cementless size No. 5 Anatomique Benoist Girard (ABG) stem (Howmedica, Allendale, NJ, USA) was merged with a bone finite element (FE) model. Analyses were carried out on the full model and on a model with the distal portion of the stem given an elastic modulus of 1.0 MPa to remove all mechanical contributions. The probability of interface failure was determined using interface stresses (Figure 14), and bone mass development was found through a remodeling simulation.

Figure 14: Pre-remodeling and post-remodeling shear stresses in the long and short-stemmed reconstructions, illustrating the implant design and short stem model⁶¹

The ABG short stem was found to provide little benefit in the reduction of stress shielding, transferring stresses more proximally instead of distributing them over the length of the diaphysis. It was also found that the probability for interface failure was very low in both cases, conferring no added benefit to the short stem design. However, while the bone remodeling simulation procedure was validated earlier, the failure criterion was not well supported through experimental validations.

Nevertheless, other literature appraises conservative hip implants much more favourably. Dabirrahmani et al.⁴⁷ undertook an FE study to determine the primary stability and long-term bone remodelling of the BMHR stem. A size 3 BMHR implant was modelled and placed in a CT scanned model femur in the correct surgical position. An intact femur model and the reconstructed models were then subjected to a 45% gait cycle loading regime, which included muscle loads and the hip contact force. A strain-adaptive remodelling theory was utilized to generate predictive bone remodelling behaviour, from which bone mineral density (BMD) could be derived. Over a predicted period of 30 months, BMD loss occurred only to the proximal neck region, with an increase in BMD at the distal neck and minimal BMD changes otherwise. Some stress shielding was noted only in the proximal neck of the reconstructed femur model, with strain distribution being otherwise comparable to the intact model (Figure 15). This study determined that long-term behaviour of the BMHR was favourable, with minimal BMD changes and good stress distribution.

Figure 15: Depiction of compressive strain and strain shielding in the intact and reconstructed femur models⁴⁷

4.4.2 Experimental Studies

There are a number of experimental studies that directly compare hip implants with a long stem versus those with a short stem. In a study by Gronewold et al.⁴² synthetic femurs were implanted with the short cementless METHA stem (Aesculap, Tuttlingen, Germany), and a traditional Bicontact stem (Aesculap, Tuttlingen, Germany). Strain gauges were affixed to key locations along the proximal femur, and the implanted femurs were loaded under 6 loading conditions that corresponded to the potential physiological range of hip-joint angles. Recorded strains were taken as a percentage of the measured intact femur strains, and these were then compared between the two stem models. The METHA stem was found to consistently elicit strains similar to the intact model, indicating reduced stress-shielding and good metaphyseal anchorage, while the Bicontact stem had dramatically decreased strains in the calcar region when compared to an intact femur.

In a similar study by Decking et al.³¹ three clinically successful implants – Alloclassic Zweymüller SL (Centerpulse, AG, Winterthur, Switzerland), OPTAN (Centerpulse AG, Winterthur, Switzerland), and CUT (ESKA Implants GmbH & Co, Luebeck, Germany) – were

implanted in pairs of human cadaver femurs affixed with rosette gauges. The femurs were loaded up to 2000N in single-leg stance and the strains were recorded both before and after implantation. The two traditional stems were found to produce a reduction in principal surface strains in the proximal femur, while the ultra-short stem CUT prosthesis exhibited increased cortical strains in the proximal subtrochanteric region of the femur with otherwise minimal changes (Figure 16). While this indicates that overall proximal stress reduction can be mitigated through use of an ultra-short stem implant, the effect of the increased subtrochanteric strains cannot be determined without long-term clinical follow-up.

Figure 16: Depiction of the changes in the absolute value of the major principal strains (in % of the intact femur value) after implantation of the Alloclassic (left), OPTAN (middle), and CUT (right)³¹

Finally, in a study by Bieger et al.,³⁶ 9 pairs of human cadaver femurs were implanted with a long, straight CBC stem (Mathys, Bettlach, Switzerland) and an optimys short stem (Mathys, Bettlach, Switzerland). Inductive miniature displacement transducers were used to measure implant motion over 100 000 axial load cycles to simulate 6 weeks *in vivo*, while rosette gauges were used to measure proximal femur stress. It was found that both implants exhibited primary stability with reversible micromotions less than 150µm; however, while both stems also showed stress shielding, the short stem induced a more physiological stress distribution when compared to an intact femur.

4.4.3 Implant Orientation Studies

With the development of these new conservative hip implants comes a change in the required surgical techniques and placement goals. For example, Santori et al.²⁵ generated a surgical implantation technique for their ultra-short stem implant dubbed “round the corner,” which

involves curving the implant into the femoral neck below the greater trochanter. Surgical implant positioning is another key component of any hip replacement surgery, thereby necessitating studies to determine the optimal placement of this new wave of conservative implants.

Speirs et al.⁶² used an FE model to determine the effect of stem position and orientation on loading of the femur when compared to an intact femur FE model. A Nanos short stem hip implant (Endoplast GmbH, Germany) was virtually implanted into an FE model of a standardized femur to yield 3 different orientations: reference, medialised and anteverted. The reference model was defined by aligning the implant with the femoral joint centre to agree with the intact model, while the medialised model placed the implant 6 mm medially, 2 mm posteriorly and 4 mm superiorly, and the anteverted model rotated the implant about the femoral shaft axis by 7° to give an 11° anteversion angle. A previously validated musculoskeletal model was then used to determine muscle and joint contact forces for walking and stair climbing, which were subsequently applied to each implanted model and an intact femur model. Resultant strain energy density was determined for each Gruen zone (Figure 17). Stress shielding was seen in the proximal femur for all implanted models when compared to the intact femur, and was found to be relatively insensitive to implant offset or anteversion. They concluded that surgical misplacement of a short-stemmed implant would have minimal effect on the dispersion of internal loads, and these variations are small when contrasted to the strain distribution in an intact femur.

Figure 17: Depiction of cortical strain energy in each Gruen zone for all solid models under walking and stair climbing loads. Intact refers to the intact femur model, Ref is the reference implanted model, Med corresponds to the medialised model, and AV is the anteverted model⁶²

Radcliffe et al.⁵⁴ determined the effect of a hip resurfacing stem orientation on load transfer in an intact femur, with a particular focus on the varus/valgus orientation. FE models of 16 subject-specific CT generated femoral solid models were used, with 4 models per femur: intact, neutral, varus by 10° and valgus by 10°. The models were then loaded with a joint contact force and an abductor muscle force, and the percentage change in strain between the intact and implanted models was determined. All implanted femurs exhibited some stress shielding in the femoral head (Figure 18); however, the varus implantation angle least resembled the strain distribution seen in the intact femur, as determined through evaluation of the percent change from the intact femur in specified proximal zones. They concluded that the valgus orientation was most comparable to the intact femur, and hence was most desirable.

Figure 18: Strain distributions for a single femur model in the: a) intact femur, b) resurfaced in varus, c) resurfaced in neutral, d) resurfaced in valgus⁵⁴

4.5 Summary of Findings

With the increasing prevalence of young patients requiring hip replacement surgery, there has been a corresponding increase in the need for more bone conserving hip implants.¹⁹ Several reviews have evaluated the conservative implants currently on the market,^{19, 20, 44, 59} while others have specifically examined the physiological and/or biomechanical advantages of conservative implants.^{25, 31, 36, 40-43, 47, 48, 54, 60-62} It was determined that ultra-short stem implants have the most favourable stress distributions and hence the most beneficial bone remodeling outcomes, with the added benefits of a less invasive surgical implantation technique and increased bone stock preservation as compared to traditional implants. Surgical placement was also evaluated, and it was shown that a valgus orientation is desirable for hip resurfacing procedures.⁵⁴ However, while the BMHR has been evaluated for its desirable physiological loading characteristics, the new and comparably designed Silent Hip has yet to undergo any biomechanical analysis. In particular, the optimal 3D surgical orientation of such an implant has not yet been studied.

CHAPTER 5

Research Methods

5.1 General Approach

This is a two part comparison study involving biomechanical testing and finite element modelling. Initially, synthetic femurs implanted with a straight short stem in a specified angle or position were instrumented with strain gauges and mechanically tested in axial compression at a subclinical load of 1000 N. The femurs with a neutrally implanted stem were further tested under extension and flexion conditions in order to capture loading throughout the range of the gait cycle. The strain gauge results of this biomechanical testing were then used to validate an FE model of the implant-femur construct with simulated loading of 1000 N. Design of experiments (DOE) was implemented using the FE model to evaluate the range of potential implant angles and positions, and determine the optimal 3D implant orientation that generated a stress distribution most similar to the intact femur model. This was completed at 3 phases of the gait cycle, and the resultant optimal orientations were compared.

5.2 Mechanical and Material Properties

5.2.1 Synthetic Femurs

This study utilized 10 large, left, fourth-generation composite femurs (Model #3406, Pacific Research Laboratories, Vashon, Washington, USA). These synthetic femurs featured cortical bone made using a mixture of short glass fibers and epoxy resin, pressure injected around a solid rigid polyurethane foam cancellous core.⁶³ The mechanical and material properties for synthetic femurs can be chosen to fully match the characteristics of a normal human femur, or to partially match the properties of an osteopaenic or osteoporotic femur. Density and strength values corresponding to a normal human femur were selected for this study (Table 2). The densities given for synthetic cortical and cancellous bone both corresponded with human femoral bone properties (Table 1). The mechanical properties have been similarly correlated with human ranges, having exhibited mechanical bending and torsional behaviour comparable to human cadaveric femurs.^{64, 65}

Table 2: Mechanical and material properties for synthetic femur bones⁶³

Bone	Loading Mode	Density (g/cm ³)	Ultimate Strength (MPa)	Young's Modulus (GPa)	Poisson's Ratio
Cortical	Longitudinal Tension	1.64	106	16	0.26
	Longitudinal Compression	1.64	157	16.7	0.26
	Transverse Tension	1.64	93	10	0.26
Cancellous ("Solid Type")	Compression	0.27	6	0.155	0.3

The chosen synthetic femurs had a standard geometry with a femoral head diameter of 52mm, a length of 485mm, a mediolateral midshaft outer diameter of 32mm and a cancellous-filled intramedullary canal diameter of 16mm (Figure 19).⁶³

Synthetic femurs have been chosen for use in this study as they have a number of benefits over human cadaveric femurs. Human cadaver femurs have large interspecimen variation in geometric and material properties, thereby decreasing experimental reproducibility. By comparison, synthetic femurs have an interspecimen variability 20-200 times lower, providing a much more reliable test sample and a standard geometry for use with accompanying FE models. Furthermore, synthetic femurs have a lower cost, no special storage requirements, and no toxicity.⁶⁴

Figure 19: Large fourth generation composite femur and femur cross-section (Model #3406, Pacific Research Laboratories, Vashon, Washington, USA). Sizing is as follows: a) 485 mm, b) 52 mm, c) 37 mm, d) 120°, e) 32 mm, f) 93 mm, g) 16 mm canal⁶³

5.2.2 Hip Implant Properties

The hip implant that was chosen for use in this study is the Silent Hip (DePuy, Leeds, UK). Given the dimensions of the synthetic femurs, an optimally sized Silent stem of 22mm diameter and 50mm length was selected. The femoral head component consisted of a 36mm 12/14 Articul/eze + 12 ceramic head.

Through scanning electron microscopy (SEM) and energy-dispersive x-ray spectroscopy (EDX), the surface features and material properties of the implant were determined. SEM allowed visualization of the surface of the implant (Figure 20), elucidating the beaded coating on the stem into which it can be assumed that bone would mechanically interdigitate.⁶⁶

Figure 20: SEM of the Silent Hip stem, depicting the sintered calcium phosphate bead coating at x30 magnification

From EDX it was found that the stem is composed of a titanium alloy, Ti-6Al-4V (Figure 21), with a beaded calcium phosphate coating (Figure 22).

Figure 21: EDX of the stem neck region, illustrating the Ti-Al-V composition of the stem

Figure 22: EDX of the stem body region, indicating the composition of the calcium phosphate beaded coating

The femoral head is comprised of an aluminum oxide matrix composite ceramic called BIOLOX[®] *delta*, which consists of 82% alumina (Al₂O₃), 17% zirconia (ZrO₂) and additional trace elements such as chromium oxide (Cr₂O₃), which imparts the pink colour (Figure 23).⁶⁷

Figure 23: Expanded scale EDX of the ceramic femoral head, indicating the aluminum oxide matrix composition

Due to sintering of the beads used for the calcium phosphate coating, there was a high stiffness modulus throughout the implant. Assuming complete mechanical interdigitation with the implant, it was therefore possible to allow the mechanical properties of the implant to be governed by the bulk properties of the material.⁶⁶ The material properties of the stem were therefore set to those of the titanium alloy from which its bulk is composed. The material properties that were used in the FEA have been further verified with the manufacturer (Table 3).

Table 3: Material properties of the ceramic femoral head and titanium alloy stem⁶⁸⁻⁷⁰

Material	Young's Modulus (GPa)	Poisson's Ratio (ν)
BIOLOX [®] <i>delta</i> ceramic (Femoral head)	358	0.24
Titanium alloy (Ti-6Al-4V) (Silent stem)	110	0.31

5.3 Biomechanical Testing

The main objective of this section was to understand the experimental strain distribution within the femur at 7 maximum implanted orientations of the Silent Hip, at 3 phases of the gait cycle. These results were then used to validate corresponding FE models of each implant orientation.

5.3.1 Orientation Definition

It was determined that 7 experimental implant positions should be tested in order to ensure that the maximum range of clinically realistic potential orientations within the FE model was valid. The stem was therefore implanted in neutral, maximum valgus, maximum varus, maximum anteversion, maximum anterior, maximum posterior, and maximum inferior. Through discussion with an experienced orthopaedic surgeon, it was concluded that a retroverted implant angle or superior positioning would seldom be surgical goals, and hence these orientations were excluded from the current study.⁷¹ The following maximum orientations were determined experimentally through ascertaining the maximum translation or angulation possible while ensuring the cortical bone was not breached. This ensured that the stem could still be implanted using a press fit, which is required for primary stability.⁴⁹

The neutral orientation was defined as a position that maintained the natural collum-caput-diaphyseal (CCD) angle of the synthetic femur in the anteroposterior (A/P) view (Figure 24), while also maintaining a neutral lateral angle and position (Figure 25).

Figure 24: A/P view of the Silent stem neutrally implanted within the femur, where the solid red line indicates the long axis of the femur and the dashed line represents the implant axis at a CCD angle of 120°

Figure 25: Lateral view of the Silent stem neutrally implanted within the femur, where its neutral angle and position are displayed through its central location between lines tangent to the cortical bone

The maximum valgus implant orientation was determined to be at a CCD angle of 130° , with a neutrally angled and positioned lateral implant orientation (Figure 26).

Figure 26: A/P view of a Silent stem implanted in 130° valgus

The maximum varus orientation was similarly oriented laterally; however, it featured an implant CCD angle of 110° (Figure 27).

Figure 27: A/P view of a Silent stem implanted in 110° varus

The maximum anteversion angle was defined as being a stem-neck angle (SNA) of 10°, with a neutral CCD angle (Figure 28).

Figure 28: Lateral view of a Silent stem implanted in 10° anteversion

The maximum anterior position was determined to be a distance of 2mm from the neutral line in the lateral view (Figure 29).

Figure 29: Lateral view of a Silent stem implanted in 2mm anterior

The maximum posterior position was similarly found to be at a location 2mm from the neutral line in the lateral view (Figure 30).

Figure 30: Lateral view of a Silent stem implanted in 2mm posterior

Finally, the maximum inferior orientation was found to be located 4mm vertically down from the neutral CCD angle line in the A/P view (Figure 31).

Figure 31: A/P view of a Silent stem implanted in 4mm inferior

A summary of the tested angles and positions can be found in Table 4, which details the chosen implant angle or position as described in each view.

Table 4: Planned angles and positions in the A/P and lateral planes for each of the 6 maximum angles and positions, and neutral

Orientation	A/P Angle (°)	A/P Position (mm)	Lateral Angle (°)	Lateral Position (mm)
Valgus	130	0	0	0
Varus	110	0	0	0
Anteversion	120	0	10	0
Anterior	120	0	0	2
Posterior	120	0	0	2
Inferior	120	4	0	0
Neutral	120	0	0	0

5.3.2 Sample Size

As the biomechanical tests in this study were conducted purely for the purpose of FEM validation, it was determined that only one femur per orientation was required. However, to quantify the repeatability of the surgical implant method, the neutral orientation was implanted in

3 femurs. Furthermore, an intact femur was prepared and tested to provide a baseline measurement and to allow for validation of an intact model. Overall, there were therefore 10 synthetic femurs used in this study.

5.3.3 Software-Guided Surgery

5.3.3.1 Alignment of Hip Resurfacing Arthroplasties

In order to get the degree of implanted accuracy required by this study, computer-assisted navigation was used in place of a goniometer for initial guidewire placement. This method has been shown to greatly decrease the error in guidewire placement when compared to conventional instrumentation during hip resurfacing.^{72, 73} Software-guided surgery is often used for implant positioning in hip resurfacing arthroplasties, which feature a femoral component with a small straight stem attached to the femoral head (Figure 32). In typical Birmingham Hip Resurfacing (BHR) (Smith & Nephew, Marl, Germany) arthroplasties, a 2.4mm guidewire is initially inserted into the femur along the planned surgical path, following which a reamer is used to enlarge the hole in preparation for implant placement.⁷⁴ Due to its similarly straight shape and a matching initial reamer size of 8mm, it was possible to use the BHR software-guided surgery system for initial orientation of the Silent stem.

Figure 32: Depiction of the Birmingham Hip Resurfacing (BHR) implant (Smith & Nephew, Marl, Germany)⁷⁵

5.3.3.2 BrainLAB System for Imageless Navigation

Guidewire placement for orientation of the Silent stem was completed using the BrainLAB Hip Essential system (BrainLAB, Feldkirchen, Germany). This system utilizes two infrared cameras

to triangulate the location of passive reflective markers in a process called optical motion capture. Initially, an optical array of 3 calibration markers was attached to the femur at the lesser trochanter using a 5mm threaded pin (Figure 33).⁷⁴ This established a global coordinate system for localization of the femur and surgical tools in 3D space.

Figure 33: Depiction of an optical array of calibration markers attached to the intact femur
The femur size was then registered using an angled pointer equipped with a marker set, which enabled identification of four particular landmarks: the medial and lateral epicondyles, the piriformis fossa, and the head-neck junction.⁷⁴ Points around the femoral head were also taken to establish the femoral head size. Through precise initial calibration, the femur size and location can be established to within $\pm 1^\circ$ of accuracy. Furthermore, following registration of the femur, guidewire placement can be tracked in real-time.

The BrainLAB software automatically generates a surgical plan based on the initial registration, which can then be modified according to user requirements. It was therefore possible to move the guidewire placement to a location or angle corresponding to each of the predetermined orientations. Within the surgical planning screen, an A/P section allowed modification of the CCD angle and superior/inferior implant position, a lateral section enabled alteration of the version angle and anterior/posterior placement, and finally a section that included a cross-section of the femoral neck allowed verification of neutral alignment (Figure 34).

Figure 34: Screenshot of the surgical planning screen provided by the BrainLAB Hip Essential software

Once the desired surgical path had been selected, the guidewire was drilled into the femur through a drill guide, which was equipped with a marker set to allow for alignment of the guidewire along the planned path (Figure 35).

Figure 35: Image of the registered femur undergoing guidewire placement using a drill and marker-equipped drill guide

A real-time feedback system on the computer screen ensured that the guidewire was drilled to within $\pm 1^\circ$ of accuracy of the planned surgical path. Once the drill had been placed to an adequate depth within the femur, a verification screen was presented which indicated both the planned and achieved surgical paths and the difference between the two (Figure 36). This provided confirmation of the guidewire placement accuracy.

Figure 36: Screenshot of the planned and achieved surgical paths for a neutrally aligned guidewire

5.3.4 Silent Stem Implantation

Following guidewire placement for each of the 7 planned orientations (including 3 placements in neutral), a cannulated reamer from the BHR surgical set was used to ream overtop of the guidewire and enlarge the initial cavity. The Silent Hip surgical technique was then followed in order to implant the Silent stem.⁷⁶ Following head resection, this technique utilizes sequential reamers in order to systematically ream the implant cavity to the desired diameter, without significantly altering the planned orientation (Figure 37).

Figure 37: Reamers included in the Silent Hip surgical technique

A depth arm in addition to a depth gauge can be utilized to ensure reaming to an adequate depth for the intended implant. In the Silent Hip surgical technique, it is recommended that the final depth be greater than the implant length, as the implant should sit proud of the neck isthmus by 3mm and should include a 3mm distal gap (Figure 38). This ensures optimal implant support proximally, and reduces the potential for distal loading.⁷⁶

Figure 38: Silent stem placement using the implant impactor instrument, indicating the final seating height and the 3mm distal gap

However, the femurs used in this experiment had to be uniformly resected in order to allow for a single validated FE model. It was therefore not possible to use the calcar mill to ensure a final resection plane at the angle of the implant. The femurs were therefore uniformly cut along a resection line located at approximately 38mm from the proximal tip of the femoral head, and at an angle of 30° from the longitudinal axis of the femur (Figure 39), which was determined to be a conservative resection line for a neutrally implanted stem. At this point, the distal condyles of the femur were also removed to allow for subsequent distal fixation (Section 5.3.6.1). For this study, the implant cavity was reamed to a diameter of 22mm, and a depth of 50mm. The Silent stems were then impacted to a point at which no portion protruded more than 3mm from the resection line. Given that the surgical technique recommends only an approximate final seating height of 3mm, it was not expected that either the protrusion of the stem or the inclination of the final resection plane would have a significant impact on the final results.

Figure 39: Model of the intact femur, where the solid red lines indicate the respective resection lines and the dashed line indicates the longitudinal axis of the femur

5.3.5 Orientation Verification

Implant placement was verified through plain film radiographs of the A/P and lateral positions of the implanted femurs. This was completed following guidewire placement to initially confirm the desired orientation, and again following implant placement to identify the final implant

orientation. The CCD angle was determined by drawing a line from the piriformis fossa down the longitudinal axis of the femur, and then measuring the angle to a line through the axis of the implant (Figure 40).

Figure 40: A/P view of a neutrally implanted femur, indicating the neutral 120° CCD angle formed by the long axis of the femur and the implant

The lateral angle of the femur was measured using the computer-assisted templating software EndoMap (Siemens AG, Medical Solutions, Erlangen, Germany). This software allows the user to load DICOM radiographs into the program, following which the images can be scaled, manipulated, and measured. A modified version of the Resubal method of measuring the stem-neck angle (SNA) was employed in this study.⁷⁷ While the Resubal method uses two lines tangent to the anterior and posterior cortical bone of the neck to determine a neutral axis, this modified method involves drawing two circles that pass through points along the cortical bone corresponding to the most proximal and distal regions of the neck. A line is then drawn between the centre points of each circle, which corresponds to the neutral axis of the neck.⁷⁸ The angle between this axis and the implanted stem axis is defined as the SNA, or the lateral angle (Figure 41).

Figure 41: Lateral view of a neutrally implanted femur, depicting the modified Resubal method for determination of implant neutrality: a) tangent circles drawn on femoral neck; b) line drawn through circle centre-points, and distance to axis of implant measured

Implant positioning within the femoral neck was also determined using the A/P and lateral radiographs of the femurs. In the A/P x-rays, a line located centrally within the isthmus of the neck was drawn at a CCD angle of 120° . The vertical distance from this line to the implant axis was then measured. Similarly, for the lateral radiographs, a neutral line was drawn centrally within the isthmus of the femoral neck and the perpendicular distance to the axis of the implant was determined.

5.3.6 Femur Preparation

5.3.6.1 Potting

In order to ensure a stable, fixed base for testing, the femurs were potted in cement-filled steel cubes measuring 88mm x 88mm x 75mm. Due to both machine constraints and to allow angling of the femur within the cube, the distal condyles of the femur were removed. Initially, lines were drawn along the base of the femurs at a level 60mm up from the base of the condyles (as determined by locating the condyles against a flat surface). A mitre saw was then used to remove the condyles at this preset resection level, along with the previously discussed proximal femoral head resection level.

Duct tape was placed along the base of the steel cubes to provide an enclosed box, and the femoral shafts were oriented vertically and held in place using a multiaxial clamping system. Each femur was angled at 7° adduction within the coronal plane, and 0° of flexion/extension in the sagittal plane in order to simulate single-legged stance.⁷⁹⁻⁸² A magnetic angle level was used to ensure accurate positioning in each plane (Figure 42). A cement mixture made using commercially available anchoring cement (Flow-stone, King Packaged Materials Company,

Burlington, ON, Canada) was then poured into each steel cube up to the brim and allowed to solidify for 24 hours, fixing the distal portion of each femur.

Figure 42: Synthetic femur located in a steel cube in preparation for potting, indicating the desired 7° of adduction

5.3.6.2 Strain Gauge Mounting

All of the specimens were instrumented with five 350Ω uniaxial linear strain gauges (Model CEA-06-125UW-350, Vishay Measurements Group, Raleigh, NC, USA) and one 350Ω rectangular rosette gauge (Model CEA-125UY-350, Vishay Measurements Group, Raleigh, NC, USA). On the medial femoral shaft, 2 linear gauges were located on the femur surface at 234mm and 189mm above the top of the cement potting cube, while the lateral shaft had 3 linear gauges placed at 234mm, 189mm and 144mm above the steel cube. The rosette gauge was attached to the anterior side of the femur at 280mm above the steel cube in order to measure the multidimensional strains of this area (Figure 43). Notwithstanding the practical problems of

proper gauge placement on the highly complex geometry of the proximal femur, experimental gauge tests using a similar number and location of gauges were found to validate the FE model in numerous prior studies.⁸³⁻⁸⁶

Figure 43: Gauged femur, indicating the location of the medial linear gauges (1 and 2), lateral linear gauges (3, 4, and 5), and the rosette gauge

To ensure a good interface between the gauge and the femur, the chosen location for each gauge was initially abraded using sandpaper, and then cleaned with the manufacturer provided cleaning solution and conditioner.⁸⁷ Each linear gauge was aligned such that its long axis was parallel to the longitudinal axis of the femoral shaft, and the rosette gauge was placed along a line perpendicular to the femoral shaft. An adhesive was used to affix each gauge to the femur. Wire leads were then soldered to the solder tab areas of each gauge, and secured to the femur using duct tape.

5.3.7 Biomechanical Testing

The direction and magnitude of loading affects stress development in the proximal femur and hence must be considered when determining the optimal stress distribution.⁸⁸ It was therefore concluded that the 3 neutrally implanted femurs would undergo testing at hip angles corresponding to the maximum bounds of the gait cycle, providing repeatability measurements. Due to Instron machine size restrictions, 15° of sagittal plane hip flexion was used to simulate the heel strike condition, 15° of hip extension was used for the toe-off condition, and 0° was used for the full single-leg stance phase. All of these values fall within or near the range limits for a typical gait cycle.^{80, 81, 89}

Initially, implanted femurs were rigidly mounted at their distal potted end into an Instron 8874 tester (Instron, Norwood, MA, USA) using an industrial vice. For axial mode testing, the load cell had a $\pm 25\text{kN}$ capacity, 0.1N resolution, and $\pm 0.5\%$ accuracy. All of the femurs were loaded axially in neutral (flexion=extension=0°) 7° adduction single-leg stance. During testing of the 3 neutral femurs, the vice was manipulated into an angle of 15°, to allow for the flexion and extension conditions. The soldered leads connected to each strain gauge were attached to an 8-channel Cronos-PL data acquisition system (IMC Mess-Systeme GmbH, Berlin, Germany) using crocodile clips (Figure 44). This system was networked with a computer for storage and data analysis.

Figure 44: Experimental set-up, depicting the femur locked into a neutral orientation within the Instron machine and attached to the Cronos data acquisition system (DAS)

For load application, a 3 block load control program was developed using WaveMaker Editor Software. The first block was set to load the femur to 1000N of compression in 10 seconds, the second block held the load at this value for 90 seconds, and the third block unloaded the femur in 8 seconds and then held for 2 seconds to allow the femur to completely unload. Load control was used to ensure each femur experienced a consistent load level, allowing direct comparison of strain gauge readings to the FE model. It was determined that a subclinical load of 1000N would be applied, as this guaranteed that the femurs would remain in their linear elastic range and so would not undergo any permanent deformation or damage.⁸³ Furthermore, the quasistatic loading scenario engendered very low strain rates, thereby mitigating the potential for fracture. Vaseline

was applied to the femoral head to decrease any friction effects at the point of load application. Loads were then applied to the apex of the femoral head using a flat stainless steel plate to allow free movement of the head and avoid generation of any unanticipated lateral loads.⁹⁰⁻⁹²

At the start of testing, femurs were subjected to a preload of 50N to minimize any mechanical slack. They were then tested using the previously defined load program. The raw data files obtained from the data acquisition system during the central 30 seconds of testing were determined to be the most stable, and hence were inputted into FAMOS V5.0 software (IMC Mess-Systeme GmbH, Berlin, Germany) to generate corresponding average strain values over this period. Each implant orientation and the intact femur were tested 3 times consecutively to give an average of the strains developed at each strain gauge location. The 3 neutrally implanted femurs were additionally tested under extension and flexion conditions. Stiffness values were determined for each loading condition by calculating the slope of the resultant force-displacement graph. Such mechanical testing protocols have been used previously in a number of similar studies with good outcomes.^{65, 83-86, 90, 93}

5.3.8 Strain Gauge Calculations

The means of the strains generated at each gauge location over the course of the 3 trials were determined for every implant orientation and the intact femur. The relative standard deviation (RSD) for each average value was below 10%, indicating a low degree of disparity. For ease of comparison to the FE model, average rosette gauge strains were first converted into principal strains (Equation 1), following which they were transformed into von Mises strains (Equation 2). The von Mises strain value for each femur was directly compared to the equivalent strain value at the rosette gauge location in the FE model, while the average linear strain results were compared to normal strains at corresponding locations in each respective FE model. Von Mises strains were used to allow for a single comparison measure, as has been done previously with von Mises stresses, increasing the consistency of manually selecting the strain gauge location within the FE model.⁹⁴

Equation 1: Calculation of the principal strains (ϵ_1 and ϵ_2), where ϵ_A , ϵ_B and ϵ_C correspond to strains from sections 1, 2 and 3 of the rosette gauge

$$\epsilon_{1,2} = \frac{\epsilon_A + \epsilon_C}{2} \pm \frac{1}{\sqrt{2}} \sqrt{(\epsilon_A - \epsilon_B)^2 + (\epsilon_B - \epsilon_C)^2}$$

Equation 2: Calculation of the von Mises strain (ε_{VM}), where ε_1 and ε_2 correspond to the previously calculated principal strains

$$\varepsilon_{VM} = \frac{2}{3} \sqrt{\varepsilon_1^2 - \varepsilon_1 \varepsilon_2 + \varepsilon_2^2}$$

5.4 Finite Element Analysis

The finite element method (FEM) is a discretization technique used for structural mechanics. It functions by subdividing a mathematical model of an object into disjointed units called elements, the geometry and degrees of freedom (DOF) for which are defined by their nodes. Each element contains its own set of discrete response equations depending on its local properties, such as imposed constraints. The overall mathematical response to a stimulus is then determined through approximation of the entirety of its finite elements.⁹⁵ This is akin to the idea of approximating a circle by connecting many small straight lines. The overall precision of a model is governed by the fineness, or number of elements/nodes, within its mesh. A finer mesh will generate more accurate modelling solutions; however, it also requires more computer processing power and takes longer to solve.⁹⁶

5.4.1 Generation of Solid Models

The first step in FE modelling is development of the geometric models. This can most easily be accomplished using 3D computer-aided design (CAD) software, such as SolidWorks. For this study, SolidWorks Standard 2014 CAD Software (SolidWorks Corp., Dassault Systemes, Concord, Waltham, MA, USA) was utilized.

5.4.1.1 CAD Model of the Implant

Due to the symmetric nature of the Silent Hip, 3D laser scanning was not required for development of the implant geometry. It was instead possible to reliably recreate the stem in SolidWorks by drawing a lengthwise, dimensioned cross-section of half of the stem, and then using the revolve feature to make the model 3D. A series of chamfers and fillets were also applied to the stem neck region to ensure that the geometry precisely matched that seen in the Silent Hip. The femoral head used in this study was similarly constructed, involving first a sketch of a dimensioned half cross-section, followed by use of the revolve feature. Cut-revolve was used to remove the centre of the head and allow for alignment with the neck of the Silent stem. Finally, a chamfer was used to attain the graded edge seen in the actual femoral head component.

In order to allow for eventual Boolean subtractions in the FE model, a body was added to the tip of the stem to represent the 3mm distal gap, while another was added to the neck of the stem to ensure the simulated surgical entry point was completely cleared of cancellous bone (Figure 45).

Figure 45: Silent Hip model fitted to the femoral head, indicating location of extra bodies in blue for eventual Boolean operations in ANSYS

5.4.1.2 CAD Model of the Synthetic Femur

The femoral model used for this study was a previously validated computed tomography (CT)-scanned model of the left, 4th generation synthetic femur.⁹³ This model was created by obtaining cross-sectional slices of the femur at every 0.5mm along its length, stocking the images in the DICOM format, and then importing them into Mimics[®] Medical Imaging Software (The Materialise Group, Leuven, Belgium) to allow for 3D model generation.^{83, 93, 97} Surfaces corresponding to both cancellous and cortical bone boundaries were obtained and hence the femur model consisted of both cancellous and cortical bone bodies. This model was imported into SolidWorks for assembly with the Silent stem and head models.

5.4.1.3 SolidWorks Assembly

Once all of the models had been assembled in SolidWorks, additional modifications were completed to match the compiled model to experimental conditions. Firstly, it was determined that the CT scan based femur model had no intramedullary canal, thus one was created by using the cut-extrude feature to remove a dimensioned portion of the cancellous bone. The cut-extrude feature was further used to remove the head and condyles of the femur at levels corresponding to

the experimental resection lines. A drawing was also projected onto the base of the femur at the height and angle of the cement cube to permit fixation of these faces during FE modelling. The stem and head were aligned and mated along their axes, then positioned neutrally within the proximal femur to simulate the neutral implant orientation. Finally, a rectangle replicating the flat loading plate was generated and placed proximal to the femur for eventual simulations of experimental loading conditions (Figure 46). The completed assembly was subsequently imported into the FEA software.

Figure 46: CAD model of the implanted femur assembly with the Silent stem in neutral

5.4.2 ANSYS Workbench for Finite Element Analysis

The software used for FEA in this study was ANSYS Workbench 15.0 (ANSYS Inc., Canonsburg, PA, USA). In general, static structural FE models were made to imitate each implant orientation for validation purposes. A number of additional models were then created to reproduce the experimental neutral, flexion, and extension conditions. These were subsequently modified to meet the run requirements specified in the DOE, allowing for individual DOE assessment of each.

5.4.2.1 ANSYS Engineering Data

Material properties can be set within the Engineering Data tab in ANSYS Workbench. This tab allows you to add materials and then define their properties, for use throughout the rest of the models. In order for ANSYS to solve linear static structural analyses, it requires the Young's modulus (E) and Poisson's ratio (ν) for each material. The values used in this study were the previously determined material properties for each material (Table 5).

Table 5: List of the Young's modulus and Poisson's ratio used for each material in the ANSYS Workbench simulations

Material	Loading Mode	Young's Modulus (GPa)	Poisson's Ratio
Cortical Bone	Longitudinal Compression	16.7	0.26
Cancellous Bone	Compression	0.155	0.3
BIOLOX [®] <i>delta</i> ceramic (Femoral head)	N/A	358	0.24
Titanium alloy (Ti-6Al-4V) (Silent stem)	N/A	110	0.31

As discussed in Section 2.1.2, bone is an inhomogeneous anisotropic material that is stronger in compression than in tension. While it is possible to assign inhomogeneous anisotropic properties to bone in an FE model, sawbones are homogeneous and transversely isotropic, and thus are better represented by assuming homogeneous linearly elastic continuum properties with isotropic behaviour.⁹⁸ Furthermore, only one Young's modulus can be assigned for any given loading direction, and thus the Young's moduli corresponding to the compression condition was selected for this model as it was assumed to be the most likely loading state. These assumptions are commonly used when modelling synthetic bone, with good results.^{79, 93, 99, 100} The inherent viscoelasticity of bone was also not included in the FE model as the static loading condition produced a strain rate so low that it had a negligible effect upon the bone stiffness.

5.4.2.2 ANSYS DesignModeler

The model generated in SolidWorks was saved as a STEP file, which is the ISO standard for representing CAD models and allows for dissociation of the ANSYS imported file from the original modifiable file. The saved CAD model was then imported into ANSYS DesignModeler, and the global coordinate system was defined to be consistent with the original SolidWorks model.

ANSYS DesignModeler utilizes a tree outline with each successive geometry modification listed in order of its activation (Figure 47). For this study, an axis was first defined along the shaft of the femur, to enable evaluation of the longitudinal strains. Body transformations corresponding to each of the potential orientations were then generated. Three translational transformations were defined along planes corresponding to the posterior, anterior and inferior positions. Three rotational transformations were then defined using components to describe the valgus, varus and anteversion angles. In each orientation definition, it was ensured that the stem protruded from the proximal femoral resection by no more than 3mm.

Boolean subtractions were used to remove any cancellous and cortical bone that overlapped with the stem. At this point, the extra bodies placed at the top and the tip of the stem were also subtracted to create a 3mm distal gap, and to remove any proximal cancellous bone that may have overlapped with the stem when its angulation caused recession.

Finally, the direction of loading was simulated by maneuvering the flat loading plate into a position representative of 7° of femoral adduction, and neutral sagittal plane angulation (except for those models verifying the experimental flexion or extension conditions). The plate was then translated to a position just above the apex of the femoral head. At this time, a small circle with a diameter of 3mm was drawn onto the base plane of the loading plate and projected onto the head to simulate the point of load application. The local coordinate system of the plate at this location was then used to define the direction of load application.

Figure 47: Screenshot of ANSYS DesignModeler, depicting the tree outline with each modification step

5.4.2.3 ANSYS Mechanical

Once the geometry had been defined, ANSYS Mechanical was used to set all of the boundary conditions and constraints, generate the mesh, and ultimately solve the model.

Geometry

Within the geometry tab, the user can define each of the bodies within the model and set their corresponding material properties. Each body was therefore assigned to its respective material as defined previously in Engineering Data. Furthermore, the load plate body was suppressed for this study, as only the load application point and its coordinate system were required for analysis.

Coordinate Systems

A number of coordinate systems were either imported from ANSYS DesignModeler or defined for meshing purposes. The longitudinal axis coordinate system and the load plane coordinate system were imported from DesignModeler, while additional coordinate systems in the neck and body of the stem were defined for eventual use in mesh generation.

Contacts

When assemblies of solid bodies are imported, contact regions between each body are automatically assigned. Contact definition allows separate bodies to have different sized and thus non-matching meshes at their boundaries. Contact between bodies can be either linear and require only 1 iteration to solve, such as bonded and no separation contact, or nonlinear and necessitate multiple iterations, which includes frictionless, rough, and frictional contact.⁹⁶ Due to their iterative nature, nonlinear contacts tend to require more processing power and take much longer to solve than linear contacts, and hence most models do not use nonlinear contact unless absolutely necessary.

For this study, it was determined that contact between the stem and the cancellous/cortical bone could be set to bonded, which would simulate the initial press-fit and long-term osseointegration that would be expected to occur. This is a common assumption when modelling the bone-implant interface.^{79, 93, 101} Contact between the cortical and cancellous bone was also set to bonded, as these bodies were assumed to be rigidly fixed to one another in the synthetic femur. Finally, contact between the stem and the ceramic femoral head was set to bonded to simulate the 12/14 taper press fit.

In experimental conditions, contact between the head and the load plate was not frictionless, despite the application of Vaseline to reduce friction. Static friction was present initially, and represented the initial force required to overcome the static cohesion of the head to the load block, following which kinetic friction was present. However, setting frictional contact within an FE model requires knowledge of the friction coefficient, and hence it is commonly initially modeled as frictionless contact.⁶⁵ A model simulating frictionless contact between the load plate and the head was therefore developed for the neutral condition, with the movement of the flat loading plate restricted in all directions except for its vertical z-plane. Frictionless contact is nonlinear in nature, and thus frequently exhibits non-converging solutions with extended computational times. It is therefore beneficial to utilize a linear model when possible. Since the flat load plate was only able to apply force in its vertical direction and sliding was assumed to be minimal, it was therefore possible to remove the plate entirely and instead apply the load directly to the femoral head, yielding a linear loading situation. Stress results from the frictionless model were compared to the linear model, and found to differ by an average of only 6% across four Regions of Interest, as defined in the Section *Named Selections* (Figure 48). Stress results for a

model defined with the most realistic frictional contact would likely fall between the nonlinear frictionless model and the linear direct load application model. Hence, given the minimal difference in the stress results between these two extremes, it can be assumed that the linear model is a good approximation for experimental conditions, and hence the direct load application approach was used in this study.

Figure 48: Comparison of linear neutral model with load applied directly to femoral head (left) versus a nonlinear neutral model with contact between head and load plate set to frictionless (right), equivalently scaled in von Mises stresses (MPa) for comparison purposes

Mesh

In FE modelling, the mesh refers to the compilation of all of the elements and nodes used to approximate the model. Selection of an appropriate mesh is an iterative process that requires selection of element types, sizes, and regions of refinement. For this study, all bodies were

meshed using 10-node tetrahedral elements, which have quadratic displacement behaviour and 3 DOF at each node (Figure 49).⁶⁵ In a previous study, these elements were shown to be preferred over mapped, hexagonal, or voxel meshes for use with the human femur.¹⁰²

Figure 49: Depiction of a 10-node quadratic tetrahedron⁹⁵

All of the element sizes used in this study were chosen based on evaluation of mesh convergence. This involved individually varying the element sizes for each method from 8mm down to 2mm, and examining the difference in the resultant max stresses, max deformations, and numbers of elements and nodes. Furthermore, mesh sensitivity analysis was completed using the relevance slider tool, which defines either a coarse mesh (negative relevance to -100) or a fine mesh (positive relevance to +100). At a relevance of 80, the modelling solution with regard to stress and deformation was found to level off, yielding values that differed by less than 1% from a mesh with a relevance of 100.¹⁰¹

For the overall mesh, element size was set to 5mm, with a relevance of 80. Additional methods were added to the cortical and cancellous bones to specify that they be meshed with tetrahedral elements. The proximal portion of the femur was determined to be an area of interest, and thus two spheres of influence were added to refine the element size of both the stem-head construct and the cortical and cancellous bone in this region to a size of 2.8mm. Individual refinements were added to the surface of the stem where it came into contact with the cancellous bone and the femoral head, and matching refinements were added to the surfaces of the head and the cancellous bone (Figure 50). This minimized the potential for mesh interferences that can arise at boundaries with material changes.⁹⁵

Finally, two sizing features were added to the cortical and cancellous bone to make their average element size approximately 4mm. Overall, each model had an average of 178000 elements, and 270000 nodes (Figure 51).

Figure 50: Proximal cross-section of the mesh used for this study, depicting the finer mesh and areas of refinement

Figure 51: Screenshot of the final mesh used in this study

Named Selections

It was determined that to best analyze the stress distribution within the proximal femur, specific Regions of Interest should be defined. These areas were based on those locations most likely to suffer catastrophic failure, such as the inferior and superior neck region.^{52, 103} The potential for distal loading was also considered significant, and hence the region lateral to the lesser trochanter was of interest.⁶¹ These regions were addressed in the FE model by generation of named nodal selections. Nodal selection involves specifying a location-based region of interest that is then converted into its meshed nodes. It was used for this study as it allows for consistent geometry selection between models regardless of the underlying mesh.

Given the much lower Young's modulus of cancellous bone, the stress distribution within this structure was found to be very low when compared to that of cortical bone. Furthermore, application of strain gauges can only evaluate the cortical surface strain, and cortical bone stress is a commonly used assessor of stress shielding.^{42, 104-106} Finally, as this was a comparative study, it is likely that similar trends in the relative stress distribution would be found in the cancellous bone. Thus for the following evaluation, only the cortical bone body was considered.

Four regions of interest were described within the nodal structure of the proximal cortical bone. The first region (Proximal Region) encompassed the entire proximal region of the implanted femur to allow for overall evaluation of the stress in this area (Figure 52). The second (Inferior Neck) and third (Superior Neck) regions of interest took into account loading of the femoral neck (Figure 53 and Figure 54). Finally, the fourth region (Lateral Region) evaluated the extent of endosteal loading present in the given implant orientation (Figure 55). Each of these regions was identically recreated in the intact femur model for comparison purposes.

Figure 52: Proximal Region of interest selected based on a global coordinate system location, where the pink dots indicate underlying nodes

Figure 53: Inferior Neck Region, depicting selected nodes

Figure 54: Superior Neck Region, depicting selected nodes

Figure 55: Lateral Region, depicting selected nodes

Constraints

The final step in generating an FE model is to select a set of constraints which govern what forces are applied and which DOF in the model are restrained. In order to replicate experimental conditions, a force of 1000N was applied along the load plane coordinate system, directly onto the femoral head at the previously defined load application point. A fixed support was then applied to the distal faces of the femur corresponding to the height of the cement block, simulating the potted femur.

Solution

The solution tab allows the user to select a number of outcome measures, including von Mises stress, von Mises strain, normal strain, and deformation. Each measure can be applied to a particular body or face, or a combination of either one. For initial validation purposes in this study, the femoral von Mises strains and the normal strains along the longitudinal axis were

generated. The von Mises stresses for each of the predefined regions of interest were also retrieved. Deflection in the z-plane at the load application point was found in order to calculate the stiffness of the model. The nodal von Mises stress results from each of the regions of interest were exported and saved into Excel files. The average nodal stress, standard deviation, and 90th percentile stress for each region was then calculated and compiled for analysis.

5.5 Design of Experiments

Design of experiments (DOE) is a statistical technique employed to evaluate whether system inputs have a significant effect on resulting outputs. It can also be used to determine whether any inputs have interaction effects, and what input levels would yield an optimal output response. Experimental Designs have 3 components: factors, levels and responses, each of which are entered by the experimenter. Factors consist of the inputs to the process or system, and are usually comprised of controllable or uncontrollable variables. Levels are the settings of each factor, and are typically defined as being high or low. The responses are the desired outcome variables for the experiment, which drive the optimization of the system (Figure 56).

Figure 56: Depiction of a black-box system with variable inputs and corresponding responses¹⁰⁷

In this study, there were 4 defined factors, which were given by the examined angles and positions and their corresponding maximum achievable values as determined experimentally (Table 6).

Table 6: DOE factors and corresponding maximum levels, as achieved in the biomechanical tests

Factor	Low Level	High Level
Varus/Valgus (°)	-10	10
Anteversion (°)	0	10
Posterior/Anterior (mm)	-2	2
Inferior (mm)	0	4

5.5.1 OFAT versus Factorial Design

The simplest method of experimental design is the one-factor-at-a-time (OFAT) method, which involves varying a single factor at a time while all of the others are kept fixed. This method is not ideal for a number of reasons, chief of which is the inability of OFAT to explore interaction effects between factors. OFAT design also requires a greater number of experimental runs when compared to some factorial designs, which can be problematic when time and resources are an issue. Factorial designs address many of these concerns by allowing the user to vary numerous factors simultaneously, which also has the advantage of increasing the inductive basis from which to draw inferences about the process. For example, a two-level factorial design might consist of 3 factors varied between high and low levels. The number of experimental runs required by a two-level factorial design is 2^k , where k is the number of factors. Thus in this case, only 8 runs would be required by a factorial design, which gives the same statistical power as 16 OFAT runs (Figure 57).¹⁰⁸

Figure 57: Two-level factorial design depicting the contents of each run, and the volume covered by the design¹⁰⁸

Through the addition of centre points, which are midpoints between the high and low factor levels, it is possible for two-level factorial designs to sense the presence of nonlinearities and curvature.¹⁰⁸ However, these designs are unable to fit a model to the given surface, and thus they must be upgraded to a more complex design.

5.5.2 Response Surface Methodology Designs

The goal of response surface methodologies (RSM) is to fit a 3D contour map to a response, thereby reflecting any significant curvature.¹⁰⁸ This can be accomplished through use of either a three-level factorial design or a central composite design (CCD).

In three-level factorial designs, the number of experimental runs required is 3^k , where k is the number of input factors.¹⁰⁷ However, depending on the amount of factors being investigated, this number can get quite large very quickly. For example, analysis of 4 factors would require 81 runs, which is an excessively large number for any given experimental budget. An alternative to this full three-level factorial is to use a CCD.

CCD differ from two-level factorial designs in that they contain axial star points in addition to centre points, which increases the number of factor combinations included in the DOE, allowing better estimation of response curvature (Figure 58).¹⁰⁷ Furthermore, the number of experimental runs required by a 4 factor CCD is only 30, which is much more feasible when compared to a three-level factorial design.¹⁰⁹

Figure 58: Depiction of a two-level CCD design space¹⁰⁷

There are a number of different types of CCD, from the basic circumscribed design in which the star points lie outside the factorial box, to the inscribed design, which treats the factor settings as star points and so defines new test points within those limits.¹⁰⁹ Finally, the face-centred CCD pulls the star points in so that they lie along the faces defined by the factorial ranges.¹⁰⁷ For this study, the defined factor ranges represent the actual orientation limits attainable by the implant, and so the circumscribed CCD is not viable. Face-centred designs provide higher-quality predictions of surface curvature over the entire design space when compared to inscribed designs, and hence the face-centred CCD was used in this study.

Figure 59: Comparison of the three types of CCD, indicating factorial ranges (blue) and star points (pink)¹⁰⁹

5.5.3 JMP for Design of Experiments

The statistical software used for analysis in this study was JMP (SAS Institute Inc., Cary, NC, USA). This software allows the user to identify responses and factors, and then select the specific form of DOE desired. A list of the required experimental runs is then generated, with a

corresponding input section for the predefined responses. The user can then run the model, which generates an array of analytic and diagnostic measures.

5.5.3.1 JMP Design Parameters

The first step in generating a CCD in JMP is to determine and input the factors to be analyzed. In order to account for variations in the units of measure, each of the predefined factor inputs used in this study was coded to a value of +1, 0 or -1 (Table 7). They were furthermore given coded names for ease of analysis.

Table 7: CCD factor levels with corresponding coded names and values

Coded Name	Factor	Low Level (-1)	Centre (0)	High Level (+1)
A	Varus/Valgus (°)	-10	0	10
B	Anteversion (°)	0	5	10
C	Posterior/Anterior (mm)	-2	0	2
D	Inferior (mm)	0	2	4

The responses chosen for use in this study were drawn from the regions identified in the Named Selections portion of Section 5.4.2.3. The exported values for the mean, standard deviation, and 90th percentile stress of each of these regions were used in a number of combinations as response inputs, which have been described in detail in Section 5.5.3.2.

Typically, multiple centre points are used in CCD, as they give an indication of the between-run response variance of the system (pure error) and hence provide the DOF required for a lack of fit test.¹¹⁰ This test uses analysis of variance (ANOVA) to determine whether the proportion of error not explained by the model is significant.¹¹¹ While this test is useful, there are numerous other means to assess the fit of the model to the data, including values for the correlation coefficient (R) and residual plots. For this study, a computer simulation was used to generate the responses of each run, and thus there was no variation between runs regardless of how many times the run was replicated. For that reason, an on face CCD with no replicates and 1 centre point was selected within the JMP DOE design box, giving rise to a design with 25 runs (Table 8). A prospective power analysis of this design ($\alpha = 0.05$) indicated that all main and two-factor interaction effects had strong predictive capabilities (power > 0.9).

Table 8: All of the runs specified by JMP given the chosen inputs, with a pattern indicating the composition of each run: “+” for high setting, “-” for low setting, “A” high axial, “a” low axial, “0” for centre point¹¹¹

Run	Pattern	A	B	C	D
1	----	-1	-1	-1	-1
2	----+	-1	-1	-1	1
3	---+-	-1	-1	1	-1
4	---++	-1	-1	1	1
5	a000	-1	0	0	0
6	-+---	-1	1	-1	-1
7	-++-	-1	1	-1	1
8	-++-	-1	1	1	-1
9	-+++	-1	1	1	1
10	0a00	0	-1	0	0
11	00a0	0	0	-1	0
12	000a	0	0	0	-1
13	0000	0	0	0	0
14	000A	0	0	0	1
15	00A0	0	0	1	0
16	0A00	0	1	0	0
17	+----	1	-1	-1	-1
18	+---+	1	-1	-1	1
19	+--+-	1	-1	1	-1
20	+---+	1	-1	1	1
21	A000	1	0	0	0
22	++--	1	1	-1	-1
23	++-+	1	1	-1	1
24	+++-	1	1	1	-1
25	++++	1	1	1	1

Another measure that can be used to assess the strength of a design is the alias matrix. Aliasing of effects occurs when a fractional design is used, which is often desirable when fewer runs are required due to experimental restrictions. A factor can have a confounding relationship, or exhibit aliasing with another factor, when its effects are attributed to that factor. For example, if a design uses the same sequence of factor settings to determine both a main effect and an interaction effect, then these two effects are aliased. This can cause problems if one of those factors happens to be a significant effect, as it could falsely label the other factor as significant as well.¹⁰⁸ JMP provides a prospective diagnostic report that allows the user to determine whether

any aliasing is present in the given model. In this study, it was found that only the third-order terms were aliased, with second-order terms exhibiting only mild correlations (Figure 60). This degree of aliasing was considered acceptable, particularly considering the rarity of three-factor interactions.^{108, 111}

Figure 60: Colour map depicting the aliasing of effects, where a blue square indicates no correlation and a red square indicates complete aliasing

5.5.3.2 Approaches to Response Selection

The goal of the chosen outcome responses variables was to capture the variance in the stress distribution arising from each orientation. Three different approaches to estimate this response were therefore developed, with the objective of producing an optimality envelope (Table 9). In order to take into account the intact stress values and thereby ensure a comparative outcome model, there were two different methods used throughout each of these approaches. The first method (Method 1) involved comparing the raw stress values obtained from each region of interest to the corresponding intact stress value. In this case, the solution was optimized to match the intact value. The second method (Method 2) involved taking a percent of the intact value for

each measure, which is a commonly used criterion when assessing femur stress in relation to a base value (Equation 3).^{31, 42} These measures were subsequently optimized to match 100, which equated to the intact femur value. Each method was input into a separate DOE to allow for independent evaluation.

Equation 3: Calculation of the percent of the intact femur, which was applied for each response measure in the second method

$$\% \text{ of Intact} = \left(\frac{\text{Response Stress Measure}}{\text{Intact Value}} \right) * 100$$

Approach 1 - Stress Measures for Proximal Region Only

The first approach assessed only the Proximal Region, as defined previously. The mean of the stress values within this region was determined to provide an idea of the central tendency of the stress level in this area. The standard deviation of these stresses was also calculated to give a sense of the spread of the data. Finally, the 90th percentile stress value was found in order to evaluate the general range of the data. This was used in place of a maximum stress value, which was considered too unstable and subject to local modelling effects to be an apt measure. Each of these measures was optimized through DOE using both Method 1 (raw value) and Method 2 (percent of intact).

Approach 2 - Mean Stress Values for All Regions

The second approach utilized the other three regions of interest in addition to the Proximal Region to evaluate stress distribution within the proximal femur. In this method, the mean stress values developed in each of the Proximal, Inferior Neck, Superior Neck and Lateral Regions were input as response variables. It was theorized that the average stress value would give a good indication of the overall response in each region, including any exceptionally high or low values. Again, both input methods were used, assessing the optimal orientation generated by both the raw stress values and the percent of intact femur values.

Approach 3 - 90th Percentile Stress Values for All Regions

The final approach utilized the 90th percentile stress values for each of the defined regions of interest. The principle behind this approach was to assess the behaviour of 90% of the stress in a given region, without any directional bias that might be caused by the presence of local

maximums. For this approach, the 90th percentile was calculated for the Proximal, Inferior Neck, Superior Neck and Lateral Regions, and input as both a raw value and a percent of the intact femur value.

Table 9: Overview of each response selection approach and the method used to generate the employed input value

Approach	Included Regions	Stress Measure	Input Value	
1	Proximal	Mean	Method 1 (Raw Values)	Method 2 (Percent of Intact)
		Standard Deviation		
		90th Percentile		
2	Proximal	Mean	Method 1 (Raw Values)	Method 2 (Percent of Intact)
	Inferior	Mean		
	Superior	Mean		
	Lateral	Mean		
3	Proximal	90th Percentile	Method 1 (Raw Values)	Method 2 (Percent of Intact)
	Inferior	90th Percentile		
	Superior	90th Percentile		
	Lateral	90th Percentile		

5.5.3.3 Model Significance and Selection

JMP offers a number of methods with which to analyze the fit of a given outcome model, and assess the significant coefficients and factor interactions. For this study, it was determined that the initial fit of the model would be established through evaluation of the model R value and root-mean-square error (RMSE) values, in conjunction with its whole model ANOVA generated *p*-value.¹¹⁰ A model that fits the data well would have an R close to 1, low RMSE values and a $p < 0.05$. The plot of residuals versus predicted values was also generated to confirm the random distribution of the data.

Individual terms were then assessed through evaluation of the significance of their discrete ANOVA generated *p*-values. Factors that had a *p*-value greater than 0.05 were considered insignificant to the response model equation and so were removed from the model. However, insignificant terms were left in the full model if they were found to be present in a higher order significant term (main effects were never removed), if it was determined that the reduced model had an inferior fit (governed by its R and RMSE values), or if the reduced model had a detrimental effect on the residual distribution.^{110, 112}

5.5.4 Optimal Orientation

Determination of the optimal set of parameters was completed for each input method within each response approach, yielding a total of 6 optimized outcomes. Evaluation of the individual contribution of each factor to its respective outcome response, and the interaction between factors, was accomplished through two main JMP diagnostic tools: parameter estimates and the desirability tool.

5.5.4.1 Factor Coefficients

Through a standard least squares fit and partial ANOVA *F*-tests on the model terms, it was possible to determine parameter coefficient estimates with corresponding 95% confidence interval (CI) and standard error (SE) terms. Due to the previous coding of factors, these coefficients could be directly compared with one another, enabling assessment of relative factor contributions. A higher negative or positive coefficient coupled with a low *p*-value indicated a greater impact on the response model, while a coefficient with a high *p*-value, regardless of size, would have a lesser effect.¹¹⁰ The CI allowed for determination of the potential range of the factor contribution.

5.5.4.2 Desirability Tool

The desirability tool offered within the prediction profiler was used to optimize the response surface for all of the included response inputs simultaneously. This was accomplished through setting a desirability scale, which ranged from the least desirable (*d*=0) to the most desirable (*d*=1) outcome.¹⁰⁸ Depending on the response input method, the most desirable outcome was set to be either the specific intact femur value or 100, and the highest and lowest response input values were given a desirability of 0. The desirability curve between these ranges was quadratic, with the peak equating to the optimal response, and each response input was given equal importance in the optimization calculations to ensure the results were not inadvertently biased. Desirability was then maximized using an internal iterative approach, and the optimal setting for each factor was produced, with the resultant response and its 95% CI indicated (Figure 61).

Figure 61: Screenshot of the desirability tool, where the lower graphs indicate the desirability response curve for each factor and the corresponding degree of desirability on the y-axis for each of the given factor settings, indicated by the numbers in red

5.5.4.3 Definition of the Optimal Orientation

As defined previously, an optimal orientation was generated from each approach and input method. A separate DOE was furthermore completed for each gait cycle phase, giving rise to 18 total optimized outcomes. A weighted average was used to determine the contribution of each optimal factor setting to the overall optimal orientation for each gait cycle phase. A weight was assigned to each factor depending upon the quantitative rise of its respective desirability response curve, in addition to the qualitative shape of its curve. Given that all desirability settings fell between 0 to 1, this ensured uniformity of comparison between the varying approaches and methods. Furthermore, the shape was taken into account by multiplying the desirability weighting by 1 for a parabolic response, and by 1.5 for a linear response curve. For example, a linear desirability curve with a measured rise of 0.3 would be given a weight of 0.45. A global optimal orientation was determined by averaging the weighted mean of each factor across all three gait cycle phases. The full weighting results for each gait cycle phase have been included in Appendix B.

A range of optimal settings for each factor was also acquired for each gait cycle phase, and used to generate optimal factor windows, within which biomechanical effects remained relatively stable. The optimal weighted average factor values for each gait cycle phase and the global optimal orientation were loaded at 3000N, representing 4 times body weight (BW) for a 75kg

person, which falls within the clinical-level range of 3 – 5 times BW for hip joint contact forces during walking and jogging.^{83, 113} The maximum stress developed under clinical-level loads was then evaluated to determine the potential for fracture at each optimal orientation.

5.5.5 DOE Validation

The success of the DOE was validated by testing the overall optimal orientation found for each gait cycle phase, and comparing the outcome stress responses to the design space corners. Each optimal orientation was therefore input into an FE model and loaded under 1000N, and the stresses developed in each Region were exported and analyzed. The generated response outcomes were then compared to the responses of each of the 25 runs in the respective gait cycle phase to ensure the optimality of the solution.

5.5.6 Global Optimal Orientation Verification

Once individual optimal orientations had been defined and validated against their respective DOE, the validity of the global optimal orientation had to be determined. This was accomplished through entering the global optimal orientation into an FE model and loading it under each of the neutral, extension, and flexion loading conditions. Resultant stress values in each Region of Interest were then compared to those values developed in the optimal orientation model for each respective gait cycle phase using the absolute value of the percentage difference (Equation 4).

Equation 4: Calculation of percentage difference (PD) using the intact (*I*) value and the equivalent implanted value (*X*).

$$PD = \left(\frac{I-X}{\frac{(I+X)}{2}} \right) * 100$$

CHAPTER 6

Results

6.1 Verification of Implant Orientation

Prior to mechanical testing, all of the implanted femurs were plain film radiographed in an A/P and lateral view for verification of their correct implant location. In the A/P plane, it was determined that all of the stems were implanted to within $\pm 1^\circ$ of the planned CCD angle, with the exception of the varus orientation, which was measured to have a CCD angle of 115° versus the planned 110° . This was therefore accounted for in the FE model, with experimental strains compared to a model with an implanted CCD angle of 115° . All of the lateral radiographs were assessed for stem version using the previously defined modified Resubal method, and were found to be within $\pm 2^\circ$ of their planned lateral angular position. Finally, all of the implanted femurs had A/P and lateral positions within $\pm 1\text{mm}$ of that planned.

6.2 Validation of FE Model

A linear regression analysis of the absolute value of the calculated experimental strains versus the absolute value of the FEM strains was completed in order to validate the model. Using JMP, a graph of experimental versus FEM strains was plotted, including strain data from all of the linear and rosette gauges, and the goodness of fit was assessed (Figure 62). Both the high correlation coefficient value of $R = 0.83$ and the small p -value of $p < 0.0001$ indicated that there was a significant relationship between the measured and predicted values.^{83, 114, 115} This was confirmed by assessing the plot of measured versus predicted residuals, which displayed a random distribution (Figure 63). It can therefore be concluded that the experimental and FEM strains displayed a linear relationship.

Figure 62: Correlation of all measured experimental and corresponding FEM strains including a line of best fit with 95% confidence limits. The equation of fit is $y = 30.52 + 1.18x$, with a correlation coefficient of $R = 0.83$, and $p < 0.0001$

Figure 63: Residual plot of measured versus predicted strains

However, it should be noted that the slope of the regression line is equal to 1.18 instead of the desired 1, which suggests that the FE model is slightly stiffer than the experimental model, resulting in lower predicted strain values overall. This was confirmed by evaluating the stiffness resulting from the vertical deflection of the load application point in the FE model against the average of that produced experimentally for each implant orientation, where stiffness is calculated as the force (N) over the displacement (mm). A bar chart of the FEM and experimental stiffness values indicated that overall, experimental stiffness values equated to only about 50% of the FEM values (Figure 64). Nevertheless, these high FEM stiffness values are within the range of those reported in the literature for axially loaded synthetic femurs.^{83, 116}

Figure 64: Bar chart of FEM and average experimental stiffness values for each implant orientation

6.3 DOE Results

For each stance condition (neutral, flexion, and extension), a list of the outcome responses for each input method has been provided, along with the terms included in the model and their coefficients and *p*-values. For a list of the response variable values input into each DOE, refer to Appendix A, and for full results of each approach see Appendices C, D, and E. It should be noted that throughout the results, each factor is referenced in its coded form, where A represents varus/valgus, B is anteversion, C denotes posterior/anterior, and D is inferior (Table 7).

6.3.1 Neutral Phase

6.3.1.1 Approach 1 Method 1 Outcomes

Responses corresponding to the mean stress, standard deviation, and 90th percentile stress in the Proximal Region were analyzed initially. The intact stress values were always lower than the response values, so this model ultimately aimed to minimize the stress response. Through model evaluation it was determined that the model with insignificant terms removed yielded an improved overall significance, with no appreciable changes in the model fit. Furthermore, all of the residual plots were randomly distributed. The reduced model was therefore selected for analysis.

The outcome of this model has been summarized, illustrating the 6 significant model terms: A, B, C, D, A*A and C*C (Table 10). Factor B, or anteversion, was a significant factor in every outcome response, and was the most significant factor in two responses. In all instances, its positive coefficients indicate that as the degree of anteversion increased, the resultant stress values also increased, which is an unfavourable outcome. Factor A and its second order term were significant in two of the outcome responses. Taking into account that the varus/valgus factor spanned 0 (with negative varus and positive valgus), these coefficients therefore indicate that valgus reduced the stress response, while varus tended to increase the response. Similarly, the coefficients for factor C indicate that a negative posterior position tended to increase the stress response while it was decreased by the positive anterior position. Finally, factor D occurred in only one outcome response, and its positive coefficient indicates that increasing the inferior position caused a stress increase.

Table 10: Summary of the reduced Proximal Region stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and *p*-values. Significant terms have been marked with an asterisk (*) and bolded.

Outcome Response	Term	Coded Coefficient	<i>p</i> -value
Mean Stress (MPa)	Overall Model		0.0014*
	B	0.40	0.0033*
	A*A	0.77	0.0134*
	A	-0.31	0.016*
	C	-0.29	0.024*
	C*C	-0.56	0.0626
	D	0.23	0.0733
Stress Standard Deviation (MPa)	Overall Model		0.0752
	B	0.84	0.0203*
	C	-0.74	0.0379*
	A*A	1.08	0.1846
	C*C	-0.79	0.3274
	D	0.26	0.4499
	A	-0.04	0.897
90th Percentile Stress (MPa)	Overall Model		<.0001*
	D	1.26	<.0001*
	C*C	2.65	<.0001*
	A	-0.59	0.0047*
	C	0.59	0.0047*
	B	0.39	0.0471*
	A*A	-0.30	0.4893

To confirm the results of the coefficient evaluation, the desirability tool was used to match the responses to the inputted intact femur values. The orientation yielding overall stress values most similar to the intact femur was found to occur at the following factor settings: $A = 0.32$, $B = -1$, $C = 0.56$, $D = -1$ (Figure 65). When converted from the coded values, this corresponds to an orientation of 3.15° valgus, 0° anteversion, 1.13mm anterior, and 0mm inferior. These results agree with the previous assessment of the effect of each factor on the stress response.

Figure 65: Output of the desirability tool for stress response, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. Starting from the top row and going down, intact femur values for this optimization were 5.13MPa, 3MPa, and 9.19MPa

6.3.1.2 Approach 1 Method 2 Outcomes

A model utilizing the percent of the intact femur values as a response measure was also completed for the Proximal Region. This model aimed to match responses to a value of 100, which would indicate the intact femur values. The reduced model was again selected for

analysis, as removal of the insignificant terms was not found to significantly influence the fit of the model or its residual distribution.

The results of this model have been summarized (Table 11). There were 7 significant factor and factor interaction terms: A, B, C, D, C*C, B*C, and C*D. The second order interaction of factor C was one of the most significant, and appeared in every outcome response. Since this term is a quadratic function with a high positive coefficient, it can therefore be concluded that any deviation into the positive or negative anterior or posterior orientations significantly increased the resulting percentage value, which indicates a higher stress level than the intact femur. Another term which appeared in every response model was factor D, or inferior. The positive coefficient in this case denotes an increasing percentage value with increasing inferior positioning. The significant B factor responded in a similar manner, with greater anteversion leading to a higher percentage value. The negative coefficient for factor A implies that any (negative) varus positioning led to an increase in the percentage value, which is again an unfavourable response. Finally, there are two interaction terms, which suggest an interaction between the anterior/posterior and inferior positions, and between the anteversion and anterior/posterior orientations. For the C*D interaction, the negative coefficient implies that a decrease in outcome percentage was achieved with either a posterior and slightly inferior position, or with an anterior and greater inferior position. Similarly for the B*C term, a decreased percentage value was only achieved at low levels of anteversion and an anterior position, or a posterior orientation and high anteversion levels.

Table 11: Summary of the reduced Proximal Region % of intact femur model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and *p*-values. Significant terms have been marked with an asterisk (*) and bolded.

Outcome Response	Term	Coded Coefficient	<i>p</i> -value
Mean Stress (%)	Overall Model		<.0001*
	C*C	23.70	<.0001*
	A	-6.98	0.0003*
	D	6.79	0.0004*
	C*D	-4.39	0.0162*
	B*C	4.30	0.0182*
	B	3.51	0.0369*
	B*B	-6.01	0.1191
	C	2.27	0.1608
Stress Standard Deviation (%)	Overall Model		<.0001*
	D	13.66	<.0001*
	C*C	29.10	<.0001*
	A	-6.38	0.0027*
	C	6.38	0.0027*
	B	4.21	0.0326*
	B*C	3.71	0.0697
	C*D	-2.49	0.2107
	B*B	-3.80	0.3861
90th Percentile Stress (%)	Overall Model		0.0002*
	C*C	95.62	<.0001*
	C*D	-21.86	0.0094*
	B*C	18.95	0.0212*
	D	15.88	0.0372*
	C	13.42	0.0728
	B*B	-19.02	0.2677
	B	7.44	0.3027
	A	1.76	0.8042

These results were confirmed using the desirability plots, which produced an optimal orientation at A = 1, B = -1, C = 0.39, and D = -1 (Figure 66). This corresponds to an orientation of 10° valgus, 0° anteversion, 0.78mm anterior, and 0mm inferior.

Figure 66: Output of the desirability tool for % of intact, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. All intact femur values for this optimization corresponded to 100

6.3.1.3 Approach 2 Method 1 Outcomes

The second approach evaluated the stress distribution in the proximal femur through inclusion of the mean stress values for each defined Region. The goal of this model was to minimize the stress response (or match the intact femur value). The reduced model for this approach was selected even though some large decreases in the R values were noted, as it was determined that the statistical benefits provided by the increased DOF were more critical. There was furthermore no noticeable difference in the distribution of the residual plots.

There were only 4 significant terms resulting from the selected model: A, B, C, and B*C. Factor B was the most significant factor in two of the response models, with positive coefficients that suggest increasing anteversion gave rise to higher resultant stress values. Factor A was present in three response outcomes, with a negative coefficient indicating that a greater degree of varus caused an increased stress response. The lone significant C term denotes higher stress in the

posterior position, while the interaction term B*C gives rise to a correlation in which stresses could only be minimized if the degree of anteversion increased when going from a posterior to anterior position.

Table 12: Summary of the reduced regional stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and *p*-values. Significant terms have been marked with an asterisk (*) and bolded.

Outcome Response	Term	Coded Coefficient	<i>p</i> -value
Proximal Region Mean Stress (MPa)	Overall Model		0.0066*
	B	0.40	0.0066*
	A	-0.31	0.0269*
	C	-0.29	0.0384*
	A*A	0.43	0.0958
	D	0.23	0.1017
	B*C	-0.03	0.8367
Inferior Region Mean Stress (MPa)	Overall Model		0.2141*
	B	1.25	0.0322*
	A	-0.88	0.1202
	C	-0.52	0.349
	A*A	0.61	0.5578
	B*C	-0.14	0.8032
	D	-0.02	0.9673
Superior Region Mean Stress (MPa)	Overall Model		0.0201*
	B*C	-0.53	0.011*
	A	-0.50	0.0111*
	D	-0.28	0.1242
	C	0.22	0.2286
	A*A	-0.15	0.6663
	B	-0.02	0.8959
Lateral Region Mean Stress (MPa)	Overall Model		0.0458*
	A	-0.37	0.0018*
	A*A	0.26	0.1901
	B	0.09	0.405
	C	-0.05	0.6317
	B*C	0.04	0.7418
	D	0.01	0.8915

These correlation evaluations were confirmed in the optimal response parameters generated by the desirability tool (Figure 67). The optimal orientation was defined as factor settings of A =

0.81, B = -1, C = 0.16, and D = -1. This equates to an orientation of 8.05° valgus, 0° anteversion, 0.32mm anterior and 0mm inferior.

Figure 67: Output of the desirability tool for mean stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. Starting from the top row and going down, the intact femur values for this optimization were 9.19MPa, 6.11 MPa, 4.06MPa and 5.13MPa

6.3.1.4 Approach 2 Method 2 Outcomes

When examining the effect of entering the percent of the intact values as a response outcome, it was determined that the reduced model had increased statistical power with no noticeable depreciation in the model fit or residual distribution. Hence insignificant terms were removed from the model.

A summary of the model coefficients has been listed, with 8 terms standing out as significant: A, B, C, D, A*B, C*C, C*D, and B*C (Table 13). The majority of these terms can be explained in a manner similar to that done previously, in which the varus orientation and increasing anteversion were found to yield higher percentage values, while any degree of anterior or posterior positioning significantly increased the percentage value. However, there were some conflicting terms with regard to the main factors C and D, which have both positive and negative coefficients depending on the outcome response model. There was furthermore a new interaction term A*B, which has a negative coefficient that indicates the percentage value decreased with increasing levels of valgus and anteversion.

These conflicting results were resolved through use of the desirability tool, which indicated the overall optimal orientation based on significance levels (Figure 68). The optimal factor settings were found to be $A = 1$, $B = -1$, $C = -0.06$, and $D = -0.94$. These correspond to an orientation with 10° of valgus, 0° of anteversion, 0.11mm posterior, and 0.12mm inferior.

Table 13: Summary of the reduced regional % of intact model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and *p*-values. Significant terms have been marked with an asterisk (*) and bolded.

Outcome Response	Term	Coded Coefficient	<i>p</i>-value
Proximal Region Mean Stress (%)	Overall Model		<.0001*
	C*C	20.07	<.0001*
	A	-6.98	0.0001*
	D	6.79	0.0002*
	C*D	-4.39	0.0086*
	B*C	4.30	0.0098*
	A*B	-3.94	0.0163*
	B	3.51	0.022*
	C	2.27	0.1211
Inferior Region Mean Stress (%)	Overall Model		0.0049*
	C*C	32.64	0.0009*
	C*D	-13.23	0.0099*
	B*C	9.32	0.056
	A	-8.16	0.0736
	A*B	-5.93	0.2081
	B	4.73	0.2838
	D	3.07	0.4823
	C	0.04	0.9924
Superior Region Mean Stress (%)	Overall Model		<.0001*
	D	-10.84	<.0001*
	A	-8.81	<.0001*
	B	5.77	0.0001*
	C	-4.92	0.0006*
	B*C	3.54	0.0111*
	C*C	-3.54	0.127
	A*B	-1.24	0.3297
	C*D	-1.07	0.3976
Lateral Region Mean Stress (%)	Overall Model		<.0001*
	A	-13.21	<.0001*
	C	3.03	0.0002*
	D	2.65	0.0008*
	B	2.50	0.0013*
	C*C	-2.08	0.106
	C*D	-0.56	0.4252
	A*B	-0.51	0.4638
	B*C	-0.15	0.8312

Figure 68: Output of the desirability tool for percent of intact stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. All intact femur values for this optimization corresponded to 100

6.3.1.5 Approach 3 Method 1 Outcomes

The third approach involved looking at all of the regions of interest again; however, this time the 90th percentile stress value was used as the response outcome. The reduced model was again selected for this approach, as there were no appreciable changes in the model fit or residuals, and the statistical power was seen to increase.

There were 8 significant main factors and factor interactions in the chosen model: A, B, C, D, C*C, B*C, C*D, and B*D (Table 14). Most of the factors again followed similar trends in terms of their output responses, where greater varus and anteversion increased the stress response,

while the anterior/posterior quadratic term produced unfavourable results in either direction. The inferior factor was also seen to have both positive and negative coefficients depending on the response region. However, there was a new significant two-factor interaction term $B*D$, for which the response stress could only be reduced if a higher inferior setting was subsequently combined with a decreased anteversion setting and vice versa.

This combination of coefficient requirements was optimized using the desirability tool, and the orientation yielding the greatest desirability was determined (Figure 69). The optimal factor settings were $A = 1$, $B = -1$, $C = -0.21$, and $D = -0.30$, which corresponds to an orientation of 10° valgus, 0° anteversion, 0.43mm posterior, and 1.39mm inferior.

Table 14: Summary of the reduced regional 90th percentile stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and *p*-values. Significant terms have been marked with an asterisk (*) and bolded.

Outcome Response	Term	Coded Coefficient	<i>p</i>-value
Proximal Region 90th Percentile Stress (MPa)	Overall Model		<.0001*
	C*C	2.46	<.0001*
	D	1.26	<.0001*
	A	-0.59	0.003*
	C	0.59	0.003*
	B	0.39	0.0351*
	B*C	0.34	0.0738
	C*D	-0.23	0.2176
	B*D	-0.09	0.6136
Inferior Region 90th Percentile Stress (MPa)	Overall Model		0.0007*
	C*C	11.77	<.0001*
	C*D	-3.93	0.0056*
	B*C	3.88	0.0061*
	C	1.72	0.1568
	B	1.36	0.2574
	D	0.76	0.5217
	B*D	-0.38	0.7632
	A	-0.31	0.7914
Superior Region 90th Percentile Stress (MPa)	Overall Model		<.0001*
	D	-1.15	<.0001*
	A	-0.81	<.0001*
	B	0.48	0.0024*
	C	-0.41	0.0067*
	B*C	0.38	0.0158*
	C*C	-0.31	0.2285
	C*D	-0.11	0.4525
	B*D	0.09	0.5298
Lateral Region 90th Percentile Stress (MPa)	Overall Model		<.0001*
	A	-0.70	<.0001*
	D	0.13	0.0013*
	B	0.09	0.0177*
	C*D	-0.08	0.0366*
	B*D	0.08	0.0481*
	C*C	-0.12	0.0773
	C	0.06	0.1138
	B*C	-0.03	0.4294

Figure 69: Output of the desirability tool for 90th percentile stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. Starting from the top row and going down, the intact femur values for this optimization were 9.19MPa, 13.61 MPa, 9.66MPa and 5.88MPa

6.3.1.6 Approach 3 Method 2 Outcomes

The final approach involved examining the 90th percentile values as a percent of the intact femur. The full and reduced models were each assessed, and it was determined that insignificant variables would be removed, as there was little depreciation in the model fit or in the dispersion of residuals.

There were 7 significant terms in the analyzed model, which included factors: A, B, C, D, C*C, B*C, and C*D (Table 15). Aside from a notably large and significant positive coefficient for the second order C term, there were no new factor effects noted in this model. In general, factor A

had negative coefficients corresponding to an increasing percentage value with the varus angle, factor B had positive coefficients equating to larger percentages with increased anteversion, and factor D had both positive and negative significant coefficients.

Table 15: Summary of the reduced regional 90th percentile % of intact stress model, sorted by significance, and indicating each response measure with its corresponding model terms, coefficients, and *p*-values. Significant terms have been marked with an asterisk (*) and bolded.

Outcome Response	Term	Coded Coefficient	<i>p</i> -value
Proximal Region 90th Percentile Stress (%)	Overall Model		<.0001*
	C*C	26.81	<.0001*
	D	13.66	<.0001*
	A	-6.38	0.0024*
	C	6.38	0.0024*
	B	4.21	0.0309*
	B*C	3.71	0.0671
	C*D	-2.49	0.207
Inferior Region 90th Percentile Stress (%)	Overall Model		0.0002*
	C*C	86.45	<.0001*
	C*D	-28.87	0.0044*
	B*C	28.50	0.0048*
	C	12.65	0.1452
	B	10.00	0.2439
	D	5.58	0.5099
	A	-2.29	0.7856
Superior Region 90th Percentile Stress (%)	Overall Model		<.0001*
	D	-11.92	<.0001*
	A	-8.38	<.0001*
	B	4.96	0.0019*
	C	-4.28	0.0057*
	B*C	3.93	0.0138*
	C*C	-3.25	0.2194
	C*D	-1.12	0.4437
Lateral Region 90th Percentile Stress (%)	Overall Model		<.0001*
	A	-11.95	<.0001*
	D	2.26	0.0026*
	B	1.54	0.0278*
	C*D	-1.41	0.0536
	C*C	-2.08	0.1046
	C	0.98	0.1463
	B*C	-0.50	0.471

The desirability tool output generally supported the discussed factor effects, recommending factor settings of $A = 1$, $B = -1$, $C = -0.16$ and $D = -0.38$ (Figure 70). This resulted in an optimal orientation at 10° valgus, 0° anteversion, 0.31mm posterior, and 1.23mm inferior.

Figure 70: Output of the desirability tool for 90th percentile % of intact femur stress, indicating ideal factor settings along the bottom row, with corresponding stress responses and CI along the left. All intact femur values for this optimization corresponded to 100

6.3.1.7 Optimal Window

In general, there were a number of factors that were seen to be consistently detrimental to either the absolute stress value or the percent of intact femur value. These crucial terms have been summarized and their implications on the outcome response described (Table 16).

Table 16: Summary of common significant neutral terms and their corresponding effects, in addition to the region which was most significantly affected by the term

Significant Term	Coefficient Sign	Implication	Proximal	Inferior	Superior	Lateral
A	Negative	Increasing varus produced higher stresses				X
B	Positive	Increasing anteversion produced higher stresses	X		X	
B*C	Positive	If any anteversion present, an anterior position produced higher stresses		X		
C*C	Positive	Any deviations from the neutral lateral position produced higher stresses	X	X		
C*D	Negative	Any combination of anterior with minimal inferior or posterior with greater inferior produced higher stresses		X		
D	Positive and Negative	Increasing inferior produced higher stresses in all Regions except for the Superior Region	X	X	X	X

In addition to the tabulated critical factors, all of the optimality settings have been compiled and the weighted means and setting ranges determined, yielding an optimality envelope (Table 17).

Table 17: Compilation of all optimal orientations generated by each neutral phase response model, specifying the weighted average values for each factor and the factor range

Response Selection	Varus/Valgus (°)	Anteversion (°)	Posterior/Anterior (mm)	Inferior (mm)
Approach 1 Method 1	3.15	0.00	1.13	0.00
Approach 1 Method 2	10.00	0.00	0.78	0.00
Approach 2 Method 1	8.05	0.00	0.32	0.00
Approach 2 Method 2	10.00	0.00	-0.11	0.12
Approach 3 Method 1	10.00	0.00	-0.43	1.39
Approach 3 Method 2	10.00	0.00	-0.31	1.23
Weighted Average	8.90	0.00	0.11	0.40
Range	3.15 - 10	0.00	-0.43 - 1.13	0 - 1.39

6.3.1.8 FE Model of Neutral Optimal Orientation

To allow for visual comparison between the specified neutral optimal orientation and the intact FE model, the weighted average optimality settings for the neutral phase (Table 17) were entered into an FE model and loaded at a clinically realistic value of 3000N, then compared to the intact femur model loaded at 3000N (Figure 71). The maximum stress developed in the optimal

orientation was evaluated in relation to the ultimate strength of the synthetic cortical bone in compression to ensure that there was no potential for failure.

Figure 71: FE model of the intact femur loaded at 3000N (top), and the neutral phase optimally implanted femur loaded at 3000N (bottom), equivalently scaled in von Mises stresses (MPa) for comparison purposes

The maximum cortical bone stress in the optimally oriented implanted model was found to be 56.61MPa, which is well below the manufacturer provided ultimate strength of synthetic cortical bone of 106MPa, indicating no risk for fracture.⁶³ Furthermore, the implanted femur exhibited a stress distribution similar to that of the intact femur, with comparable regions of maximum and minimum stress and a Proximal Region average stress of 115.7% of the intact femur value.

6.3.2 Extension Phase

In order to evaluate the behaviour of the implanted stem throughout the range of the gait cycle, the stress distribution at the toe-off phase, represented by 15° of extension, was assessed. In a manner identical to the neutral stance, a DOE for each response selection approach and input method was completed, and the resulting factor coefficients evaluated (Table 18). Reduced models without insignificant terms were used for every response model except for Approach 2 Method 1, for which the removal of insignificant terms was found to have a drastic effect on the model fit. Using the match target function, all of the models were also optimized to match either the intact femur values or 100, depending on the method in use. The optimal orientation window was then determined (Table 19). For complete results see Appendix D.

Table 18: Summary of common significant extension terms and their corresponding effects, in addition to the region which was most significantly affected by the term

Significant Term	Coefficient Sign	Implication	Proximal	Inferior	Superior	Lateral
A	Negative	Increasing varus orientation produced higher stresses			X	X
B	Positive	Increasing anteversion produced higher stresses			X	X
B*C	Positive	Increasing anteversion with increasing anterior produced higher stresses		X		
C	Positive	Increasing anterior produced higher stresses	X			X
C*C	Positive	Any deviations from the neutral lateral position produced higher stresses	X	X		
D	Positive and Negative	Increasing inferior produced higher stresses in all Regions except for the Superior Region	X	X	X	X

Table 19: Compilation of all optimal orientations generated by each extension phase response model, specifying the weighted average values for each factor and the factor range

Response Selection	Varus/Valgus (°)	Anteversion (°)	Posterior/Anterior (mm)	Inferior (mm)
Approach 1 Method 1	-0.98	0.00	-0.17	0.00
Approach 1 Method 2	-0.88	0.00	-0.17	0.00
Approach 2 Method 1	10.00	0.00	-0.82	0.00
Approach 2 Method 2	10.00	1.89	-0.44	0.00
Approach 3 Method 1	10.00	0.00	-0.93	0.00
Approach 3 Method 2	10.00	0.00	-0.91	0.00
Weighted Average	7.99	0.27	-0.46	0.00
Range	-0.98 - 10	0.00 – 1.89	-0.93 – -0.17	0.00

In order to evaluate the behaviour of the implant under clinical level loads in the extension condition, the optimal orientation was entered into an FE model and loaded at 3000N. The maximum stress developed was found to be 70.11MPa, which is again well below the 106MPa ultimate strength value of synthetic cortical bone, as provided by the manufacturer.⁶³ The implanted femur furthermore exhibited an average Proximal Region stress value of 112.4% of the intact femur, indicating good similarity to the intact condition.

6.3.3 Flexion Phase

The flexion phase represented the heel strike condition within the gait cycle, and corresponded to 15° of flexion. Similar to the previous two phases, FEA was used to complete a DOE for the flexion condition and the resultant stress values were used as outcome responses in their raw form and as percent of the intact femur. All of the models used for this phase were reduced and so had their whole model insignificant variables removed. Resulting factor coefficients and terms were evaluated for their significance and implication in the overall stress reduction objective (Table 20). Using the desirability tool, an optimal orientation was determined for each model and a subsequent optimality range obtained (Table 21). For full results see Appendix E.

Table 20: Summary of common significant flexion terms and their corresponding effects, in addition to the region which was most significantly affected by the term

Significant Term	Coefficient Sign	Implication	Proximal	Inferior	Superior	Lateral
A	Negative	Increasing varus orientation produced higher stresses	X		X	X
A*D	Negative	Increasing valgus with lower levels of inferior, or varus with higher levels of inferior, produced increased stresses	X			X
B*C	Positive	Increasing anteversion with increasing anterior produced higher stresses		X	X	
C	Negative	Increasing posterior produced higher stresses			X	
C*C	Positive	Any deviation from neutral produced higher stress	X	X		
D	Positive and Negative	Increasing inferior produced higher stresses in all Regions except for the Superior Region			X	

Table 21: Compilation of all optimal orientations generated by each flexion phase response model, specifying the weighted average values for each factor and the factor range

Response Selection	Varus/Valgus (°)	Anteversion (°)	Posterior/Anterior (mm)	Inferior (mm)
Approach 1 Method 1	10.00	10.00	0.00	1.88
Approach 1 Method 2	10.00	10.00	0.01	1.85
Approach 2 Method 1	10.00	10.00	-1.55	0.00
Approach 2 Method 2	10.00	10.00	-1.55	0.00
Approach 3 Method 1	10.00	10.00	-1.38	0.90
Approach 3 Method 2	10.00	10.00	-1.38	0.89
Weighted Average	10.00	10.00	-0.74	0.41
Range	10.00	10.00	-1.55 – 0.00	0.00 – 1.88

The optimal orientation for the flexion condition was entered into an FE model and loaded under 3000N to assess the stress resulting under clinical-level loads. The maximum stress was found to be 53.59MPa, which is below the manufacturer-stated ultimate strength of synthetic cortical bone of 106MPa, signifying no risk for fracture.⁶³ The Proximal Region was also found to yield an average stress value of 103.3% of the intact femur, indicating good similarity to the intact condition.

6.3.4 Global Optimal Orientation

The global optimal orientation was calculated by taking an average of all of the output values for each factor setting, generating an orientation that features settings from the entire range of the

gait cycle (Table 22). For comparison purposes, this global optimal orientation was entered into an FE model and loaded in single-leg stance at 3000N. The resultant stress distribution was again compared to the intact femur, while the maximum stress developed in the optimally implanted model was compared to the ultimate strength limits provided by the synthetic femur manufacturer (Figure 72).

Table 22: Global optimal orientation, including weighted average values for each factor and the total acceptable factor range

Measure	Varus/Valgus (°)	Anteversion (°)	Posterior/Anterior (mm)	Inferior (mm)
Global Weighted Average	9.14	2.49	-0.48	0.23
Range	-0.98 - 10	0 - 10	-1.55 - 1.13	0 - 1.88

The maximum cortical bone stress developed in the global optimally implanted femur was found to be 56.01MPa, which is again well below the 106MPa ultimate strength limit for synthetic cortical bone, indicating no potential for failure.⁶³ In addition to the neutral phase, the global optimal orientation was also entered into flexion and extension phase FE models and compared to corresponding intact femur models. It was found that the average stress value in the Proximal Region was 115.8% of intact under the neutral gait phase condition, 112.3% of intact under extension, and 108.1% of intact in the flexion condition, signifying a strong similarity to the intact femur throughout the range of the gait cycle.

Figure 72: FE model of the intact femur loaded at 3000N (top), and the global optimally implanted femur loaded at 3000N (bottom), equivalently scaled in von Mises stresses (MPa) for comparison purposes

6.4 Validation of DOE

In order to evaluate the success of the DOE, the optimal orientation for each gait cycle phase was entered into an FE model and loaded under 1000N. The stress values produced in each Region of Interest were then retrieved and compared to those developed during each run of the respective gait phase DOE (Table 23).

Table 23: Response outcome measures for each Region of Interest across all gait phase optimal orientations, including the global optimal orientation, for loading under 1000N. Bolded and starred terms indicate responses that are more similar to the intact femur value than any other response outcome measure for that particular DOE

Gait Phase	Proximal 90th	Proximal Average	Proximal Standard Deviation	Inferior 90th	Inferior Average	Superior 90th	Superior Average	Lateral 90th	Lateral Average
Neutral (MPa)	9.85*	5.93*	2.78*	12.75*	8.77*	10.54	6.37	8.29	5.79
Neutral (%)	107.20*	115.72*	92.84	93.64	95.44	109.09	104.20	141.05	142.65
Extension (MPa)	11.65*	6.85*	3.47*	14.84	10.08	11.55*	7.24*	8.81	6.02
Extension (%)	108.43*	112.35*	102.33*	90.08	95.40	101.62	101.00	131.40	129.70
Flexion (MPa)	9.68*	5.63*	2.94	13.06	8.69	10.07	6.10	7.90	5.25
Flexion (%)	98.61*	103.33*	87.92	90.76	86.97	99.03	96.07	127.22	130.30
Global Neutral (MPa)	9.87	5.94	2.78	12.69	8.75	10.93	6.64	8.25	5.71
Global Neutral (%)	107.35	115.81	92.74	93.20	95.24	113.12	108.63	140.46	140.63
Global Extension (MPa)	11.70	6.84	3.50	14.94	10.14	11.18	7.07	8.76	6.00
Global Extension (%)	108.92	112.29	103.32	90.70	95.95	98.39	98.59	130.69	129.23
Global Flexion (MPa)	10.09	5.89	3.11	13.49	9.16	10.58	6.35	8.24	5.32
Global Flexion (%)	102.80	108.12	92.90	93.79	91.62	104.08	100.13	132.58	131.93

It was found that the optimal orientation for the neutral gait phase yielded stress values more similar to the intact femur in the overall proximal region when compared to results from the complete DOE design space. However, when examining the individual Regions of Interest, the neutral phase optimal orientation did not necessarily produce optimal regional results when compared to the rest of the design space. This was due to the inherent compromises in the optimal orientation calculations required to achieve maximal desirability, in addition to the compromises caused by the weighted averages for each overall optimal orientation. In the neutral phase, the Superior Region was most similar to the intact femur when the stem was implanted in 4mm of inferior positioning, while the Lateral Region developed the lowest stress values under 10° valgus; both of which are extremes that were not selected for in the optimal orientation calculations. However, while the regional stress values may not be most similar to the intact femur values when compared to the extremes of the design space, on the whole the stress developed in the proximal femur was optimal. This indicates successful optimization of the neutral gait phase DOE.

In the extension gait phase DOE, the Proximal Region was again found to yield stress values most similar to the intact femur when compared to the DOE design space. Similar to the neutral gait phase, the optimal orientation under the extension phase did not yield stress values most analogous to the intact femur across all regions, due to specific regional requirements for optimality. For example the Inferior Region favoured a more anteverted stem with no anterior or posterior positioning, while the Lateral Region yielded optimal stress values under maximal valgus angulation. However, given that the Proximal Region produced stress values most akin to the intact femur, it can again be concluded that overall the extension phase optimal orientation was in fact optimal, and hence the DOE was successful.

Within the flexion gait phase DOE, the optimal orientation FE model generated stress values most comparable to the intact femur on the whole proximally when compared to the entirety of the design space. There were again regional differences that caused specific areas to favour more extreme implant orientations not included in the desirability calculations. The Inferior Region benefited most from a neutrally angulated and positioned stem while the Superior Region favoured a varus and slightly inferior orientation, and the Lateral Region was optimal under maximal valgus angulation. Since the stresses developed in the optimally implanted flexion

femur were most similar to the intact femur in the Proximal Region when compared to the rest of the design space, the flexion phase DOE was therefore successfully optimized.

6.5 Validation of Global Optimal Orientation

The global optimal orientation FE model had no direct DOE with which to compare, given that it was generated using a weighted average of all of the gait phase optimal orientations. It was therefore loaded under the neutral, extension, and flexion gait phase conditions, and the regional stress values were contrasted with those developed in the respective FE models for each gait phase optimal orientation. Under the neutral and extension loading conditions, values across all global Regions were found to differ by an average of 1% (Equation 4) from the stress output of the neutral and extension optimal orientation FE models, while global values in flexion differed by 4% from those in the flexion optimal orientation. This high degree of similarity between the individual optimal orientation models and the global model indicates that this global optimal orientation is a good representative of each gait cycle optimal orientation. The global optimal orientation was therefore defined as the optimal orientation of a straight short stem hip implant for the range of the gait cycle.

CHAPTER 7

Discussion

Conservative total hip replacements are becoming increasingly popular in the younger, more active population, as they allow for a minimally invasive surgical approach, conserve proximal bone stock, and result in more physiological stress distributions.^{25, 45, 46} However, there is a lack of biomechanical data to support these implants, in particular data pertaining to the optimal implant angle and position. This study endeavoured to ascertain the optimal orientation for a straight short stem hip implant within the proximal femur, as determined by its stress difference from the intact femur. It was hypothesized that a valgus and inferior orientation would be optimal, based on previous studies of similarly shaped implants.⁵²⁻⁵⁶ Through evaluation of the optimal orientations generated using DOE for each gait cycle loading condition, it was found that increasing valgus yields a stress distribution more similar to an intact femur, while greater inferior positioning is only beneficial for the superior neck region.

7.1 Optimal Orientation

The optimal orientation for the neutral hip phase during walking was given by parameter settings of 8.9° valgus, 0° anteversion, 0.11mm anterior position and 0.40mm inferior position. The hip extension phase during walking yielded similar values for the optimal orientation, with 7.99° valgus, 0.27° anteversion, 0.46mm posterior position and 0mm inferior position. Finally, the hip flexion phase during walking exhibited a slightly different optimal orientation with 10° valgus, 10° anteversion, 0.74mm posterior position and 0.41mm inferior position. Each of these orientations was seen to minimize the difference from the intact femur in terms of either raw stress values or percent of intact femur values. A global optimal orientation was also determined, and was defined by parameter settings of 9.14° valgus, 2.49° anteversion, 0.48mm posterior position, and 0.23mm inferior position. The stress distribution resulting from an FE model implanted with a stem in the global optimal orientation and loaded under the range of potential gait cycle conditions was found to be very similar to the respective stress distributions resulting from a stem implanted in each gait phase optimal orientation. The global optimal orientation was therefore defined as the optimal orientation for a straight short stem hip implant.

All of these orientation factor levels fall within optimal ranges for the surgical implantation of hip implants as determined in the literature. It has been found that for femurs with a relatively varus native CCD angle, placement of the femoral component of hip resurfacings in up to 10° valgus is optimal.^{52-54, 117} According to Yoshimine et al., the clinically relevant range of neck anteversion angles measured from the coronal plane for total hip replacements is between 0 - 30° (equivalent SNA range of 0 - 25° for the synthetic femur), with a lower anteversion level recommended if anteversion of the acetabular cup is increased.¹¹⁸ Higher levels of anteversion have been found to increase the risk for posterior impingement, suggesting lower levels are optimal.¹¹⁹ Since implant positioning is governed by both the size of the implant and its anchoring location within the femur, there are very few studies that assess the optimal position of any type of implant. However, a study by Speirs et al. examined the effect of moving an implant into an offset position with 2mm of posterior positioning and 4mm of superior positioning, and found that the relative changes in proximal cortical strain due to this orientation were negligible when compared to overall changes from the intact femur.⁶² This suggests that for minor changes in position within the proximal femur, the resultant strain distribution is only minimally affected, and hence the small range of implant positions suggested by the optimal orientations would not significantly affect the outcome response.

7.1.1 Common Significant Terms

Through analysis of each DOE, it was determined that there were a number of significant terms which appeared in many of the models, regardless of the loading situation. The first such term was factor A, corresponding to the varus/valgus orientation, which commonly exhibited high significance ($p < 0.05$) alongside a large negative coefficient. This indicated that any deviation into the varus (negative) orientation would act to increase the factor's contribution to the outcome response, thereby increasing the difference from the intact femur value. For this reason, none of the optimal orientations included any appreciable varus angulation, and instead tended toward the maximum valgus range. This is in line with previous studies on optimal varus/valgus implant orientation, which determined a valgus position yielded strain values more similar to an intact femur, and furthermore increased the load to failure.^{52, 54, 117}

Another term that was found to be significant ($p < 0.05$) throughout the range of the gait cycle was the two-factor interaction between B and C, or anteversion and the posterior/anterior position.

The large positive coefficient arising for this term in many of the models suggests a positive correlation between anteversion and the anterior position. This is not surprising when considering the anatomy of the femoral neck, for which the combination of anteversion and an anterior position would tend to bring the implant closer to the anterior cortex. Any interaction between the implant stem and the cortical bone was noted to greatly increase the local stress, which would have subsequently affected the mean stresses developed in the Inferior, Superior and Proximal Regions.

The second order interaction of factor C was significant ($p < 0.0001$) throughout the range of the gait cycle, usually occurring alongside a highly positive coefficient. This would have the effect of yielding a positive contribution to the response outcome regardless of the sign of the factor value. Indeed, the response surface for this factor combination usually resembled an inverse quadratic, which suggests that any large deviation from the neutral lateral position would tend to increase the resulting stresses in the proximal femur. In a study by Kuzyk et al. it was found that anterior or posterior positioning of a femoral head lag screw decreased the load to failure, which indicates that such positioning yields a detrimental biomechanical response in the bone.⁵⁶

7.1.2 Region- and Phase-Specific Significant Terms

Many of the response outcome models resulted in region-specific significant terms that exhibited a high impact on the stress developed in that particular region. For example factor D, or inferior positioning, was noted to be significant ($p < 0.05$) in many of the response outcome models; however, the sign of its complementary coefficient altered between positive and negative. This was due to the fact that one of the examined regions was the superior neck, for which an increased distance to the implant would result in a correspondingly decreased amount of superior stress propagation. This was supported by the surface profile for this region, which exhibited a linear decrease in the resultant stress or percentage value with increasing inferior positioning. For every other region, the inferior factor had a positive coefficient, indicating a tendency to increase stress in that particular region.

Another factor that regularly displayed a high degree of region-specific impact was factor A, or the varus/valgus angle, within the Lateral Region. It was found that this term tended to have a large negative coefficient, with a high level of significance ($p < 0.0001$). This implies that distal

loading of the femur is strongly related to the varus/valgus orientation of the stem, with valgus tending to lessen the stress developed in the lateral proximal region.

A number of factors were also noted to gain significance or undergo a coefficient sign reversal depending on the phase of the gait cycle. Factor C was significant in both the extension and flexion conditions, however in opposite directions. Under extension, factor C generally had a positive coefficient that indicated higher levels of stress with the anterior position, while the negative coefficient in the flexion condition suggested a posterior position would increase stress. These opposing results are to be expected if one considers the orientation of loading for each condition. When the femur is loaded under extension, its angulation is such that the applied load would tend to result in compression on the anterior aspect of the femur, while flexion would yield a posterior compression. As such, an anterior stem positioning would be disposed toward loading of the anterior cortex under extension, thereby increasing the stress in this region, with a similar effect for posterior positioning in the flexion condition.

It is also interesting to note that factor B, or anteversion, was significant ($p < 0.05$) in the neutral and extension conditions, possessing large positive coefficients that acted to decrease the levels of anteversion recommended by the optimal orientation. Yet under the flexion condition, for which 10° of anteversion was recommended, it was rarely found to be a significant term. This result can be explained by the positive two-factor interaction of B with C, which was found to be highly significant in flexion. This interaction suggests that when C is negative, a positive input for B is required to minimize the output response. Hence a posterior orientation under flexion must also be combined with increased anteversion, which would act to rotate the stem away from the posterior cortex and thereby aid in minimizing stresses in this region. Another effect contributing to this high level of anteversion was the relatively flat response curve seen in many of the desirability profiles for the flexion condition (Appendix E). This implies that any changes in anteversion would only minimally affect the outcome response, and hence a high or low level of anteversion, as governed by its interaction effects, is irrelevant biomechanically for the flexion condition.

7.2 Clinical Relevance

The global optimal orientation can provide orthopaedic surgeons with a surgical goal when implanting a straight short stem; however, the maximal accuracy that can be expected when using software-guided navigation is only $\pm 1^\circ$ and $\pm 1\text{mm}$. Since the established global optimal positions are both less than half a millimeter in any direction, it is therefore recommended that surgeons aim to implant the stem in a neutral position. Both the anterior/posterior and inferior positions are relatively sensitive to small positional changes, and thus a neutral position goal would allow for some slight positional inaccuracies without the risk of deviating too far into any one plane. Furthermore, given that the only orientation noted to consistently minimize proximal femur stress was the valgus orientation, it is therefore recommended that this be a primary surgical goal when implanting short stem hip implants. A valgus hip implant angulation is consistent with prior clinical and biomechanical recommendations.^{52-55, 76, 117} Aside from these recommendations, placement of the stem in any other deviation from neutral should be based upon patient-specific physiology and anatomy. For example, a patient with decreased BMD in the superior proximal region of the femur might benefit from an inferiorly placed stem to minimize the potential for superior neck fracture. Likewise, a patient whose anatomy is such that the anteversion angle is increased could see optimal results from a slightly posterior positioning.

Perhaps the more useful measure for orthopaedic surgeons is the optimality window given by the range of optimal orientation settings for each gait cycle condition (Table 22). While the surgeon should still aim for the global optimal orientation, this optimality window allows for some leeway with regard to stem placement. Since values within this envelope all resulted in optimal orientations, it can be assumed that any factor combination therein would result in minimal changes biomechanically. That said, any orientation that results in contact with the cortical bone will increase the stress developed in that region, and so should be avoided.⁷⁶ Furthermore, given that the high level for the anteversion range was generated by only the flexion condition through an interaction effect, while both of the other conditions experienced detrimental response outcomes directly pertaining to high levels of anteversion, it is therefore recommended that a more neutral version angle be the ultimate surgical goal.

The present study found the superior and inferior femoral neck to be regions of high stress, which is in accordance with the findings of similar studies. Shah et al. used infrared (IR) thermography to record the principal stress map of an axially loaded synthetic femur, and found the superior femoral neck to be a region of high tensile stress, while the inferior neck exhibited high compressive stresses.⁸⁶ This is further supported by Van Rietbergen et al., who used a micro-finite element model to quantify the strain distribution in healthy and osteoporotic human femurs, and found peak stresses occurred around the femoral neck for both.¹²⁰ Such regions of high stress indicate potential areas of failure, and indeed, fracture along the femoral neck was found to be the most common failure mode in a study by Nicayenzi et al., which axially loaded intact synthetic femurs to failure.¹¹⁶ These findings are particularly relevant to the placement of a straight short stem hip implant, as a study by Bishop et al. found that an axially loaded human femur implanted with the Silent Hip most commonly failed due to superior separation of the implant from the cancellous bone, followed by medial cut-out of the femoral neck.¹⁰³ This further supports implanting the femur in a valgus orientation, which acts to align the implant axis with the joint force vector and so reduce the resultant moment, which consequently minimizes the stresses developed in the neck in response to the applied moment.¹⁰³

As with all conservative cementless implants, there are two main reaction stresses that act at the implant-bone interface: the rotation stress caused by the transverse component of the joint force, and the press-fit stress that resists axial subsidence caused by the axial component of the joint force.¹⁰³ These stresses regulate the primary stability achievable in a cementless implant such as the Silent Hip. In addition to the wedging effect generated by the tapered design, the initial press fit of the stem within the proximal cancellous bone introduces radial stresses that provide enough friction to resist relative shear movement.¹⁰³ When the rotational stress is greater than the press fit stress, the contact stresses between the bone and the implant decrease and the implant undergoes separation and subsequent catastrophic failure, as described previously.¹⁰³ A more valgus implant angle would therefore act to decrease the moment arm of the transverse component of the joint force and so reduce the rotational stress; ensuring contact stresses at the implant-bone interface remain high. According to Woolson et al., cementless stems that achieve a stable initial press fit can undergo full weight bearing immediately with good results.¹²¹ However, a parametric analysis of the implanted Silent Hip established that interface stresses developed linearly with the magnitude of the joint load.¹⁰³ Given the high peak joint forces

developed during fast walking and especially stumbling, which can be greater than 800% of BW, it is therefore recommended that patients undergo only partial weight-bearing post-operatively, in order to minimize the risk of the rotational stress exceeding the press fit stress.¹¹³

Orthopaedic surgeons should be aware that implantation of any hip implant in an angle more valgus than the natural CCD angle will tend to change the centre of rotation of the hip joint and increase the patient leg-length. Modifications in the hip centre of rotation may affect the lengths and moment arms of the muscles acting about the hip joint, and so decrease their ability to generate forces and moments adequate for the activities of daily living.^{122, 123} Furthermore, leg-length discrepancies following THA can give rise to a number of postoperative complications, such as nerve palsy, low-back pain, and abnormal gait.¹²⁴ The Silent Hip is a modular hip implant with both femoral stem and head components. As such, if the implant is placed in the recommended valgus orientation, the surgeon has the option of choosing from four different possible head offsets to account for the modified joint centre and leg-length.⁷⁶ It has been shown that a leg-length discrepancy of 2cm or greater has a significant effect on physiological parameters postoperatively, such as oxygen consumption and the rate of perceived exertion.¹²⁵ However, patients can be aware and disturbed by an inequality of only 1cm, and hence orthopaedic surgeons should aim to minimize any discrepancies.¹²⁶ Given that the present study validated a maximum valgus orientation of up to only 10° valgus, it is unlikely that even a stem implanted in the maximum potential orientation will yield a leg-length discrepancy of greater than 1cm. A study evaluating the effect of valgus orientation on fracture load also found that the biomechanical benefits of a valgus orientation may decrease for implant placements of greater than 10° valgus.⁵² It is therefore recommended that orthopaedic surgeons implant straight short stem hip implants in up to, but not exceeding, 10° valgus.

7.3 Study Limitations and Future Directions

One primary limitation of this study was that the contribution of muscle forces to the loads developed in the proximal femur were not considered. It was therefore not possible to predict the degree to which stress shielding might occur, as resultant stresses were not directly comparable to physiological values. However, this was determined to be a reasonable exclusion for this study, as results were based on minimizing the differences to an identically loaded intact femur. Furthermore, studies have shown that inclusion of muscle forces does not significantly affect the

stress distribution in the proximal metaphysis.^{127, 128} Future work might include developing and validating a muscle model to examine the magnitude and localization of stresses under a physiologically loaded femur.

Another limitation of this study was the use of synthetic femurs instead of human cadaver bones, as synthetic bones are less able to mimic the biological response of human bone. Specifically, cancellous architecture in human femurs follows lines of applied stress, whereas cancellous material in synthetic femurs is isotropic and homogeneous. Moreover, no commercially-available synthetic femurs exist on the market that replicate the thinning of human femoral cortical bone during osteoporosis. However, due to the interspecimen variability introduced by cadaver bones, many comparison studies have determined that the physiological benefits of using human bones were outweighed by the disadvantage of variable specimens, which would negatively affect the results.^{79, 83-85, 97} Synthetic femurs have furthermore been experimentally validated against human femurs and have been used successfully in prior biomechanical studies.^{56, 65, 129, 130}

It should be recognized that the femurs in this study were loaded quasistatically instead of in a cyclic or dynamic manner, which would be more representative of physiological loading conditions.^{113, 131} Cyclic loading would additionally provide information as to the fatigue failure of the implant, and stiffness changes over time.¹³¹ However, the goal of this study was only to compare the relative stress distributions between implant orientations, the relationship between which should be constant over time, and hence quasistatic loading conditions were deemed sufficient.⁸³ Furthermore, this loading mode facilitated use of FEA, which is commonly validated against quasistatic results.^{52, 65, 83, 93, 97}

Experimental protocol did not require loading of the femurs to failure, which is frequently completed in evaluations of fracture fixation devices to determine their behaviours under failure conditions.^{91, 92, 131} In this study, subclinical loading was used predominantly because the main goal of the experimental tests was to validate the FE model. However, loading femurs until catastrophic structural failure occurs is useful in determining failure modes, and assessing peak failure forces and displacements.¹¹⁶ For that reason, future studies could include loading the optimal implant orientations for each gait phase to failure, allowing for evaluation of their behaviours under failure conditions.

Furthermore, such future fracture models could evaluate the effect of strain rate on the apparent stiffness and ultimate fracture load of implanted femurs. Strain rate is relevant both under cyclic loading conditions, and when considering loading conditions representative of a fall.^{132, 133} The strain rate during normal activities such as walking and jogging is typically too low to cause bone damage in and of itself; however, cyclic loading can lead to the development of microdamage that acts to decrease the apparent strength and modulus of bone, potentially leading to failure.¹³⁴ Under fall conditions, the strain rate is significantly higher and so engages the viscoelastic nature of bone, acting to both increase the stiffness and the load to failure of the femur.¹³² Future failure models of the implanted stem could therefore take into account the viscoelastic nature of bone under varying strain rates by assessing both low-level cyclic loading of the femur and high strain rate loading conditions.

Another experimental drawback was the use of a flat loading plate instead of an acetabular-like loading cup with a polymer liner, which simulates the hip joint in a THA.^{79, 83, 93, 101} Firstly, a polymer liner was not used to ensure that the true stiffness of the femur-implant construct could be measured without the influence of a soft polymer.¹⁰¹ With regard to the loading plate, it can prove difficult to ensure consistent alignment of the femur within the Instron machine when using a loading cup, and thus lateral loads might be transmitted under some loading scenarios but not others. It was therefore determined that for the subclinical loads employed in this study, a flat loading plate would suffice, and would eliminate the potential for any lateral load transmission that might not be accounted for in the FE model. Furthermore, flat loading plates are often used in the literature with good results.^{85, 91, 92, 94} It is not expected that use of a loading cup would significantly modify the optimally determined orientations, which were based on relative stress values. However, it might affect the actual stress distribution within the femur, and thus should be considered in future studies to allow for direct comparison to clinical results.

Finally, only axial compression was considered, and so this study did not evaluate the strain distribution under lateral or torsional loading. Axial compression is the primary loading mechanism in the hip joint, and it also gives rise to higher strain values than either lateral or torsional loading.¹³⁵ Axial loading is therefore the most relevant loading mechanism when evaluating stress distributions that would affect long-term stress shielding, as with the current study. Furthermore, axial loading is one of the most widespread loading mechanisms seen in the literature, and hence there is good precedent for using this loading mode.^{85, 105, 116, 136} However,

given that fractures due to falls are commonly seen in clinical cases, it may therefore be beneficial for future studies to examine the behaviour of the femur under loading conditions representative of a fall.^{88, 137, 138}

For FEA, bonded contact was assigned between the stem and the cancellous bone, and the joint load was applied directly to the femoral head. In actuality, contact between the implant and the cancellous bone was not bonded under experimental conditions, and load was applied to the superior femoral head through a flat loading plate. Contact at both of these regions would have been most similar to a frictional condition, as has been done in many previous studies.¹³⁹⁻¹⁴¹ However, frictional contact is a nonlinear contact that would act to greatly increase the computational time, and also requires knowledge of the specific friction coefficients throughout the model. Bonded contact is therefore a more common assumption in FE modelling, in particular when the goal is to simulate full bony osseointegration with the implant.^{83, 90, 93} It has furthermore been shown that motion at the stem-bone interface is minimal during static axial loading.⁷⁹ Given the number of models required for DOE and the static loading condition, bonded contact was therefore considered to be appropriate for this study. However, this assumption also has implications for the stiffness of the model, as it would have acted to increase the FE model stiffness. Further work could therefore include experimental analyses to determine adequate friction coefficients at the implant-bone interface and between the superior femoral head and the flat loading plate, for eventual use of frictional contact at these regions. If the computational time of such a model is determined to be too great, then bonded contact could be used along with an intermediary stiffness layer at the implant-bone interface. This is a particular property of the studied Silent stem, as this region would actually be composed of both the sintered titanium beads and bone, which would therefore yield a boundary layer containing the averaged stiffness properties of both.⁶⁸

The slope of the validation graph for the FE model was 1.18 instead of 1 (Figure 62). This suggests that the FE model was stiffer than the experimental implant-femur constructs, as the strains experienced by the FE model were lower than experimental strains. This could have been due to a number of reasons, including micromotions between the implant-bone interface and bone-cement cube interface that were not accounted for in the FE model, both of which could have led to a reduction in the stiffness of the experimental femurs. The synthetic femur was also modelled as a linear elastic material with isotropic behaviour for FEA, which is not truly

representative of synthetic femurs, as they possess both longitudinal and transverse elastic moduli.¹¹⁵ Furthermore, the boundary conditions used to simulate experimental conditions were approximations only, in particular the direct application of force versus a more realistic frictional contact with a flat loading plate. However, given the high value of the correlation coefficient in the validation graph, it can be concluded that the FE model produced consistently lower strain values, and thus a correction factor could have been applied to generate a validation graph with a slope of 1. Since the study was comparative in nature, with various potential implant orientations directly compared, the multiplication of their stresses by the same constant would not have altered the relative trends observed and final conclusions reached.

CHAPTER 8

Conclusions

This study endeavoured to determine the optimal orientation of a straight short stem hip implant within the proximal femur for improved physiological stress distribution. A global optimal orientation of 9.14° valgus, 2.49° anteversion, 0.48mm posterior position, and 0.23mm inferior position was found to yield stress distributions most similar to the intact femur across the range of the gait cycle. In general, it was determined that the valgus orientation was optimal throughout the range of the gait cycle, consistently exhibiting a stress distribution more similar to that of the intact femur. Minimal levels of anterior/posterior and inferior positioning were seen to be beneficial in achieving more physiological stresses in specific regions of interest within the proximal femur, while the anteverted orientation was only beneficial in loading under flexion. Overall, orthopaedic surgeons should aim to implant the stem in valgus up to 10°, with an otherwise neutral position and version, unless some degree of deviation would be beneficial for a patient-specific reason.

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Appendix A: DOE Outcome Responses

Neutral Phase Stress Outcome Responses:

Run	Pattern	A	B	C	D	Proximal Region (MPa)	Inferior Neck (MPa)	Superior Neck (MPa)	Lateral Neck (MPa)	Proximal Region Standard Deviation (MPa)	Proximal Region 90th Percentile (MPa)	Inferior Neck 90th Percentile (MPa)	Superior Neck 90th Percentile (MPa)	Lateral Neck 90th Percentile (MPa)
1	----	-1	-1	-1	-1	6.786	11.107	6.969	6.796	3.829	11.156	20.04	12.561	9.372
2	----+	-1	-1	-1	1	7.719	13.865	4.695	6.934	5.949	14.459	32.031	10.61	9.507
3	---+-	-1	-1	1	-1	6.977	12.726	7.14	7.066	3.579	11.497	17.394	11.547	9.623
4	--++	-1	-1	1	1	6.98	11.521	8.363	6.573	3.443	15.156	25.327	7.557	9.566
5	a000	-1	0	0	0	8.538	16.248	6.611	7.048	6.516	11.91	18.265	11.331	9.828
6	-+--	-1	1	-1	-1	8.54	19.232	7.739	7.054	8.396	11.787	15.798	13.445	9.286
7	-+++	-1	1	-1	1	8.609	13.349	6.838	7.292	6.62	15.607	30.357	10.206	9.927
8	-++-	-1	1	1	-1	6.389	10.539	6.5	6.36	3.052	14.787	44.671	12.436	9.693
9	-+++	-1	1	1	1	8.013	15.968	6.551	6.543	6.593	16.445	29.498	11.158	10.004
10	0a00	0	-1	0	0	6.452	9.982	7.411	6.313	3.103	10.742	14.501	10.585	9.038
11	00a0	0	0	-1	0	6.485	10.628	6.818	6.459	3.175	13.772	30.963	10.733	9.289
12	000a	0	0	0	-1	6.477	11.22	6.086	6.538	3.523	10.788	14.198	12.221	8.89
13	0000	0	0	0	0	7.902	15.737	6.345	6.592	7.018	10.957	14.965	11.042	9.081
14	000A	0	0	0	1	6.478	10.879	6.583	6.479	3.692	11.076	16.819	9.713	9.25
15	00A0	0	0	1	0	6.831	14.45	6.311	5.441	6.163	14.021	35.792	10.461	9.256
16	0A00	0	1	0	0	7.458	13.585	5.736	5.687	6.087	11.004	17.637	10.726	9.075
17	+---	1	-1	-1	-1	7.004	13.154	5.863	5.814	6.2	11.086	30.13	10.244	7.848
18	+--+	1	-1	-1	1	7.09	10.484	4.331	5.932	5.522	13.704	28.659	9.351	8.172
19	++--	1	-1	1	-1	5.952	9.094	5.803	5.899	3.163	11.933	30.049	10.021	8.244
20	++++	1	-1	1	1	6.125	9.856	7.02	5.71	3.248	14.24	20.134	7.017	8.235
21	A000	1	0	0	0	7.429	14.576	5.599	5.917	6.573	9.982	13.992	9.498	8.413
22	++--	1	1	-1	-1	7.279	14.315	6.634	5.878	6.771	10.196	16.232	11.24	8.07

23	++-+	1	1	-1	1	7.507	12.305	5.512	6.058	5.938	13.438	28.116	9.077	8.65
24	+++-	1	1	1	-1	6.786	11.107	6.969	6.796	3.829	12.988	34.61	10.771	8.213
25	++++	1	1	1	1	7.719	13.865	4.695	6.934	5.949	14.686	25.849	9.061	8.32
Intact Values						5.128	9.190	6.113	4.057	2.995	9.191	13.612	9.663	5.877

Neutral Phase % of Intact Outcome Responses:

Run	Pattern	A	B	C	D	Proximal Region (%)	Inferior Neck (%)	Superior Neck (%)	Lateral Neck (%)	Proximal Region Standard Deviation (%)	Proximal Region 90th Percentile (%)	Inferior Neck 90th Percentile (%)	Superior Neck 90th Percentile (%)	Lateral Neck 90th Percentile (%)
1	----	-1	-1	-1	-1	132.02	131.75	124.14	161.57	123.49	121.38	147.22	129.98	159.48
2	----+	-1	-1	-1	1	159.96	165.04	107.28	164.22	223.17	157.33	235.31	109.80	161.78
3	--+-	-1	-1	1	-1	132.33	120.86	113.99	167.51	127.84	125.10	127.78	119.50	163.74
4	--++	-1	-1	1	1	150.53	150.87	76.80	170.91	198.62	164.91	186.06	78.20	162.79
5	a000	-1	0	0	0	136.05	138.47	116.79	174.17	119.48	129.59	134.19	117.26	167.23
6	-+---	-1	1	-1	-1	136.12	125.36	136.80	162.01	114.95	128.25	116.06	139.13	158.02
7	-+-+	-1	1	-1	1	166.49	176.80	108.14	173.73	217.57	169.82	223.02	105.61	168.93
8	-++-	-1	1	1	-1	166.55	209.27	126.60	173.87	280.33	160.89	328.17	128.69	164.94
9	-+++	-1	1	1	1	167.89	145.25	111.85	179.74	221.04	178.93	216.71	115.47	170.23
10	0a00	0	-1	0	0	124.59	114.68	106.33	156.76	101.89	116.88	106.53	109.53	153.79
11	00a0	0	0	-1	0	156.26	173.76	107.16	161.28	220.13	149.85	227.47	111.06	158.07
12	000a	0	0	0	-1	125.83	108.62	121.23	155.62	103.60	117.38	104.31	126.47	151.27
13	0000	0	0	0	0	126.47	115.64	111.53	159.20	106.02	119.22	109.94	114.27	154.53
14	000A	0	0	0	1	126.32	122.09	99.55	161.14	117.64	120.52	123.56	100.51	157.40
15	00A0	0	0	1	0	154.10	171.24	103.78	162.47	234.32	152.55	262.94	108.25	157.51
16	0A00	0	1	0	0	126.34	118.38	107.68	159.70	123.28	119.73	129.57	111.00	154.42
17	+----	1	-1	-1	-1	133.21	157.24	103.23	134.12	205.79	120.63	221.35	106.01	133.54
18	+---+	1	-1	-1	1	145.44	147.83	93.83	140.19	203.23	149.10	210.54	96.76	139.06
19	+--+	1	-1	1	-1	136.58	143.14	95.90	143.32	207.01	129.83	220.76	103.70	140.29
20	+---+	1	-1	1	1	138.26	114.08	70.84	146.22	184.36	154.94	147.91	72.61	140.13
21	A000	1	0	0	0	116.07	98.95	94.93	145.40	105.61	108.61	102.79	98.29	143.16
22	++--	1	1	-1	-1	119.44	107.24	114.83	140.74	108.43	110.94	119.25	116.31	137.32

23	++-+	1	1	-1	1	144.88	158.61	91.58	145.85	219.45	146.21	206.55	93.93	147.19
24	+++-	1	1	1	-1	141.96	155.76	108.52	144.88	226.06	141.31	254.26	111.46	139.76
25	++++	1	1	1	1	146.41	133.90	90.17	149.32	198.27	159.79	189.90	93.76	141.58
Intact Values						100	100	100	100	100	100	100	100	100

Extension Phase Stress Outcome Responses:

Run	Pattern	A	B	C	D	Proximal Region (MPa)	Inferior Neck (MPa)	Superior Neck (MPa)	Lateral Neck (MPa)	Proximal Region Standard Deviation (MPa)	Proximal Region 90th Percentile (MPa)	Inferior Neck 90th Percentile (MPa)	Superior Neck 90th Percentile (MPa)	Lateral Neck 90th Percentile (MPa)
1	----	-1	-1	-1	-1	7.451	13.331	7.932	6.624	4.174	12.816	21.809	13.108	9.523
2	----+	-1	-1	-1	1	9.118	16.416	6.834	6.841	8.220	16.923	34.723	11.407	9.767
3	--+-	-1	-1	1	-1	7.626	11.969	7.536	6.942	4.446	13.145	18.141	12.013	9.784
4	--++	-1	-1	1	1	9.684	17.681	6.236	7.095	8.197	18.528	30.632	9.978	9.877
5	a000	-1	0	0	0	7.766	13.868	7.680	7.137	4.274	13.752	20.544	12.177	10.086
6	-+--	-1	1	-1	-1	7.797	12.852	8.906	6.634	4.173	13.619	18.443	14.824	9.509
7	-+-+	-1	1	-1	1	9.636	18.946	6.888	7.114	7.981	18.547	34.953	11.152	10.128
8	-++-	-1	1	1	-1	9.821	20.631	8.643	7.180	9.633	17.335	44.750	13.445	10.291
9	-+++	-1	1	1	1	9.942	15.878	7.287	7.433	8.442	18.628	31.564	11.976	10.435
10	0a00	0	-1	0	0	7.165	11.562	6.902	6.564	3.719	12.413	16.463	11.041	9.357
11	00a0	0	0	-1	0	9.037	17.608	6.822	6.671	8.031	16.607	33.629	11.712	9.593
12	000a	0	0	0	-1	7.416	11.379	8.119	6.479	3.834	12.733	16.380	13.089	9.451
13	0000	0	0	0	0	7.375	11.917	7.383	6.676	3.944	12.897	17.347	11.828	9.611
14	000A	0	0	0	1	7.299	12.177	6.442	6.724	4.290	13.019	18.645	10.268	9.653
15	00A0	0	0	1	0	8.993	16.963	7.050	6.823	8.156	16.037	36.438	11.027	9.642
16	0A00	0	1	0	0	7.483	12.143	7.216	6.705	4.505	13.220	19.108	11.750	9.685
17	+----	1	-1	-1	-1	7.965	15.379	6.757	5.715	7.327	13.657	27.581	10.653	8.392
18	+---+	1	-1	-1	1	8.391	15.272	5.797	5.915	7.491	16.178	32.166	10.067	8.450
19	+--+	1	-1	1	-1	8.423	13.928	6.522	6.165	8.412	14.639	29.165	10.578	8.801
20	+---+	1	-1	1	1	8.165	11.830	4.994	6.325	6.613	16.302	22.015	7.743	8.793
21	A000	1	0	0	0	6.897	10.085	6.286	6.210	4.006	12.056	15.816	10.228	8.943
22	++--	1	1	-1	-1	7.324	12.045	7.903	6.013	4.111	12.782	19.289	12.708	8.906

23	++-+	1	1	-1	1	7.772	13.820	5.997	6.087	5.660	14.172	22.496	9.855	8.789
24	+++-	1	1	1	-1	8.891	15.584	7.608	6.266	8.780	15.911	34.618	11.911	9.164
25	++++	1	1	1	1	8.787	14.266	6.252	6.425	7.344	16.900	28.270	10.123	9.124
Intact Values						6.096	10.564	7.168	4.640	3.391	10.740	16.470	11.361	6.703

Extension Phase % of Intact Outcome Responses:

Run	Pattern	A	B	C	D	Proximal Region (%)	Inferior Neck (%)	Superior Neck (%)	Lateral Neck (%)	Proximal Region Standard Deviation (%)	Proximal Region 90th Percentile (%)	Inferior Neck 90th Percentile (%)	Superior Neck 90th Percentile (%)	Lateral Neck 90th Percentile (%)
1	----	-1	-1	-1	-1	122.23	126.20	110.66	142.75	123.10	119.32	132.42	115.38	142.07
2	----+	-1	-1	-1	1	149.58	155.40	95.34	147.43	242.43	157.57	210.83	100.40	145.70
3	---+-	-1	-1	1	-1	125.10	113.30	105.13	149.60	131.14	122.39	110.15	105.74	145.96
4	---++	-1	-1	1	1	158.86	167.38	86.99	152.91	241.77	172.51	185.99	87.82	147.34
5	a000	-1	0	0	0	127.39	131.28	107.14	153.81	126.07	128.04	124.74	107.18	150.46
6	-+++	-1	1	-1	-1	127.91	121.66	124.24	142.96	123.07	126.80	111.98	130.48	141.86
7	-+++	-1	1	-1	1	158.08	179.35	96.08	153.30	235.38	172.69	212.23	98.16	151.09
8	-++-	-1	1	1	-1	161.10	195.30	120.57	154.74	284.10	161.40	271.71	118.34	153.53
9	-+++	-1	1	1	1	163.10	150.30	101.65	160.18	248.97	173.44	191.65	105.41	155.68
10	0a00	0	-1	0	0	117.54	109.45	96.28	141.46	109.69	115.57	99.96	97.18	139.60
11	00a0	0	0	-1	0	148.24	166.68	95.16	143.77	236.87	154.62	204.19	103.08	143.11
12	000a	0	0	0	-1	121.65	107.72	113.25	139.63	113.09	118.55	99.45	115.21	140.99
13	0000	0	0	0	0	120.99	112.81	102.99	143.88	116.34	120.08	105.33	104.10	143.37
14	000A	0	0	0	1	119.74	115.27	89.86	144.90	126.52	121.21	113.21	90.37	144.00
15	00A0	0	0	1	0	147.53	160.58	98.35	147.03	240.56	149.32	221.25	97.05	143.84
16	0A00	0	1	0	0	122.76	114.95	100.66	144.49	132.88	123.09	116.02	103.42	144.49
17	+----	1	-1	-1	-1	130.67	145.59	94.27	123.15	216.11	127.16	167.47	93.77	125.20
18	+---+	1	-1	-1	1	137.64	144.57	80.87	127.47	220.93	150.62	195.30	88.60	126.06
19	+--+	1	-1	1	-1	138.18	131.85	90.99	132.85	248.10	136.30	177.08	93.10	131.29
20	+--+	1	-1	1	1	133.93	111.99	69.67	136.30	195.03	151.79	133.67	68.15	131.17
21	A000	1	0	0	0	113.14	95.47	87.69	133.82	118.16	112.25	96.03	90.03	133.42
22	++--	1	1	-1	-1	120.15	114.02	110.25	129.57	121.26	119.01	117.12	111.85	132.86

23	++-+	1	1	-1	1	127.49	130.82	83.66	131.17	166.92	131.95	136.59	86.74	131.12
24	+++-	1	1	1	-1	145.85	147.52	106.13	135.03	258.94	148.15	210.19	104.84	136.70
25	++++	1	1	1	1	144.14	135.05	87.21	138.46	216.59	157.35	171.65	89.10	136.11
Intact Values						100	100	100	100	100	100	100	100	100

Flexion Phase Stress Outcome Responses:

Run	Pattern	A	B	C	D	Proximal Region (MPa)	Inferior Neck (MPa)	Superior Neck (MPa)	Lateral Neck (MPa)	Proximal Region Standard Deviation (MPa)	Proximal Region 90th Percentile (MPa)	Inferior Neck 90th Percentile (MPa)	Superior Neck 90th Percentile (MPa)	Lateral Neck 90th Percentile (MPa)
1	----	-1	-1	-1	-1	6.728	12.129	7.337	6.157	4.102	11.628	20.634	11.966	9.304
2	----+	-1	-1	-1	1	8.600	17.290	6.363	6.329	7.890	15.083	32.304	9.953	9.683
3	--+-	-1	-1	1	-1	6.603	11.744	6.537	6.339	4.041	11.543	17.517	11.834	9.553
4	--++	-1	-1	1	1	8.729	15.836	5.090	6.507	8.629	17.030	30.373	8.340	9.711
5	a000	-1	0	0	0	6.642	12.738	6.544	6.591	3.773	11.689	18.303	10.632	9.588
6	-+--	-1	1	-1	-1	6.668	11.215	7.785	6.117	3.526	11.263	15.865	12.507	9.063
7	-+-+	-1	1	-1	1	8.666	18.508	6.093	6.542	7.780	16.002	33.852	9.467	9.543
8	-++-	-1	1	1	-1	8.282	20.038	6.695	6.511	9.144	14.133	44.457	12.034	9.317
9	-+++	-1	1	1	1	8.510	13.445	6.108	6.704	8.276	16.678	30.371	10.094	9.464
10	0a00	0	-1	0	0	6.277	10.901	6.265	5.884	3.430	10.985	15.290	10.227	9.073
11	00a0	0	0	-1	0	8.042	17.050	6.047	6.051	7.561	13.570	32.398	9.743	8.891
12	000a	0	0	0	-1	6.205	10.030	6.884	5.849	3.276	10.597	14.714	12.151	8.806
13	0000	0	0	0	0	6.211	10.681	6.350	5.910	3.356	10.793	15.452	10.422	8.812
14	000A	0	0	0	1	6.190	11.610	5.689	5.994	3.700	11.012	17.386	9.090	8.965
15	00A0	0	0	1	0	7.807	17.350	5.617	5.975	8.083	13.848	36.754	10.156	8.964
16	0A00	0	1	0	0	6.069	11.048	6.064	5.927	3.650	10.700	16.967	10.243	8.822
17	+----	1	-1	-1	-1	7.297	16.043	6.239	5.084	7.732	11.859	35.636	10.091	7.948
18	+---+	1	-1	-1	1	7.589	14.306	5.619	5.222	7.059	13.453	28.281	8.945	8.280
19	+--+	1	-1	1	-1	7.175	14.669	5.691	5.355	7.230	12.091	30.063	10.320	8.357
20	+--+	1	-1	1	1	6.121	12.522	3.824	4.877	5.341	11.321	23.967	6.278	7.475
21	A000	1	0	0	0	5.752	9.602	5.532	5.334	3.276	10.029	14.407	8.963	8.300
22	++--	1	1	-1	-1	5.722	8.908	6.269	5.160	3.264	9.701	14.390	10.494	7.719

23	++-+	1	1	-1	1	5.845	11.621	4.743	5.236	4.753	10.068	22.153	7.482	7.852
24	+++-	1	1	1	-1	7.153	15.569	5.783	5.278	7.559	12.468	34.029	11.175	7.893
25	++++	1	1	1	1	7.461	13.525	4.777	5.411	7.221	15.159	26.963	7.924	8.127
Intact Values						5.452	9.997	6.345	4.029	3.347	9.813	14.388	10.163	6.212

Flexion Phase % of Intact Outcome Responses:

Run	Pattern	A	B	C	D	Proximal Region (%)	Inferior Neck (%)	Superior Neck (%)	Lateral Neck (%)	Proximal Region Standard Deviation (%)	Proximal Region 90th Percentile (%)	Inferior Neck 90th Percentile (%)	Superior Neck 90th Percentile (%)	Lateral Neck 90th Percentile (%)
1	----	-1	-1	-1	-1	123.41	121.32	115.63	152.80	122.57	118.49	143.41	117.74	149.76
2	----+	-1	-1	-1	1	157.75	172.95	100.28	157.08	235.77	153.70	224.52	97.93	155.87
3	---+-	-1	-1	1	-1	121.12	117.48	103.02	157.32	120.74	117.62	121.75	116.44	153.78
4	--++	-1	-1	1	1	160.12	158.41	80.21	161.49	257.83	173.54	211.10	82.06	156.32
5	a000	-1	0	0	0	121.83	127.42	103.14	163.58	112.74	119.11	127.21	104.61	154.33
6	-+---	-1	1	-1	-1	122.30	112.19	122.69	151.82	105.36	114.77	110.27	123.06	145.90
7	-+-+	-1	1	-1	1	158.96	185.13	96.02	162.37	232.47	163.06	235.28	93.14	153.61
8	-++-	-1	1	1	-1	151.91	200.44	105.52	161.59	273.23	144.02	308.99	118.40	149.98
9	-+++	-1	1	1	1	156.09	134.49	96.26	166.40	247.30	169.95	211.09	99.31	152.34
10	0a00	0	-1	0	0	115.14	109.04	98.74	146.03	102.49	111.94	106.27	100.63	146.05
11	00a0	0	0	-1	0	147.52	170.55	95.29	150.18	225.94	138.28	225.18	95.86	143.12
12	000a	0	0	0	-1	113.82	100.33	108.49	145.16	97.88	107.98	102.27	119.55	141.75
13	0000	0	0	0	0	113.93	106.85	100.07	146.69	100.29	109.98	107.40	102.54	141.86
14	000A	0	0	0	1	113.54	116.14	89.66	148.76	110.55	112.21	120.84	89.44	144.31
15	00A0	0	0	1	0	143.20	173.55	88.52	148.29	241.52	141.12	255.45	99.93	144.30
16	0A00	0	1	0	0	111.33	110.52	95.56	147.09	109.07	109.03	117.92	100.78	142.01
17	+----	1	-1	-1	-1	133.86	160.48	98.33	126.17	231.03	120.85	247.68	99.29	127.94
18	+---+	1	-1	-1	1	139.20	143.10	88.55	129.60	210.93	137.09	196.57	88.01	133.28
19	+--+-	1	-1	1	-1	131.61	146.74	89.69	132.91	216.04	123.20	208.95	101.54	134.53
20	+---+	1	-1	1	1	112.27	125.26	60.27	121.05	159.59	115.36	166.58	61.77	120.33
21	A000	1	0	0	0	105.52	96.04	87.19	132.39	97.90	102.20	100.13	88.19	133.61
22	++--	1	1	-1	-1	104.97	89.11	98.79	128.06	97.52	98.85	100.01	103.25	124.26

23	++-+	1	1	-1	1	107.21	116.25	74.76	129.96	142.02	102.59	153.97	73.62	126.39
24	+++-	1	1	1	-1	131.21	155.74	91.14	131.00	225.88	127.05	236.51	109.96	127.05
25	++++	1	1	1	1	136.86	135.29	75.28	134.29	215.77	154.47	187.40	77.96	130.82
Intact Values						100	100	100	100	100	100	100	100	100

Appendix B: Weighted Averages for Each Gait Phase

Neutral Phase:

Response Selection	A Weight	A Factor Setting (°)	A Weighted Value	B Weight	B Factor Setting (°)	B Weighted Value	C Weight	C Factor Setting (mm)	C Weighted Value	D Weight	D Factor Setting (mm)	D Weighted Value
Approach 1 Method 1	0.25	3.15	0.79	0.38	0.00	0.00	0.10	1.13	0.11	0.38	0.00	0.00
Approach 1 Method 2	0.19	10.00	1.88	0.13	0.00	0.00	0.25	0.78	0.19	0.30	0.00	0.00
Approach 2 Method 1	0.30	8.05	2.42	0.30	0.00	0.00	0.10	0.32	0.03	0.05	0.00	0.00
Approach 2 Method 2	0.45	10.00	4.50	0.15	0.00	0.00	0.25	-0.11	-0.03	0.25	0.12	0.03
Approach 3 Method 1	0.45	10.00	4.50	0.15	0.00	0.00	0.25	-0.43	-0.11	0.20	1.39	0.28
Approach 3 Method 2	0.45	10.00	4.50	0.23	0.00	0.00	0.25	-0.31	-0.08	0.20	1.23	0.25
Total	2.09	---	8.90°	1.33	---	0.00°	1.20	---	0.11mm	1.38	---	0.40mm

Extension Phase:

Response Selection	A Weight	A Factor Setting (°)	A Weighted Value	B Weight	B Factor Setting (°)	B Weighted Value	C Weight	C Factor Setting (mm)	C Weighted Value	D Weight	D Factor Setting (mm)	D Weighted Value
Approach 1 Method 1	0.20	-0.98	-0.20	0.15	0.00	0.00	0.65	-0.17	-0.11	0.45	0.00	0.00
Approach 1 Method 2	0.20	-0.88	-0.18	0.15	0.00	0.00	0.60	-0.17	-0.10	0.42	0.00	0.00
Approach 2 Method 1	0.45	10.00	4.50	0.23	0.00	0.00	0.35	-0.82	-0.29	0.35	0.00	0.00
Approach 2 Method 2	0.45	10.00	4.50	0.15	1.89	0.28	0.25	-0.44	-0.11	0.38	0.00	0.00
Approach 3 Method 1	0.45	10.00	4.50	0.23	0.00	0.00	0.25	-0.93	-0.23	0.30	0.00	0.00
Approach 3 Method 2	0.42	10.00	4.20	0.15	0.00	0.00	0.25	-0.91	-0.23	0.30	0.00	0.00
Total	2.17	---	7.99°	1.05	---	0.27°	2.35	---	-0.46mm	2.20	---	0.00mm

Flexion Phase:

Response Selection	A Weight	A Factor Setting (°)	A Weighted Value	B Weight	B Factor Setting (°)	B Weighted Value	C Weight	C Factor Setting (mm)	C Weighted Value	D Weight	D Factor Setting (mm)	D Weighted Value
Approach 1 Method 1	0.42	10.00	4.20	0.00	10.00	0.00	0.70	0.00	0.00	0.00	1.88	0.00
Approach 1 Method 2	0.41	10.00	4.05	0.00	10.00	0.00	0.73	0.01	0.01	0.00	1.85	0.00
Approach 2 Method 1	0.75	10.00	7.50	0.15	10.00	1.50	0.42	-1.55	-0.65	0.30	0.00	0.00
Approach 2 Method 2	0.68	10.00	6.75	0.15	10.00	1.50	0.35	-1.55	-0.54	0.45	0.00	0.00
Approach 3 Method 1	0.68	10.00	6.75	0.23	10.00	2.25	0.35	-1.38	-0.48	0.30	0.90	0.27
Approach 3 Method 2	0.60	10.00	6.00	0.23	10.00	2.25	0.35	-1.38	-0.48	0.33	0.89	0.29
Total	3.53	---	10.00°	0.75	---	10.00°	2.90	---	-0.74mm	1.38	---	0.41mm

Appendix C: Complete Neutral Phase DOE Results

Approach 1 Outcomes – Proximal Region

Method 1 - Mean Stress Model:

Summary of Fit

RSquare	0.666204
RSquare Adj	0.554939
Root Mean Square Error	0.501962
Mean of Response	7.181
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	9.051900	1.50865	5.9875
Error	18	4.535380	0.25197	
C. Total	24	13.587280		

Prob > F
0.0014*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	7.028	0.205937	34.13	<.0001*	6.5953424	7.4606576
A	-0.314444	0.118314	-2.66	0.0160*	-0.563012	-0.065877
B	0.4008333	0.118314	3.39	0.0033*	0.1522659	0.6494008
C	-0.2915	0.118314	-2.46	0.0240*	-0.540067	-0.042933
D	0.225	0.118314	1.90	0.0733	-0.023567	0.4735675
A*A	0.769	0.280327	2.74	0.0134*	0.1800545	1.3579455
C*C	-0.5565	0.280327	-1.99	0.0626	-1.145446	0.0324455

Method 1 - Stress Standard Deviation Model:

Summary of Fit

RSquare	0.43872
RSquare Adj	0.251626
Root Mean Square Error	1.407224
Mean of Response	5.11724
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	27.861598	4.64360	2.3449
Error	18	35.645031	1.98028	Prob > F
C. Total	24	63.506629		0.0752

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	4.906198	0.577334	8.50	<.0001*	3.6932646	6.1191314
A	-0.043556	0.331686	-0.13	0.8970	-0.740402	0.6532906
B	0.8443889	0.331686	2.55	0.0203*	0.1475427	1.5412351
C	-0.743389	0.331686	-2.24	0.0379*	-1.440235	-0.046543
D	0.2562222	0.331686	0.77	0.4499	-0.440624	0.9530684
A*A	1.0843069	0.785883	1.38	0.1846	-0.566772	2.7353856
C*C	-0.791193	0.785883	-1.01	0.3274	-2.442272	0.8598856

Method 1 - 90th Percentile Stress Model:

Summary of Fit

RSquare	0.8744
RSquare Adj	0.832534
Root Mean Square Error	0.770062
Mean of Response	12.69668
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	74.309670	12.3849	20.8854
Error	18	10.673927	0.5930	Prob > F
C. Total	24	84.983597		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	11.009446	0.315929	34.85	<.0001*	10.345703	11.673188
A	-0.586167	0.181505	-3.23	0.0047*	-0.967495	-0.204838
B	0.3869444	0.181505	2.13	0.0471*	0.0056157	0.7682732
C	0.586	0.181505	3.23	0.0047*	0.2046713	0.9673287
D	1.2551667	0.181505	6.92	<.0001*	0.8738379	1.6364954
A*A	-0.303559	0.430051	-0.71	0.4893	-1.207064	0.5999452
C*C	2.6469406	0.430051	6.15	<.0001*	1.7434359	3.5504452

Method 2 - Mean % of Intact Stress Model:

Summary of Fit

RSquare	0.874638
RSquare Adj	0.811957
Root Mean Square Error	6.538135
Mean of Response	140.4032
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	4771.8929	596.487	13.9538
Error	16	683.9554	42.747	Prob > F
C. Total	24	5455.8484		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	127.67018	2.682364	47.60	<.0001*	121.98382	133.35654
A	-6.984	1.541053	-4.53	0.0003*	-10.25089	-3.717113
B	3.5087778	1.541053	2.28	0.0369*	0.2418907	6.7756649
C	2.2661111	1.541053	1.47	0.1608	-1.000776	5.5329982
D	6.7857778	1.541053	4.40	0.0004*	3.5188907	10.052665
B*C	4.29975	1.634534	2.63	0.0182*	0.834693	7.764807
C*D	-4.394	1.634534	-2.69	0.0162*	-7.859057	-0.928943
B*B	-6.013624	3.651308	-1.65	0.1191	-13.75405	1.7268041
C*C	23.698376	3.651308	6.49	<.0001*	15.957948	31.438804

Method 2 - % of Intact Stress Standard Deviation Model:

Summary of Fit

RSquare	0.907275
RSquare Adj	0.860913
Root Mean Square Error	7.635869
Mean of Response	138.1478
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	9128.101	1141.01	19.5692
Error	16	932.904	58.31	Prob > F
C. Total	24	10061.005		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	119.93304	3.132725	38.28	<.0001*	113.29196	126.57412
A	-6.379056	1.799792	-3.54	0.0027*	-10.19444	-2.563668
B	4.2104444	1.799792	2.34	0.0326*	0.3950566	8.0258323
C	6.3755	1.799792	3.54	0.0027*	2.5601122	10.190888
D	13.657333	1.799792	7.59	<.0001*	9.8419455	17.472721
B*C	3.711625	1.908967	1.94	0.0697	-0.335205	7.7584549
C*D	-2.489375	1.908967	-1.30	0.2107	-6.536205	1.5574549
B*B	-3.800139	4.264352	-0.89	0.3861	-12.84016	5.2398843
C*C	29.098361	4.264352	6.82	<.0001*	20.058338	38.138384

Method 2 - 90th Percentile % of Intact Stress Model:

Summary of Fit

RSquare	0.807652
RSquare Adj	0.711478
Root Mean Square Error	29.65338
Mean of Response	171.6625
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	59075.304	7384.41	8.3978
Error	16	14069.170	879.32	Prob > F
C. Total	24	73144.474		0.0002*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	116.5055	12.16573	9.58	<.0001*	90.715318	142.29569
A	1.7622222	6.989369	0.25	0.8042	-13.05458	16.579024
B	7.4430556	6.989369	1.06	0.3027	-7.373746	22.259857
C	13.424667	6.989369	1.92	0.0728	-1.392135	28.241468
D	15.881389	6.989369	2.27	0.0372*	1.0645875	30.69819
B*C	18.946063	7.413346	2.56	0.0212*	3.2304714	34.661654
C*D	-21.85656	7.413346	-2.95	0.0094*	-37.57215	-6.140971
B*B	-19.01677	16.56032	-1.15	0.2677	-54.12309	16.089552
C*C	95.623733	16.56032	5.77	<.0001*	60.517413	130.73005

Approach 2 Outcomes – All Regions Mean Stress

Method 1 - Proximal Region Stress:

Summary of Fit

RSquare	0.594108
RSquare Adj	0.458811
Root Mean Square Error	0.553522
Mean of Response	7.181
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	8.072317	1.34539	4.3911
Error	18	5.514963	0.30639	Prob > F
C. Total	24	13.587280		0.0066*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.869	0.209212	32.83	<.0001*	6.4294625	7.3085375
A	-0.314444	0.130466	-2.41	0.0269*	-0.588544	-0.040345
B	0.4008333	0.130466	3.07	0.0066*	0.1267335	0.6749332
C	-0.2915	0.130466	-2.23	0.0384*	-0.5656	-0.0174
D	0.225	0.130466	1.72	0.1017	-0.0491	0.4990998
B*C	-0.028938	0.138381	-0.21	0.8367	-0.319664	0.2617893
A*A	0.4333333	0.246558	1.76	0.0958	-0.084667	0.9513333

Method 1 - Inferior Region Stress:

Summary of Fit

RSquare	0.342931
RSquare Adj	0.123908
Root Mean Square Error	2.282704
Mean of Response	12.79168
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	48.95158	8.15860	1.5657
Error	18	93.79324	5.21074	Prob > F
C. Total	24	142.74482		0.2141

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	12.354429	0.862781	14.32	<.0001*	10.541793	14.167064
A	-0.877722	0.538038	-1.63	0.1202	-2.008099	0.2526545
B	1.2486667	0.538038	2.32	0.0322*	0.11829	2.3790434
C	-0.517389	0.538038	-0.96	0.3490	-1.647766	0.6129878
D	-0.022333	0.538038	-0.04	0.9673	-1.15271	1.1080434
B*C	-0.144313	0.570676	-0.25	0.8032	-1.343258	1.054633
A*A	0.6072937	1.016797	0.60	0.5578	-1.528918	2.7435048

Method 1 - Superior Region Stress:

Summary of Fit

RSquare	0.531412
RSquare Adj	0.375216
Root Mean Square Error	0.7476
Mean of Response	6.36488
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	11.409098	1.90152	3.4022
Error	18	10.060307	0.55891	Prob > F
C. Total	24	21.469405		0.0201*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.47	0.282566	22.90	<.0001*	5.8763503	7.0636497
A	-0.498889	0.176211	-2.83	0.0111*	-0.869095	-0.128683
B	-0.023389	0.176211	-0.13	0.8959	-0.393595	0.3468168
C	0.2196111	0.176211	1.25	0.2286	-0.150595	0.5898168
D	-0.284167	0.176211	-1.61	0.1242	-0.654372	0.086039
B*C	-0.52975	0.1869	-2.83	0.0110*	-0.922412	-0.137088
A*A	-0.146	0.333008	-0.44	0.6663	-0.845623	0.5536229

Method 1 - Lateral Region Stress:

Summary of Fit

RSquare	0.476553
RSquare Adj	0.302071
Root Mean Square Error	0.432613
Mean of Response	6.40452
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	6	3.0669816	0.511164	2.7312
Error	18	3.3687766	0.187154	Prob > F
C. Total	24	6.4357582		0.0458*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.2155714	0.163512	38.01	<.0001*	5.8720445	6.5590983
A	-0.373778	0.101968	-3.67	0.0018*	-0.588004	-0.159551
B	0.0869444	0.101968	0.85	0.4050	-0.127282	0.3011711
C	-0.049722	0.101968	-0.49	0.6317	-0.263949	0.1645044
D	0.0141111	0.101968	0.14	0.8915	-0.200116	0.2283378
B*C	0.0361875	0.108153	0.33	0.7418	-0.191034	0.2634092
A*A	0.2624286	0.192701	1.36	0.1901	-0.142422	0.6672789

Method 2 - Proximal Region % of Intact:

Summary of Fit

RSquare	0.898899
RSquare Adj	0.848348
Root Mean Square Error	5.871511
Mean of Response	140.4032
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	4904.2541	613.032	17.7821
Error	16	551.5942	34.475	Prob > F
C. Total	24	5455.8484		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	125.952	2.219223	56.76	<.0001*	121.24746	130.65654
A	-6.984	1.383928	-5.05	0.0001*	-9.917797	-4.050203
B	3.5087778	1.383928	2.54	0.0220*	0.5749807	6.4425749
C	2.2661111	1.383928	1.64	0.1211	-0.667686	5.1999082
D	6.7857778	1.383928	4.90	0.0002*	3.8519807	9.7195749
A*B	-3.9395	1.467878	-2.68	0.0163*	-7.051262	-0.827738
B*C	4.29975	1.467878	2.93	0.0098*	1.1879882	7.4115118
C*D	-4.394	1.467878	-2.99	0.0086*	-7.505762	-1.282238
C*C	20.071111	2.615379	7.67	<.0001*	14.526756	25.615467

Method 2 - Inferior Region % of Intact:

Summary of Fit

RSquare	0.694296
RSquare Adj	0.541444
Root Mean Square Error	18.0902
Mean of Response	140.193
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	11891.868	1486.48	4.5423
Error	16	5236.083	327.26	Prob > F
C. Total	24	17127.951		0.0049*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	116.692	6.837451	17.07	<.0001*	102.19725	131.18675
A	-8.162944	4.2639	-1.91	0.0736	-17.20201	0.8761195
B	4.7283889	4.2639	1.11	0.2838	-4.310675	13.767453
C	0.0414444	4.2639	0.01	0.9924	-8.99762	9.0805084
D	3.0672222	4.2639	0.72	0.4823	-5.971842	12.106286
A*B	-5.93325	4.522549	-1.31	0.2081	-15.52063	3.6541251
B*C	9.318125	4.522549	2.06	0.0560	-0.26925	18.9055
C*D	-13.2255	4.522549	-2.92	0.0099*	-22.81288	-3.638125
C*C	32.640222	8.058013	4.05	0.0009*	15.557997	49.722447

Method 2 - Superior Region % of Intact:

Summary of Fit

RSquare	0.925494
RSquare Adj	0.888242
Root Mean Square Error	4.942392
Mean of Response	105.7401
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	4854.8861	606.861	24.8436
Error	16	390.8359	24.427	Prob > F
C. Total	24	5245.7220		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	108.292	1.868049	57.97	<.0001*	104.33191	112.25209
A	-8.808056	1.164933	-7.56	<.0001*	-11.2776	-6.338508
B	5.7685556	1.164933	4.95	0.0001*	3.2990078	8.2381033
C	-4.919111	1.164933	-4.22	0.0006*	-7.388659	-2.449563
D	-10.84333	1.164933	-9.31	<.0001*	-13.31288	-8.373786
A*B	-1.242125	1.235598	-1.01	0.3297	-3.861476	1.3772259
B*C	3.544625	1.235598	2.87	0.0111*	0.9252741	6.1639759
C*D	-1.073875	1.235598	-0.87	0.3976	-3.693226	1.5454759
C*C	-3.544333	2.201517	-1.61	0.1270	-8.21134	1.1226732

Method 2 - Lateral Region % of Intact:

Summary of Fit

RSquare	0.967758
RSquare Adj	0.951638
Root Mean Square Error	2.728333
Mean of Response	157.3572
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	3574.9084	446.864	60.0316
Error	16	119.1008	7.444	Prob > F
C. Total	24	3694.0092		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	158.85629	1.031213	154.05	<.0001*	156.67021	161.04236
A	-13.20617	0.643074	-20.54	<.0001*	-14.56942	-11.84291
B	2.5018889	0.643074	3.89	0.0013*	1.1386322	3.8651455
C	3.0302222	0.643074	4.71	0.0002*	1.6669656	4.3934789
D	2.6492222	0.643074	4.12	0.0008*	1.2859656	4.0124789
A*B	-0.512	0.682083	-0.75	0.4638	-1.957952	0.933952
B*C	-0.14775	0.682083	-0.22	0.8312	-1.593702	1.298202
C*D	-0.558125	0.682083	-0.82	0.4252	-2.004077	0.887827
C*C	-2.082063	1.215296	-1.71	0.1060	-4.658376	0.4942494

Approach 3 Outcomes – All Regions 90th Percentile

Method 1 - Proximal Region Stress:

Summary of Fit

RSquare	0.904268
RSquare Adj	0.856402
Root Mean Square Error	0.713077
Mean of Response	12.69668
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	76.847944	9.60599	18.8916
Error	16	8.135654	0.50848	Prob > F
C. Total	24	84.983597		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	10.922714	0.269518	40.53	<.0001*	10.351362	11.494066
A	-0.586167	0.168074	-3.49	0.0030*	-0.942467	-0.229866
B	0.3869444	0.168074	2.30	0.0351*	0.0306439	0.743245
C	0.586	0.168074	3.49	0.0030*	0.2296995	0.9423005
D	1.2551667	0.168074	7.47	<.0001*	0.8988662	1.6114672
B*C	0.3410625	0.178269	1.91	0.0738	-0.036851	0.7189763
B*D	-0.091813	0.178269	-0.52	0.6136	-0.469726	0.2861013
C*D	-0.228812	0.178269	-1.28	0.2176	-0.606726	0.1491013
C*C	2.4638413	0.31763	7.76	<.0001*	1.7904966	3.1371859

Method 1 - Inferior Region Stress:

Summary of Fit

RSquare	0.768869
RSquare Adj	0.653304
Root Mean Square Error	4.917527
Mean of Response	24.24108
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	1287.0897	160.886	6.6531
Error	16	386.9131	24.182	Prob > F
C. Total	24	1674.0028		0.0007*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	15.768143	1.85865	8.48	<.0001*	11.82798	19.708306
A	-0.311667	1.159072	-0.27	0.7914	-2.76879	2.1454567
B	1.3612778	1.159072	1.17	0.2574	-1.095846	3.8184011
C	1.7221111	1.159072	1.49	0.1568	-0.735012	4.1792344
D	0.7593333	1.159072	0.66	0.5217	-1.69779	3.2164567
B*C	3.8800625	1.229382	3.16	0.0061*	1.2738896	6.4862354
B*D	-0.376813	1.229382	-0.31	0.7632	-2.982985	2.2293604
C*D	-3.929938	1.229382	-3.20	0.0056*	-6.53611	-1.323765
C*C	11.767968	2.190441	5.37	<.0001*	7.1244416	16.411495

Method 1 - Superior Region Stress:

Summary of Fit

RSquare	0.900533
RSquare Adj	0.850799
Root Mean Square Error	0.563648
Mean of Response	10.50448
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	46.020924	5.75262	18.1071
Error	16	5.083182	0.31770	Prob > F
C. Total	24	51.104106		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	10.730857	0.213039	50.37	<.0001*	10.279235	11.182479
A	-0.8095	0.132853	-6.09	<.0001*	-1.091136	-0.527864
B	0.4792778	0.132853	3.61	0.0024*	0.1976418	0.7609137
C	-0.413222	0.132853	-3.11	0.0067*	-0.694858	-0.131586
D	-1.152	0.132853	-8.67	<.0001*	-1.433636	-0.870364
B*C	0.380125	0.140912	2.70	0.0158*	0.081405	0.678845
B*D	0.0905	0.140912	0.64	0.5298	-0.20822	0.38922
C*D	-0.1085	0.140912	-0.77	0.4525	-0.40722	0.19022
C*C	-0.314413	0.251069	-1.25	0.2285	-0.846655	0.2178292

Method 1 - Lateral Region Stress:

Summary of Fit

RSquare	0.966234
RSquare Adj	0.949351
Root Mean Square Error	0.14555
Mean of Response	8.994
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	9.699510	1.21244	57.2316
Error	16	0.338956	0.02118	Prob > F
C. Total	24	10.038466		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	9.0821429	0.055013	165.09	<.0001*	8.9655211	9.1987646
A	-0.702278	0.034306	-20.47	<.0001*	-0.775004	-0.629551
B	0.0907222	0.034306	2.64	0.0177*	0.0179958	0.1634486
C	0.0573889	0.034306	1.67	0.1138	-0.015338	0.1301153
D	0.1328889	0.034306	3.87	0.0013*	0.0601625	0.2056153
B*C	-0.0295	0.036387	-0.81	0.4294	-0.106638	0.047638
B*D	0.077875	0.036387	2.14	0.0481*	0.000737	0.155013
C*D	-0.083	0.036387	-2.28	0.0366*	-0.160138	-0.005862
C*C	-0.122421	0.064833	-1.89	0.0773	-0.259861	0.0150194

Method 2 - Proximal Region % of Intact:

Summary of Fit

RSquare	0.902673
RSquare Adj	0.862597
Root Mean Square Error	7.589494
Mean of Response	138.1478
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	9081.798	1297.40	22.5241
Error	17	979.207	57.60	Prob > F
C. Total	24	10061.005		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	118.84729	2.868559	41.43	<.0001*	112.79516	124.89942
A	-6.379056	1.788861	-3.57	0.0024*	-10.15322	-2.604889
B	4.2104444	1.788861	2.35	0.0309*	0.4362781	7.9846108
C	6.3755	1.788861	3.56	0.0024*	2.6013336	10.149666
D	13.657333	1.788861	7.63	<.0001*	9.8831669	17.4315
B*C	3.711625	1.897373	1.96	0.0671	-0.291483	7.714733
C*D	-2.489375	1.897373	-1.31	0.2070	-6.492483	1.513733
C*C	26.806214	3.380629	7.93	<.0001*	19.67371	33.938718

Method 2 - Inferior Region % of Intact:

Summary of Fit

RSquare	0.767503
RSquare Adj	0.67177
Root Mean Square Error	35.15092
Mean of Response	178.0861
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	69340.374	9905.77	8.0171
Error	17	21004.983	1235.59	Prob > F
C. Total	24	90345.357		0.0002*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	115.841	13.2858	8.72	<.0001*	87.810414	143.87159
A	-2.289833	8.285151	-0.28	0.7856	-19.76997	15.190308
B	10.0005	8.285151	1.21	0.2439	-7.479642	27.480642
C	12.651222	8.285151	1.53	0.1452	-4.828919	30.131364
D	5.5777222	8.285151	0.67	0.5099	-11.90242	23.057864
B*C	28.504562	8.78773	3.24	0.0048*	9.9640725	47.045053
C*D	-28.87081	8.78773	-3.29	0.0044*	-47.4113	-10.33032
C*C	86.451556	15.65746	5.52	<.0001*	53.417193	119.48592

Method 2 - Superior Region % of Intact:

Summary of Fit

RSquare	0.898005
RSquare Adj	0.856008
Root Mean Square Error	5.730297
Mean of Response	108.7023
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	4914.7894	702.113	21.3822
Error	17	558.2172	32.836	Prob > F
C. Total	24	5473.0066		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	111.04586	2.165849	51.27	<.0001*	106.47632	115.6154
A	-8.377889	1.350644	-6.20	<.0001*	-11.2275	-5.528279
B	4.9595	1.350644	3.67	0.0019*	2.1098902	7.8091098
C	-4.275222	1.350644	-3.17	0.0057*	-7.124832	-1.425612
D	-11.92183	1.350644	-8.83	<.0001*	-14.77144	-9.072224
B*C	3.93475	1.432574	2.75	0.0138*	0.9122824	6.9572176
C*D	-1.1235	1.432574	-0.78	0.4437	-4.145968	1.8989676
C*C	-3.254968	2.552477	-1.28	0.2194	-8.640225	2.130288

Method 2 - Lateral Region % of Intact:

Summary of Fit

RSquare	0.956606
RSquare Adj	0.938738
Root Mean Square Error	2.724056
Mean of Response	153.0454
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	2780.9021	397.272	53.5372
Error	17	126.1482	7.420	Prob > F
C. Total	24	2907.0503		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	154.54314	1.029597	150.10	<.0001*	152.37088	156.7154
A	-11.95094	0.642066	-18.61	<.0001*	-13.30559	-10.5963
B	1.5449444	0.642066	2.41	0.0278*	0.190303	2.8995859
C	0.9774444	0.642066	1.52	0.1463	-0.377197	2.3320859
D	2.2623889	0.642066	3.52	0.0026*	0.9077475	3.6170303
B*C	-0.502125	0.681014	-0.74	0.4710	-1.938939	0.9346892
C*D	-1.412375	0.681014	-2.07	0.0536	-2.849189	0.0244392
C*C	-2.080143	1.213391	-1.71	0.1046	-4.640175	0.4798888

Appendix D: Complete Extension Phase DOE Results

Approach 1 Outcomes – Proximal Region

Method 1 - Mean Stress Model:

Summary of Fit

RSquare	0.901703
RSquare Adj	0.842724
Root Mean Square Error	0.367287
Mean of Response	8.24896
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	9	18.561992	2.06244	15.2887
Error	15	2.023497	0.13490	Prob > F
C. Total	24	20.585489		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	7.343	0.138821	52.90	<.0001*	7.0471091	7.6388909
A	-0.345889	0.08657	-4.00	0.0012*	-0.530409	-0.161368
B	0.1925	0.08657	2.22	0.0420*	0.0079796	0.3770204
C	0.3245	0.08657	3.75	0.0019*	0.1399796	0.5090204
D	0.3377778	0.08657	3.90	0.0014*	0.1532574	0.5222982
A*B	-0.217937	0.091822	-2.37	0.0314*	-0.413651	-0.022224
B*C	0.2461875	0.091822	2.68	0.0171*	0.050474	0.441901
A*D	-0.323313	0.091822	-3.52	0.0031*	-0.519026	-0.127599
C*D	-0.160188	0.091822	-1.74	0.1015	-0.355901	0.035526
C*C	1.2582778	0.163603	7.69	<.0001*	0.909567	1.6069886

Method 1 - Stress Standard Deviation Model:

Summary of Fit

RSquare	0.894579
RSquare Adj	0.831327
Root Mean Square Error	0.834157
Mean of Response	6.23052
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	9	88.568361	9.84093	14.1430
Error	15	10.437263	0.69582	Prob > F
C. Total	24	99.005624		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	4.0817143	0.315282	12.95	<.0001*	3.4097074	4.7537212
A	0.0113333	0.196613	0.06	0.9548	-0.407737	0.4304033
B	0.1127778	0.196613	0.57	0.5747	-0.306292	0.5318477
C	0.7141667	0.196613	3.63	0.0025*	0.2950967	1.1332366
D	0.5193333	0.196613	2.64	0.0185*	0.1002634	0.9384033
A*B	-0.57125	0.208539	-2.74	0.0152*	-1.015741	-0.126759
B*C	0.738625	0.208539	3.54	0.0030*	0.2941342	1.1831158
A*D	-0.746	0.208539	-3.58	0.0028*	-1.190491	-0.301509
C*D	-0.640125	0.208539	-3.07	0.0078*	-1.084616	-0.195634
C*C	2.9844524	0.371563	8.03	<.0001*	2.1924847	3.7764201

Method 1 - 90th Percentile Stress Model:

Summary of Fit

RSquare	0.896626
RSquare Adj	0.834602
Root Mean Square Error	0.86893
Mean of Response	14.91264
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	9	98.23377	10.9149	14.4560
Error	15	11.32558	0.7550	Prob > F
C. Total	24	109.55935		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	12.87	0.328425	39.19	<.0001*	12.16998	13.57002
A	-0.594222	0.204809	-2.90	0.0110*	-1.030762	-0.157683
B	0.3618333	0.204809	1.77	0.0976	-0.074706	0.7983727
C	0.6735556	0.204809	3.29	0.0050*	0.2370162	1.1100949
D	1.2533333	0.204809	6.12	<.0001*	0.816794	1.6898727
A*B	-0.483	0.217232	-2.22	0.0420*	-0.94602	-0.01998
B*C	0.413375	0.217232	1.90	0.0764	-0.049645	0.8763949
A*D	-0.57175	0.217232	-2.63	0.0189*	-1.03477	-0.10873
C*D	-0.226125	0.217232	-1.04	0.3144	-0.689145	0.2368949
C*C	2.837	0.387052	7.33	<.0001*	2.0120182	3.6619818

Proximal Region Stress Desirability:

Method 2 - Mean % of Intact Stress Model:

Summary of Fit

RSquare	0.901786
RSquare Adj	0.842857
Root Mean Square Error	6.02254
Mean of Response	135.3211
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	9	4995.0235	555.003	15.3030
Error	15	544.0131	36.268	Prob > F
C. Total	24	5539.0366		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	120.46051	2.276198	52.92	<.0001*	115.60891	125.31212
A	-5.675418	1.419459	-4.00	0.0012*	-8.700923	-2.649913
B	3.1578305	1.419459	2.22	0.0419*	0.1323256	6.1833353
C	5.3220977	1.419459	3.75	0.0019*	2.2965928	8.3476025
D	5.5409396	1.419459	3.90	0.0014*	2.5154348	8.5664445
A*B	-3.574648	1.505563	-2.37	0.0314*	-6.78368	-0.365615
B*C	4.0375332	1.505563	2.68	0.0171*	0.8285007	7.2465657
A*D	-5.306571	1.505563	-3.52	0.0031*	-8.515603	-2.097538
C*D	-2.626808	1.505563	-1.74	0.1015	-5.83584	0.5822246
C*C	20.63971	2.682525	7.69	<.0001*	14.922043	26.357376

Method 2 - % of Intact Stress Standard Deviation Model:

Summary of Fit

RSquare	0.894546
RSquare Adj	0.831273
Root Mean Square Error	24.60414
Mean of Response	183.7604
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	9	77027.672	8558.63	14.1380
Error	15	9080.454	605.36	Prob > F
C. Total	24	86108.126		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	120.39211	9.29949	12.95	<.0001*	100.57071	140.2135
A	0.3352111	5.799251	0.06	0.9547	-12.0256	12.696022
B	3.3237075	5.799251	0.57	0.5751	-9.037103	15.684518
C	21.063228	5.799251	3.63	0.0025*	8.7024174	33.424039
D	15.312249	5.799251	2.64	0.0185*	2.9514387	27.67306
A*B	-16.8452	6.151034	-2.74	0.0152*	-29.95582	-3.734584
B*C	21.78173	6.151034	3.54	0.0030*	8.6711103	34.89235
A*D	-22.00447	6.151034	-3.58	0.0028*	-35.11509	-8.893853
C*D	-18.87825	6.151034	-3.07	0.0078*	-31.98887	-5.767628
C*C	88.01146	10.95955	8.03	<.0001*	64.651724	111.3712

Method 2 - 90th Percentile % of Intact Stress Model:

Summary of Fit

RSquare	0.896623
RSquare Adj	0.834597
Root Mean Square Error	8.090644
Mean of Response	138.8471
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	9	8516.1715	946.241	14.4556
Error	15	981.8777	65.459	Prob > F
C. Total	24	9498.0492		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	119.82892	3.057976	39.19	<.0001*	113.311	126.34684
A	-5.532796	1.906983	-2.90	0.0110*	-9.597434	-1.468158
B	3.3690348	1.906983	1.77	0.0976	-0.695603	7.4336728
C	6.2721712	1.906983	3.29	0.0050*	2.2075332	10.336809
D	11.669444	1.906983	6.12	<.0001*	7.6048064	15.734082
A*B	-4.497139	2.022661	-2.22	0.0420*	-8.808339	-0.18594
B*C	3.8488799	2.022661	1.90	0.0764	-0.46232	8.1600795
A*D	-5.324281	2.022661	-2.63	0.0188*	-9.635481	-1.013081
C*D	-2.105446	2.022661	-1.04	0.3144	-6.416646	2.2057533
C*C	26.4142	3.603859	7.33	<.0001*	18.732756	34.095643

Proximal Region % of Intact Desirability:

Approach 2 Outcomes – All Regions Mean Stress

Method 1 - Proximal Region Stress:

Summary of Fit

RSquare	0.927847
RSquare Adj	0.826834
Root Mean Square Error	0.385396
Mean of Response	8.24896
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	14	19.100190	1.36430	9.1854
Error	10	1.485298	0.14853	
C. Total	24	20.585489		

Prob > F
0.0006*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	7.4708475	0.166409	44.89	<.0001*	7.1000648	7.8416302
A	-0.345889	0.090839	-3.81	0.0034*	-0.54829	-0.143488
B	0.1925	0.090839	2.12	0.0601	-0.009901	0.3949011
C	0.3245	0.090839	3.57	0.0051*	0.1220989	0.5269011
D	0.3377778	0.090839	3.72	0.0040*	0.1353767	0.5401789
A*B	-0.217937	0.096349	-2.26	0.0472*	-0.432616	-0.003259
A*C	-0.016063	0.096349	-0.17	0.8709	-0.230741	0.1986163
B*C	0.2461875	0.096349	2.56	0.0286*	0.0315087	0.4608663
A*D	-0.323313	0.096349	-3.36	0.0073*	-0.537991	-0.108634
B*D	-0.099312	0.096349	-1.03	0.3269	-0.313991	0.1153663
C*D	-0.160188	0.096349	-1.66	0.1274	-0.374866	0.0544913
A*A	-0.155322	0.241497	-0.64	0.5346	-0.693412	0.3827679
B*B	-0.162822	0.241497	-0.67	0.5155	-0.700912	0.3752679
C*C	1.528178	0.241497	6.33	<.0001*	0.990088	2.0662679
D*D	-0.129322	0.241497	-0.54	0.6040	-0.667412	0.4087679

Method 1 - Inferior Region Stress:

Summary of Fit

RSquare	0.803916
RSquare Adj	0.5294
Root Mean Square Error	1.829536
Mean of Response	14.3012
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	14	137.23092	9.80221	2.9285
Error	10	33.47203	3.34720	Prob > F
C. Total	24	170.70295		0.0467*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	12.500661	0.789971	15.82	<.0001*	10.740495	14.260827
A	-1.075722	0.431226	-2.49	0.0317*	-2.036553	-0.114891
B	0.4887222	0.431226	1.13	0.2835	-0.472109	1.4495533
C	0.1700556	0.431226	0.39	0.7016	-0.790776	1.1308866
D	0.5104444	0.431226	1.18	0.2639	-0.450387	1.4712755
A*B	-0.60025	0.457384	-1.31	0.2187	-1.619365	0.4188653
A*C	-0.345125	0.457384	-0.75	0.4679	-1.36424	0.6739903
B*C	0.855375	0.457384	1.87	0.0910	-0.16374	1.8744903
A*D	-0.742875	0.457384	-1.62	0.1354	-1.76199	0.2762403
B*D	-0.299625	0.457384	-0.66	0.5272	-1.31874	0.7194903
C*D	-0.8315	0.457384	-1.82	0.0991	-1.850615	0.1876153
A*A	-0.621438	1.146428	-0.54	0.5996	-3.175839	1.9329631
B*B	-0.745438	1.146428	-0.65	0.5302	-3.299839	1.8089631
C*C	4.6875621	1.146428	4.09	0.0022*	2.1331612	7.2419631
D*D	-0.819938	1.146428	-0.72	0.4908	-3.374339	1.7344631

Method 1 - Superior Region Stress:

Summary of Fit

RSquare	0.966723
RSquare Adj	0.920136
Root Mean Square Error	0.254323
Mean of Response	7.03968
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	14	18.790422	1.34217	20.7509
Error	10	0.646804	0.06468	Prob > F
C. Total	24	19.437225		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	7.1399492	0.109814	65.02	<.0001*	6.8952689	7.3846294
A	-0.545889	0.059945	-9.11	<.0001*	-0.679454	-0.412324
B	0.3994444	0.059945	6.66	<.0001*	0.2658796	0.5330093
C	-0.094889	0.059945	-1.58	0.1445	-0.228454	0.038676
D	-0.733278	0.059945	-12.23	<.0001*	-0.866843	-0.599713
A*B	0.0315	0.063581	0.50	0.6310	-0.110167	0.1731669
A*C	-0.01375	0.063581	-0.22	0.8331	-0.155417	0.1279169
B*C	0.133	0.063581	2.09	0.0629	-0.008667	0.2746669
A*D	0.001375	0.063581	0.02	0.9832	-0.140292	0.1430419
B*D	-0.109375	0.063581	-1.72	0.1161	-0.251042	0.0322919
C*D	0.027625	0.063581	0.43	0.6732	-0.114042	0.1692919
A*A	-0.116441	0.159365	-0.73	0.4818	-0.471527	0.2386459
B*B	-0.040441	0.159365	-0.25	0.8048	-0.395527	0.3146459
C*C	-0.163441	0.159365	-1.03	0.3293	-0.518527	0.1916459
D*D	0.1810593	0.159365	1.14	0.2824	-0.174027	0.5361459

Method 1 - Lateral Region Stress:

Summary of Fit

RSquare	0.984491
RSquare Adj	0.962778
Root Mean Square Error	0.083769
Mean of Response	6.59052
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	14	4.4544400	0.318174	45.3419
Error	10	0.0701722	0.007017	Prob > F
C. Total	24	4.5246122		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.6977288	0.03617	185.17	<.0001*	6.6171362	6.7783214
A	-0.437722	0.019745	-22.17	<.0001*	-0.481716	-0.393729
B	0.0928333	0.019745	4.70	0.0008*	0.0488398	0.1368268
C	0.1688889	0.019745	8.55	<.0001*	0.1248954	0.2128824
D	0.1078333	0.019745	5.46	0.0003*	0.0638398	0.1518268
A*B	-0.01175	0.020942	-0.56	0.5871	-0.058412	0.0349122
A*C	0.000875	0.020942	0.04	0.9675	-0.045787	0.0475372
B*C	0.0015	0.020942	0.07	0.9443	-0.045162	0.0481622
A*D	-0.031875	0.020942	-1.52	0.1590	-0.078537	0.0147872
B*D	0.01475	0.020942	0.70	0.4973	-0.031912	0.0614122
C*D	-0.015375	0.020942	-0.73	0.4797	-0.062037	0.0312872
A*A	-0.02785	0.052491	-0.53	0.6073	-0.144808	0.0891079
B*B	-0.06685	0.052491	-1.27	0.2316	-0.183808	0.0501079
C*C	0.0456497	0.052491	0.87	0.4049	-0.071308	0.1626079
D*D	-0.09985	0.052491	-1.90	0.0863	-0.216808	0.0171079

All Regions Mean Stress Desirability:

Method 2 - Proximal Region % of Intact:

Summary of Fit

RSquare	0.881854
RSquare Adj	0.822781
Root Mean Square Error	6.395384
Mean of Response	135.3211
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	4884.6215	610.578	14.9282
Error	16	654.4150	40.901	Prob > F
C. Total	24	5539.0366		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	120.46051	2.417228	49.83	<.0001*	115.33622	125.58481
A	-5.675418	1.507406	-3.77	0.0017*	-8.870977	-2.479859
B	3.1578305	1.507406	2.09	0.0525	-0.037729	6.3533895
C	5.3220977	1.507406	3.53	0.0028*	2.1265386	8.5176567
D	5.5409396	1.507406	3.68	0.0020*	2.3453806	8.7364986
A*B	-3.574648	1.598846	-2.24	0.0400*	-6.96405	-0.185245
B*C	4.0375332	1.598846	2.53	0.0225*	0.648131	7.4269354
A*D	-5.306571	1.598846	-3.32	0.0043*	-8.695973	-1.917169
C*C	20.63971	2.848731	7.25	<.0001*	14.600671	26.678749

Method 2 - Inferior Region % of Intact:

Summary of Fit

RSquare	0.666833
RSquare Adj	0.50025
Root Mean Square Error	17.84768
Mean of Response	135.3797
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of		F Ratio
		Squares	Mean Square	
Model	8	10200.905	1275.11	4.0030
Error	16	5096.635	318.54	Prob > F
C. Total	24	15297.540		0.0088*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	112.4194	6.745789	16.67	<.0001*	98.118967	126.71983
A	-10.18306	4.206739	-2.42	0.0278*	-19.10095	-1.265172
B	4.62477	4.206739	1.10	0.2879	-4.293117	13.542658
C	1.6101907	4.206739	0.38	0.7069	-7.307697	10.528078
D	4.8322092	4.206739	1.15	0.2676	-4.085678	13.750097
A*B	-5.681824	4.46192	-1.27	0.2211	-15.14067	3.7770237
B*C	8.0973817	4.46192	1.81	0.0884	-1.361466	17.55623
A*D	-7.033502	4.46192	-1.58	0.1345	-16.49235	2.4253456
C*C	31.889256	7.949989	4.01	0.0010*	15.036033	48.742479

Method 2 - Superior Region % of Intact:

Summary of Fit

RSquare	0.951125
RSquare Adj	0.926688
Root Mean Square Error	3.399263
Mean of Response	98.2038
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	3597.8466	449.731	38.9209
Error	16	184.8798	11.555	Prob > F
C. Total	24	3782.7264		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	99.696805	1.284801	77.60	<.0001*	96.973149	102.42046
A	-7.614389	0.801214	-9.50	<.0001*	-9.312887	-5.915892
B	5.570768	0.801214	6.95	<.0001*	3.8722702	7.2692657
C	-1.324214	0.801214	-1.65	0.1179	-3.022712	0.3742839
D	-10.23123	0.801214	-12.77	<.0001*	-11.92973	-8.532731
A*B	0.4389487	0.849816	0.52	0.6126	-1.36258	2.2404776
B*C	1.8556427	0.849816	2.18	0.0442*	0.0541138	3.6571716
A*D	0.0199979	0.849816	0.02	0.9815	-1.781531	1.8215268
C*C	-2.073619	1.514152	-1.37	0.1898	-5.283478	1.1362404

Method 2 - Lateral Region % of Intact:

Summary of Fit

RSquare	0.966058
RSquare Adj	0.949087
Root Mean Square Error	2.111947
Mean of Response	142.027
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	2031.1935	253.899	56.9240
Error	16	71.3651	4.460	Prob > F
C. Total	24	2102.5586		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	143.14207	0.798241	179.32	<.0001*	141.44988	144.83427
A	-9.436322	0.497791	-18.96	<.0001*	-10.49159	-8.381053
B	1.9991581	0.497791	4.02	0.0010*	0.943889	3.0544272
C	3.6398423	0.497791	7.31	<.0001*	2.5845733	4.6951114
D	2.325121	0.497791	4.67	0.0003*	1.2698519	3.3803901
A*B	-0.252964	0.527987	-0.48	0.6383	-1.372246	0.8663174
B*C	0.0341589	0.527987	0.06	0.9492	-1.085123	1.1534408
A*D	-0.686675	0.527987	-1.30	0.2118	-1.805957	0.432607
C*C	-1.548713	0.940736	-1.65	0.1192	-3.542984	0.445558

All Regions % of Intact Desirability:

Approach 3 Outcomes – All Regions 90th Percentile

Method 1 - Proximal Region Stress:

Summary of Fit

RSquare	0.862557
RSquare Adj	0.793835
Root Mean Square Error	0.970123
Mean of Response	14.91264
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	94.50115	11.8126	12.5515
Error	16	15.05820	0.9411	Prob > F
C. Total	24	109.55935		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	12.87	0.366672	35.10	<.0001*	12.09269	13.64731
A	-0.594222	0.22866	-2.60	0.0194*	-1.07896	-0.109485
B	0.3618333	0.22866	1.58	0.1331	-0.122904	0.846571
C	0.6735556	0.22866	2.95	0.0095*	0.1888179	1.1582933
D	1.2533333	0.22866	5.48	<.0001*	0.7685956	1.738071
B*C	0.413375	0.242531	1.70	0.1076	-0.100767	0.927517
A*D	-0.57175	0.242531	-2.36	0.0315*	-1.085892	-0.057608
C*D	-0.226125	0.242531	-0.93	0.3650	-0.740267	0.288017
C*C	2.837	0.432127	6.57	<.0001*	1.9209319	3.7530681

Method 1 - Inferior Region Stress:

Summary of Fit

RSquare	0.794707
RSquare Adj	0.692061
Root Mean Square Error	4.473087
Mean of Response	25.7994
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	1239.2766	154.910	7.7422
Error	16	320.1361	20.009	Prob > F
C. Total	24	1559.4126		0.0003*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	17.757571	1.690668	10.50	<.0001*	14.173516	21.341627
A	-1.341278	1.054317	-1.27	0.2215	-3.576329	0.8937737
B	1.1553333	1.054317	1.10	0.2894	-1.079718	3.3903848
C	1.6946667	1.054317	1.61	0.1275	-0.540385	3.9297181
D	1.4048889	1.054317	1.33	0.2014	-0.830163	3.6399403
B*C	3.7716875	1.118272	3.37	0.0039*	1.4010574	6.1423176
A*D	-2.152188	1.118272	-1.92	0.0723	-4.522818	0.2184426
C*D	-3.213063	1.118272	-2.87	0.0110*	-5.583693	-0.842432
C*C	11.169206	1.992471	5.61	<.0001*	6.9453561	15.393057

Method 1 - Superior Region Stress:

Summary of Fit

RSquare	0.921522
RSquare Adj	0.882283
Root Mean Square Error	0.499001
Mean of Response	11.38644
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	46.782087	5.84776	23.4848
Error	16	3.984030	0.24900	Prob > F
C. Total	24	50.766116		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	11.483	0.188605	60.88	<.0001*	11.083176	11.882824
A	-0.900778	0.117616	-7.66	<.0001*	-1.150112	-0.651444
B	0.6197778	0.117616	5.27	<.0001*	0.3704438	0.8691118
C	-0.371778	0.117616	-3.16	0.0061*	-0.621112	-0.122444
D	-1.097778	0.117616	-9.33	<.0001*	-1.347112	-0.848444
B*C	0.2399375	0.12475	1.92	0.0724	-0.024521	0.5043961
A*D	0.0509375	0.12475	0.41	0.6885	-0.213521	0.3153961
C*D	0.0428125	0.12475	0.34	0.7359	-0.221646	0.3072711
C*C	-0.134111	0.222273	-0.60	0.5547	-0.605308	0.3370858

Method 1 - Lateral Region Stress:

Summary of Fit

RSquare	0.960098
RSquare Adj	0.940147
Root Mean Square Error	0.134682
Mean of Response	9.43016
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	6.9833099	0.872914	48.1227
Error	16	0.2902294	0.018139	Prob > F
C. Total	24	7.2735394		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	9.5408571	0.050905	187.42	<.0001*	9.432943	9.6487712
A	-0.557667	0.031745	-17.57	<.0001*	-0.624963	-0.49037
B	0.1826111	0.031745	5.75	<.0001*	0.1153148	0.2499074
C	0.1585556	0.031745	4.99	0.0001*	0.0912593	0.2258518
D	0.0663889	0.031745	2.09	0.0528	-0.000907	0.1336852
B*C	0.0349375	0.033671	1.04	0.3149	-0.036441	0.106316
A*D	-0.075438	0.033671	-2.24	0.0396*	-0.146816	-0.004059
C*D	-0.038437	0.033671	-1.14	0.2704	-0.109816	0.032941
C*C	-0.153746	0.059992	-2.56	0.0209*	-0.280924	-0.026568

All Regions 90th Percentile Stress Desirability:

Method 2 - Proximal Region % of Intact:

Summary of Fit

RSquare	0.862554
RSquare Adj	0.793831
Root Mean Square Error	9.032808
Mean of Response	138.8471
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	8192.5833	1024.07	12.5512
Error	16	1305.4659	81.59	Prob > F
C. Total	24	9498.0492		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	119.82892	3.41408	35.10	<.0001*	112.59139	127.06645
A	-5.532796	2.129053	-2.60	0.0194*	-10.04619	-1.019405
B	3.3690348	2.129053	1.58	0.1331	-1.144356	7.882426
C	6.2721712	2.129053	2.95	0.0095*	1.75878	10.785562
D	11.669444	2.129053	5.48	<.0001*	7.1560532	16.182836
B*C	3.8488799	2.258202	1.70	0.1076	-0.938294	8.6360542
A*D	-5.324281	2.258202	-2.36	0.0314*	-10.11146	-0.537107
C*D	-2.105446	2.258202	-0.93	0.3650	-6.892621	2.681728
C*C	26.4142	4.023532	6.56	<.0001*	17.884692	34.943707

Method 2 - Inferior Region % of Intact:

Summary of Fit

RSquare	0.7947
RSquare Adj	0.69205
Root Mean Square Error	27.15992
Mean of Response	156.6488
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	45686.839	5710.85	7.7418
Error	16	11802.576	737.66	Prob > F
C. Total	24	57489.415		0.0003*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	107.82121	10.26548	10.50	<.0001*	86.059359	129.58306
A	-8.143707	6.401653	-1.27	0.2215	-21.71461	5.4271924
B	7.0150548	6.401653	1.10	0.2894	-6.555844	20.585954
C	10.28966	6.401653	1.61	0.1275	-3.281239	23.860559
D	8.5309208	6.401653	1.33	0.2013	-5.039978	22.10182
B*C	22.900513	6.789979	3.37	0.0039*	8.5064009	37.294625
A*D	-13.06749	6.789979	-1.92	0.0723	-27.4616	1.3266237
C*D	-19.50864	6.789979	-2.87	0.0110*	-33.90275	-5.114525
C*C	67.816045	12.09799	5.61	<.0001*	42.169456	93.462633

Method 2 - Superior Region % of Intact:

Summary of Fit

RSquare	0.921578
RSquare Adj	0.882367
Root Mean Square Error	4.390826
Mean of Response	100.2198
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	3624.9734	453.122	23.5030
Error	16	308.4696	19.279	Prob > F
C. Total	24	3933.4430		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	101.07004	1.659576	60.90	<.0001*	97.551895	104.58818
A	-7.929001	1.034928	-7.66	<.0001*	-10.12295	-5.735053
B	5.4550202	1.034928	5.27	<.0001*	3.2610717	7.6489687
C	-3.273169	1.034928	-3.16	0.0060*	-5.467117	-1.07922
D	-9.663764	1.034928	-9.34	<.0001*	-11.85771	-7.469816
B*C	2.1115355	1.097706	1.92	0.0724	-0.215498	4.4385693
A*D	0.4487673	1.097706	0.41	0.6881	-1.878266	2.7758011
C*D	0.3771542	1.097706	0.34	0.7356	-1.94988	2.704188
C*C	-1.180941	1.955829	-0.60	0.5544	-5.327114	2.965232

Method 2 - Lateral Region % of Intact:

Summary of Fit

RSquare	0.96013
RSquare Adj	0.940195
Root Mean Square Error	2.008354
Mean of Response	140.6806
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	1554.1151	194.264	48.1629
Error	16	64.5358	4.033	Prob > F
C. Total	24	1618.6509		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	142.33183	0.759086	187.50	<.0001*	140.72264	143.94102
A	-8.319758	0.473374	-17.58	<.0001*	-9.323265	-7.316251
B	2.7252738	0.473374	5.76	<.0001*	1.7217668	3.7287808
C	2.3630256	0.473374	4.99	0.0001*	1.3595186	3.3665326
D	0.9896931	0.473374	2.09	0.0529	-0.013814	1.9932001
B*C	0.5220268	0.502088	1.04	0.3139	-0.542353	1.5864067
A*D	-1.12519	0.502088	-2.24	0.0396*	-2.18957	-0.06081
C*D	-0.571816	0.502088	-1.14	0.2715	-1.636196	0.4925635
C*C	-2.293315	0.894592	-2.56	0.0208*	-4.189765	-0.396865

All Regions 90th Percentile % of Intact Desirability:

Appendix E: Complete Flexion Phase DOE Results

Approach 1 Outcomes – Proximal Region

Method 1 - Mean Stress Model:

Summary of Fit

RSquare	0.818672
RSquare Adj	0.744008
Root Mean Square Error	0.5026
Mean of Response	7.05376
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	19.388319	2.76976	10.9647
Error	17	4.294318	0.25261	Prob > F
C. Total	24	23.682637		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.1922857	0.189965	32.60	<.0001*	5.7914946	6.5930768
A	-0.517389	0.118464	-4.37	0.0004*	-0.767326	-0.267452
B	-0.041278	0.118464	-0.35	0.7318	-0.291215	0.2086594
C	0.1491111	0.118464	1.26	0.2251	-0.100826	0.3990483
D	0.3265556	0.118464	2.76	0.0135*	0.0766184	0.5764927
B*C	0.3806875	0.12565	3.03	0.0076*	0.1155891	0.6457859
A*D	-0.409687	0.12565	-3.26	0.0046*	-0.674786	-0.144589
C*C	1.1964921	0.223876	5.34	<.0001*	0.7241552	1.6688289

Method 1 - Stress Standard Deviation Model:

Summary of Fit

RSquare	0.796593
RSquare Adj	0.712837
Root Mean Square Error	1.150784
Mean of Response	5.74608
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	88.16692	12.5953	9.5109
Error	17	22.51315	1.3243	Prob > F
C. Total	24	110.68007		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	3.4944286	0.434955	8.03	<.0001*	2.576753	4.4121041
A	-0.207	0.271242	-0.76	0.4558	-0.779271	0.3652713
B	-0.015611	0.271242	-0.06	0.9548	-0.587882	0.5566602
C	0.6587222	0.271242	2.43	0.0265*	0.086451	1.2309935
D	0.5986111	0.271242	2.21	0.0414*	0.0263398	1.1708824
B*C	0.9011875	0.287696	3.13	0.0061*	0.2942022	1.5081728
A*D	-0.823313	0.287696	-2.86	0.0108*	-1.430298	-0.216327
C*C	3.1272937	0.5126	6.10	<.0001*	2.0458026	4.2087847

Method 1 - 90th Percentile Stress Model:

Summary of Fit

RSquare	0.868507
RSquare Adj	0.814362
Root Mean Square Error	0.924466
Mean of Response	12.50812
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of		F Ratio
		Squares	Mean Square	
Model	7	95.96203	13.7089	16.0406
Error	17	14.52883	0.8546	Prob > F
C. Total	24	110.49085		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	10.829286	0.349415	30.99	<.0001*	10.092084	11.566487
A	-1.05	0.217899	-4.82	0.0002*	-1.509726	-0.590274
B	0.0655	0.217899	0.30	0.7674	-0.394226	0.525226
C	0.6468889	0.217899	2.97	0.0086*	0.1871629	1.1066149
D	1.1401667	0.217899	5.23	<.0001*	0.6804407	1.5998926
B*C	0.715125	0.231116	3.09	0.0066*	0.227512	1.202738
A*D	-0.7715	0.231116	-3.34	0.0039*	-1.259113	-0.283887
C*C	2.3317143	0.41179	5.66	<.0001*	1.4629139	3.2005147

Proximal Region Stress Desirability:

Method 2 - Mean % of Intact Stress Model:

Summary of Fit

RSquare	0.818626
RSquare Adj	0.743942
Root Mean Square Error	9.219953
Mean of Response	129.3869
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	6522.5195	931.789	10.9612
Error	17	1445.1280	85.008	Prob > F
C. Total	24	7967.6475		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	113.58719	3.484815	32.59	<.0001*	106.23487	120.93951
A	-9.488682	2.173164	-4.37	0.0004*	-14.07366	-4.903707
B	-0.757854	2.173164	-0.35	0.7316	-5.342829	3.8271203
C	2.7334636	2.173164	1.26	0.2255	-1.851511	7.3184381
D	5.9894365	2.173164	2.76	0.0135*	1.404462	10.574411
B*C	6.9829582	2.304988	3.03	0.0076*	2.1198583	11.846058
A*D	-7.517357	2.304988	-3.26	0.0046*	-12.38046	-2.654257
C*C	21.944098	4.106893	5.34	<.0001*	13.279311	30.608886

Method 2 - % of Intact Stress Standard Deviation Model:

Summary of Fit

RSquare	0.796599
RSquare Adj	0.712846
Root Mean Square Error	34.38594
Mean of Response	171.6974
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	78722.478	11246.1	9.5113
Error	17	20100.681	1182.4	Prob > F
C. Total	24	98823.159		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	104.41718	12.99666	8.03	<.0001*	76.996614	131.83774
A	-6.184021	8.104844	-0.76	0.4559	-23.28375	10.915706
B	-0.465196	8.104844	-0.06	0.9549	-17.56492	16.63453
C	19.682267	8.104844	2.43	0.0266*	2.5825407	36.781994
D	17.887652	8.104844	2.21	0.0413*	0.7879251	34.987378
B*C	26.931472	8.596485	3.13	0.0061*	8.7944731	45.068471
A*D	-24.60178	8.596485	-2.86	0.0108*	-42.73878	-6.464784
C*C	93.444789	15.31672	6.10	<.0001*	61.129343	125.76023

Method 2 - 90th Percentile % of Intact Stress Model:

Summary of Fit

RSquare	0.868494
RSquare Adj	0.814344
Root Mean Square Error	9.421224
Mean of Response	127.459
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	9965.162	1423.59	16.0388
Error	17	1508.911	88.76	Prob > F
C. Total	24	11474.073		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	110.35179	3.560888	30.99	<.0001*	102.83897	117.86461
A	-10.69962	2.220604	-4.82	0.0002*	-15.38468	-6.014554
B	0.6684413	2.220604	0.30	0.7671	-4.016623	5.3535059
C	6.5917684	2.220604	2.97	0.0086*	1.9067038	11.276833
D	11.619217	2.220604	5.23	<.0001*	6.9341525	16.304282
B*C	7.288236	2.355306	3.09	0.0066*	2.3189746	12.257497
A*D	-7.862256	2.355306	-3.34	0.0039*	-12.83152	-2.892995
C*C	23.760017	4.196547	5.66	<.0001*	14.906077	32.613957

Proximal Region % of Intact Desirability:

Approach 2 Outcomes – All Regions Mean Stress

Method 1 - Proximal Region Stress:

Summary of Fit

RSquare	0.837585
RSquare Adj	0.756377
Root Mean Square Error	0.490307
Mean of Response	7.05376
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	19.836215	2.47953	10.3141
Error	16	3.846422	0.24040	Prob > F
C. Total	24	23.682637		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.1922857	0.185319	33.41	<.0001*	5.7994274	6.585144
A	-0.517389	0.115567	-4.48	0.0004*	-0.762379	-0.272399
B	-0.041278	0.115567	-0.36	0.7256	-0.286268	0.2037124
C	0.1491111	0.115567	1.29	0.2153	-0.095879	0.3941013
D	0.3265556	0.115567	2.83	0.0122*	0.0815654	0.5715457
B*C	0.3806875	0.122577	3.11	0.0068*	0.1208362	0.6405388
A*D	-0.409687	0.122577	-3.34	0.0041*	-0.669539	-0.149836
C*D	-0.167313	0.122577	-1.36	0.1912	-0.427164	0.0925388
C*C	1.1964921	0.2184	5.48	<.0001*	0.7335041	1.65948

Method 1 - Inferior Region Stress:

Summary of Fit

RSquare	0.657996
RSquare Adj	0.486994
Root Mean Square Error	2.147082
Mean of Response	13.53512
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	141.90865	17.7386	3.8479
Error	16	73.75938	4.6100	Prob > F
C. Total	24	215.66803		0.0105*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	10.944286	0.811521	13.49	<.0001*	9.2239386	12.664633
A	-0.898778	0.506072	-1.78	0.0948	-1.971603	0.1740471
B	-0.086833	0.506072	-0.17	0.8659	-1.159658	0.9859916
C	0.4237778	0.506072	0.84	0.4147	-0.649047	1.4966027
D	0.4621111	0.506072	0.91	0.3747	-0.610714	1.534936
B*C	1.082625	0.536771	2.02	0.0608	-0.055278	2.2205276
A*D	-0.823	0.536771	-1.53	0.1447	-1.960903	0.3149026
C*D	-1.257625	0.536771	-2.34	0.0324*	-2.395528	-0.119722
C*C	3.598381	0.956386	3.76	0.0017*	1.5709325	5.6258295

Method 1 - Superior Region Stress:

Summary of Fit

RSquare	0.928664
RSquare Adj	0.892995
Root Mean Square Error	0.272804
Mean of Response	5.9978
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	15.501319	1.93766	26.0362
Error	16	1.190753	0.07442	Prob > F
C. Total	24	16.692072		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	6.1897143	0.10311	60.03	<.0001*	5.9711303	6.4082983
A	-0.559722	0.064301	-8.70	<.0001*	-0.696033	-0.423411
B	0.0751111	0.064301	1.17	0.2599	-0.0612	0.2114222
C	-0.354056	0.064301	-5.51	<.0001*	-0.490367	-0.217745
D	-0.606333	0.064301	-9.43	<.0001*	-0.742644	-0.470022
B*C	0.1805625	0.068201	2.65	0.0176*	0.0359828	0.3251422
A*D	-0.019938	0.068201	-0.29	0.7738	-0.164517	0.1246422
C*D	-0.005938	0.068201	-0.09	0.9317	-0.150517	0.1386422
C*C	-0.266548	0.121517	-2.19	0.0434*	-0.524151	-0.008944

Method 1 - Lateral Region Stress:

Summary of Fit

RSquare	0.96375
RSquare Adj	0.945625
Root Mean Square Error	0.127358
Mean of Response	5.85376
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	6.8996547	0.862457	53.1721
Error	16	0.2595218	0.016220	Prob > F
C. Total	24	7.1591766		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	5.927	0.048137	123.13	<.0001*	5.8249544	6.0290456
A	-0.602222	0.030019	-20.06	<.0001*	-0.665859	-0.538586
B	0.0628889	0.030019	2.09	0.0524	-0.000748	0.1265255
C	0.0588333	0.030019	1.96	0.0677	-0.004803	0.12247
D	0.054	0.030019	1.80	0.0909	-0.009637	0.1176366
B*C	0.0351875	0.03184	1.11	0.2854	-0.032309	0.1026843
A*D	-0.068062	0.03184	-2.14	0.0483*	-0.135559	-0.000566
C*D	-0.049688	0.03184	-1.56	0.1382	-0.117184	0.0178093
C*C	-0.101722	0.05673	-1.79	0.0919	-0.221984	0.0185397

All Regions Mean Stress Desirability:

Method 2 - Proximal Region % of Intact:

Summary of Fit

RSquare	0.837529
RSquare Adj	0.756294
Root Mean Square Error	8.994829
Mean of Response	129.3869
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	6673.1363	834.142	10.3099
Error	16	1294.5112	80.907	Prob > F
C. Total	24	7967.6475		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	113.58719	3.399726	33.41	<.0001*	106.38009	120.79429
A	-9.488682	2.120102	-4.48	0.0004*	-13.9831	-4.994267
B	-0.757854	2.120102	-0.36	0.7254	-5.252269	3.7365602
C	2.7334636	2.120102	1.29	0.2156	-1.760951	7.227878
D	5.9894365	2.120102	2.83	0.0122*	1.4950221	10.483851
B*C	6.9829582	2.248707	3.11	0.0068*	2.2159118	11.750005
A*D	-7.517357	2.248707	-3.34	0.0041*	-12.2844	-2.750311
C*D	-3.068151	2.248707	-1.36	0.1913	-7.835197	1.6988957
C*C	21.944098	4.006615	5.48	<.0001*	13.450453	30.437743

Method 2 - Inferior Region % of Intact:

Summary of Fit

RSquare	0.658007
RSquare Adj	0.48701
Root Mean Square Error	21.47684
Mean of Response	135.3918
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	14199.525	1774.94	3.8481
Error	16	7380.072	461.25	Prob > F
C. Total	24	21579.597		0.0105*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	109.47553	8.117481	13.49	<.0001*	92.267239	126.68382
A	-8.990583	5.062139	-1.78	0.0947	-19.72184	1.7406717
B	-0.868092	5.062139	-0.17	0.8660	-11.59935	9.8631632
C	4.239716	5.062139	0.84	0.4146	-6.491539	14.970971
D	4.6233002	5.062139	0.91	0.3746	-6.107955	15.354555
B*C	10.829175	5.369209	2.02	0.0608	-0.55304	22.21139
A*D	-8.232864	5.369209	-1.53	0.1447	-19.61508	3.1493505
C*D	-12.57956	5.369209	-2.34	0.0324*	-23.96177	-1.197341
C*C	35.994838	9.566543	3.76	0.0017*	15.714672	56.275004

Method 2 - Superior Region % of Intact:

Summary of Fit

RSquare	0.928681
RSquare Adj	0.893022
Root Mean Square Error	4.298623
Mean of Response	94.52402
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	3849.8264	481.228	26.0431
Error	16	295.6505	18.478	Prob > F
C. Total	24	4145.4769		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	97.549312	1.624727	60.04	<.0001*	94.105045	100.99358
A	-8.820771	1.013195	-8.71	<.0001*	-10.96865	-6.672893
B	1.1824112	1.013195	1.17	0.2603	-0.965466	3.3302888
C	-5.579482	1.013195	-5.51	<.0001*	-7.727359	-3.431604
D	-9.555477	1.013195	-9.43	<.0001*	-11.70335	-7.4076
B*C	2.8458994	1.074656	2.65	0.0175*	0.5677312	5.1240676
A*D	-0.312467	1.074656	-0.29	0.7750	-2.590635	1.9657012
C*D	-0.094539	1.074656	-0.09	0.9310	-2.372707	2.1836294
C*C	-4.201792	1.914759	-2.19	0.0433*	-8.2609	-0.142685

Method 2 - Lateral Region % of Intact:

Summary of Fit

RSquare	0.963747
RSquare Adj	0.94562
Root Mean Square Error	3.160714
Mean of Response	145.2828
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	8	4249.2133	531.152	53.1677
Error	16	159.8419	9.990	Prob > F
C. Total	24	4409.0551		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	147.09972	1.194638	123.13	<.0001*	144.5672	149.63224
A	-14.94444	0.744988	-20.06	<.0001*	-16.52374	-13.36513
B	1.5620332	0.744988	2.10	0.0523	-0.01727	3.1413363
C	1.4612548	0.744988	1.96	0.0675	-0.118048	3.0405578
D	1.3431307	0.744988	1.80	0.0903	-0.236172	2.9224338
B*C	0.8717527	0.790179	1.10	0.2862	-0.803351	2.5468566
A*D	-1.69032	0.790179	-2.14	0.0482*	-3.365424	-0.015216
C*D	-1.234099	0.790179	-1.56	0.1379	-2.909203	0.4410051
C*C	-2.523473	1.407894	-1.79	0.0920	-5.508076	0.4611289

All Regions % of Intact Desirability:

Approach 3 Outcomes – All Regions 90th Percentile

Method 1 - Proximal Region Stress:

Summary of Fit

RSquare	0.868507
RSquare Adj	0.814362
Root Mean Square Error	0.924466
Mean of Response	12.50812
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	95.96203	13.7089	16.0406
Error	17	14.52883	0.8546	Prob > F
C. Total	24	110.49085		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	10.829286	0.349415	30.99	<.0001*	10.092084	11.566487
A	-1.05	0.217899	-4.82	0.0002*	-1.509726	-0.590274
B	0.0655	0.217899	0.30	0.7674	-0.394226	0.525226
C	0.6468889	0.217899	2.97	0.0086*	0.1871629	1.1066149
D	1.1401667	0.217899	5.23	<.0001*	0.6804407	1.5998926
B*C	0.715125	0.231116	3.09	0.0066*	0.227512	1.202738
A*D	-0.7715	0.231116	-3.34	0.0039*	-1.259113	-0.283887
C*C	2.3317143	0.41179	5.66	<.0001*	1.4629139	3.2005147

Method 1 - Inferior Region Stress:

Summary of Fit

RSquare	0.670701
RSquare Adj	0.535108
Root Mean Square Error	5.980775
Mean of Response	24.90104
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	1238.5191	176.931	4.9464
Error	17	608.0844	35.770	Prob > F
C. Total	24	1846.6035		0.0034*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	16.074143	2.26052	7.11	<.0001*	11.304862	20.843424
A	-0.765944	1.409682	-0.54	0.5939	-3.740114	2.208225
B	0.2767778	1.409682	0.20	0.8467	-2.697392	3.2509472
C	2.1656111	1.409682	1.54	0.1429	-0.808558	5.1397806
D	1.0191667	1.409682	0.72	0.4795	-1.955003	3.9933361
B*C	4.0309375	1.495194	2.70	0.0153*	0.8763544	7.1855206
A*D	-2.573813	1.495194	-1.72	0.1033	-5.728396	0.5807706
C*C	12.259579	2.664049	4.60	0.0003*	6.6389274	17.880231

Method 1 - Superior Region Stress:

Summary of Fit

RSquare	0.919732
RSquare Adj	0.886681
Root Mean Square Error	0.51122
Mean of Response	10.02124
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	50.907988	7.27257	27.8274
Error	17	4.442877	0.26135	Prob > F
C. Total	24	55.350865		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	10.246857	0.193223	53.03	<.0001*	9.8391924	10.654522
A	-0.841944	0.120496	-6.99	<.0001*	-1.096168	-0.587721
B	0.1925556	0.120496	1.60	0.1285	-0.061668	0.4467792
C	-0.1385	0.120496	-1.15	0.2663	-0.392724	0.1157236
D	-1.388833	0.120496	-11.53	<.0001*	-1.643057	-1.13461
B*C	0.34125	0.127805	2.67	0.0162*	0.0716051	0.6108949
A*D	-0.06025	0.127805	-0.47	0.6433	-0.329895	0.2093949
C*C	-0.313357	0.227715	-1.38	0.1867	-0.793795	0.1670804

Method 1 - Lateral Region Stress:

Summary of Fit

RSquare	0.934863
RSquare Adj	0.908041
Root Mean Square Error	0.205765
Mean of Response	8.7804
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	10.330228	1.47575	34.8553
Error	17	0.719766	0.04234	Prob > F
C. Total	24	11.049994		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	8.9094286	0.077772	114.56	<.0001*	8.7453443	9.0735128
A	-0.7375	0.048499	-15.21	<.0001*	-0.839824	-0.635176
B	-0.088	0.048499	-1.81	0.0873	-0.190324	0.0143245
C	0.0321111	0.048499	0.66	0.5168	-0.070213	0.1344356
D	0.0633333	0.048499	1.31	0.2090	-0.038991	0.1656578
B*C	0.0464375	0.051441	0.90	0.3793	-0.062094	0.154969
A*D	-0.084187	0.051441	-1.64	0.1201	-0.192719	0.024344
C*C	-0.179206	0.091655	-1.96	0.0672	-0.372581	0.0141688

All Regions 90th Percentile Stress Desirability:

Method 2 - Proximal Region % of Intact:

Summary of Fit

RSquare	0.868494
RSquare Adj	0.814344
Root Mean Square Error	9.421224
Mean of Response	127.459
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	9965.162	1423.59	16.0388
Error	17	1508.911	88.76	Prob > F
C. Total	24	11474.073		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	110.35179	3.560888	30.99	<.0001*	102.83897	117.86461
A	-10.69962	2.220604	-4.82	0.0002*	-15.38468	-6.014554
B	0.6684413	2.220604	0.30	0.7671	-4.016623	5.3535059
C	6.5917684	2.220604	2.97	0.0086*	1.9067038	11.276833
D	11.619217	2.220604	5.23	<.0001*	6.9341525	16.304282
B*C	7.288236	2.355306	3.09	0.0066*	2.3189746	12.257497
A*D	-7.862256	2.355306	-3.34	0.0039*	-12.83152	-2.892995
C*C	23.760017	4.196547	5.66	<.0001*	14.906077	32.613957

Method 2 - Inferior Region % of Intact:

Summary of Fit

RSquare	0.670702
RSquare Adj	0.535108
Root Mean Square Error	41.56827
Mean of Response	173.0702
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	59829.117	8547.02	4.9464
Error	17	29374.660	1727.92	Prob > F
C. Total	24	89203.777		0.0034*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	111.72064	15.71133	7.11	<.0001*	78.572636	144.86865
A	-5.323608	9.797735	-0.54	0.5939	-25.99502	15.347807
B	1.9240838	9.797735	0.20	0.8466	-18.74733	22.595499
C	15.051911	9.797735	1.54	0.1429	-5.619504	35.723326
D	7.0838565	9.797735	0.72	0.4795	-13.58756	27.755271
B*C	28.016401	10.39207	2.70	0.0153*	6.0910546	49.941747
A*D	-17.88872	10.39207	-1.72	0.1033	-39.81407	4.036623
C*C	85.207733	18.51598	4.60	0.0003*	46.142431	124.27304

Method 2 - Superior Region % of Intact:

Summary of Fit

RSquare	0.919737
RSquare Adj	0.886687
Root Mean Square Error	5.030196
Mean of Response	98.60017
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	4929.0606	704.152	27.8289
Error	17	430.1488	25.303	Prob > F
C. Total	24	5359.2095		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	100.82011	1.901235	53.03	<.0001*	96.808858	104.83137
A	-8.284487	1.185629	-6.99	<.0001*	-10.78594	-5.783029
B	1.8935374	1.185629	1.60	0.1287	-0.60792	4.3949951
C	-1.362181	1.185629	-1.15	0.2665	-3.863639	1.1392768
D	-13.6663	1.185629	-11.53	<.0001*	-16.16776	-11.16485
B*C	3.3575624	1.257549	2.67	0.0162*	0.7043659	6.010759
A*D	-0.593072	1.257549	-0.47	0.6432	-3.246268	2.0601248
C*C	-3.083251	2.240627	-1.38	0.1867	-7.810561	1.64406

Method 2 - Lateral Region % of Intact:

Summary of Fit

RSquare	0.934869
RSquare Adj	0.90805
Root Mean Square Error	3.311608
Mean of Response	141.3405
Observations (or Sum Wgts)	25

Analysis of Variance

Source	DF	Sum of Squares	Mean Square	F Ratio
Model	7	2676.0113	382.287	34.8588
Error	17	186.4348	10.967	Prob > F
C. Total	24	2862.4461		<.0001*

Parameter Estimates

Term	Estimate	Std Error	t Ratio	Prob> t	Lower 95%	Upper 95%
Intercept	143.41829	1.25167	114.58	<.0001*	140.7775	146.05909
A	-11.87025	0.780554	-15.21	<.0001*	-13.51707	-10.22342
B	-1.416084	0.780554	-1.81	0.0873	-3.062908	0.2307398
C	0.5170974	0.780554	0.66	0.5165	-1.129727	2.1639216
D	1.0174805	0.780554	1.30	0.2098	-0.629344	2.6643047
B*C	0.7455946	0.827902	0.90	0.3804	-1.001126	2.4923154
A*D	-1.354433	0.827902	-1.64	0.1202	-3.101154	0.3922878
C*C	-2.885767	1.475108	-1.96	0.0671	-5.997972	0.2264377

All Regions 90th Percentile % of Intact Desirability: