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Design and Analysis of a Compliant 3D Printed Energy Harvester Package for Knee Implants

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Supervisor: Willing, Ryan T., *The University of Western Ontario* A thesis submitted in partial fulfillment of the requirements for the Master of Engineering Science degree in Mechanical and Materials Engineering © Geofrey Yamomo 2019

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Abstract

Instrumented implants provide the potential to measure the *in vivo* tibiofemoral forces that are transmitted through total knee replacements (TKR). The continuous feedback from instrumented implants can be used to objectively justify actions to reduce the risk of implant failure. The main obstacle in developing "smart implants" is reliably powering such devices. Energy harvesting mechanisms, such as the triboelectric effect, can be leveraged to produce usable electricity and measure the transmitted loads in TKRs. A compliant package that interlocks with commercially available TKR components was designed to house triboelectric generators (TEG). Prototypes were more compliant than what was expected from the computational models. During fatigue testing, the prototype failed prematurely due to inherent issues with additive manufacturing. However, these issues can be mitigated with improved post-processing techniques. This package serves as a novel approach to integrating self-powering load sensors in currently available knee implants.

Keywords

energy harvesting, knee implant, additive manufacturing, load sensor, triboelectric

Summary for Lay Audience

Osteoarthritis (OA), or the cartilage degradation in joints, can lead to pain and joint dysfunction. In severe cases of OA, the diseased joint may need to be reconstructed. Knee implants, consisting of metal components resurfacing the shin bone and thigh bone with a plastic insert in-between, replace the diseased bone, relieve pain and restore function. Over time, knee implants can fail for several reasons, such as implant loosening from the surrounding bone and abnormal motion between the thigh bone and the shinbone. Devices that measure forces transmitted through knee implants can improve our understanding of what a knee implant undergoes daily, thus providing information on how to prevent implant failure.

Currently, devices that monitor a patient's knee loads are unavailable. The main reason for this is because of the difficulty of powering these devices. Sensors that can generate power from human motion can be used to measure the loads acting on knee implants. Load sensors have been developed to generate power from static electricity. These sensors require a compliant package to cushion the forces acting on them when placed within a knee implant.

This thesis outlines the design of such a package. The package was designed using computer simulations and then its performance was measured through lab experiments. Prototypes of the package design were made with 3D printed titanium. In one lab experiment where the applied load was intentionally shifted from one side of the package to the other, the prototype predictably deformed more in the location where forces were concentrated. However, the 3D printed package was softer than what was predicted in the computer simulations. During durability testing, the package prototype underwent loading that simulates walking. Implant components should last for millions of cycles, but the current prototype failed prematurely. 3D printed titanium parts may have internal holes and defects that reduce the longevity of the parts. The fatigue strength of the package could be improved with heat treatment and removal of surface defects. The use of this package with embedded load sensors is a novel perspective on measuring the forces that act on knee implants.

ii

Co-Authorship Statement

Chapter 1: Geofrey Yamomo - wrote manuscript Ryan Willing – reviewed manuscript Chapter 2: Geofrey Yamomo - study design, data collection, interpreted results, wrote manuscript Ryan Willing – study design, interpreted results, reviewed manuscript Chapter 3: Geofrey Yamomo - study design, data collection, interpreted results, wrote manuscript Ryan Willing – study design, interpreted results, reviewed manuscript Chapter 4: Geofrey Yamomo – study design, data collection, interpreted results, wrote manuscript Ryan Willing – study design, interpreted results, reviewed manuscript Chapter 5: Geofrey Yamomo – study design, data collection, interpreted results, wrote manuscript Ryan Willing – study design, interpreted results, reviewed manuscript Chapter 6: Geofrey Yamomo - wrote manuscript

Ryan Willing - reviewed manuscript

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| Table of | Contents |
|----------|----------|
|----------|----------|

| bstract | i |
|--|----|
| Immary for Lay Audience | ii |
| o-Authorship Statementi | ii |
| cknowledgmentsi | V |
| able of Contents | v |
| st of Tables | x |
| st of Figures | ĸi |
| st of Appendices | /i |
| st of Abbreviations, Symbols and Nomenclaturexv | ii |
| napter 1 | 1 |
| Introduction | 1 |
| 1.1 Anatomical Movement Descriptors | 1 |
| 1.1.1 Anatomical Terms Used to Describe Relative Position or Direction | 1 |
| 1.1.2 Basic Movements | 2 |
| 1.1.3 Cardinal Planes of the Body | 3 |
| 1.2 Knee Anatomy | 3 |
| 1.2.1 Osseous Anatomy of the Knee | 4 |
| 1.2.2 Soft Tissue Anatomy of the Knee | 5 |
| 1.2.2.1 Knee Ligaments | 5 |
| 1.2.2.2 Muscles | 7 |
| 1.3 Knee Osteoarthritis | 8 |
| 1.3.1 Prevalence of Knee Osteoarthritis | 9 |
| 1.4 Total Knee Replacements | 0 |
| 1.4.1 Prevalence of Total Knee Replacements 1 | 1 |

| | | 1.4.2 | Complications in Total Knee Arthoplasty | 12 |
|---|-------|--|--|----------------------------------|
| | | | 1.4.2.1 Instability | 13 |
| | | | 1.4.2.2 Aseptic Loosening | 14 |
| | | 1.4.3 | Revision Total Knee Arthoplasty | 15 |
| | 1.5 | Instru | mented Implants | 17 |
| | | 1.5.1 | Intraoperative Load Sensors | 17 |
| | | 1.5.2 | Postoperative Load Sensors | 19 |
| | | 1.5.3 | Triboelectric Effect | 23 |
| | 1.6 | Thesis | Rationale | 26 |
| | 1.7 | Resear | rch Objectives | 26 |
| | 1.8 | Refere | ences | 28 |
| C | hapte | er 2 | | 38 |
| 2 | Tun | ning of | a 3D Printed Energy Harvester Package Design | 38 |
| | 2.1 | Introd | uction | 38 |
| | 2.2 | Mater | ials and Methods | 40 |
| | | 2.2.1 | Dimensions of Elastic Bodies | 40 |
| | | 2.2.2 | Analytical Model Design | 41 |
| | | 2.2.3 | Computational Model Design | 43 |
| | | 2.2.4 | Placement of Elastic Bodies in Package Prototype | 46 |
| | | 225 | Coving Thistmans A divergent | |
| | | 2.2.3 | Spring Thickness Aujustinent | 47 |
| | | 2.2.3 | Reverse Engineering Interlocking Mechanism of TKR Components | 47 49 |
| | 2.3 | 2.2.3 2.2.6 Result | Reverse Engineering Interlocking Mechanism of TKR Components | 47 49 50 |
| | 2.3 | 2.2.3 2.2.6 Result 2.3.1 | Reverse Engineering Interlocking Mechanism of TKR Components | 47 49 50 50 |
| | 2.3 | 2.2.3 2.2.6 Result 2.3.1 2.3.2 | Spring Thickness Adjustment Reverse Engineering Interlocking Mechanism of TKR Components S Analytical Model Design Computational Model Design | 47 49 50 50 52 |
| | 2.3 | 2.2.3 2.2.6 Result 2.3.1 2.3.2 | Spring Thickness Adjustment Reverse Engineering Interlocking Mechanism of TKR Components s Analytical Model Design Computational Model Design 2.3.2.1 Simplified Design Simulations | 47 49 50 50 52 52 |

| | | 2.3.3 | Simulation Results of Prototype Design | 56 |
|----|------------|---------------|---|-------------------|
| | | 2.3.4 | Dimensions of Elastic Bodies | 58 |
| | 2.4 | Discus | sion | 60 |
| | 2.5 | Conclu | usion | 64 |
| | 2.6 | Refere | nces | 65 |
| Cl | napte | er 3 | | 69 |
| 3 | Exp Des | erimen ign | tal Validation and Load Imbalance Measurements of a Compliant Packa | <u>.g</u> e 69 |
| | 3.1 | Introdu | action | 69 |
| | 3.2 | Materi | als and Methods | 70 |
| | | 3.2.1 | Manufactured Prototype | 70 |
| | | 3.2.2 | Axial Loading of Package Prototype | 72 |
| | | 3.2.3 | Imbalance Testing | 73 |
| | | | 3.2.3.1 Flexiforce Sensor Calibration | 76 |
| | | | 3.2.3.2 Experimental Set-Up | 77 |
| | 3.3 | Result | s | 79 |
| | | 3.3.1 | 11Axial Loading of Package Prototype | 79 |
| | | 3.3.2 | Imbalance Testing | 80 |
| | 3.4 | Discus | sion | 81 |
| | | 3.4.1 | Stiffness of Additively Manufactured Parts | 81 |
| | | 3.4.2 | Intercompartmental Load Measurements | 82 |
| | 3.5 | Conclu | ision | 83 |
| | 3.6 | Refere | nces | 85 |
| Cl | napte | er 4 | | 87 |
| 4 | Dur | ability | Testing of Energy Harvester Package | 87 |
| | 4.1 | Introdu | action | 87 |

| | 4.2 | Materi | als and Methods | 88 |
|---|------------|---------------------|---|-------|
| | | 4.2.1 | Experimental Set-Up | 88 |
| | | 4.2.2 | Loading Scenario at Maximum Vertical Load | 90 |
| | | 4.2.3 | Reduced Vertical Load | 91 |
| | 4.3 | Results | S | 92 |
| | | 4.3.1 | Maximum Vertical Load | 92 |
| | | 4.3.2 | Reduced Vertical Load | 94 |
| | 4.4 | Discus | sion | 95 |
| | 4.5 | Conclu | ision | 100 |
| | 4.6 | Refere | nces | 102 |
| C | hapte | er 5 | | 105 |
| 5 | Sen Loa | sitivity ding As | Analysis of Polyethylene Thickness and the Effect of Physiological ssumptions | . 105 |
| | 5.1 | Introdu | action | 105 |
| | 5.2 | Materi | als and Methods | 106 |
| | 5.3 | Results | S | 107 |
| | | 5.3.1 | Axial Force Only versus Six Component Loading | 107 |
| | | 5.3.2 | Effect of Polyethylene Insert Thickness | 110 |
| | 5.4 | Discus | sion | . 111 |
| | 5.5 | Conclu | ision | 113 |
| | 5.6 | Refere | nces | 114 |
| C | hapte | er 6 | | 115 |
| 6 | Ger | eral Di | scussion and Conclusion | 115 |
| | 6.1 | Summ | ary | 115 |
| | 6.2 | Streng | ths and Limitations | 116 |
| | 6.3 | Future | Directions | . 117 |

| 6.4 Significance | |
|------------------|--|
| Appendices | |
| Curriculum Vitae | |

List of Tables

| Table 2-1-Stress and deflection comparison between computational and analytical model. |
|--|
| The computational models presented in this table had dimensions similar to what was used in |
| the analytical model |
| Table 2-2- Stress and deflections resulting from dimensions altered using the One-Factor-at- |
| a-Time method |
| Table 2-3- Summary of stresses and resultant vertical force after altering beam midsection |
| thickness using Bezier curves |
| Table 2-4- Maximum Stress and Displacements from Maximum Loads in ISO 14243 58 |
| Table 3-1- Selective Laser Melting Parameters 71 |

List of Figures

| Figure 1-1- Primary movements of the knee (tibia with respect to the femur) |
|--|
| Figure 1-2-Cardinal Planes of the Human Body |
| Figure 1-3- Osseous Anatomy of the Knee |
| Figure 1-4- Cruciate and Collateral Ligaments7 |
| Figure 1-5- (a) Quadriceps: vastus lateralis, rectus femoris, vastus medialis and vastus |
| intermedius (<i>under rectus femoris</i>) (Anterior View of the Knee), and (b) Hamstrings: biceps |
| femoris, semimembranosus and semitendinosus (Posterior View of the Knee) |
| Figure 1-6- (A) The joint space narrowing and osteophyte formation in the anterior-posterior |
| radiograph are indicative of bilateral medial osteoarthritis where joint space narrowing is |
| greater in the right knee. (B) A magnified view of the right knee. (Braun and Gold, 2012) |
| (Image use permitted by Elsevier) 9 |
| (image use permitted by Lise (iei) |
| Figure 1-7- Total Knee Replacement Components 11 |
| Figure 1-8-Radiolucent Lines Surrounding the Tibial Stem (Kutzner et al., 2018) (Image use |
| permitted by Taylor and Francis) |
| |
| Figure 1-9- Verasense Display (Manning et al., 2018) (Image use permitted by Creative |
| Commons (https://creativecommons.org/licenses/by/4.0/)) |
| Figure 1-10- Instrumented implant developed by D'Lima et al. with four transducers located |
| Tigure 1-10- instrumented implant developed by D Enna et al. with four transducers located |
| at the four corners of the tibial tray (D'lima et al., 2005b) (Image use permitted by Elsevier) |
| |
| Figure 1, 11 (a) Tibial tray with talemetry system (b) magnified view of the microprocessor |
| I de la de |
| and the internal power induction coil, (c) patient with external coil for powering the onboard |
| electronics and telemetry system. (Kirking et al., 2006) (Image use permitted by Elsevier). 21 |
| Figure 1-12- Triboelectric Working Mechanism (Ibrahim et al., 2019a) (Image use permitted |
| by IOP Publishing) |

| Figure 1-13- Saw-tooth ridges of metal and dielectric layers (Ibrahim et al., 2019a) (Image |
|--|
| use permitted by IOP Publishing) |
| Figure 2-1-Design Process Flowchart |
| Figure 2-2- CAD model of the preliminary prototype concept design that feature elastic |
| bodies along the periphery: (a) superior view, (b) isometric view, (c) posterior view, (d) |
| section view that shows the internal void for TEG |
| Figure 2-3- A magnified view of the beam structures located on the posterior of the package |
| design. Length, height and fillet radius of the beams are denoted by L, h and r, respectively. |
| Note: The base (b) is orthogonal to h and L |
| Figure 2-4- Simplified, rectangular model used for further computational model iterations on |
| the beam design. (a) isometric view, (b) front view |
| Figure 2-5- Boundary and Loading Conditions: (a) Fixed restraint on corner on bottom |
| surface (b) Boller/elider on bottom surface. (a) Force applied on ton surface. |
| surface, (b) Koner/shder on bottom surface, (c) Porce appried on top surface |
| Figure 2-6- Bezier Curve Dimensions |
| Figure 2-7- CAD Progression of Package Prototype: (a) STL file of tibial tray, (b) base of |
| prototype, (c) extrusion of elastic bodies along periphery, (d) prototype with top plate 47 |
| Figure 2-8-A posterior view of the package-PE bearing assembly. Reference point, RP-1. |
| was placed inferior to the assembly. Reference point, RP-2, was placed superior to the |
| assembly and offset medially to replicate a 60:40 (medial:lateral) load bias |
| |
| Figure 2-9- Effects of length and base parameters on the bending stress |
| Figure 2-10- Effect of length and base on the height |
| Figure 2-11- An isometric view of the FE model for Test Run 2 (Table 2-1). Peak stress |
| locations were located at the base of the beams (denoted by red arrow) |
| Figure 2-12- Stress Plot from Maximum Axial Load |

| Figure 2-13- Vertical Displacement Plot from Maximum Axial Load |
|--|
| Figure 2-14- Shear displacements due to maximum posteriorly directed force (F_y) and internal moment (M_z) in ISO 14243 |
| Figure 2-15-Dimensions of stacked beam structure |
| Figure 2-16- Cross-sectional view of the elastic bodies. The prototype was symmetrical along the sagittal plane, therefore dimensions were denoted only on one side |
| Figure 3-1- Package Prototype: (a) top view, (b) isometric view, (c) front view, (d) bottom view, (e) back view and (f) disassembled prototype |
| Figure 3-2- The VIVO joint simulator operates in six degrees of freedom. There are three translational (medial/lateral, anterior/posterior, superior/inferior) and three rotational degrees of freedom (internal/external, abduction/adduction, flexion/extension) |
| Figure 3-3- Experimental set-up (the implant flexed about 60° for clarity) |
| Figure 3-4- Package Prototype with Embedded Flexiforce Sensors in the Medial and Lateral Compartments |
| Figure 3-5- Experimental set-up configurations: a) TKR mounted on VIVO joint simulator and b) TKR mounted on VIVO joint simulator with package prototype embedded with Flexiforce sensors |
| Figure 3-6- Force vs. displacement plot of package prototype subjected to uniaxial vertical loading |
| Figure 3-7- Flexiforce measurements during cyclic abduction/adduction |
| Figure 4-1-Experimental Set-up for Fatigue Testing on AMTI Boston: (a) Medial View, (b) Anteromedial View |
| Figure 4-2- von Mises Stress Plots of Maximum Load in Gait Cycle: a) Vertical Load of 2600 N, and b) Vertical Load of 2600 N Superimposed with Corresponding Shear Forces in |

Percent Gait Cycle (Anteriorly Directed Load of 109 N and Internal Torque of 0.9 Nm)..... 91

| Figure 4-3- Stress plot of the package at a reduced compressive load of 1400 N. The peak |
|--|
| stress was below 550 MPa. The red arrow denotes the location of the peak stress |
| Figure 4-4- Locations of cracks on the lateral side of the prototype after fatigue testing was |
| terminated. Cracks were located in the same locations on the medial side |
| Figure 4-5- Vertical position of lower actuator at the maximum applied load of 2600 N in 10 |
| 000 cycle intervals |
| Figure 4-6- Vertical position of lower actuator at the maximum applied load of 1400 N in |
| 2000 cycle intervals. A decrease in the actuator position becomes consistent after 70 000 |
| cycles |
| Figure 4-7- Comparison between the Results from Fatigue Testing and the FE Model 96 |
| Figure 4-8- Posterior view of the computational model and a prototype that illustrate the |
| distortion that was present in some of the elastic bodies: a) the area of interest is highlighted |
| in the red circle, b) magnified view of computational model, and c) magnified view of |
| prototype |
| Figure 4-9- Example of a shift in the printing layers: a) posteromedial view of computational |
| |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |
| model, b) magnified view of computational model, and c) magnified view of prototype 100 Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest. |

| Figure 5-5- Vertical Displacement Plot of Maximum Vertical Fz: a) superior, b) isometric, c) |
|--|
| anterior, d) lateral |
| |
| Figure 5-6- Comparison of the maximum von Mises stresses calculated in six component |
| loading between a 9 mm and 11 mm PE bearing 111 |

List of Appendices

| Appendix A- Analytical Model Calculations 1 | 119 |
|--|-----|
| Appendix B- Simplified Design Iterations using OFAT 1 | 123 |
| Appendix C: Axial Loading of Rectangular Models 1 | 127 |
| Appendix D: Stiffness Calculation of Single Silicone Spacer 1 | 132 |
| Appendix E: AP Translation and IE Rotation Testing 1 | 132 |
| Appendix F: Copyright Approval 1 | 135 |
| Appendix G- Mechanical Testing Machine Technical Data Sheets 1 | 141 |

List of Abbreviations, Symbols and Nomenclature

| А | area |
|-------|--------------------------------------|
| AA | abduction/adduction |
| ACL | anterior cruciate ligament |
| ADL | activities of daily living |
| AM | additive manufacturing |
| AP | anterior/posterior |
| b | base |
| BMI | body mass index |
| CAD | computer aided design |
| COP | centre of pressure |
| CP-Ti | commercially pure titanium |
| CR | cruciate retaining |
| CS | condylar stabilized |
| d | displacement |
| DAQ | data acquisition |
| DOE | design of experiments |
| DOF | degree of freedom(s) |
| E | Young's modulus |
| FEA | finite element analysis |
| h | height |
| HIP | hot isostatic pressing |
| Ι | moment of inertia |
| IE | internal/external |
| JSN | joint space narrowing |
| k | stiffness |
| KSS | American Knee Society Scoring System |
| L | length |
| LCL | lateral collateral ligament |
| Μ | bending moment |
| MCL | medial collateral ligament |
| MEMS | micro-electromechanical system |

| ML | medial/lateral |
|------------------|---|
| NEMS | nano-electromechanical system |
| OA | osteoarthritis |
| OFAT | one-factor-at-a-time |
| PCL | posterior cruciate ligament |
| PDMS | polydimethylsiloxane |
| PE | polyethylene |
| PMMA | polymethylmethacrylate |
| PROMs | patient reported outcome measures |
| PS | posterior stabilized |
| PZT | piezoelectric transducer |
| SLM | selective laser melting |
| STL | standard tessellation language |
| Та | tantalum |
| Ti6Al4V | titanium aluminum vanadium |
| TiTa | titanium tantalum |
| TEG | triboelectric generator |
| TJA | total joint arthoplasty |
| TKR | total knee replacement |
| UHMWPE | ultra high molecular weight polyethylene |
| V | shear force |
| WOMAC | Western Ontario and McMaster Universities Arthritis Index |
| Δ | deflection |
| ν | Poisson's ratio |
| σ_{vm} | von Mises stress |
| $\sigma_{\rm y}$ | yield strength |

Chapter 1

1 Introduction

In this chapter, aspects of how the bones and soft tissue of the human knee joint achieve motion will be described. Diseases such as osteoarthritis (OA) can hinder joint function by affecting the structure and stability of the joint. Severe cases of OA can necessitate surgical treatments such as total knee replacements (TKR). Despite the restoration of joint function and reduction in joint pain due to TKR, complications can arise thus rendering the implant dysfunctional. Instrumented implants can provide quantitative feedback that can be used to direct actions towards the improvement of implant longevity and patient outcomes. The thesis rationale and research objectives will be outlined after an overview of the anatomy of the knee, TKR failure modes, current smart implant technologies, and how energy harvesting mechanisms, particularly the triboelectric effect, can be leveraged in measuring the loads transmitted in a total knee replacement.

1.1 Anatomical Movement Descriptors

1.1.1 Anatomical Terms Used to Describe Relative Position or Direction

The terms that are used to describe relative anatomical positions or directions are reviewed here. The term medial refers to a position relatively close to the midline of the body or a movement that moves toward the midline. Conversely, the term lateral refers to a position relatively far from the midline or a movement away from the midline. The term proximal refers to a position that is closer relative to a reference point, whereas the term distal refers to a position that is farther from a reference point. A segment or anatomical landmark is termed superior if it is above a reference point. If a segment or anatomical landmark is inferior. The position of an object or a movement that is relative to the front or back is anterior or posterior, respectively (Hamill and Knutzen, 2014).

1.1.2 Basic Movements

Flexion occurs when the relative angle of a joint between two adjacent segments decreases. Extension is the increase of the relative angle of a joint between two adjacent segments. Abduction is when a segment's movement is away from the midline. Adduction is the movement of a segment towards the midline. Internal and external rotations rotate about a vertical axis running through the segment. Internal rotation is the rotation directed toward the midline. External rotation is the rotation directed away from the midline (Hamill and Knutzen, 2014). Flexion/extension and internal/external rotation of the knee are illustrated in Figure 1-1.



Figure 1-1- Primary movements of the knee (tibia with respect to the femur).

1.1.3 Cardinal Planes of the Body

Three cardinal planes intersect at the centre of mass of the body (Figure 1-2). The sagittal plane bisects the left and right sides of the body. The coronal or frontal plane bisects the body to create front and back halves. The transverse plane bisects the body to create top and bottom halves (Hamill and Knutzen, 2014).



Figure 1-2-Cardinal Planes of the Human Body

1.2 Knee Anatomy

The knee primarily flexes and extends. Flexion is accompanied with a small, but significant amount of rotation. The knee is a mobile joint that is stabilized passively by

the ligaments, joint geometry, the active muscles, and the compressive forces pushing the bones together (Hamill and Knutzen, 2014). Joint function and stability are directly affected by the degeneration of the articulating surfaces of the joint, bones, ligaments, and the muscles. An overview of the aforementioned components of the knee will be discussed further in this chapter.

1.2.1 Osseous Anatomy of the Knee

The human knee joint is comprised of three articulating surfaces. They include the tibiofemoral joint, the patellofemoral joint, and the tibiofibular joint. Although all three of these articulations play a role in the kinematics and kinetics of the human knee joint, the tibiofemoral articulation will be emphasized. It is the largest articulation that transmits the largest loads. Therefore, the effects of musculoskeletal diseases in this articulation can be the most detrimental.

Tibiofemoral Joint

The tibiofemoral joint, as the name suggests, is where the tibia and the femur meet. On the distal femur, there are two large convex surfaces, the medial and lateral condyles. The medial and lateral condyles are separated by the intercondylar notch in the posterior region of the distal femur, whereas the patellar or trochlear groove is the anterior separation between the condyles (Hamill and Knutzen, 2014).

The features that differentiate the lateral condyle from its medial counterpart are the larger surface area, the flatter surface and the more prominent anterior extension (Hamill and Knutzen, 2014). The epicondyles, located above the condyles, serve as attachment points for ligaments, muscles and capsule. The condyles on the distal femur rest on the tibial plateau of the proximal tibia. The tibial plateau is separated into two regions: the medial and lateral tibial plateau. The oval-shaped, concave shape of the medial plateau articulates with the convex shape of the medial condyle. The shape of the lateral condyle can be described as circular and slightly convex. The surfaces of the medial compartment have a convex-concave relationship, whereas both articulating surfaces in the lateral

compartment are convex (Hamill and Knutzen, 2014). The osseous structure of the knee is shown in Figure 1-3. The geometric differences between the compartments allow the lateral condyle to translate along the anteroposterior direction during flexion and extension; this accommodates femoral roll-back during flexion and the screw-home mechanism during extension.



Figure 1-3- Osseous Anatomy of the Knee

1.2.2 Soft Tissue Anatomy of the Knee

The soft tissues, such as capsules, muscles and ligaments, are responsible for maintaining mechanical stability in the joint.

1.2.2.1 Knee Ligaments

The cruciate and collateral ligaments of the knee serve as passive restraints along the transverse and coronal planes (Figure 1-4). The patellar ligament connects the patella and

the tibia at the tibial tuberosity. This ligament is essential for the extensor mechanism to function properly (Hamill and Knutzen, 2014).

Cruciate Ligaments

There are two cruciate ligaments that are in the intercondylar space. The anterior cruciate ligament (ACL) restrains anterior translation of the femur relative to the tibia and internal-external rotation. The posterior cruciate ligament (PCL) restrains posterior translation of the femur relative to the tibia and internal rotation. The length of the PCL remains constant in external rotation (Hamill and Knutzen, 2014).

Collateral Ligaments

The collateral ligaments, located on the sides of the knee joint, restrain varus-valgus torques and provide some restraint to internal-external rotation. The medial collateral ligament (MCL) resists against valgus moments, or forces that act in the medial direction on the lateral side of the knee. The lateral collateral ligament (LCL) supports the knee against varus moments, or forces that are directed laterally on the medial side of the joint (Hamill and Knutzen, 2014).



Figure 1-4- Cruciate and Collateral Ligaments

1.2.2.2 Muscles

The muscles surrounding the knee joint act as secondary stabilizers and produce motion (Figure 1-5). The quadriceps femoris, composed of the vastus lateralis, rectus femoris, vastus medialis and vastus intermedius, is connected to the patellar tendon. Extension is achieved from the contraction of these muscles. The biceps femoris, semimembranosus and semitendinosus make up the hamstrings. The hamstrings are responsible for knee flexion (Hamill and Knutzen, 2014).



Figure 1-5- (a) Quadriceps: vastus lateralis, rectus femoris, vastus medialis and vastus intermedius (*under rectus femoris*) (Anterior View of the Knee), and (b) Hamstrings: biceps femoris, semimembranosus and semitendinosus (Posterior View of the Knee)

1.3 Knee Osteoarthritis

Osteoarthritis (OA) is a joint disease characterized by the degeneration of the articular cartilage covering the bones. Cartilage loss leads to pain and reduced joint function. This joint disease is usually understood to result from aging and wear-and-tear. However, the onset of arthritis is multifactorial. Although mechanical loading is one of the main factors, joint integrity, genetic predisposition, local inflammation, and cellular and biochemical processes contribute to the severity of OA (Lespasio et al., 2017).

There are signs and symptoms that are indicative of OA. Pain in the affected joint can vary from being dull, sharp, constant, and intermittent. In addition to this pain, the range of motion of the joint can be limited, and movement may not be smooth as a result of an absence of articular cartilage and the presence of osteophytes, or bone spurs. This, in turn, affects joint function. Radiographic images may be used to classify the severity of a patient's OA based on the identification of osteophytes, and joint space narrowing (JSN) (Figure 1-6). The presence of JSN is a prime indicator of an absence of articular cartilage. The severity of OA can be graded using the Kellgren-Lawrence scale (KELLGREN and LAWRENCE, 1957). The grading scale ranges from 1 to 4. Kellgren and Lawrence grades from 1 to 3 can be treated with non-operative treatments such as weight loss, assistive devices, medications, and intra-articular knee injections. For Grade 4 OA, surgical options need to be considered. OA at this state has large osteophytes, marked JSN, severe sclerosis, and definite bone deformity (Lespasio et al., 2017). Joint reconstruction can restore joint function, alleviate pain, and improve quality of life.



Figure 1-6- (A) The joint space narrowing and osteophyte formation in the anteriorposterior radiograph are indicative of bilateral medial osteoarthritis where joint space narrowing is greater in the right knee. (B) A magnified view of the right knee. (Braun and Gold, 2012) (Image use permitted by Elsevier)

1.3.1 Prevalence of Knee Osteoarthritis

The knee is the most susceptible joint to OA (Bliddal and Christensen, 2009). Knee OA is more prevalent in the older population and in individuals with a higher body mass index (BMI). Increased joint loading due to obesity is not the only factor that contributes to the onset of OA in weight-bearing joints. Changes in body composition, negative

effects related to inflammation, the decrease of physical activity, and resulting loss of muscle strength from a sedentary lifestyle increase the risk of OA (Wluka et al., 2013). Obesity rates are expected to increase thereby increasing the incidence of knee OA and the need for knee arthroplasty (Bryan et al., 2013).

1.4 Total Knee Replacements

For Grade 4 OA, based on the Kellgren-Lawrence scale, total knee replacements (TKR) remains the most viable option for restoring joint function and pain relief. Implant designs can vary based on the required constraint. However, a TKR generally consists of a cobalt-chromium femoral component, a titanium tibial tray, and an ultra-high molecular weight polyethylene (UHMWPE) bearing (Figure 1-7). Most TKRs function with the removal of the ACL, but on rare occasions, it is not removed when a bi-cruciate retaining implant is used. The PCL is removed about half of the time. Therefore, there are different implant designs that compensate for the contributions of the removed ligaments. TKR does not solely remove the diseased tissue, but it also realigns the mechanical axes of the femur and the tibia in a way that replicates the load distribution in a healthy joint. The alignment and fixation of the TKR is imperative for implant longevity and ensuring patient satisfaction (Sikorski, 2008).

Material strength and durability of implants should be high enough to prevent yielding and fatigue failure from the stresses transmitted through the knee during activities of daily living (ADL). Knee implants, generally, should be durable enough to last about 15-20 years (Rönn et al., 2011).



Figure 1-7- Total Knee Replacement Components

1.4.1 Prevalence of Total Knee Replacements

TKR cases are expected to increase in the next decade. In Canada, based on the Canadian Joint Registry, there has been a 17.0% increase in the TKR procedure volume from 2012 to 2017 (Canadian Institute for Health Information, 2018). In the United States, procedural volume for all total joint arthroplasties (TJA) rose by 38% between 2016 and 2017 (American Joint Replacement Registry, 2018). In both the United States and Canada, TKRs represents about 60% of all TJA procedures. Kurtz et al. projected a 673% increase in TKRs by 2030 when compared to the annual number of procedures in 2005. Based on their projection methodology, in 2030, 3.48 million procedures can be expected in the United States (Kurtz et al., 2007). In more recent studies, the projected increase in the number of TKRs is not as high. Sloan et al. projected that TKR procedures

in 2030 will be about half of what was envisioned by Kurtz et al.; 1.26 million procedures are expected based on a linear estimate (Sloan et al., 2018). Nevertheless, TKR procedures are expected to be in high demand in the future.

There are several factors that are contributing to the increase of TKR procedures. With medical and technological improvements, life expectancies have increased in developed countries. In countries such as the UK, Japan, Canada, and Australia, life expectancies are 80 and 83 years for men and women, respectively (Hamilton et al., 2015). Another trend that may be connected to the increase in TKR procedures is the increasing incidence in active, younger adults (Weinstein et al., 2013). Another risk factor that is contributing to the projected increase in TKR is the higher prevalence of obesity. Obesity has been associated with earlier cases of moderate to severe OA (Coggon et al., 2001; Doherty, 2001; Gillespie and Porteous, 2007; Harms et al., 2007; Vasarhelyi and MacDonald, 2012). In Canada and the United States, from the late 1980s to 2009, there was approximately a 10% increase in the prevalence of obesity (Shields et al., 2011; Vasarhelyi and MacDonald, 2012). Factors such as longer life expectancies, obesity and TKR performed earlier in life can result in not only in an increase in primary TKR procedures, but can warrant the need for a revision, and possibly a re-revision surgery, as patients outlive their knee prostheses.

1.4.2 Complications in Total Knee Arthoplasty

Despite the success of the majority of primary TKRs, some patients may experience pain, stiffness, and instability. In some cases, based on clinical diagnoses, revision surgery may be considered. Infection, instability, and loosening have been the most common reasons for revision surgery (Gaizo et al., 2011; Vince et al., 2006; Wilson et al., 2017). Joint instability and aseptic loosening are both mechanical in nature and are both sensitive to implant design and surgical technique.

Instability can result from muscle weakness, improper implant sizing, and ligament imbalance which can alter the loading of components (Abdel and Haas, 2014; Petrie and

Haidukewych, 2016; Vince et al., 2006). Changes to the loading pattern of implant components can lead to excessive stress and motion which can lead to component damage. Osteolysis can affect the integrity of the surrounding bone structures which can lead to implant loosening. Misalignment of TKR components resulting from surgical error can contribute to physiologically incorrect loading thus introducing high stresses that can lead to the mechanical failure of implant components. Abnormal loading patterns and biological changes can affect a TKR's articulation which may result in premature prosthesis failure (Gaizo et al., 2011; Vince et al., 2006; Wilson et al., 2017).

1.4.2.1 Instability

Instability consistently remains one of the most common modes of implant failure (Dalury et al., 2013; Sharkey et al., 2014; Vince et al., 2006; Wilson et al., 2017). Instability accounts for 10 to 22% of revisions (Abdel and Haas, 2014; Callaghan et al., 2004; Parratte and Pagnano, 2008; Vince, 2003; Yercan et al., 2005a, 2005b). A reconstructed knee is unstable when the implant and surrounding tissue do not provide sufficient restraint against the secondary joint motion when compared to a healthy, intact joint. The lack of restraint can lead to excessive relative motion between the articulating surfaces of the knee. Obvious visual signs of dislocation are indicative of instability. Dislocations can take the form of varus, valgus, or recurvatum deformities. Symptoms of instability include pain, recurrent knee effusion, restricted motion, giving-way, a sense of rubbing between the components, or "locking" of the knee (Yercan et al., 2005b).

Component loosening, prosthetic breakage, component size or position, fracture, polyethylene bearing wear, and collateral ligament failure can lead to instability (Vince et al., 2006). There are three types of instability proposed by Parratte and Pagnano: extension instability, flexion instability, and genu recurvatum (Chang et al., 2014; Parratte and Pagnano, 2008). When soft tissue balancing is performed, the objective is to balance the tibial forces along the mediolateral and anteroposterior axes while not over constraining the joint.

1.4.2.2 Aseptic Loosening

Successful prostheses have a solid fixation, and thus no relative motion between the implant and bone they are implanted in. The infection of surrounding tissues in a TKR can cause components to loosen by weakening the periprosthetic bond at the implantbone interface. However, aseptic loosening can arise when the implant-bone fixation is weakened, not from necrotic tissue, but what is commonly believed to result from wearinduced osteolysis. In TKRs, particulate debris usually originates from the UHMWPE bearing. The presence of wear debris elicits a foreign-body reaction. Wear debris triggers the creation of cytokines. The higher concentration of cytokines promotes osteoclast differentiation and activity thus perpetuating bone resorption. Additionally, wear debris inhibits bone formation by affecting osteoblast progenitor cells (Jiang et al., 2013; Saleh et al., 2004). This leads to periprosthetic bone loss or osteolysis. However, in recent years, the technological advancements in the design and manufacturing of highly crosslinked UHMWPE have reduced the degradation and wear of UHMWPE bearings, but fatigue remains a concern (Collier et al., 2005; Medel et al., 2009). Signs, or the presence, of aseptic loosening, can be identified as radiolucent zones at the implant-bone interface (Rosenthall, 1997) (Figure 1-8).



Figure 1-8-Radiolucent Lines Surrounding the Tibial Stem (Kutzner et al., 2018) (Image use permitted by Taylor and Francis)

Aseptic loosening is not always associated with osteolysis. Micromotion can lead to insufficient primary fixation or impede osseointegration. Large micromotion can lead to fibrous tissue formation in the implant-bone interface. Increased relative motion between the implant and bone can enlarge the effective joint space and increase the amount of cement and metal wear particles. Stress shielding, high fluid pressure due to the inflammatory response, materials of articulating surfaces, and individual variations can also contribute to aseptic loosening (Sundfeldt et al., 2006). Continuous measurement of the tibial forces can detect the effect of load imbalances have on bone remodeling and the progression of failure modes such as instability and aseptic loosening.

1.4.3 Revision Total Knee Arthoplasty

The revision burden for TKR, in 2017, was 6.9% and 5% in Canada and the United States, respectively (American Joint Replacement Registry, 2018; Canadian Institute for Health Information, 2018). Revision burden is the proportion of revision surgeries performed out of all TKR. The revision burden in both countries has remained constant from 2012 to 2017. With the increasing prevalence of TKR, an increase in the procedure volume of revision TKR will be expected despite the consistent revision burden in recent years.

Reiterating the impact associated with obesity on OA, obesity can lead to complications that can affect the outcome of a primary TKR procedure. Obese patients, younger than 60 years of age, are more likely to undergo a revision TKR procedure. Heavier, active patients tend to have decreased implant survivorship (Foran et al., 2004; Vazquez-Vela Johnson et al., 2003). Overall, the increased joint loading in obese patients can decrease the success of a TKR procedure thus warranting revision TKRs.

Based on the prevalence of obesity, the projected increase in TKR procedures, an aging population, and younger individuals requiring a TKR, revision TKR will become more prevalent. Revision TKR is more complex than a primary TKR. It is a more costly and complex procedure (Kurtz et al., 2005). Revision TKR presents challenges that can adversely affect the outcome of the surgery. Detailed planning based on the implant failure mode determined from clinical history, examination, analysis of radiographs and laboratory tests must be undertaken (Hamilton et al., 2015). Removal of the primary TKR components can be challenging if there isn't adequate exposure. Long stems, augments, and osseous-integrative materials are commonly used to compensate for tissue loss when the primary TKR is removed (Dennis et al., 2018; Mason and Fehring, 2006). Due to the challenges associated with revision TKR, revision TKR do not replicate the same implant survival rates and patient outcomes as the primary TKR.

Based on the undesirable aspects of revision surgeries, the ability to acquire quantitative feedback of the forces transmitted through a load-sensing implant can possibly mitigate implant failures that are mechanical in nature such as instability and aseptic loosening. By understanding the *in vivo* tibial forces, suitable interventions that restore the load distribution within the knee to an established, objective target, can be implemented to improve implant survivorship, thus reducing the procedure volume of revision surgeries.

1.5 Instrumented Implants

In order to address the complications that occur with TKR, instrumented implants can be utilized to better understand the etiology of implant failure. Implants equipped with load sensors can provide quantitative data that can be used to corroborate a patient's history when issues arise, or more importantly, predict precursors of implant failure. Instrumented implants have a potential of supplementing current diagnostic tools such as radiographic images and laboratory tests.

1.5.1 Intraoperative Load Sensors

Soft tissue imbalances in TKR can be detrimental to implant survivorship and patient outcomes (Babazadeh et al., 2009; Parratte and Pagnano, 2008; Sharkey et al., 2014; Unitt et al., 2008; Whiteside, 2002). Inadequate soft tissue balance can be the precursor to instability, premature implant wear, and aseptic loosening. Required bone resections, ligament releases, and rotations are judged based on the subjective tactile assessment of the surgeon (Gustke et al., 2014a). Despite the importance of a correctly aligned and well-balanced TKR, a quantitative standard that defines a well-balanced knee has not been established (Gustke et al., 2014b).

The Verasense Knee System (Orthosensor, Dania FL) was developed for the purpose of measuring the forces transmitted in the medial and lateral compartments of the joint during the preliminary sizing and positioning of components during TKR surgery. This instrumented tibial tray insert trial provides real-time force feedback and the locations of peak tibiofemoral forces throughout the knee's full range of motion. With the quantitative feedback of the mediolateral force distribution, soft tissue balancing can be evaluated and corrected accordingly based on an objective basis.


Figure 1-9- Verasense Display (Manning et al., 2018) (Image use permitted by Creative Commons (https://creativecommons.org/licenses/by/4.0/))

Measuring the intercompartmental loads using the Verasense Knee System, for acquiring a desired mediolateral load distribution, has shown favourable short-term outcomes (Cho et al., 2018; Elmallah et al., 2016; Gustke et al., 2014b; Gustke et al., 2014; Meneghini et al., 2016a; Risitano et al., 2017a). The Western Ontario and McMaster Universities Arthritis Index (WOMAC) (Bellamy et al., 1988) and the American Knee Society Scoring System (KSS) (Noble et al., 2012; Scuderi et al., 2012) have been commonly used to measure patient reported outcome measures (PROMs) before and after TKR. Both scoring systems evaluate pain, stiffness, and physical function. A higher WOMAC score indicates worse pain, stiffness, and functional limitations, whereas a higher KSS score reflects improvement. Gustke et al. observed that KSS and WOMAC scores indicated better improvement in the quantifiably balanced (intercompartmental load difference of < 15 lbs) group versus the quantifiably unbalanced group six months post-operatively. The improvements in the PROMs of the balanced group are reflected by the increased activity levels. This may be the result of better performance in post-operative physiotherapy from the more favourable biomechanics stemming from the intraoperative

compartmental measurements (Gustke et al., 2014b). However, the course of healing can change over time, and because of the lack of *in vivo* load measurements, the objective targets used in soft tissue balancing with the aid of an intraoperative sensor like the Verasense cannot be validated.

Intraoperative sensors, in conjunction with an established objective load target, may be beneficial in preventing instability, loosening, and implant wear in the short term. However, the load balance, even when determined objectively with an intraoperative sensor, may change over the course of healing. Successfully achieving the optimal intercompartmental load distribution intraoperatively does not guarantee immunity from complications. Embedded sensors in TKR components can monitor patients' activity levels and improve our understanding of *in vivo* knee mechanics. With instrumented implants, quantitative evidence of patient outcomes can possibly support how intraoperative sensors are used for TKR.

1.5.2 Postoperative Load Sensors

Load sensors have been used to measure forces and moments in the shoulder, hip, knee, and spine. Due to the scope of this project, relevant work related to the instrumentation of knee implants measuring tibial forces will be outlined.

The first instrumented knee implant that was used to measure forces *in vivo* was developed by D'Lima et al (Figure 1-10). Four transducers were placed at the four corners of the tibial tray in order to measure the total axial force and determining the center of pressure. The microtransmitter, power induction coil, and antenna were housed within the stem of the tibial component. During use, inductive coupling was used to power the transducers and telemetry system wirelessly. The participant was instructed to perform a set of activities of daily living including gait, stair ascent, stair descent, sitstand, stand-sit, and cycling, during which knee loads were recorded (D'lima et al., 2005a).



Figure 1-10- Instrumented implant developed by D'Lima et al. with four transducers located at the four corners of the tibial tray (D'lima et al., 2005b) (Image use permitted by Elsevier)

Kirking et al. developed an instrumented implant capable of measuring all six load components (Kirking et al., 2006). The telemetry system, and mechanism for powering the microtransmitter and load cells used an identical set-up as D'Lima et al (D'lima et al., 2005a). The instrumented design was accurate with a highly linear response ($\mathbb{R}^2 > 0.997$) when comparing the measured loads of the instrumented implant and the applied external loads. Power was consistently supplied at approximately 40 mW via induction coupling. The net signal integrity, of the wireless transmission of all components of loading, was greater than 98%. Overall, this instrumented implant design was able to wirelessly transmit accurate load data based on the applied uniaxial loading along each axis, and during a sinusoidal dynamic loading scenario.



Figure 1-11- (a) Tibial tray with telemetry system, (b) magnified view of the microprocessor and the internal power induction coil, (c) patient with external coil for powering the onboard electronics and telemetry system. (Kirking et al., 2006) (Image use permitted by Elsevier)

Another design by Bergmann et al., similar to what was developed by Kirking et al., was tested *in vivo*. Strain gauges were embedded in a tibial component that transmitted load data using an inductively powered telemetry system. The design of their instrumented tibial component was able to measure the six load components.

Although powering strain gauges and telemetry systems using induction coupling has proved to be an effective means of measuring and transmitting tibiofemoral contact forces accurately and reliably, this method of power transmission is inherently cumbersome. An external coil must be donned by the individual. Therefore, powering the electronics for smart implants remains a limiting factor. With the advancements in microelectromechanical and nano-electromechanical systems (MEMS and NEMS, respectively), more compact methods of measurement, and power generation can be developed. Energy harvesting mechanisms can be used for load acquisition and power generation.

In recent years, piezoelectric transduction has been a mechanism that has been explored. Safaei et al. conceptualized the idea of embedding piezoelectric transducers (PZT) in the polyethylene insert. By locating transducers in the anteromedial, anterolateral, posteromedial, and posterolateral regions of the polyethylene bearing, the center of pressure can be determined on the tibiofemoral joint. Safaei et al. performed preliminary testing with a simplified model of this concept by embedding a single PZT in a polyethylene disk (Safaei et al., 2017). The vertical component of the gait cycle was applied to this simplified prototype. Biomechanical modeling, finite element analysis and electromechanical modeling were conducted in the design phase preceding the experiments (Safaei et al., 2018).

Platt et al. were the first to place piezoelectric ceramics in an altered tibial tray. PZTs were placed in the anterior, posteromedial, and posterolateral locations of the tibial plateau. The prototype was able to generate 850 μ W of continuous regulated power. The PZT elements were deemed to have the longevity required in TKR applications. A single PZT element was axially compressed with an ISO knee load profile for up to 20 million cycles. A 17% decrease in power output was measured when compared to the initial power output. Therefore, the use of PZT elements integrated into TKR components in the hopes of powering low power microprocessors and sensors has shown some promise (Platt et al., 2005).

Almouahed et al. developed a concept that echoes aspects of Safei et al. and Platt et al's work. The prototype had four PZT placed in the four corners of a custom-made tibial component, instead of the PE bearing. This design also measured the COP of the applied axial load. In previous iterations of this prototype, there were some drawbacks present. The top layer of the PZTs was torn off due to shear forces. Power generation was not optimized due to the geometry of the PZTs. The power generated was proportional to the

height, and inversely proportional to the surface area (Almouahed et al., 2017). One limitation in Platt et al.'s design was the overall thickness of the tibial baseplate that measured at least 22 mm in height. Almouahed et al reduced the thickness of the baseplate with the PZTs to 6.65 mm in their design. Removing a large amount of bone during TKR would jeopardize the success of possible revision TKRs (Almouahed et al., 2011).

Electromagnetic induction was another method used for energy harvesting purposes. Luciano et al.'s design consisted of an altered rotating hinge TKR. A coil of wire was spooled around the hinge of the TKR. In order to leverage Faraday's law of induction, magnets were embedded in a femoral component. During the swing phase, the femoral component would translate along a curvilinear path during flexion and extension. This design could generate 1.7 mW of power every 7.6 s (Luciano et al., 2014). The main limitations are the feasibility of making the necessary changes to the TKR components to recreate this design and the use of an overly constrained TKR system that is generally used for revision surgeries.

1.5.3 Triboelectric Effect

Another mechanism that can be utilized in energy harvesting is the triboelectric effect. The triboelectric effect is electrification induced by contact and friction between two interfacing materials (Wang, 2013). This phenomenon can be used for energy harvesting and load sensing applications (F.-R. R. Fan et al., 2012; Lin et al., 2013; Zhang et al., 2014). For the purposes of measuring loads in a TKR, a design that utilizes the contactseparation between a metal and a dielectric is the most suitable for harvesting energy while measuring *in vivo* loads during activities of daily living.

The contact-separation mode operates based on the cyclic compression applied to the triboelectric generator (TEG). Stacked structures are generally used for this triboelectrification modality. Materials selected for the stacked structure is an important consideration. A larger disparity in the polarities between the contacting materials

increases the charge transferred in the TEG (Wang, 2013). In other words, contacting materials with opposite polarities on the triboelectric series will generate more power. The top and bottom layers of the stacked structure act as electrodes. The material of choice for the electrodes tends towards a positive polarity like aluminum, for instance. A material with a negative polarity, like polydimethylsiloxane (PDMS), is secured to the top surface of the bottom electrode (F.-R. R. Fan et al., 2012; Wang, 2013, 2014). Therefore, the top layer comes into contact with the dielectric when the TEG is loaded, and separation between the layers occurs when the TEG is unloaded.

The contact and separation of the metal and dielectric layers allow the electron transfer between the electrodes. As the metal layer comes into contact with the dielectric, electrons from the metal layer are transferred to the dielectric layer. At full compression of the layers, the metal layer is positively charged whereas the bottom layer becomes negatively charged. As the metal layer separates from the dielectric, electrons transfer back to the positively charged metal layer. At full separation, the TEG is restored to a neutral state as the triboelectric and electrostatic charges equalize. The cyclic change of electron flow direction creates an alternating current (Figure 1-12) (Ibrahim et al., 2019a; Wang, 2013).



Figure 1-12- Triboelectric Working Mechanism (Ibrahim et al., 2019a) (Image use permitted by IOP Publishing)

In addition to using a metal-to-dielectric surface interface with materials on the opposite end of the triboelectric scale, the volumetric charge density can also be enhanced by changing the geometry of the contacting surfaces. Micro- and nano-patterns on the contacting surfaces can increase the surface area thus improving triboelectrification (F.-R. Fan et al., 2012; Wang, 2013).

Ibrahim et al. developed a TEG with the intention of instrumenting a TKR (Ibrahim et al., 2019a, 2019b, 2018). Material selection and patterns of the contacting surfaces were considered to maximize the electric charge between a metal-to-dielectric interface in a vertical contact-separation mode-based TEG. Titanium and PDMS were used in the TEG due to the opposite polarities of the said materials on the triboelectric series. As a means of increasing the contact surface area, matching saw-tooth ridges were fabricated on the contacting surfaces of the titanium and PDMS layers (Figure 1-13) (Ibrahim et al., 2019a). However, these TEGs require a relatively large compression distance (0.2 mm) to be able to operate, therefore a compliant package is needed to integrate the TEGs with a TKR.



Figure 1-13- Saw-tooth ridges of metal and dielectric layers (Ibrahim et al., 2019a) (Image use permitted by IOP Publishing)

1.6 Thesis Rationale

Currently, there are no commercially available means of continuously monitoring load transfer through TKR knees. Energy harvesting mechanisms can be of utility in powering instrumented knee implants. The triboelectric effect is the energy harvesting mechanism investigated by our group for the purpose of measuring tibiofemoral contact forces. Measuring these forces in uninhibited ADL can provide valuable quantitative feedback on the condition of a patient's knee years after a TKR procedure. There are several benefits of acquiring the *in vivo* loads postoperatively. The load data can be used to refine surgical techniques, such as soft-tissue balancing and implant alignment, which is critical to a well-functioning TKR. By monitoring patient activity levels, patients, therapists, and surgeons can be alerted of problems that can possibly be addressed early to avoid a revision surgery. With a better understanding of the loads and demands placed on implants, future implant designs can be improved. Therefore, there is a need for self-powering systems capable of quantifying tibiofemoral forces after TKR procedures.

Theoretically, TEGs in an instrumented knee implant has shown promise. However, there is still a need for further development before these devices can be integrated into commercially available TKR systems. The shear forces in ADL can damage the contacting surfaces of the TEGs. The TEGs are capable of generating electricity only when cyclic contact and full separation between the contacting layers occurs. A compliant durable package, designed to interlock between the UHMWPE bearing and tibial tray, can ensure the contact and separation of the TEG layers during the cyclic loading in ADL. In this thesis, the design and the analysis of the mechanical behaviour of a 3D printed titanium package prototype is outlined.

1.7 Research Objectives

The objectives of this study were:

1. Design a compliant package that houses the TEGs, and interlocks with a commercially available polyethylene insert and tibial tray.

2. Characterize and validate the stiffness of the compliant package when subjected to gait loading. The package's resistance to shear forces was measured.

3. Assess possible damage in the energy harvester package resulting from durability testing under simulated gait.

4. Develop computational models of the package that quantify its sensitivity to different size PE bearing thicknesses. Differences in the deflection and peak stresses were also measured between loading scenarios that account for all six components of loading and uniaxial loading.

1.8 References

- Abdel, M.P., Haas, S.B., 2014. The unstable knee: Wobble and Buckle. Bone Jt. J. 96B, 112–114. https://doi.org/10.1302/0301-620X.96B11.34325
- Almouahed, S., Gouriou, M., Hamitouche, C., Stindel, E., Roux, C., 2011. The use of piezoceramics as electrical energy harvesters within instrumented knee implant during walking. IEEE/ASME Trans. Mechatronics 16, 799–807. https://doi.org/10.1109/TMECH.2011.2159512
- Almouahed, S., Hamitouche, C., Stindel, E., 2017. Optimized Prototype of Instrumented Knee Implant: Experimental Validation. IRBM 38, 250–255. https://doi.org/10.1016/J.IRBM.2017.06.005
- American Joint Replacement Registry, 2018. Fifth AJRR Annual Report on Hip and Knee Arthroplasty Data 2018, American Joint Replacement Registry. https://doi.org/1.1
- Babazadeh, S., Stoney, J.D., Lim, K., Choong, P.F.M., 2009. The relevance of ligament balancing in total knee arthroplasty: how important is it? A systematic review of the literature. Orthop. Rev. (Pavia). 1, e26. https://doi.org/10.4081/or.2009.e26
- Bellamy, N., Buchanan, W.W., Goldsmith, C.H., Campbell, J., Stitt, L.W., 1988. Validation study of WOMAC: a health status instrument for measuring clinically important patient relevant outcomes to antirheumatic drug therapy in patients with osteoarthritis of the hip or knee. J. Rheumatol. 15, 1833–40.
- Bliddal, H., Christensen, R., 2009. The treatment and prevention of knee osteoarthritis: a tool for clinical decision-making. Expert Opin. Pharmacother. 10, 1793–1804. https://doi.org/10.1517/14656560903018911
- Braun, H.J., Gold, G.E., 2012. Diagnosis of osteoarthritis: Imaging. Bone 51, 278–288. https://doi.org/10.1016/j.bone.2011.11.019

- Bryan, D., Parvizi, J., Austin, M., Backe, H., Valle, C. Della, Kolessar, D.J., Kreuzer, S., Malinzak, R., Masri, B., McGrory, B.J., Mochel, D., Yates, A., 2013. Obesity and total joint arthroplasty. A literature based review. J. Arthroplasty. https://doi.org/10.1016/j.arth.2013.02.011
- Callaghan, J.J., O'Rourke, M.R., Saleh, K.J., 2004. Why knees fail. J. Arthroplasty 19, 31–34. https://doi.org/10.1016/j.arth.2004.02.015
- Canadian Institute for Health Information, 2018. Hip and Knee Replacements in Canada, 2017 2018: Canadian Joint Replacement Registry Annual Report. Ottawa, ON.
- Chang, M.J., Lim, H., Lee, N.R., Moon, Y.-W., 2014. Diagnosis, Causes and Treatments of Instability Following Total Knee Arthroplasty. Knee Surg. Relat. Res. 26, 61–67. https://doi.org/10.5792/ksrr.2014.26.2.61
- Cho, K.-J., Seon, J.-K., Jang, W.-Y., Park, C.-G., Song, E.-K., 2018. Objective quantification of ligament balancing using VERASENSE in measured resection and modified gap balance total knee arthroplasty. BMC Musculoskelet. Disord. 19, 266. https://doi.org/10.1186/S12891-018-2190-8
- Coggon, D., Reading, I., Croft, P., McLaren, M., Barrett, D., Cooper, C., 2001. Knee osteoarthritis and obesity. Int. J. Obes. 25, 622–627. https://doi.org/10.1038/sj.ijo.0801585
- Collier, M., EnghJr, C., Mcauley, J., 2005. Osteolysis After Total Knee Arthroplasty: Influence of Tibial Baseplate Surface Finish and Sterilization of Polyethylene InsertFindings at Five to Ten Years. J. Bone ... 2702–2708.
- D'lima, D.D., Patil, S., Steklov, N., Slamin, J.E., Colwell, C.W., 2005a. THE CHITRANJAN RANAWAT AWARD In Vivo Knee Forces after Total Knee Arthroplasty. https://doi.org/10.1097/01.blo.0000186559.62942.8c

D'lima, D.D., Townsend, C.P., Arms, S.W., Morris, B.A., Colwell, C.W., 2005b. An

implantable telemetry device to measure intra-articular tibial forces. J. Biomech. 38, 299–304. https://doi.org/10.1016/j.jbiomech.2004.02.011

- Dalury, D.F., Pomeroy, D.L., Gorab, R.S., Adams, M.J., 2013. Why are Total Knee Arthroplasties Being Revised? J. Arthroplasty 28, 120–121. https://doi.org/10.1016/J.ARTH.2013.04.051
- Dennis, D.A., Berry, D.J., Engh, G., Fehring, T., MacDonald, S.J., Rosenberg, A.G., Scuderi, G., 2018. Revision total knee arthroplasty. Orthop. Knowl. Updat. Hip Knee Reconstr. 5 267–278.
- Doherty, M., 2001. Risk factors for progression of knee osteoarthritis. Lancet 358, 775– 776. https://doi.org/10.1016/S0140-6736(01)06006-8
- Elmallah, R.K., Mistry, J.B., Cherian, J.J., Chughtai, M., Bhave, A., Roche, M.W., Mont, M.A., 2016. Can We Really "Feel" a Balanced Total Knee Arthroplasty? J. Arthroplasty 31, 102–105. https://doi.org/10.1016/J.ARTH.2016.03.054
- Fan, F.-R., Lin, L., Zhu, G., Wu, W., Zhang, R., Wang, Z.L., 2012. Transparent Triboelectric Nanogenerators and Self-Powered Pressure Sensors Based on Micropatterned Plastic Films. Nano Lett 12, 3114. https://doi.org/10.1021/nl300988z
- Fan, F.-R.R., Lin, L., Zhu, G., Wu, W., Zhang, R., Wang, Z.L., 2012. Transparent triboelectric nanogenerators and self-powered pressure sensors based on micropatterned plastic films. Nano Lett. 12, 3109–3114. https://doi.org/10.1021/nl300988z
- Foran, J.R.H., Mont, M.A., Etienne, G., Jones, L.C., Hungerford, D.S., 2004. The outcome of total knee arthroplasty in obese patients. J. Bone Joint Surg. Am. 86, 1609–15. https://doi.org/10.2106/00004623-200408000-00002

Gaizo, D.J. Del, Craig, ;, Della Valle, J., Della Valle, C.J., 2011. Instability in Primary

Total Knee Arthroplasty 34. https://doi.org/10.3928/01477447-20110714-46

- Gillespie, G.N., Porteous, A.J., 2007. Obesity and knee arthroplasty. Knee 14, 81–86. https://doi.org/10.1016/J.KNEE.2006.11.004
- Gustke, Kenneth A, Golladay, G.J., Roche, M.W., Elson, L.C., Anderson, C.R., 2014. Primary TKA patients with quantifiably balanced soft-tissue achieve significant clinical gains sooner than unbalanced patients. Adv. Orthop. 2014, 628695. https://doi.org/10.1155/2014/628695
- Gustke, Kenneth A., Golladay, G.J., Roche, M.W., Elson, L.C., Anderson, C.R., 2014. A new method for defining balance: Promising Short-Term Clinical Outcomes of Sensor-Guided TKA. J. Arthroplasty 29, 955–960. https://doi.org/10.1016/j.arth.2013.10.020
- Hamill, J., Knutzen, K.M., 2014. Biomechanical Basis of Human Movement, Fourth. ed. Lippincott Williams & Wilkins, Baltimore, MD, USA.
- Hamilton, D.F., Howie, C.R., Burnett, R., Simpson, A.H.R.W., Patton, J.T., Hamilton, □ D F, 2015. Dealing with the predicted increase in demand for revision total knee arthroplasty CHALLENGES, RISKS AND OPPORTUNITIES. https://doi.org/10.1302/0301-620X.97B6
- Harms, S., Larson, R., Sahmoun, A.E., Beal, J.R., 2007. Obesity increases the likelihood of total joint replacement surgery among younger adults. Int. Orthop. 31, 23–26. https://doi.org/10.1007/s00264-006-0130-y
- Ibrahim, A., Jain, M., Salman, E., Willing, R., Towfighian, S., 2019a. A smart knee implant using triboelectric energy harvesters. Smart Mater. Struct. 28, 025040. https://doi.org/10.1088/1361-665X/aaf3f1
- Ibrahim, A., Ramini, A., Towfighian, S., 2018. Experimental and theoretical investigation of an impact vibration harvester with triboelectric transduction. J. Sound Vib. 416,

111-124. https://doi.org/10.1016/J.JSV.2017.11.036

- Ibrahim, A., Yamomo, G., Willing, R., Towfighian, S., 2019b. Analysis of a triboelectric energy harvester for total knee replacements under gait loading 47. https://doi.org/10.1117/12.2515351
- Jiang, Y., Jia, T., Wooley, P.H., Yang, S.Y., 2013. Current research in the pathogenesis of aseptic implant loosening associated with particulate wear debris. Acta Orthop. Belg. 79, 1–9.
- KELLGREN, J.H., LAWRENCE, J.S., 1957. RADIOLOGICAL ASSESSMENT OF OSTEO-ARTHROSIS, Ann. rheum. Dis. BMJ Publishing Group. https://doi.org/10.1136/ard.16.4.494
- Kirking, B., Krevolin, J., Townsend, C., Colwell, C.W., D'Lima, D.D., 2006. A multiaxial force-sensing implantable tibial prosthesis. J. Biomech. 39, 1744–1751. https://doi.org/10.1016/J.JBIOMECH.2005.05.023
- Kurtz, S., Mowat, F., Ong, K., Chan, N., Lau, E., Halpern, M., 2005. Prevalence of primary and revision total hip and knee arthroplasty in the United States from 1990 through 2002. J. Bone Joint Surg. Am. 87, 1487–97. https://doi.org/10.2106/JBJS.D.02441
- Kurtz, S., Ong, K., Lau, E., Mowat, F., Halpern, M., 2007. Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. J. Bone Joint Surg. Am. 89, 780–5. https://doi.org/10.2106/JBJS.F.00222
- Kutzner, I., Hallan, G., Høl, P.J., Furnes, O., Gøthesen, Ø., Figved, W., Ellison, P., 2018.
 Early aseptic loosening of a mobile-bearing total knee replacement. Acta Orthop. 89, 77–83. https://doi.org/10.1080/17453674.2017.1398012
- Lespasio, M.J., Piuzzi, N.S., Husni, M.E., Muschler, G.F., Guarino, A., Mont, M.A., 2017. Knee Osteoarthritis: A Primer. Perm. J. 21, 16–183.

https://doi.org/10.7812/TPP/16-183

- Lin, L., Xie, Y., Wang, S., Wu, W., Niu, S., Wen, X., Wang, Z.L., 2013. Triboelectric active sensor array for self-powered static and dynamic pressure detection and tactile imaging. ACS Nano 7, 8266–8274. https://doi.org/10.1021/nn4037514
- Luciano, V., Sardini, E., Serpelloni, M., Baronio, G., 2014. An energy harvesting converter to power sensorized total human knee prosthesis. Meas. Sci. Technol. 25, 025702. https://doi.org/10.1088/0957-0233/25/2/025702
- Manning, W.A., Ghosh, K.M., Blain, A., Longstaff, L., Rushton, S.P., Deehan, D.J., 2018. Internal femoral component rotation adversely influences load transfer in total knee arthroplasty: a cadaveric navigated study using the Verasense device. Knee Surg. Sports Traumatol. Arthrosc. 26, 1577–1585. https://doi.org/10.1007/s00167-017-4640-5
- Mason, J.B., Fehring, T.K., 2006. Removing well-fixed total knee arthroplasty implants, in: Clinical Orthopaedics and Related Research. pp. 76–82. https://doi.org/10.1097/01.blo.0000214413.06464.ce
- Medel, F.J., Kurtz, S.M., Hozack, W.J., Parvizi, J., Purtill, J.J., Sharkey, P.F.,
 MacDonald, D., Kraay, M.J., Goldberg, V., Rimnac, C.M., 2009. Gamma inert
 sterilization: a solution to polyethylene oxidation? J. Bone Joint Surg. Am. 91, 839–
 49. https://doi.org/10.2106/JBJS.H.00538
- Meneghini, R.M., Ziemba-Davis, M.M., Lovro, L.R., Ireland, P.H., Damer, B.M., 2016.
 Can Intraoperative Sensors Determine the "Target" Ligament Balance? Early
 Outcomes in Total Knee Arthroplasty. J. Arthroplasty 31, 2181–2187.
 https://doi.org/10.1016/J.ARTH.2016.03.046
- Noble, P.C., Scuderi, G.R., Brekke, A.C., Sikorskii, A., Benjamin, J.B., Lonner, J.H., Chadha, P., Daylamani, D.A., Scott, W.N., Bourne, R.B., 2012. Development of a new Knee Society scoring system. Clin. Orthop. Relat. Res. 470, 20–32.

https://doi.org/10.1007/s11999-011-2152-z

Parratte, B.S., Pagnano, M.W., 2008. Instability After Total Knee Arthroplasty 184–194.

- Petrie, J.R., Haidukewych, G.J., 2016. Instability in total knee arthroplasty. Bone Joint J. 98-B, 116–119. https://doi.org/10.1302/0301-620X.98B1.36371
- Platt, S.R., Farritor, S., Garvin, K., Haider, H., 2005. The use of piezoelectric ceramics for electric power generation within orthopedic implants. IEEE/ASME Trans. Mechatronics 10, 455–461. https://doi.org/10.1109/TMECH.2005.852482
- Risitano, S., Karamian, B., Indelli, P.F., 2017. Intraoperative load-sensing drives the level of constraint in primary total knee arthroplasty: Surgical technique and review of the literature. J. Clin. Orthop. trauma 8, 265–269. https://doi.org/10.1016/j.jcot.2017.06.004
- Rönn, K., Reischl, N., Gautier, E., Jacobi, M., 2011. Current surgical treatment of knee osteoarthritis. Arthritis 2011, 454873. https://doi.org/10.1155/2011/454873
- Rosenthall, L., 1997. Hip and knee prostheses: Evaluation of the natural history of periprosthetic bone changes. Semin. Nucl. Med. 27, 346–354. https://doi.org/10.1016/S0001-2998(97)80007-4
- Safaei, M., Meneghini, R.M., Anton, S.R., 2018. Energy Harvesting and Sensing with Embedded Piezoelectric Ceramics in Knee Implants. IEEE/ASME Trans. Mechatronics 23, 864–874. https://doi.org/10.1109/TMECH.2018.2794182
- Safaei, M., Meneghini, R.M., Anton, S.R., 2017. Parametric analysis of electromechanical and fatigue performance of total knee replacement bearing with embedded piezoelectric transducers. Smart Mater. Struct. 26. https://doi.org/10.1088/1361-665X/aa814e
- Saleh, K.J., Thongtrangan, I., Schwarz, E.M., 2004. Osteolysis: Medical and surgical

approaches. Clin. Orthop. Relat. Res. 138–147. https://doi.org/10.1097/01.blo.0000142288.66246.4d

- Scuderi, G.R., Bourne, R.B., Noble, P.C., Benjamin, J.B., Lonner, J.H., Scott, W.N., 2012. The New Knee Society Knee Scoring System. Clin. Orthop. Relat. Res. 470, 3. https://doi.org/10.1007/S11999-011-2135-0
- Sharkey, P.F., Lichstein, P.M., Shen, C., Tokarski, A.T., Parvizi, J., 2014. Why Are Total Knee Arthroplasties Failing Today—Has Anything Changed After 10 Years? J. Arthroplasty 29, 1774–1778. https://doi.org/10.1016/J.ARTH.2013.07.024
- Shields, M., Carroll, M.D., Ogden, C.L., 2011. Adult obesity prevalence in Canada and the United States. NCHS Data Brief 1–8.
- Sikorski, J.M., 2008. Alignment in total knee replacement. J. Bone Joint Surg. Br. 90-B, 1121–1127. https://doi.org/10.1302/0301-620X.90B9.20793
- Sloan, M., Premkumar, A., Sheth, N.P., 2018. Projected volume of primary total joint arthroplasty in the u.s., 2014 to 2030. J. Bone Jt. Surg. - Am. Vol. 100, 1455–1460. https://doi.org/10.2106/JBJS.17.01617
- Sundfeldt, M., Carlsson, L. V., Johansson, C.B., Thomsen, P., Gretzer, C., 2006. Aseptic loosening, not only a question of wear: A review of different theories. Acta Orthop. 77, 177–197. https://doi.org/10.1080/17453670610045902
- Unitt, L., Sambatakakis, A., Johnstone, D., Briggs, T.W.R., 2008. Short-term outcome in total knee replacement after soft-tissue release and balancing. J. Bone Joint Surg. Br. 90-B, 159–165. https://doi.org/10.1302/0301-620X.90B2.19327
- Vasarhelyi, E.M., MacDonald, S.J., 2012. The influence of obesity on total joint arthroplasty. J. Bone Jt. Surg. - Ser. B 94 B, 100–102. https://doi.org/10.1302/0301-620X.94B11.30619

- Vazquez-Vela Johnson, G., Worland, R.L., Keenan, J., Norambuena, N., 2003. Patient demographics as a predictor of the ten-year survival rate in primary total knee replacement. J. Bone Joint Surg. Br. 85, 52–6. https://doi.org/10.1302/0301-620x.85b1.12992
- Vince, K.G., 2003. Why knees fail. J. Arthroplasty 18, 39–44. https://doi.org/10.1054/ARTH.2003.50102
- Vince, K.G., Abdeen, A., Sugimori, T., 2006. The Unstable Total Knee Arthroplasty: Causes and Cures. J. Arthroplasty 21, 44–49. https://doi.org/10.1016/J.ARTH.2006.02.101
- Wang, Z.L., 2014. Triboelectric nanogenerators as new energy technology and selfpowered sensors – Principles, problems and perspectives. Faraday Discuss. 176, 447–458. https://doi.org/10.1039/C4FD00159A
- Wang, Z.L., 2013. Triboelectric nanogenerators as new energy technology for selfpowered systems and as active mechanical and chemical sensors. ACS Nano 7, 9533–9557. https://doi.org/10.1021/nn404614z
- Weinstein, A.M., Rome, B.N., Reichmann, W.M., Collins, J.E., Burbine, S.A., Thornhill, T.S., Wright, J., Katz, J.N., Losina, E., 2013. Estimating the burden of total knee replacement in the United States. J. Bone Jt. Surg. - Ser. A 95, 385–392. https://doi.org/10.2106/JBJS.L.00206
- Whiteside, L.A., 2002. Soft tissue balancing: The knee. J. Arthroplasty 17, 23–27. https://doi.org/10.1054/ARTH.2002.33264
- Wilson, C.J., Theodoulou, A., Damarell, R.A., Krishnan, J., 2017. Knee instability as the primary cause of failure following Total Knee Arthroplasty (TKA): A systematic review on the patient, surgical and implant characteristics of revised TKA patients. Knee 24, 1271–1281. https://doi.org/10.1016/j.knee.2017.08.060

- Wluka, A.E., Lombard, C.B., Cicuttini, F.M., 2013. Tackling obesity in knee osteoarthritis. Nat. Rev. Rheumatol. 9, 225–235. https://doi.org/10.1038/nrrheum.2012.224
- Yercan, H.S., Ait Si Selmi, T., Sugun, T.S., Neyret, P., 2005a. Tibiofemoral instability in primary total knee replacement: A review, Part 1: Basic principles and classification. Knee 12, 257–266. https://doi.org/10.1016/J.KNEE.2005.01.004
- Yercan, H.S., Ait Si Selmi, T., Sugun, T.S., Neyret, P., 2005b. Tibiofemoral instability in primary total knee replacement: A review: Part 2: Diagnosis, patient evaluation, and treatment. Knee 12, 336–340. https://doi.org/10.1016/J.KNEE.2005.01.005
- Zhang, H., Yang, Y., Su, Y., Chen, J., Adams, K., Lee, S., Hu, C., Wang, Z.L., 2014. Triboelectric nanogenerator for harvesting vibration energy in full space and as selfpowered acceleration sensor. Adv. Funct. Mater. 24, 1401–1407. https://doi.org/10.1002/adfm.201302453

Chapter 2

2 Tuning of a 3D Printed Energy Harvester Package Design

Overview: In this chapter, the design of a compliant package prototype will be discussed. The triboelectric generators were designed to generate electricity based on the cyclic contact and separation of two components. These generators harness the electron exchange resulting from the triboelectric effect. The TEGs, in their role of measuring loads in TKRs, require a package that ensures gap closure which is proportional to the amount of load transferred. The package was designed to achieve an axial force-displacement relationship that is compatible with the maximum allowable compression of the triboelectric generators while minimizing shear displacements resulting from shear forces during gait, which can otherwise cause TEG damage.

2.1 Introduction

Despite the satisfactory success rate of modern TKR systems, implant failure still poses a problem (Baker et al., 2013; Mannion et al., 2009; Scott et al., 2010; Toms et al., 2009). Complications can be mitigated intraoperatively by ensuring optimal implant alignment and soft tissue balancing. Intraoperative tools such as the Verasense (Orthosensor, Inc., Dania Beach, FL) have proved to be an effective tool in soft tissue balancing during surgery. Some instrumented implants have been used in small cohorts of patients for research, however, clinically available instrumented implants are non-existent. Instrumented implants have the potential for providing important feedback in determining causes of implant failure which is unique to individual TKR recipients (Almouahed et al., 2017).

D'Lima et al. were the first to report the *in vivo* loads in a knee implant. In their first design iteration, a modified tibial tray with load cells located at the four corners of the tibial plateau, a power coil, a microtransmitter with an antenna were used to measure and relay the total force, anteroposterior and mediolateral load distribution, and the center of

pressure in the joint (D'lima et al., 2005a). The next iteration was capable of accurately measuring all six force components (Kirking et al., 2006). Bergmann et al. used a similar modified tibial tray as Kirking et al. to measure *in vivo* loads that may serve as a more accurate alternative to previously established standardized loads for pre-clinical testing of TKR such as ISO 14243 (Bergmann et al., 2014).

The main challenge of developing an instrumented implant is reliably powering these devices for their entire lifetime. The limitation of inductive coupling is the use of an external coil that is wrapped around the patient's knee. The presence of an external coil can impede mobility and thus alter patient biomechanics. In response to the need for a reliable, compact method of powering instrumented implants, the application of energy harvesting mechanisms has been investigated. Piezoelectric transducers have been embedded in UHMWPE bearings and tibial trays (Almouahed et al., 2017; Platt et al., 2005; Safaei et al., 2017). Luciano et al. presented work that uses an electromagnetic generator (Luciano et al., 2014). The alternative energy harvesting mechanisms, for load sensing applications in orthopaedic implants, have not been investigated thoroughly yet.

The triboelectric generator (TEG) developed by Ibrahim et al. is a novel approach in measuring tibial forces and harvesting energy from activities of daily living (Ibrahim et al., 2019a). Ibrahim et al. developed an energy harvesting system that uses the phenomenon known as the triboelectric effect. The TEGs are composed of specially designed contact interfaces that slide with respect to one another. The contact and separation between these interfacing surfaces create an electric charge that can be turned into usable electricity.

The objective of this study is to design a compliant interpositional device that interlocks between the UHMWPE bearing and the tibial tray, which can house the TEG components and elastically deform under physiological loading to provide an ideal amount of TEG contact and separation. Furthermore, the package must resist shear forces that can lead to permanent damage of the TEGs.



2.2 Materials and Methods

Figure 2-1-Design Process Flowchart

The overall design process is outlined in Figure 2-1. An analytical model was initially used to determine spring dimensions for the package design. Design iterations of the spring structures were performed with computer aided design (CAD) modelling and finite element analysis (FEA) on simplified models. The spring design that had the deflection and stress response was exported to a tibial tray shaped part. This part underwent subsequent design iterations to adjust the stiffness in different locations of the package design by altering the spring thickness.

2.2.1 Dimensions of Elastic Bodies

The geometry of a size 7 Triathlon tibial tray (Stryker, Kalamazoo, MI) was used for the prototype design. The geometries of the implant components were provided in the form

of polygonal surface models in the standard tessellation language (STL) file format. The inferior and superior geometry of the package was designed to interface with the existing locking mechanism between the PE bearing and tibial tray. The perimeter shape of the package was recreated based on that of the tibial tray and the PE insert. Based on the operating conditions of the TEGs, the package was designed to achieve a vertical displacement of 0.2 mm at the maximum vertical load of 2600 N as defined in ISO 14243 (ISO, 2009) while having a linear-elastic behavior across the entire force range. Elastic bodies were located along the periphery of the package to ensure an internal void to accommodate the 3 mm thick TEGs and the accompanying electronics.



2.2.2 Analytical Model Design

Figure 2-2- CAD model of the preliminary prototype concept design that feature elastic bodies along the periphery: (a) superior view, (b) isometric view, (c) posterior view, (d) section view that shows the internal void for TEG

For this prototype, a series of stacked beam structures were placed along the periphery. The design of the beam structures in the prototype can be calculated using the known perimeter of the tibial tray and the length of the beams (Figure 2-2). The force per beam structure can be calculated based on the number of beams that can fit within the finite perimeter of the package. If the theoretical amount of load transmitted in each beam structure is known, the moment of inertia can be altered to yield a desired deflection within the elastic region.

A net deflection of 0.2 mm was desired at a force of 2600 N, approximately equaling the maximum vertical contact force during gait. Ti6Al4V was the material used for the design due to its biocompatibility and the availability of selective laser melting for future manufacturing.



Figure 2-3- A magnified view of the beam structures located on the posterior of the package design. Length, height and fillet radius of the beams are denoted by L, h and r, respectively. *Note: The base (b) is orthogonal to h and L*.

A preliminary analytical solution was used to determine appropriate beam dimensions for the length (L), base (b), height (h), and fillet radii (r) (Figure 2-3). Equations 1 served as a starting point for calculating the moment of inertia (I) of a rectangular cross-sectional

area. Equations 1 calculate the height (*h*) in a beam fixed at both ends with a prescribed transverse displacement, Δ . Since two beams were stacked to decrease the strain of individual beams in the elastic bodies, a prescribed displacement of 0.1 mm was used in the calculations. Combinations of *b*, *h*, and *L* that resulted in calculated bending stresses and shear stresses less than the fatigue strength of Ti6Al4V (550 MPa) were considered in the design. Raw data and calculations of *h* are available in the Appendix.

$$h = \sqrt[3]{(\frac{2FL^3}{Eb\Delta})}$$
 Equation 1

2.2.3 Computational Model Design

Simplified models were constructed using CAD software (Solidworks, Dassault Systemes, Vélizy-Villacoublay, France). The simplified models consisted of two parallel, rectangular plates with elastic beam structures along two opposing sides (Figure 2-4). The stacked beam structures were modelled with the b, h, and L dimensions from the analytical solution.



Figure 2-4- Simplified, rectangular model used for further computational model iterations on the beam design. (a) isometric view, (b) front view

FEA was conducted on the simplified models using Solidworks Simulation. A static study where one of the corners on the bottom plate was fully restrained, and the bottom surface had a roller/slider constraint was performed. A compressive force equal to the theoretical magnitude which, based on the analytical model, would lead to the model deflecting 0.2 mm was applied to the top surface of the model (Figure 2-5).



Figure 2-5- Boundary and Loading Conditions: (a) Fixed restraint on corner on bottom surface, (b) Roller/slider on bottom surface, (c) Force applied on top surface

Tetrahedral elements with a global mesh size of 1 mm were used for the analyses. An hadaptive solution method was used for local mesh refinement in high stress regions in the model. H-adaptive meshing iteratively alters the mesh density in consecutive loops. In this study, the maximum number of 5 loops was used in the analysis. Convergence was achieved when the strain energy between subsequent loops yielded a 1% difference. The elastic modulus (*E*) was set to 128 GPa, yield strength (σ_y) equal to 1000 MPa (Renishaw plc, 2017) and the Poisson's ratio (v) was defined as 0.31 (AZoMaterials, n.d.). Stacked beam structure dimensions (base (*b*), length (*L*), height (*h*), and fillet radii (*r*)) from the analytical model that yielded the target response values closest to the desired values (σ_{vm} = 550 ±100 MPa, d = 0.2 ± 0.05 mm) were considered for further iterations.

The One-Factor-at-a-Time (OFAT) method was used to determine suitable dimensions for decreasing the midsection thickness of the beam using Bezier curves. The dimensions that were altered were *a* and h_{middle} (Figure 2-6). Reducing the thickness of the beam's midsection increased the compliance of the model. The dimensions that provided the desired deflection of 0.2 mm at the maximum load determined from the analytical model while having peak stresses below the fatigue strength of Ti6Al4V (550 MPa) were used in the final prototype design.





2.2.4 Placement of Elastic Bodies in Package Prototype

Using the acquired STL file of a size 7 Triathlon tibial tray and the *Scanto3D* add-in in Solidworks, the surface mesh of the tibial tray's perimeter shape was used to create a 5 mm thick solid geometry of the prototype base. For this prototype, there would be anticipated prohibitively high stresses in the elastic bodies if they contoured the posterior notch of the tibial tray. Therefore, this prototype disregards the posterior notch and as a result, the prototype is limited to the use for TKRs that resect the cruciate ligaments (cruciate sacrificing designs). The geometry of the elastic bodies, determined by the OFAT method in the simplified models, were sketched along the periphery of the base and were extruded to an initial thickness of 3 mm. The base of the package was reproduced on the top of the elastic bodies to create a top plate (Figure 2-7).



Figure 2-7- CAD Progression of Package Prototype: (a) STL file of tibial tray, (b) base of prototype, (c) extrusion of elastic bodies along periphery, (d) prototype with top plate

2.2.5 Spring Thickness Adjustment

In the following computational models used for the spring thickness adjustment, the model consisted of the package design from the previous section and a defeatured UHMWPE bearing. The UHMWPE bearing was modeled by sketching the perimeter shape of the tibial tray and extruding the resulting sketch by 6.25 mm— the thickness of a 9 mm thick UHMWPE bearing at its lowest point. The UHMWPE bearing model was mated to the superior surface of the top plate to create an assembly. This assembly was used in subsequent FEA. The CAD model of the package and UHMWPE bearing assembly was exported, as a .STEP file, to FEA software (Abaqus, Simulia, Johnston, RI) for further elastic body design iterations. A tie constraint was defined to connect the package prototype and the UHMWPE bearing instances. Two reference points were created: one was placed inferior of the assembly, and the other one was placed superior to the package offset from the sagittal plane in order to simulate a 60:40 medial load bias, and above the lowest points of the UHMWPE bearing (Figure 2-8). The inferior reference point was rigidly connected to the inferior surface of the package and fully restrained using an Encastre boundary condition. The superior reference point was rigidly connected to the superior surface of the UHMWPE bearing. Loads were applied to the superior reference point. The global mesh size for the package prototype was 0.7 mm, whereas the UHMWPE bearing had a global mesh size of 6.3 mm. Tetrahedral elements were used for both components. The Ti6Al4V package had the following material properties: $E = 130\ 000\ \text{MPa}$, and v = 0.31. The UHMWPE bearing component was assigned $E = 830\ \text{MPa}$, and v = 0.42.



Figure 2-8-A posterior view of the package-PE bearing assembly. Reference point, RP-1, was placed inferior to the assembly. Reference point, RP-2, was placed superior to the assembly and offset medially to replicate a 60:40 (medial:lateral) load bias.

A vertical compressive load of 2600 N, the maximum load in gait based on ISO 14243, was applied to the superior reference point. At this point, the thickness of the elastic bodies had a thickness of 3 mm along the entire periphery of the package. Due to the asymmetric geometry along the anteroposterior axis and the load being applied in the posterior part of the package, the thicknesses of the elastic bodies were altered in regions along the package to ensure a uniform deflection of 0.2 mm was achieved. Deflections that exceed 0.2 mm could exceed the gap distance of the TEGs thus possibly resulting in

damage to the TEGs' contacting surfaces. To compound the effect of an uneven deflection behaviour, the improper vertical alignment of the sawtooth ridges of the TEG interfacing surfaces can also contribute to TEG damage. Another reason for having the package deflect uniformly along the anteroposterior axis is to maximize the contact between the TEG surfaces which in turn maximizes the energy harvested.

If the target deflection behaviour was not achieved, the thickness of the package was altered to adjust the localized stiffness in different regions in the package. After elastic body adjustments were performed in Solidworks, FEA was conducted using the same parameters in the compression simulation. The elastic body thickness underwent further iterations until a uniform deflection of 0.2 mm was achieved.

As eluded previously, the TEG contacting surfaces are susceptible to damage from shear displacements and operate solely from contact and separation in the vertical direction. Therefore, the package must be able to eliminate shear displacements. The package's resistance to shear forces in gait was analyzed by applying a posteriorly-directed load of 265 N and an internal torque of 6 Nm from ISO 14243 to the superior rigid point to calculate the shear displacements in separate simulations. Compression was not superimposed on the shear forces.

2.2.6 Reverse Engineering Interlocking Mechanism of TKR Components

In order to install the TEGs inside the package, the package was separated into a bottom and top part with an interlocking geometry to resist relative transverse motion. To secure the two parts together, three M3 x 0.5mm, 6 mm flat head screws were located in threads created in the medial, lateral, and posterior regions of the prototype.

On the superior surface of the top plate, the interlocking geometry of the tibial tray was created using manufactured-provided surface mesh data. On the inferior surface of the bottom plate, the interlocking geometry of a UHMWPE bearing was replicated using the mesh data. Therefore, the package prototype was designed to fit in-between the tibial tray and UHMWPE bearing of a size 7 Stryker Triathlon system.

2.3 Results

2.3.1 Analytical Model Design

Based on the analytical model of a single beam, the σ_b (bending stress) decreased as the length and base of the beams increased. Contrarily, the calculated height, which would result in the desired deflection of 0.1 mm, exhibited a proportional relationship with σ_b (Figure 2-9 and Figure 2-10).

Due to the volumetric constraints of the package where the overall thickness should be minimized and the internal volume should be maximized, some dimensions for L, b and h were not considered. The gap distance between the top and bottom plates of the package was limited to a maximum of 6 mm. Enough vertical space should be reserved for the deflection of the beams. Also, having a small gap between beams would lead to sharp reentrant edges that can serve as stress concentrations. In this design, enough space was desired to relieve stress concentrations with fillets at the corners. In regards to the internal volume, encroachment of the elastic bodies towards the interior of the package was minimized by minimizing b of the beams.



Figure 2-9- Effects of length and base parameters on the bending stress.



Figure 2-10- Effect of length and base on the height

2.3.2 Computational Model Design

2.3.2.1 Simplified Design Simulations

The computational models that simulate compression of the simplified designs did not demonstrate a close agreement with the analytical model. The computational models had larger stresses and deflections that were calculated in the analytical model, as summarized in Table 2-1. A variable that was overlooked in the analytical model was the influence the fillet radii had on the stresses and deflections. When comparing test runs 3 and 4 in Table 2-1, there was approximately a 600 MPa decrease in $\sigma_{vm,max}$ and a 0.04 mm decrease in the vertical displacement when the fillet radius was increased by 0.25 mm. The peak stresses were located at the base of the beams in the computational

model (Figure 2-11). The exclusion of considering the effects of including the fillets in the analytical led to an underestimation of the expected stresses and deflections.

Table 2-1-Stress and deflection comparison between computational and analyticalmodel. The computational models presented in this table had dimensions similar towhat was used in the analytical model.

| | | | | Computational | | Analytical |
|----------|-----------------|--------|--------|------------------------|---------------------|------------------|
| | Beam Dimensions | | | Responses | | |
| Test Run | h [mm] | L [mm] | r [mm] | $\sigma_{ m vm}$ [MPa] | d _y [mm] | σ_b [MPa] |
| 1 | 1 | 8 | 0.25 | 1774 | 0.49 | 640 |
| 2 | 1.5 | 10 | 0.25 | 1330 | 0.39 | 552 |
| 3 | 1.5 | 12 | 0.25 | 1896 | 0.69 | 489 |
| 4 | 1.5 | 12 | 0.5 | 1300 | 0.65 | 489 |

Note: b = 3 mm, $d_{y,analytical}$ = 0.2 mm



Figure 2-11- An isometric view of the FE model for Test Run 2 (Table 2-1). Peak stress locations were located at the base of the beams (denoted by red arrow).
Combinations of dimensions were not limited to what was determined from the analytical model especially in light of the discrepancies that it had with the aforementioned computational models presented in Table 2-2. The shorter length beams, despite having lower stresses, were too stiff. The opposite was observed for longer beams. The stiffness was proportional to the height. The peak stresses and deflections occurred with the longer slender beams (i.e. Test Run 3 in Table 2-2). For the prototype design, Test 16 was used for further iterations in acquiring the desired mechanical response. These dimensions had

response values close to the target values; the stiffness and the stress could be reduced by reducing the thickness of the beam's midsection.

| | Beam Dimensions | | | Responses | | |
|----------|-----------------|--------|--------|-------------------------|---------------------|--|
| Test Run | h [mm] | L [mm] | r [mm] | $\sigma_{\rm vm}$ [MPa] | d _v [mm] | |
| 1 | 1 | 6 | 0.25 | 1113 | 0.19 | |
| 2 | 1 | 8 | 0.25 | 1774 | 0.49 | |
| 3 | 1 | 10 | 0.25 | 2200 | 1.00 | |
| 4 | 1.5 | 6 | 0.25 | 632 | 0.08 | |
| 5 | 1.5 | 8 | 0.25 | 984 | 0.19 | |
| 6 | 1.5 | 10 | 0.25 | 1330 | 0.39 | |
| 7 | 1.5 | 12 | 0.25 | 1896 | 0.69 | |
| 8 | 2 | 6 | 0.25 | 400 | 0.04 | |
| 9 | 2 | 8 | 0.25 | 608 | 0.10 | |
| 10 | 2 | 10 | 0.25 | 815 | 0.20 | |
| 11 | 2 | 12 | 0.25 | 594 | 0.17 | |
| 12 | 1 | 6 | 0.5 | 839 | 0.17 | |
| 13 | 1 | 8 | 0.5 | 1358 | 0.45 | |
| 14 | 1 | 10 | 0.5 | 1999 | 0.97 | |
| 15 | 1.5 | 6 | 0.5 | 389 | 0.07 | |
| 16 | 1.5 | 8 | 0.5 | 632 | 0.17 | |
| 17 | 1.5 | 10 | 0.5 | 1919 | 0.35 | |
| 18 | 1.5 | 12 | 0.5 | 1300 | 0.65 | |

 Table 2-2- Stress and deflections resulting from dimensions altered using the One-Factor-at-a-Time method.

```
Note: b = 3 mm
```

2.3.2.2 Tuning of Midsection Thickness

After selecting the dimensions from Test Run 16 in Table 2-2, Bezier curves were used to create a tapered midsection in the beams. Thinning the midsection increased the compliance and reduced the stresses when compared to a beam with a uniform cross-section. After several changes to the altered parameters, the dimensions from Test Run 8, in Table 2-3, were exported to the tibial tray shaped prototype since it had a peak von Mises stress less than 550 MPa and the resultant force from a prescribed vertical

displacement of 0.2 mm that was close to the theoretical target compressive load. The progressive beam dimension changes are outlined below in Table 2-3.

| Table 2-3- Summary of | f stresses and res | sultant vertical | force after | altering bean | n |
|-----------------------|--------------------|------------------|-------------|---------------|---|
| mid | section thicknes | s using Bezier c | urves. | | |

| | Beam Dimensions | | | | | Responses | | |
|----------|-----------------|--------|--------|--------|---------------------------------|------------------------|----------------------|--|
| Test Run | h [mm] | L [mm] | r [mm] | a [mm] | h _{midsection} [mm] | $\sigma_{ m vm}$ [MPa] | F _{R.z} [N] | |
| 1 | 1.5 | 8.5 | 0.5 | 2 | 1 | 560 | 816 | |
| 2 | 1.5 | 8.5 | 0.5 | 3 | 1 | 598 | 854 | |
| 3 | 1.5 | 8.5 | 0.5 | 3.5 | 1 | 611.3 | 872 | |
| 3 | 1.75 | 8.5 | 0.5 | 3.5 | 1 | 719.2 | 1129 | |
| 4 | 1.75 | 8.5 | 0.5 | 3 | 1 | 702.8 | 1100 | |
| 5 | 1.75 | 8.5 | 0.5 | 3.5 | 1 | 712.8 | 1129 | |
| 6 | 1.75 | 8.5 | 0.5 | 3.5 | 0.75 | 637 | 983 | |
| 7 | 1.75 | 8.5 | 0.6 | 3.5 | 0.5 | 513 | 830 | |
| 8 | 1.75 | 8.5 | 0.6 | 3.75 | 0.5 | 527.5 | 855 | |

Note: b = 3 mm; Prescribed displacement was applied instead of compressive force

2.3.3 Simulation Results of Prototype Design

From the vertical load simulation, stresses were generally below the 550 MPa threshold. The highest stresses were located at the base of the beams, where bending moments would dominate, in the medial and lateral regions of the package. The high stresses exceeded the fatigue strength of Ti6Al4V in a small region (about 5.7% of the beam's volume) (Figure 2-12). The peak stress was 857 MPa located in the anteromedial portion of the package. The deflection of the top plate with respect to the bottom plate was 0.22 mm, slightly greater than the target of 0.2 mm (Figure 2-13).



Figure 2-12- Stress Plot from Maximum Axial Load





From the posteriorly directed load of 265 N, the shear displacements of the top plate with respect to the bottom plate and the stresses were negligible. Similar stresses and displacements were calculated when an internal torque of 6 Nm was applied to the computational model. The maximum stresses and displacements from the ISO 14243 loads are outlined in Table 2-4. In Figure 2-14, the maximum shear displacements from the shear loads are shown.



Figure 2-14- Shear displacements due to maximum posteriorly directed force (F_y) and internal moment (M_z) in ISO 14243

| Table 2-4- Maximum Stress a | and Displac | ements from | Maximum | Loads in | ISO 14243 |
|-----------------------------|-------------|-------------|----------------|----------|------------------|
|-----------------------------|-------------|-------------|----------------|----------|------------------|

| Applied Load | σ _{vm} [MPa] | u _x [mm] | u _v [mm] | u _z [mm] |
|-------------------------|-----------------------|---------------------|---------------------|---------------------|
| F _z = 2600 N | 550 | 0.01 | 0.013 | 0.22 |
| F _y = -265 N | 52.6 | 0.00061 | 0.0038 | 0.0039 |
| $M_z = 6 Nm$ | 46.4 | 0.0017 | 0.0016 | 0.0017 |

2.3.4 Dimensions of Elastic Bodies

The dimensions of the stacked beam structures for the compliant package are outlined in (Figure 2-15). The thickness of the elastic bodies was altered because of the larger load distribution in the posterior of the package. Therefore, beam thickness gradually increased from the anterior to the posterior. This gradual change in the thickness is illustrated in Figure 2-16.



Figure 2-15-Dimensions of stacked beam structure



Figure 2-16- Cross-sectional view of the elastic bodies. The prototype was symmetrical along the sagittal plane, therefore dimensions were denoted only on one side.

2.4 Discussion

In this current study, a compliant, linearly elastic package was designed for housing the TEGs developed by Ibrahim et al (Ibrahim et al., 2019a). Springs were positioned along the periphery of a tibial tray shaped package to achieve a maximum deflection of 0.2 mm (gap distance between the TEG layers) under the maximum vertical load according to ISO 14243. The package had peak stresses that surpassed the fatigue strength of Ti6Al4V but were lower than the yield strength.

The beam dimensions, initially determined using an analytical model, were used for creating rectangular, simplified designs that underwent simulated compression. The computational models calculated the peak stress and stiffness of the elastic body design iterations. Modifications were made to the dimensions of the springs until the target stress and deflection were achieved. The springs that had the desired mechanical response from the computational models of the simplified designs were exported along the periphery of a tibial tray shaped package. Further tuning was done to this model by varying the thickness of the beams until a uniform vertical displacement was calculated. After applying the shear forces occurring during gait, the package was verified to have a negligible amount of shear displacement.

The reason behind using a large TKR was to maximize the surface area of the TEGs. The TEG's energy harvesting capability is partly limited to the surface area. An increase of the surface area would increase the area for contact and separation between the interfacing surfaces thus increasing the power generated. Another benefit to deriving a design from a larger tibial tray is that the volume available for a compliant geometry within the package would be maximized. The technique used in this chapter can be implemented in all other implant sizes and different designs.

The dimensions of the current design may not be the optimal dimensions that would result in the desired stress and deflections. There may be other combinations not considered that may be more suitable. Localized changes to all the dimensions may be better suited in different areas of the package as opposed to altering the base (b) of the beams where force distributions differ. The rationale behind changing only the base of the beams was the efficiency to update the model and undergo further design iterations.

Higher stresses were calculated in the tibial tray shaped prototype when compared to the simplified model. This is a limitation of projecting an idealized beam structure onto a curved profile. After projecting the sketch and extruding the beams towards the interior of the package, the beams do not maintain the length that was tested in the simplified models. This led to increased stress magnitudes in sections of the beams adjacent to the internal void. The primary objective was to create a prototype with the desired stiffness. By limiting the elastic bodies along the periphery, the beams did not maintain the geometry that was used in the simplified models thus resulting in stresses and stiffness that differ from the simplified models.

High stresses resulting from the projection of the beam structure to the package periphery explains the absence of the posterior notch. The idealized beam structure was not replicated along the posterior notch. The shorter beams of the projected beam structure increased the stresses. With the means of reducing the stress and maximizing the internal surface area for the TEGs, the notch was disregarded in this initial prototype. Although some TKR designs require the resection of the cruciate ligaments, cruciate retaining (CR) designs will require the PCL. Future design iterations will require the need to include the posterior notch while possessing the mechanical properties to allow the TEGs to operate optimally.

The height (h) would have a more profound effect on the stresses in the package because of its cubic relationship with the moment of inertia (I). In addition, b could be reduced as h increases thus further maximizing the internal volume. The localized tuning of h can optimize the stiffness and reduce the stress. However, altering the projected elastic body sketch along the tibial tray shaped periphery proved to be difficult. Varying even just a single beam parameter would require changes to the sketch originating from the simplified design. Performing this for all elastic body units of the package design would be time consuming and arduous. For future consideration, the use of a tibial tray shaped prototype with elastic bodies with parameters that can be readily changed can facilitate localized tuning.

Longer beams in the analytical model resulted in lower calculated stresses while achieving the desired stiffness. The main limitation behind designing beams that are longer than what was used is that the required beam height h was limited by the height of the internal void. In other words, the longer beams could not be accommodated in a void with a 6 mm vertical height.

There are inherent limitations associated with using OFAT in designing the compliant package. OFAT can be an inefficient method in design optimization. It can require more time, experiments and resources (Telford, 2007). The estimates of each factor that explains the response aren't always precise. The main effects can be determined from OFAT, but the effects of interactions between input variables cannot be quantified. For instance, using OFAT, the effect of simultaneously changing the *L* and *b* on the deflection cannot be estimated. Since OFAT consists largely on "trial and error", the entire factor space is not considered thus increasing the possibility of false optimal conditions (Antony et al., 2003). Design of experiments (DOE) is a more efficient method of quantifying the sensitivity of the input variables and it provides a mathematical model that accounts for the estimation of main effects and interactions (Antony et al., 2003; Czitrom, 1999; Telford, 2007; Wahid and Nadir, 2013).

The peak stresses in the package design were below the yield strength of Ti6Al4V, but higher than the fatigue strength. For this prototype, a package with a linearly elastic behaviour was prioritized over the fatigue behaviour. The multi-objective nature of this application can prove to be difficult to satisfy all the desired engineering constraints, but this package had a stiffness that has the potential to work with the TEGs, maximized the space for the TEGs, and the package remained in the elastic region under 2600 N of compression.

Another limitation to this study is that other motions like jogging were not tested. Ideally, the package should be designed to withstand physiological loading with the largest magnitudes. For the case of preliminary evaluation, gait loading from ISO 14243 was assumed to be acceptable.

The modular design of the package provides surgeons intraoperative choices regarding component thickness and implant constraint. Another advantage of using modular components is that a worn PE bearing can be replaced without affecting the bony fixation of the tibial component during a revision procedure (Barrack, 1994). However, the modularity of implant components can be problematic. Wear and tribocorrosion can occur between interfacing surfaces thus resulting in wear induced synovitis, osteolysis or adverse local soft tissue reactions (ALTRs) (Berry et al., 2014; Jacobs et al., 2009). The PE bearing-package and package-tibial tray interfaces can elicit an increase in wear particles.

The micromotion between the underside of the PE bearing and the top plate of the package can lead to larger amounts of polyethylene wear and debris particles (Li et al., 2002; Sisko et al., 2017; Wasielewski et al., 1997). Two factors that greatly affect this PE backside wear are the surface finish of the tibial tray and the interlocking mechanism (Rao et al., 2002; Sisko et al., 2017). The use of TKR systems with highly polished tibial tray surfaces can reduce backside damage and linear wear (Berry et al., 2012; Teeter et al., 2015). A robust interlocking mechanism that restricts tibial tray motion also contributes to PE bearing longevity (Conditt et al., 2004; Jayabalan et al., 2007). The current prototype's design is derived from the Stryker Triathlon TKR system. The Triathlon interlocking mechanism, which features a central anti-rotational island to the peripheral capture, limits PE backside motion (Sisko et al., 2017). Despite the Triathlon interlocking mechanism's effectiveness in reducing PE backside wear, future iterations of this package design can limit PE damage by polishing the package surfaces.

The package-tibial tray interface may be more concerning than the PE bearing-package interface. This metal-on-metal interface can introduce metal ions which can lead to

synovitis and local soft tissue reactions (Berry et al., 2014; Jacobs et al., 2009). The wear and tribocorrosion at this interface can be exacerbated by the rough surfaces of both components. Future designs should ensure minimal relative micromotion between these surfaces. A high tolerance press-fit of the package and the tibial tray may be a viable option in reducing the micromotion. Otherwise, a custom tibial component with the integrated package may be required to eliminate the metal-on-metal surface interaction. Overall, when considering a package design that is compatible with other TKR systems, surface finish and the interlocking mechanism are still of a concern in terms of mitigating implant failure.

Future research can be done to optimize the design of the elastic bodies. DOE and topological optimization may provide other geometric alternatives that will provide the desired stiffness and reduce the peak stresses. This current study only considered the loads in ISO 14243. In future iterations, the stiffness and stresses should be tuned with consideration of other ADL such as jogging, stair ascent and stair descent where higher magnitude loads can be transmitted through the package.

2.5 Conclusion

In this chapter, the process of designing a compliant package intended for the integration of TEGS was outlined. The deflection of the package was slightly larger than the target of 0.2 mm at a compressive load of 2600 N. The package, based on the computational models, was able to resist shear forces that can lead to permanent damage of the TEGs. The calculated peak stresses exceeded the fatigue strength but remained within the elastic region of Ti6Al4V. This suggests that failure is possible due to fatigue. The following chapters will examine a 3D printed prototype where the stiffness of the prototype will be compared with the computational results and the fatigue life of the package will be quantified.

2.6 References

- Almouahed, S., Hamitouche, C., Stindel, E., 2017. Optimized Prototype of Instrumented Knee Implant: Experimental Validation. IRBM 38, 250–255. https://doi.org/10.1016/J.IRBM.2017.06.005
- Antony, J., Chou, T., Ghosh, S., 2003. Training for design of experiments. Work Study 52, 341–346. https://doi.org/10.1108/00438020310502642
- AZoMaterials, n.d. Properties: Titanium Alloys Ti6Al4V Grade 5 [WWW Document]. URL https://www.azom.com/properties.aspx?ArticleID=1547 (accessed 11.4.19).
- Baker, P.N., Rushton, S., Jameson, S.S., Reed, M., Gregg, P., Deehan, D.J., 2013. Patient satisfaction with total knee replacement cannot be predicted from pre-operative variables alone: A cohort study from the national joint registry for england and wales. Bone Jt. J. 95 B, 1359–1365. https://doi.org/10.1302/0301-620X.95B10.32281
- Barrack, R.L., 1994. Modularity of Prosthetic Implants, Journal of the American Academy of Orthopaedic Surgeons.
- Bergmann, G., Bender, A., Graichen, F., Dymke, J., Rohlmann, A., Trepczynski, A., Heller, M.O., Kutzner, I., 2014. Standardized Loads Acting in Knee Implants. PLoS One 9, 86035. https://doi.org/10.1371/journal.pone.0086035
- Berry, D.J., Abdel, M.P., Callaghan, J.J., 2014. What Are the Current Clinical Issues in Wear and Tribocorrosion? Clin. Orthop. Relat. Res. 472, 3659–3664. https://doi.org/10.1007/s11999-014-3610-1
- Berry, D.J., Currier, J.H., Mayor, M.B., Collier, J.P., 2012. Knee wear measured in retrievals: A polished tray reduces insert wear, in: Clinical Orthopaedics and Related Research. Springer New York LLC, pp. 1860–1868. https://doi.org/10.1007/s11999-012-2248-0

- Conditt, M.A., Stein, J.A., Noble, P.C., 2004. Factors affecting the severity of backside wear of modular tibial inserts. J. Bone Joint Surg. Am. 86, 305–11. https://doi.org/10.2106/00004623-200402000-00013
- Czitrom, V., 1999. One-factor-at-a-time versus designed experiments. Am. Stat. 53, 126– 131. https://doi.org/10.1080/00031305.1999.10474445
- D'lima, D.D., Patil, S., Steklov, N., Slamin, J.E., Colwell, C.W., 2005. THE CHITRANJAN RANAWAT AWARD In Vivo Knee Forces after Total Knee Arthroplasty. https://doi.org/10.1097/01.blo.0000186559.62942.8c
- Ibrahim, A., Jain, M., Salman, E., Willing, R., Towfighian, S., 2019. A smart knee implant using triboelectric energy harvesters. Smart Mater. Struct. 28, 025040. https://doi.org/10.1088/1361-665X/aaf3f1
- ISO, 2009. ISO 14243-1, Implants for surgery Wear of total knee-joint prostheses.
- Jacobs, J.J., Urban, R.M., Hallab, N.J., Skipor, A.K., Fischer, A., Wimmer, M.A., 2009. Metal-on-metal bearing surfaces. J. Am. Acad. Orthop. Surg. 17, 69–76. https://doi.org/10.5435/00124635-200902000-00003
- Jayabalan, P., Furman, B.D., Cottrell, J.M., Wright, T.M., 2007. Backside wear in modern total knee designs. HSS J. 3, 30–34. https://doi.org/10.1007/s11420-006-9033-0
- Kirking, B., Krevolin, J., Townsend, C., Colwell, C.W., D'Lima, D.D., 2006. A multiaxial force-sensing implantable tibial prosthesis. J. Biomech. 39, 1744–1751. https://doi.org/10.1016/J.JBIOMECH.2005.05.023
- Li, S., Scuderi, G., Furman, B.D., Bhattacharyya, S., Schmieg, J.J., Insall, J.N., 2002. Assessment of backside wear from the analysis of 55 retrieved tibial inserts, in: Clinical Orthopaedics and Related Research. Lippincott Williams and Wilkins, pp. 75–82. https://doi.org/10.1097/00003086-200211000-00013

- Luciano, V., Sardini, E., Serpelloni, M., Baronio, G., 2014. An energy harvesting converter to power sensorized total human knee prosthesis. Meas. Sci. Technol. 25, 025702. https://doi.org/10.1088/0957-0233/25/2/025702
- Mannion, A.F., Kämpfen, S., Munzinger, U., Kramers-de Quervain, I., 2009. The role of patient expectations in predicting outcome after total knee arthroplasty. Arthritis Res. Ther. 11. https://doi.org/10.1186/ar2811
- Platt, S.R., Farritor, S., Garvin, K., Haider, H., 2005. The use of piezoelectric ceramics for electric power generation within orthopedic implants. IEEE/ASME Trans. Mechatronics 10, 455–461. https://doi.org/10.1109/TMECH.2005.852482
- Rao, A.R., Engh, G.A., Collier, M.B., Lounici, S., 2002. Tibial interface wear in retrieved total knee components and correlations with modular insert motion. J. Bone Joint Surg. Am. 84, 1849–55. https://doi.org/10.2106/00004623-200210000-00017
- Renishaw plc, 2017. Ti6Al4V ELI-0406 powder for additive manufacturing [WWW Document]. URL https://resources.renishaw.com/en/details/data-sheet-ti6al4v-eli-0406-powder-for-additive-manufacturing--94700
- Safaei, M., Meneghini, R.M., Anton, S.R., 2017. Parametric analysis of electromechanical and fatigue performance of total knee replacement bearing with embedded piezoelectric transducers. Smart Mater. Struct. 26. https://doi.org/10.1088/1361-665X/aa814e
- Scott, C.E.H., Howie, C.R., MacDonald, D., Biant, L.C., 2010. Predicting dissatisfaction following total knee replacement: A prospective study of 1217 patients. J. Bone Jt. Surg. - Ser. B 92, 1253–1258. https://doi.org/10.1302/0301-620X.92B9.24394
- Sisko, Z.W., Teeter, M.G., Lanting, B.A., Howard, J.L., McCalden, R.W., Naudie, D.D., MacDonald, S.J., Vasarhelyi, E.M., 2017. Current Total Knee Designs: Does Baseplate Roughness or Locking Mechanism Design Affect Polyethylene Backside Wear? Clin. Orthop. Relat. Res. 475, 2970–2980. https://doi.org/10.1007/s11999-

- Teeter, M.G., Lanting, B.A., Shrestha, K.R., Howard, J.L., Vasarhelyi, E.M., 2015. Contribution of Surface Polishing and Sterilization Method to Backside Wear in Total Knee Arthroplasty. J. Arthroplasty 30, 2320–2322. https://doi.org/10.1016/j.arth.2015.06.036
- Telford, J.K., 2007. A brief introduction to design of experiments. Johns Hopkins APL Tech. Dig. (Applied Phys. Lab. 27, 224–232.
- Toms, A.D., Mandalia, V., Haigh, R., Hopwood, B., 2009. The management of patients with painful total knee replacement. J. Bone Jt. Surg. - Ser. B 91, 143–150. https://doi.org/10.1302/0301-620X.91B2.20995
- Wahid, Z., Nadir, N., 2013. Improvement of one factor at a time through design of experiments. World Appl. Sci. J. 21, 56–61. https://doi.org/10.5829/idosi.wasj.2013.21.mae.99919
- Wasielewski, R.C., Parks, N., Williams, I., Surprenant, H., Collier, J.P., Engh, G., 1997.Tibial insert undersurface as a contributing source of polyethylene wear debris. Clin.Orthop. Relat. Res. 53–9.

Chapter 3

3 Experimental Validation and Load Imbalance Measurements of a Compliant Package Design

Overview: The stiffness of the package was determined from axial compressive loads applied to the package. The package also underwent simulated load imbalances to measure differences between the medial and lateral compartmental forces measured by surrogate sensors.

3.1 Introduction

Proper implant alignment and soft tissue balancing are imperative for favourable patient outcomes and implant longevity. Intraoperative sensors have been used to objectively ensure that implant components are positioned correctly, and as a useful tool for ligament balancing by providing real-time quantitative feedback (Cho et al., 2018; Elmallah et al., 2016; Kenneth A Gustke et al., 2014; Kenneth A. Gustke et al., 2014). All possible preemptive measures can be taken to minimize future implant failures intraoperatively, but the condition of an implant can change over the course of healing. Currently, there is no quantitative method of monitoring the postoperative loads transmitted in one's prosthetic knee available for routine clinical use.

The benefit of instrumented implants is the ability to better understand the *in vivo* mechanics of a patient's TKR. The ability to measure tibial forces can help identify signs of instability, and aseptic loosening. Measured loads can be used to corroborate patient histories so that the most suitable actions can be made to improve implant longevity.

The triboelectric effect can be used for both load sensing and energy harvesting applications. TEGs developed by Ibrahim et al. are designed to operate between the UHMWPE bearing and tibial tray. The motivation behind the design of the compliant package described in Chapter 2 was to integrate the TEG in a commercially available TKR system.

This objective of this chapter is the experimental validation of the computational model presented in Chapter 2, and to measure the effect of intentional load imbalances on the deflection of a package prototype. A manufactured prototype underwent uniaxial compression testing to determine the stiffness of the prototype and how this value deviates from the stiffness determined from the computational model. In addition, several simplified structures underwent compression testing. The stiffness of the manufactured simplified structures was also compared to their corresponding computational models. Using embedded load sensors, the medial and lateral compartment forces were measured as the mediolateral load distribution oscillated between a medial and a lateral load bias during simulated abduction/adduction.

3.2 Materials and Methods

3.2.1 Manufactured Prototype

Alterations were made to the prototype CAD model in Chapter 2 prior to manufacturing and mechanical testing. Since the package was intended to accommodate the TEGs, access to the interior of the package was achieved by printing the prototype into two interlocking, separate parts. The interlocking geometry between the parts provides resistance to relative transverse displacements. Additionally, using the acquired STL file of a size 7 Triathlon tibial tray and PE bearing, the *Scanto3D* add-in in Solidworks was used to recreate the interlocking geometry of the TKR components on the top and bottom surfaces of the package. The parts were fastened together using three M3 x 0.05 6mm long flat head screws to prevent removal of the top plate. Threaded holes were located at the medial, lateral and posterior locations of the prototype. Once these alterations were completed, the prototype was manufactured using selective laser melting (Figure 3-1). The printing parameters are outlined in Table 3-1. The resulting prototype assembly was used in the subsequent experiments.

| Laser Power [W] | 400 |
|---------------------------|--------------------------------------|
| Laser Diameter [µm] | 70 |
| Layer Thickness [µm] | 40 |
| Powder Size Diameter [µm] | 15-45 |
| Heat Treatment Process | With argon supply on, 60 |
| | minutes to 350°C, hold at |
| | 350°C for 30 minutes. 60 |
| | minutes to 850°C, hold at |
| | 850°C for 60 minutes. Cool the |
| | furnace down slowly. |
| Surface Finish | Grit Blast (~5-10µm R _a) |

Table 3-1- Selective Laser Melting Parameters

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Figure 3-1- Package Prototype: (a) top view, (b) isometric view, (c) front view, (d) bottom view, (e) back view and (f) disassembled prototype.

3.2.2 Axial Loading of Package Prototype

In order to compare the stiffness of the computational model and the manufactured prototype, an Instron 8874 mechanical testing machine (Instron, Norwood, MA) equipped with a 5 kN load cell was used to measure the stiffness of the prototype. The prototype was placed along the vertical axis of the actuator on a metal stand used to raise the specimen within the actuator's line of travel. The actuator was translated downwards in displacement until in contact with the prototype, as indicated by changes in force

measured by the load cell in series with the prototype. In force control, contact between the load cell and the prototype was initially set by applying 500 N of compression. The further axial compressive load was then applied at a rate of 26 N/s up to a maximum of 1800 N. The displacement of the crosshead was acquired at a sampling rate of 1000 samples/s. Afterwards, the stiffness of the stand and load frame was acquired by performing the same loading protocol that was applied to the prototype with the only exception of swapping the prototype with a rigid, metal block. In order to correct for the compliance of the experimental set-up, the compliance of the prototype was isolated by subtracting the compliance determined from compressing the rigid block from the system that contained the prototype. The stiffness of the prototype was compared to the stiffness of the computational model presented in the previous chapter to quantify if there is agreement between the experimental and computational models.

3.2.3 Imbalance Testing

Compression and simulated load imbalances were applied using a VIVO joint motion simulator (Advanced Mechanical Technology, Inc, Watertown, MA) capable of applying loads and displacements in all 6 degrees-of-freedom (DOF). A Triathlon Knee System (Stryker, Kalamazoo, MI) was used for the design and testing of the package. The femoral component was cemented on a femoral component holder secured to the abduction arm of the VIVO using polymethylmethacrylate (PMMA) bone cement (Bosworth Fastray, Keystone Industries, Myerstown, PA). While the abduction arm of the VIVO was positioned at 0° of abduction/adduction and with the femoral component holder attached to the abduction arm, a sagittal plane defined midway between the femoral component's condyles was positioned to intersect the origin of the flexion and abduction arm along the mediolateral axis (Figure 3-2). On the lower actuator of the VIVO, the tibial tray was cemented on a custom fixture using dental cement (Dentstone, Kulzer, LLC, South Bend, IN).



Figure 3-2- The VIVO joint simulator operates in six degrees of freedom. There are three translational (medial/lateral, anterior/posterior, superior/inferior) and three rotational degrees of freedom (internal/external, abduction/adduction, flexion/extension).



Figure 3-3- Experimental set-up (the implant flexed about 60° for clarity)

At the time of the experiment, suitably capable TEG components were not available for experimental testing. Thus, ultra-thin Flexiforce sensors (Tekscan, Boston, MA), rated for up to 111 N (25 lbs.), were employed. Castable silicone (Amazing Remelt, Alumilite Corporation, Kalamazoo, MI) with the shape of the sensing element of the Flexiforce sensors was used to occupy the excess vertical space within the package and transmit loads through the Flexiforce sensors. The height of the silicone spacers was made to have a height slightly greater than the height of the internal void. This was done to ensure

contact between the Flexiforce sensors and the top plate. Based on the Shore A hardness of the silicone, provided by the manufacturer's specifications, the stiffness of a silicone spacer was calculated to be 43 N/mm. With the stiffness of the package determined from the computational model in Chapter 2 (11 818 N/mm) and the calculated silicone stiffness, the approximate amount of load that would be transmitted through the Flexiforce sensors at the maximum load of 2600 N was determined. Assuming equal loading is applied between the compartments, approximately 10 N would be transmitted through one of the Flexiforce sensors. The package transmits the majority of the load since it is stiffer than the silicone. The 111 N (25 lbs.) sensors were chosen because the low-load (4.4 N) rated sensors would most likely be saturated whereas the resolution may be too large with the high-load (445 N) rated sensors at the expected load of 10 N.

3.2.3.1 Flexiforce Sensor Calibration

Calibration of the Fexiforce sensors was performed using weights ranging from 0.45 kg (1 lbs.) to 5 kg (11 lbs.). The maximum test weight of 5 kg, which would exceed the expected Flexiforce load of 10 N, was selected to avoid possible output voltage saturation. A data acquisition (DAQ) card (National Instruments Corporation, Austin, TX) that was connected to the output voltage terminal of a Flexiforce Quickstart Board (Tekscan, Boston, MA) and a custom Labview (National Instruments Corporation, Austin, TX) program were used to acquire the output voltage from the Quickstart Board. Power to the Flexiforce Quickstart Board was supplied by a 9 V battery connected to the positive and negative input terminals. With a 111 N (25 lbs.) rated Flexiforce sensor connected to the Flexiforce Quickstart Board, the largest weight of 5 kg was placed on the Flexiforce's sensing element. The feedback resistance was adjusted with the potentiometer located on the Flexiforce Quickstart Board until the output voltage was 80% to 90% of the 5 V maximum output voltage. Afterwards, the full test weight was removed. The corresponding output voltages for 0.45, 0.9, 1.36 and 2.26 kg weights were acquired. A linear plot of the voltage-force relationship and the resulting line of best fit was acquired for subsequent experiments. The same calibration procedure was conducted with another 111 N (25 lbs.) rated Flexiforce sensor.



Figure 3-4- Package Prototype with Embedded Flexiforce Sensors in the Medial and Lateral Compartments

3.2.3.2 Experimental Set-Up

The vertical displacement from the VIVO joint simulator during axial loading was used to measure the deflection of the prototype. However, machine and fixture compliance are expected during testing. In order to eliminate the effects of the machine and fixture compliance, the TKR was tested with and without the prototype. By knowing the stiffness of both test configurations, the stiffness of the prototype could be isolated. The force and kinematic data from the VIVO were acquired at a rate of 100 Hz. The actuator position resolution is ± 0.1 mm.



Figure 3-5- Experimental set-up configurations: a) TKR mounted on VIVO joint simulator and b) TKR mounted on VIVO joint simulator with package prototype embedded with Flexiforce sensors.

In both testing configurations, the initial pose of the experimental set-up was determined by reducing the joint with a 50 N compressive load, while maintaining 0 N of force (or 0 Nm of torque) for all other remaining DOFs other than flexion. Flexion was set to 0°, or full extension. Once the initial position of the joint was defined, a sinusoidal compressive load ranging from 50 N to 2600 N was applied at 0.5 Hz. The stiffness of the experimental set-ups with and without the package prototype were calculated using the vertical displacement acquired from the VIVO and the applied force. The stiffness of the package was then calculated by subtracting the stiffness of the experimental set-up without the package from the stiffness of the TKR with the package. In addition to the prototype's deflection measured by the VIVO, the Flexiforce sensors measured the compartmental loads. A static compressive load of 2600 N was applied to the Flexiforce embedded prototype. Loading perturbations were superimposed atop the 2600 N compressive load. To simulate a load imbalance along the coronal plane, a sinusoidal abduction-adduction (AA) load of ± 20 Nm was applied. Using the medial force ratio distribution reported by Halder et al and Kutzner et al (Halder et al., 2014; Kutzner et al., 2013), 20 Nm of tibial adduction and an axial load of 2600 N, 65% of the total axial load would be transmitted through the medial compartment. An abduction moment of 20 Nm would shift the majority of the axial load laterally in which the medial compartment carries 35% of the mediolateral force distribution. All remaining DOF in each loading scenario maintained 0 N of force, or 0 Nm of torque. The cyclic loads were applied at a frequency of 0.5 Hz. Loads measured by the Flexiforce sensors and the VIVO joint simulator were recorded during loading.

3.3 Results

3.3.1 11Axial Loading of Package Prototype

During the cyclic compression testing, using the VIVO, the package had a stiffness of 4555 N/mm. With the Instron, the stiffness of the test fixture was 8712 N/mm. The fixture with the prototype mounted had a stiffness of 2615 N/mm (Figure 3-6). In order to nullify the stiffness of the test fixture, it was assumed the test fixture and the prototype were compressed in series. The prototype's stiffness was calculated to be 3190 N/mm. This is 8628 N/mm, or 73%, less than what was predicted by the computational model in Chapter 2.



Figure 3-6- Force vs. displacement plot of package prototype subjected to uniaxial vertical loading.

3.3.2 Imbalance Testing

During simulated coronal imbalance testing, as expected, the Flexiforce sensors showed opposite responses to the AA moment, however the magnitudes were not identical. This was probably due to the geometric and positional variance of the silicone spacers beneath the Flexiforce sensors. When a 20 Nm adduction moment was applied, the medial compartment force sensor experienced a 23% load increase versus a 33% decrease at the lateral compartment force sensor with respect to the static 2600 N compression. When a 20 Nm abduction moment was applied, the medial compartment load decreased by 8%, while the lateral compartment load increased by 8%.





3.4 Discussion

3.4.1 Stiffness of Additively Manufactured Parts

In this study, the stiffness of an additively manufactured prototype was experimentally determined for the purpose of measuring the difference with respect to the stiffness calculated in the computational model presented in the previous chapter. The prototype's stiffness, under axial loading, was 73% less than what was predicted in the computational model.

Additively manufactured structures have been shown to have stiffness that vary greatly from the stiffness determined from computational models which may be due to geometric discrepancies. The internal porosity, unfused material, and surface roughness can reduce the effective cross-sectional area of the 3D printed prototypes. As a result, additively manufactured parts may not possess the stiffness suggested by their respective

computational models. In some cases, computational models can have stiffness values that are greater than additively manufactured structures by a factor of 4 (Harrysson et al., 2008). Additional manufactured simplified designs with a similar elastic body structure underwent uniaxial vertical loading. Discrepancies between the computational and experimental models were observed with these simplified designs. These results are outlined in Appendix C. Despite the difference between the computational and experimental models, it can be valuable to understand the implications of considering SLM Ti6Al4V as a manufacturing method.

3.4.2 Intercompartmental Load Measurements

The prototype was also subjected to cyclic AA loading to quantify differences in the package's deflection as the tibiofemoral load distribution changes. During AA testing, compartmental loads from the Flexiforce sensors did indicate a shift in the load along the mediolateral axis. However, the measured loads at peak abduction and adduction did not show a symmetric response. The medial compartment measured 64% and 46% of the total load in full adduction and abduction, respectively. The measured compartmental loads did not display a symmetrical force response at the maximum and minimum loads, or displacements, of the perturbed loading scenarios. This may be attributed to the load sensor calibration, inconsistent silicone fill levels, and the placement of the load sensors within the package.

The ability to measure the tibiofemoral forces about the anteroposterior axis can help detect signs of future implant complications such as malalignment. Large abduction or adduction angles, greater than 3°, has been shown to increase the intercompartmental load difference (Bäthis et al., 2004; Halder et al., 2014). Higher AA moments may entail increased stresses in the surrounding implant components and underlying bones, PE bearing wear (Collier et al., 2007; Srivastava et al., 2012) and aseptic loosening (Gromov et al., 2014; Ritter et al., 2011). Kutzner et al. have demonstrated a correlation between tibiofemoral alignment and mediolateral force distribution where varus alignment leads to increased medial compartment loads. Medial force ratios of 70% to 80% can be expected

with a knee that is 3° varus (Kutzner et al., 2017). The prototype with embedded Flexiforce sensors has demonstrated the ability to measure the compartmental loads during the application of a cyclic AA moment. The package was compliant enough in the medial and lateral compartments to measure loads as the AA angle oscillated between $\pm 2^{\circ}$ during AA loading. The package most likely will be able to detect the increased coronal planeload imbalances associated with AA angles greater than 3° which can be beneficial in restoring a more even mediolateral load distribution through interventions such as soft tissue balancing.

A limitation to this study is that the performance of the TEGs, with this current package design, was not quantified. In spite of this, thin Flexiforce sensors were used in lieu of the TEGs due to their unavailability at the time of testing. Silicone spacers were placed underneath the sensors. Silicone was selected because of its low stiffness relative to the stiffness of the package. The low stiffness silicone reduced the interference the spacers had on the force transmitted through the package. However, silicone may not have been an ideal choice. Visual signs of plastic deformation, or indentations of the Flexiforce sensors, were seen on the silicone spacers after testing. This means that the measured loads acquired by the Flexiforce sensors may be inaccurate since a portion of the load was transmitted to the silicone that was not covered by the sensing element of the Flexiforce sensors.

Future testing can be performed to quantify the measured compartmental loads and power generated by the TEGs embedded in the package. Further tuning of the package may need to be done to maximize the electricity generated.

3.5 Conclusion

From uniaxial compression testing of the prototype, the AM Ti6Al4V components were more compliant as determined from their corresponding computational models. Despite this stiffness discrepancy between the experimental and computational models, the prototype deflected proportionally to shifts in the mediolateral load distribution. The compartmental loads measured with this package design in conjunction with embedded sensors can be a means of better understanding the effects of surgical technique and implant design on stability after implantation.

3.6 References

- Bäthis, H., Perlick, L., Tingart, M., Lüring, C., Zurakowski, D., Grifka, J., 2004. Alignment in total knee arthroplasty. A comparison of computer-assisted surgery with the conventional technique. J. Bone Jt. Surg. - Ser. B 86, 682–687. https://doi.org/10.1302/0301-620X.86B5.14927
- Cho, K.-J., Seon, J.-K., Jang, W.-Y., Park, C.-G., Song, E.-K., 2018. Objective quantification of ligament balancing using VERASENSE in measured resection and modified gap balance total knee arthroplasty. BMC Musculoskelet. Disord. 19, 266. https://doi.org/10.1186/S12891-018-2190-8
- Collier, M.B., Engh, C.A., McAuley, J.P., Engh, G.A., 2007. Factors Associated with the Loss of Thickness of Polyethylene Tibial Bearings After Knee Arthroplasty. J. Bone Jt. Surg. 89, 1306–1314. https://doi.org/10.2106/JBJS.F.00667
- Elmallah, R.K., Mistry, J.B., Cherian, J.J., Chughtai, M., Bhave, A., Roche, M.W., Mont, M.A., 2016. Can We Really "Feel" a Balanced Total Knee Arthroplasty? J. Arthroplasty 31, 102–105. https://doi.org/10.1016/J.ARTH.2016.03.054
- Gromov, K., Korchi, M., Thomsen, M.G., Husted, H., Troelsen, A., 2014. What is the optimal alignment of the tibial and femoral components in knee arthroplasty? An overview of the literature. Acta Orthop. 85, 480–487. https://doi.org/10.3109/17453674.2014.940573
- Gustke, Kenneth A., Golladay, G.J., Roche, M.W., Elson, L.C., Anderson, C.R., 2014. A new method for defining balance: Promising Short-Term Clinical Outcomes of Sensor-Guided TKA. J. Arthroplasty 29, 955–960. https://doi.org/10.1016/j.arth.2013.10.020
- Gustke, Kenneth A, Golladay, G.J., Roche, M.W., Elson, L.C., Anderson, C.R., 2014. Primary TKA patients with quantifiably balanced soft-tissue achieve significant clinical gains sooner than unbalanced patients. Adv. Orthop. 2014, 628695.

https://doi.org/10.1155/2014/628695

- Halder, A., Kutzner, I., Graichen, F., Heinlein, B., 2014. Influence of Limb Alignment on Mediolateral. J. Bone Jt. Surg. 94, 1023–1029.
- Harrysson, O.L.A., Cansizoglu, O., Marcellin-Little, D.J., Cormier, D.R., West, H.A., 2008. Direct metal fabrication of titanium implants with tailored materials and mechanical properties using electron beam melting technology. Mater. Sci. Eng. C 28, 366–373. https://doi.org/10.1016/j.msec.2007.04.022
- Kutzner, I., Bender, A., Dymke, J., Duda, G., Von Roth, P., Bergmann, G., 2017. Mediolateral force distribution at the knee joint shifts across activities and is driven by tibiofemoral alignment. Bone Jt. J. 99B, 779–787. https://doi.org/10.1302/0301-620X.99B6.BJJ-2016-0713.R1
- Kutzner, I., Trepczynski, A., Heller, M.O., Bergmann, G., 2013. Knee adduction moment and medial contact force-facts about their correlation during gait. PLoS One 8, 8–15. https://doi.org/10.1371/journal.pone.0081036
- Ritter, M.A., Davis, K.E., Meding, J.B., Pierson, J.L., Berend, M.E., Malinzak, R.A., 2011. The Effect of Alignment and BMI on Failure of Total Knee Replacement. J. Bone Jt. Surgery-American Vol. 93, 1588–1596. https://doi.org/10.2106/JBJS.J.00772
- Srivastava, A., Lee, G.Y., Steklov, N., Colwell, C.W., Ezzet, K.A., D'Lima, D.D., 2012. Effect of tibial component varus on wear in total knee arthroplasty. Knee 19, 560– 563. https://doi.org/10.1016/j.knee.2011.11.003

Chapter 4

4 Durability Testing of Energy Harvester Package

Overview: The fatigue strength is an important mechanical property to consider in implant design. Ideally, TKRs should withstand repetitive physiological loading for up to several decades of use. In this chapter, the fatigue life of the package prototype was quantified when subjected to a sinusoidal vertical load with maximum and minimum loads that correspond with ISO 14243.

4.1 Introduction

Ti6Al4V is a common material used for orthopaedic applications because of its high yield strength, low stiffness relative to other metal alloys, biocompatibility, and high fatigue strength. One of the main requirements the compliant package must satisfy is that it must be durable enough to withstand the repetitive loading of activities of daily living. The failure of the package will not only render the energy harvester damaged and unusable, but this can also lead to an introduction of foreign wear particles, and affect the stresses transmitted to the surrounding implant components that may elicit a biological response.

Complex geometries can be manufactured through additive manufacturing (AM) methods such as selective laser melting (SLM). The mechanical properties of AM Ti6Al4V are comparable to its wrought form (Baufeld et al., 2011; Brandl et al., 2010; Chan et al., 2013; Facchini et al., 2010; Koike et al., 2011; Leuders et al., 2013; Murr et al., 2009; Qiu et al., 2013), but AM Ti6Al4V may possess anisotropic microstructure and properties due to the directional nature of AM (Baufeld et al., 2011; Harrysson et al., 2008). However, the fatigue performance of AM Ti6Al4V is adversely affected by its inherent internal porosity, residual stress, build orientation and surface condition (Baufeld et al., 2011; Brandl et al., 2010; Chan et al., 2013; Edwards and Ramulu, 2014; Ghouse et al., 2018; Leuders et al., 2013; Shiomi et al., 2004; Sterling et al., 2015). The poor notch sensitivity of Ti6Al4V makes AM parts susceptible to defects and stress concentrations (Ahmadi et al., 2018; Hrabe et al., 2011). Internal pores and surface defects can become potential sites for crack initiation and propagation (Ghouse et al., 2018; Yadollahi and Shamsaei, 2017).

Despite the poor fatigue performance of AM Ti6Al4V compared to its wrought form, the post processing performed can improve the fatigue strength by relieving residual stress, improving the microstructure and surface finish (Edwards and Ramulu, 2014; Ghouse et al., 2018; Yadollahi and Shamsaei, 2017). Heat treatment and hot isostatic pressing (HIP) have been shown to homogenize the microstructure through recrystallizing the material, relieving residual stress and fusing un-melted particles (Edwards and Ramulu, 2014; Wycisk et al., 2015; Yadollahi and Shamsaei, 2017). Machined or polished surfaces of specimens that have underwent heat treatment or HIP have increased the fatigue performance by mitigating crack initiation due to surface defects (Edwards and Ramulu, 2014; Wycisk et al., 2015).

As an initial attempt at quantifying the fatigue performance of the compliant package design, two prototypes underwent cyclic axial loading at the maximum vertical load as per ISO 14243 and a reduced vertical load determined from the computational model. Visual indications of fatigue failure were compared to high stress locations in the computational model presented in Chapter 2. AM Ti6Al4V has been observed to have a brittle failure mode as a porous material because of its poor notch sensitivity (Ahmadi et al., 2018; Ghouse et al., 2018; Hrabe et al., 2011). Therefore, cracks may initiate and propagate in the high stress areas indicated in the computational model.

4.2 Materials and Methods

4.2.1 Experimental Set-Up

The same prototype that was used in Chapter 3 underwent fatigue testing (Figure 3-1). A similar experimental set-up as in Chapter 3 was used, however a different joint simulator was utilized for fatigue testing. The Boston joint simulator (AMTI, Watertown, MA) was used to apply a sinusoidal, vertical load to the package prototype. The AMTI Boston

operates in 4 DOF. The vertical actuator of the AMTI Boston operates only in force control. The AP, IE and flexion DOFs operate in displacement control. The remaining DOF, AA and ML, are left unconstrained or constrained. The femoral component was cemented, using PMMA, on a femoral component holder designed to interface with the AMTI Boston. Dentstone cemented the tibial tray on a custom fixture positioned on the lower actuator of the test station. Flexion, AP, and IE DOF were set to zero displacement. The ML DOF was left to translate freely along its axis. The flexion angle was set to 0° (Figure 4-1).



Figure 4-1-Experimental Set-up for Fatigue Testing on AMTI Boston: (a) Medial View, (b) Anteromedial View
4.2.2 Loading Scenario at Maximum Vertical Load

The amplitude of the load waveform was based on the maximum and minimum vertical forces in ISO 14243. The minimum compressive load was 265 N, and the maximum compressive load was 2600 N (ISO, 2009). The sinusoidal load was applied at a rate of 2 Hz. Load and displacement data, from the AMTI Boston, were acquired every 10 000 cycles for 2.5 s at a sampling rate of 200 samples/second. The Boston joint simulator was set to perform a total of 5M cycles in 500k cycle intervals. Because of the brittle fatigue failure of AM Ti6Al4V (Ahmadi et al., 2018; Ghouse et al., 2018; Hrabe et al., 2011), the beams on the prototype were assessed for cracks after removal every 500k cycles. However, the prototype was periodically monitored for cracks during testing. Further fatigue testing was terminated if cracks were detected. The loading scenario used is assumed to be an approximation of the "double-humped", gait waveform in ISO 14243 at 1 Hz. Because the vertical force in gait has the largest component, and the energy harvester's electrification mechanism is dependent on the vertical displacement of the package, the shear forces in ISO 14243 were neglected. Furthermore, when the shear forces were considered along with the maximum vertical load in the gait cycle, the shear forces had little effect on the calculated stresses (Figure 4-2). The vertical position of the lower actuator, at the maximum load, was analyzed for changes in the displacement of the package, or for the permanent collapse of the package. The vertical actuator of the AMTI Boston has a resolution of 0.025 mm.



Figure 4-2- von Mises Stress Plots of Maximum Load in Gait Cycle: a) Vertical Load of 2600 N, and b) Vertical Load of 2600 N Superimposed with Corresponding Shear Forces in Percent Gait Cycle (Anteriorly Directed Load of 109 N and Internal Torque of 0.9 Nm).

4.2.3 Reduced Vertical Load

In the computational model of the prototype presented in Chapter 2, the peak stresses were about 800 MPa at the maximum vertical load of 2600 N. Considering that this stress exceeded the fatigue threshold of 550 MPa, an additional prototype was subjected to the same cyclical experiment as the previous prototype but at a reduced load. Additionally, load and displacement data were acquired every 2k cycles. The maximum load was reduced to 1400 N where the peak von Mises stresses did not exceed 500 MPa based on the computational model (Figure 4-3).



Figure 4-3- Stress plot of the package at a reduced compressive load of 1400 N. The peak stress was below 550 MPa. The red arrow denotes the location of the peak stress.

4.3 Results

4.3.1 Maximum Vertical Load

Fatigue testing was terminated after 140 000 cycles. Cracks were visible at the medial and lateral locations of the package (Figure 4-4). The vertical position of the lower actuator is relative to the flexion axis of the AMTI Boston's upper actuator. Therefore, a decrease in the vertical position at the maximum load indicates the lower actuator translated to a more proximal position relative to the upper actuator. The vertical position, at the maximum applied load, of the lower actuator, decreased by 0.1 mm at 30k cycles. This is assumed to be when the cracks developed (Figure 4-5). The gradual decrease prior to 30k cycles may be due to creep deformation of the PE bearing and plastic deformation of the prototype.



Figure 4-4- Locations of cracks on the lateral side of the prototype after fatigue testing was terminated. Cracks were located in the same locations on the medial side.



Figure 4-5- Vertical position of lower actuator at the maximum applied load of 2600 N in 10 000 cycle intervals.

4.3.2 Reduced Vertical Load

Despite the reduction in the vertical load, failure occurred around 70k cycles based on the decrease in the vertical position of the actuator's position (Figure 4-6). Cracks were located at the anteromedial and anterolateral regions where the peak stresses were calculated in the computational model (Figure 4-3).





4.4 Discussion

In this study, the fatigue performance of the prototype was quantified at the maximum vertical load of the gait cycle and at a load below the fatigue threshold of 550 MPa. When the prototype was subjected to the maximum axial load of 2600 N, cracks propagated in the high stress areas of the computational model in Chapter 2 at 30k cycles (Figure 4-7). At the decreased compressive load of 1400 N, a consistent decrease in the actuator position occurred at 70k cycles. Although the fatigue life was extended by decreasing the applied force, the fatigue strength of the prototype appears to be less than 550 MPa. However, there is a lack of knowledge of the fatigue performance of small beam structures similar to what was employed in this package design. The fatigue strength of a

material is determined experimentally and dependent on the parameters related to the applied load. For instance, a beam undergoing reversed bending will have a different fatigue strength as when the same part is subjected to unidirectional bending. The previously assumed fatigue strength of 550 MPa may not be applicable. Although the computational and experimental models lack absolute agreement, the patterns and crack locations are useful to know.



Figure 4-7- Comparison between the Results from Fatigue Testing and the FE

Model

In its solid form, Ti6Al4V has a high fatigue strength thereby making it a popular material for implants. However, there can be a decrease in fatigue strength of Ti6Al4V when additively manufactured. The premature failure of the prototype may be attributed to factors inherent to AM Ti6Al4V. The internal porosity, the presence of unfused material, residual stresses, sensitivity to defects and the microstructure can impact the fatigue performance of AM Ti6Al4V (Ghouse et al., 2018; Sterling et al., 2015; Yadollahi and Shamsaei, 2017).

Parameters in the AM process, such as alloy used and scanning parameters, can be optimized to improve the fatigue performance of the prototype. Ghouse et al. compared the fatigue strength differences between commercially-pure titanium (CP-Ti), Ti6Al4V, tantalum (Ta), and TiTa. Ti6Al4V had the lowest fatigue strength: modulus ratio when compared to the other aforementioned alloys. Ta and TiTa had fatigue strength: modulus ratios 8% greater than CP-Ti, and 19% greater than Ti6Al4V (Ghouse et al., 2018). The laser parameters and scanning strategies can be considered for the fatigue strength optimization of Ti6Al4V. Ghouse et al. observed that a low laser power of 50 W increased the fatigue strength of Ti6Al4V by 7% when compared to samples manufactured using a high laser power of 200 W. A contour scan strategy improved the fatigue strength of Ti6Al4V samples by 8% compared to samples manufactured using points or pulsed scanning strategies (Ghouse et al., 2018).

Post processing of the AM Ti6Al4V can increase the fatigue strength. Post-process heat treatment has proven to increase the fatigue strength of Ti6Al4V by eliminating or minimizing the residual stresses and porosity (Edwards and Ramulu, 2014; Nicoletto et al., 2017; Wycisk et al., 2015). However, the temperature, time, and pressure for both annealing and cooling can greatly affect the fatigue strength of the sample. HIP has also shown to eliminate internal voids, homogenize the microstructure and relieving residual stresses thus improving the fatigue strength of AM Ti6Al4V (Kasperovich and Hausmann, 2015; Leuders et al., 2015). In addition to heat treatment, a smooth surface finish, achieved by machining or polishing, can mitigate the detrimental effects of surface

defects that can be sites for stress concentrations, and crack development (Sterling et al., 2015; Yadollahi and Shamsaei, 2017). By considering the alloy used, laser parameters, scanning strategies, surface finish, and heat treatment, the adverse effects of the inherent anisotropic mechanical properties of as-built SLM Ti6Al4V can be mitigated.

Nicoletto et al. were able to achieve fatigue strengths of AM Ti6Al4V that were comparable to its wrought form. By heat treating the specimen at a temperature of 740°C for more than an hour, followed by a period of vacuum cooling at 530°C for an additional hour and cooling to room temperature in argon, and subsequent surface machining had a fatigue strength of 500 MPa at 10⁷ cycles (Nicoletto et al., 2017). The heat treatment used by Nicoletto et al. vary from what was used in the fabrication of the current prototype. HIP or a heat treatment procedure employed by Nicoletto et al. can improve the fatigue strength of future package designs. However, depending on the complexity of the geometry being additively manufactured, surface machining may not be feasible thus rendering the structure susceptible to crack initiation from surface defects. Surface machining of the current package design may not be feasible with the thin gaps between the beams.

There are discrepancies between the computational and the experimental model, mainly in the geometry. There were defects in the beams that were not present in the computational model. For instance, some of the beams were distorted from the manufacturing process (Figure 4-8e) and layers may have shifted during printing (Figure 4-9). The defects can introduce stress concentrations and alter the load transmission through the elastic bodies. These differences can introduce stresses in the experimental model that may not be accounted for in the computational model.



Figure 4-8- Posterior view of the computational model and a prototype that illustrate the distortion that was present in some of the elastic bodies: a) the area of interest is highlighted in the red circle, b) magnified view of computational model, and c) magnified view of prototype.



Figure 4-9- Example of a shift in the printing layers: a) posteromedial view of computational model, b) magnified view of computational model, and c) magnified view of prototype.

4.5 Conclusion

The current prototype may not have the fatigue strength to withstand high cycle fatigue $(>10^6 \text{ cycles})$. However, this can be attributed to the inherent porosity, surface defects, and unfused material in AM Ti6Al4V. There are several factors that can address the microstructure of the material and AM Ti6Al4V's poor notch sensitivity. The choice of alloy, heat treatment, laser parameters, scanning strategies, and surface finish are variables that can be optimized to improve fatigue strength immensely. In this study, the

high stress areas in the computational models corresponded with where cracks were initiated.

4.6 References

- Ahmadi, S.M., Hedayati, R., Li, Y., Lietaert, K., Tümer, N., Fatemi, A., Rans, C.D., Pouran, B., Weinans, H., Zadpoor, A.A., 2018. Fatigue performance of additively manufactured meta-biomaterials: The effects of topology and material type. Acta Biomater. 65, 292–304. https://doi.org/10.1016/j.actbio.2017.11.014
- Baufeld, B., Brandl, E., Van Der Biest, O., 2011. Wire based additive layer manufacturing: Comparison of microstructure and mechanical properties of Ti-6Al-4V components fabricated by laser-beam deposition and shaped metal deposition. J. Mater. Process. Technol. 211, 1146–1158. https://doi.org/10.1016/j.jmatprotec.2011.01.018
- Brandl, E., Baufeld, B., Leyens, C., Gault, R., 2010. Additive manufactured Ti-6A1-4V using welding wire: Comparison of laser and arc beam deposition and evaluation with respect to aerospace material specifications, in: Physics Procedia. Elsevier B.V., pp. 595–606. https://doi.org/10.1016/j.phpro.2010.08.087
- Chan, K.S., Koike, M., Mason, R.L., Okabe, T., 2013. Fatigue life of titanium alloys fabricated by additive layer manufacturing techniques for dental implants. Metall. Mater. Trans. A Phys. Metall. Mater. Sci. 44, 1010–1022. https://doi.org/10.1007/s11661-012-1470-4
- Edwards, P., Ramulu, M., 2014. Fatigue performance evaluation of selective laser melted Ti-6Al-4V. Mater. Sci. Eng. A 598, 327–337. https://doi.org/10.1016/j.msea.2014.01.041
- Facchini, L., Magalini, E., Robotti, P., Molinari, A., Höges, S., Wissenbach, K., 2010.
 Ductility of a Ti-6Al-4V alloy produced by selective laser melting of prealloyed powders. Rapid Prototyp. J. 16, 450–459.
 https://doi.org/10.1108/13552541011083371
- Ghouse, S., Babu, S., Nai, K., Hooper, P.A., Jeffers, J.R.T., 2018. The influence of laser parameters, scanning strategies and material on the fatigue strength of a stochastic porous structure. Addit. Manuf. 22, 290–301.

https://doi.org/10.1016/J.ADDMA.2018.05.024

- Harrysson, O.L.A., Cansizoglu, O., Marcellin-Little, D.J., Cormier, D.R., West, H.A., 2008. Direct metal fabrication of titanium implants with tailored materials and mechanical properties using electron beam melting technology. Mater. Sci. Eng. C 28, 366–373. https://doi.org/10.1016/j.msec.2007.04.022
- Hrabe, N.W., Heinl, P., Flinn, B., Körner, C., Bordia, R.K., 2011. Compressioncompression fatigue of selective electron beam melted cellular titanium (Ti-6Al-4V). J. Biomed. Mater. Res. Part B Appl. Biomater. 99B, 313–320. https://doi.org/10.1002/jbm.b.31901
- ISO, 2009. ISO 14243-1, Implants for surgery Wear of total knee-joint prostheses.
- Kasperovich, G., Hausmann, J., 2015. Improvement of fatigue resistance and ductility of TiAl6V4 processed by selective laser melting. J. Mater. Process. Technol. 220, 202– 214. https://doi.org/10.1016/j.jmatprotec.2015.01.025
- Koike, M., Greer, P., Owen, K., Lilly, G., Murr, L.E., Gaytan, S.M., Martinez, E., Okabe, T., 2011. Evaluation of Titanium Alloys Fabricated Using Rapid Prototyping Technologies—Electron Beam Melting and Laser Beam Melting. Materials (Basel).
 4, 1776–1792. https://doi.org/10.3390/ma4101776
- Leuders, S., Thöne, M., Riemer, A., Niendorf, T., Tröster, T., Richard, H.A., Maier, H.J., 2013. On the mechanical behaviour of titanium alloy TiAl6V4 manufactured by selective laser melting: Fatigue resistance and crack growth performance. Int. J. Fatigue 48, 300–307. https://doi.org/10.1016/j.ijfatigue.2012.11.011
- Leuders, S., Vollmer, M., Brenne, F., Tröster, T., Niendorf, T., 2015. Fatigue Strength Prediction for Titanium Alloy TiAl6V4 Manufactured by Selective Laser Melting. Metall. Mater. Trans. A Phys. Metall. Mater. Sci. 46, 3816–3823. https://doi.org/10.1007/s11661-015-2864-x
- Murr, L.E., Esquivel, E. V., Quinones, S.A., Gaytan, S.M., Lopez, M.I., Martinez, E.Y., Medina, F., Hernandez, D.H., Martinez, E., Martinez, J.L., Stafford, S.W., Brown,

D.K., Hoppe, T., Meyers, W., Lindhe, U., Wicker, R.B., 2009. Microstructures and mechanical properties of electron beam-rapid manufactured Ti-6Al-4V biomedical prototypes compared to wrought Ti-6Al-4V. Mater. Charact. 60, 96–105. https://doi.org/10.1016/j.matchar.2008.07.006

- Nicoletto, G., Maisano, S., Antolotti, M., Dall'aglio, F., 2017. Influence of post fabrication heat treatments on the fatigue behavior of Ti-6Al-4V produced by selective laser melting, in: Procedia Structural Integrity. Elsevier B.V., pp. 133–140. https://doi.org/10.1016/j.prostr.2017.11.070
- Qiu, C., Adkins, N.J.E., Attallah, M.M., 2013. Microstructure and tensile properties of selectively laser-melted and of HIPed laser-melted Ti-6Al-4V. Mater. Sci. Eng. A 578, 230–239. https://doi.org/10.1016/j.msea.2013.04.099
- Shiomi, M., Osakada, K., Nakamura, K., Yamashita, T., Abe, F., 2004. Residual stress within metallic model made by selective laser melting process. CIRP Ann. - Manuf. Technol. 53, 195–198. https://doi.org/10.1016/S0007-8506(07)60677-5
- Sterling, A., Shamsaei, N., Torries, B., Thompson, S.M., 2015. Fatigue Behaviour of Additively Manufactured Ti-6Al-4 v. Procedia Eng. 133, 576–589. https://doi.org/10.1016/j.proeng.2015.12.632
- Wycisk, E., Siddique, S., Herzog, D., Walther, F., Emmelmann, C., 2015. Fatigue performance of laser additive manufactured Ti–6Al–4V in very high cycle fatigue regime up to 10 9 cycles. Front. Mater. 2. https://doi.org/10.3389/fmats.2015.00072
- Yadollahi, A., Shamsaei, N., 2017. Additive manufacturing of fatigue resistant materials: Challenges and opportunities. Int. J. Fatigue 98, 14–31. https://doi.org/10.1016/j.ijfatigue.2017.01.001

Chapter 5

5 Sensitivity Analysis of Polyethylene Thickness and the Effect of Physiological Loading Assumptions

Overview: In the operating room, the TKR components used are modular in design to ensure correct sizing, and joint stability. With this in mind, the influence of different implant sizes on the mechanical behaviour of the package should be considered. This chapter quantifies the package's sensitivity to PE insert thickness. In addition to considerations behind PE insert thickness, the assumption that the tibiofemoral forces in the gait cycle can be accurately approximated with a vertical compression loading scenario will be examined.

5.1 Introduction

Implant sizing is an important factor that is considered to ensure stability of a TKR procedure. Different PE bearing thicknesses need to be considered to restore the joint line and equalizing the extension and flexion gaps. If the proximal tibia was excessively resected during the TKR procedure, which affects the flexion and extension gaps, a thicker PE bearing can restore the joint line (Abdel and Haas, 2014). Therefore, the package's mechanical behaviour and stress from physiological loading should not be adversely affected by the PE bearing thickness deemed suitable by the surgeon.

In the previous studies in this thesis, it was assumed that the consideration of the vertical load of the gait cycle would be sufficient for analyzing the stresses and mechanical behaviour of the package design. This was assumed because the vertical component contributes the largest proportion of force relative to the other components in the gait cycle. This assumption overlooks the significance of the other components of loading thus overlooking how they affect the stresses in the package. Simulations or experiments that only consider vertical loading may underestimate the stresses an instrumented implant would actually be subjected to under physiological loading.

In this chapter, computational modeling was performed to elucidate the effects PE bearing thickness and multiaxial loading have on the current package design. The effect of the PE bearing thickness on the mechanical behaviour in the package were quantified. In other words, the changes in the stresses and displacements in the package were calculated when the PE bearing thickness was altered during loading. Furthermore, considering that the vertical component is the largest contributor to the resultant force in gait, the viability of assuming a vertical compression loading scenario is representative of a loading scenario that considers all load components was examined.

5.2 Materials and Methods

Finite element analysis was conducted using Abaqus. A defeatured version of the prototype was used for analysis. Simulations with loads based on Orthoload were applied to the model (Bergmann et al., 2014). A rigidly connected reference point was defined inferiorly to the bottom surface of the package. An Encastre boundary condition fully restrained this reference point. The thickness of the lowest points of a 9 mm and 11 mm size 7 Stryker Triathlon Cruciate Retaining (CR) UHMWPE bearing were measured for the purpose of determining the position of the reference point where loads were resolved in the computational model to replicate the coordinate system used in (Kutzner et al., 2010). The Ti6Al4V package had the following material properties $E = 130\ 000\ MPa$ (Renishaw plc, 2017) and v = 0.31 (AZoMaterials, n.d.). The package was meshed with a global mesh size of 1 mm. Tetrahedral elements were used to mesh the part. For all simulations, a self-contact interaction, nonlinear effects of large deformations and displacements were defined.

The Orthoload dataset for AVER75 in the gait cycle was used in the analyses (Bergmann et al., 2014). All six components of force were applied to models simulating a 9 mm and 11 mm thick PE bearing in 1% gait cycle intervals. In two separate models that replicate a 9 mm and 11 mm PE bearing, the maximum vertical load in AVER75 of 1960 N was applied in 0.01 step increments. Stresses and displacements were analyzed to quantify the differences when the shear forces, moments, and the PE insert thickness are considered.

5.3 Results

5.3.1 Axial Force Only versus Six Component Loading

The deflection and stress were different between the two loading scenarios. The six component loading scenario had a larger peak stress than the uniaxial compressive scenario (851 MPa and 733 MPa, respectively). Although, the peak stresses occurred at 48% in the gait cycle, or at the time maximum loads were applied to the package. Throughout the stance phase, the stress was greater in the six component loading than the vertical loading scenario (Figure 5-1).



Figure 5-1- Comparison between uniaxial compressive loading and six component loading. Peak stresses occurred at 48% gait cycle for both loading scenarios. There was a 15% difference between load scenarios where the six component loading scenario was greatest.

When all six components were simulated, the superimposed shear forces created an asymmetrical stress and deflection response. The greatest stress and maximum deflection were located at the anterolateral portion of the package (Figure 5-2). The maximum deflection, under simulated gait, was 0.22 mm (Figure 5-3).



Figure 5-2-Stress Plot of Maximum Six Component Loading: a) anterior, b) isometric, c) medial, d) lateral



Figure 5-3- Vertical Displacement Plot of Maximum Six Component Loading: a) superior, b) isometric, c) anterior, d) medial, e) lateral

Under the loading scenario where only the vertical force was considered, the maximum deflection was 0.23 mm in the anterior aspect of the package (Figure 5-4 and Figure 5-5). Due to the symmetry of the package and the location of where the load was applied, the peak stresses were located in the anteromedial and anterolateral regions of the package.



Figure 5-4- Stress Plot of Maximum Vertical Load: a) anterior, b) lateral, c) posterior, d) isometric



Figure 5-5- Vertical Displacement Plot of Maximum Vertical Fz: a) superior, b) isometric, c) anterior, d) lateral

5.3.2 Effect of Polyethylene Insert Thickness

There was little difference in the stresses and displacements between the models with 9 mm and 11 mm PE thicknesses. In the simulations with only a vertical load applied, the maximum stress was 733 MPa, and the shear displacements of the top plate's bottom surface did not exceed 10 μ m. The largest magnitudes in displacements and stresses were located in the anterior portion of the package.

The mechanical behaviour in the gait cycle simulations were very similar (Figure 5-6). The largest stress and displacements were located in the anterolateral region of the package. The peak stress was 851 MPa. The percent difference between the peak stress in the 9 mm and 11 mm bearing simulations was 0.02%. The shear displacement, along the mediolateral axis, ranged from 3 to 5 μ m directed in the lateral direction where the largest displacement was in the posterior section of the top plate. Along the anteroposterior axis, the shear displacements were greatest in the lateral part of the top plate (about 6 μ m) and decreased towards the medial direction to a minimum of about 2 μ m.





5.4 Discussion

In this chapter, the package design was not sensitive to the effects of changing the PE bearing thickness– there was little difference in the stress and deflection response between the 9 mm and 11 mm thick PE bearings. When considering all force and moment components in gait, the calculated stresses were greater throughout the stance phase compared to when the vertical component was considered. The effects of the shear forces and moments appear to have a significant effect on the stresses in the package design.

The stress distribution and displacements in this chapter are different in the computational model presented in Chapter 2. In Chapter 2, the load was applied to a reference point offset from the sagittal plane to achieve a medial bias and positioned more posteriorly than the one used in this chapter. Additionally, the forces were derived

from the tibial forces presented by Bergmann et al (Bergmann et al., 2014). The computational models simulate the loading scenario used by Bergmann et al. since the design of this package would require the use of a PS or CS TKR system, and the *in vivo* measurements gathered by Bergmann et al. serves as a more accurate representation of the forces transmitted in the knee.

The TKR components were excluded from the simulations in order to simplify the analyses. The main objective of this study was to quantify the mechanical behaviour of the package design under physiological loading. Therefore, the consideration of the contact mechanics of the articulating surfaces, and stresses in the interlocking mechanism were omitted due to the increased complexity these factors would introduce. Future work can investigate how the presence of the package may affect the interfacing TKR components.

When comparing the effect of PE insert thickness on the mechanical behaviour of the package, there was no difference observed. The stresses and displacements of the computational model are not sensitive to a 2 mm difference in thickness. The use of a thicker PE insert, thicker than the ones used in these analyses, may not reproduce the same trend. It can be hypothesized that a thicker PE insert will magnify the effects of the shear forces due to the increased moment arm. However, because the computational model simulates a linear-elastic model, peak stresses in thicker PE bearings are proportional to the change in the PE bearing thickness. Considering the thickness of the current package design, the use of a PE insert thicker than 11 mm is highly unlikely because of the amount of the proximal tibia required to be resected.

Although the vertical component in gait has the largest contribution to the resultant force, the exclusion of the shear forces and moments may not be advisable. The asymmetrical load distribution, due to the shear forces and moments associated with gait, resulted in localized increases in stress and displacements. Dismissing all the load components, aside from the vertical component, can produce results that may not be representative of what would occur in reality.

Future iterations should increase the stiffness of the anterior region of the package. Based on the location of the coordinate system used in (Bergmann et al., 2014), the magnitude of the load transmitted through the anterior region of the package was greater than expected in the design process using the loads from ISO 14243. The increased load thereby increases the stresses and the likelihood of premature failure.

5.5 Conclusion

In this chapter, the sensitivity of the mechanical response of the package was determined between computational models that compared the effect of different PE insert thicknesses and quantified the differences in the stresses and displacements between a simplified simulation where only the vertical component of gait was considered versus a simulation with all six load components in gait were considered. The increase in PE insert thickness did not change the mechanical behaviour of the package. However, this cannot be said when comparing the two loading scenarios. Applying only the vertical force to the package did not result in an accurate approximation of the physiological loading in gait despite the vertical component having the largest magnitude relative to the other components.

5.6 References

- Abdel, M.P., Haas, S.B., 2014. The unstable knee: Wobble and Buckle. Bone Jt. J. 96B, 112–114. https://doi.org/10.1302/0301-620X.96B11.34325
- AZoMaterials, n.d. Properties: Titanium Alloys Ti6Al4V Grade 5 [WWW Document]. URL https://www.azom.com/properties.aspx?ArticleID=1547 (accessed 11.4.19).
- Bergmann, G., Bender, A., Graichen, F., Dymke, J., Rohlmann, A., Trepczynski, A., Heller, M.O., Kutzner, I., 2014. Standardized Loads Acting in Knee Implants. PLoS One 9, 86035. https://doi.org/10.1371/journal.pone.0086035
- Kutzner, I., Heinlein, B., Graichen, F., Bender, A., Rohlmann, A., Halder, A., Beier, A., Bergmann, G., 2010. Loading of the knee joint during activities of daily living measured in vivo in five subjects. https://doi.org/10.1016/j.jbiomech.2010.03.046
- Renishaw plc, 2017. Ti6Al4V ELI-0406 powder for additive manufacturing [WWW Document]. URL https://resources.renishaw.com/en/details/data-sheet-ti6al4v-eli-0406-powder-for-additive-manufacturing--94700

Chapter 6

6 General Discussion and Conclusion

Overview: This chapter summarizes the conclusions made to address the research objectives presented in Chapter 1. The strengths and limitations of the current thesis work are discussed. Future directions in the improvement and development of an energy harvesting load sensor for the use of measuring tibiofemoral forces are outlined. Lastly, the significance of the current work is highlighted.

6.1 Summary

This present work presents a novel method in measuring *in vivo* tibial forces by using a 3D printed package designed to house load sensing and energy harvesting elements in between the PE insert and tibial tray. This eliminates the need to alter existing TKR components. Instead, the package can be designed to be compatible with commercially available TKR components. The stacked beam structure of the elastic bodies placed along the periphery of the package has shown to provide the deflection response required for the TEGs to operate from gait loading.

In Chapter 2, the iterative design process of a package was outlined. The parameters of the stacked beam structures were tuned to provide the desired force-displacement relationship when subjected to loads associated with the gait cycle. The design of the package was also designed to resist shear forces that could damage the embedded TEGs. The package design was also tuned to minimize the stresses when the maximum vertical load in gait was applied.

In Chapter 3, it was noted that the package prototype was more compliant than what was expected based on computational models. However, experiments demonstrated that the package prototype was capable of deflecting accordingly to the cyclic abduction/adduction applied to the package. In other words, the package, in conjunction with embedded load sensors, is capable of measuring coronal imbalances. Interpositional

load sensing devices, such as this package design, can measure the compartmental loads transmitted through a TKR.

Chapter 4 outlines the fatigue performance of the current prototype. Unfortunately, despite the capabilities of manufacturing complex geometries from SLM, the package prototype did not have the durability to prove that it can withstand years of use. The internal porosity and the defect sensitivity of SLM Ti6Al4V are notable reasons for the prototype's poor fatigue strength. However, the fatigue strength of SLM Ti6Al4V can be improved to address the inherent obstacles with SLM Ti6Al4V. HIP has been shown to improve the microstructure and density and machining the surfaces of parts can eliminate surface defects.

In Chapter 5, from the computational models, the package is not sensitive to the PE insert thickness. So, this package design can most likely be used with varying PE insert thicknesses to ensure joint line restoration and equalized extension and flexion gaps. It was also deduced that applying a cyclic sinusoidal vertical load is not an accurate representation of a loading scenario where all six load components are considered. The package sustained higher stresses when all components of loading were considered in the analyses.

6.2 Strengths and Limitations

Additive manufacturing was used to create a prototype with a complex geometry. Despite the ability to manufacture parts that are not feasible with subtractive manufacturing techniques, there were some limitations associated with AM Ti6Al4V. Manufacturing errors were present in the prototypes such as distorted beams and shifts in print layers. AM Ti6Al4V is also notorious for its internal porosity, unfused material, and poor defect sensitivity. This may explain the large stiffness discrepancy between the computational and experimental models and low fatigue life.

The prototypes were subjected to vertical loading. It is expected that the other forces and moments in gait will have an effect on the mechanical behaviour and stress of the

prototype. To address this shortcoming, computational models were developed to determine how a six-component loading differs from a loading scenario where only the vertical force was considered. Also, the vertical load is the largest force component in gait. With this in mind, the TEGs were designed to operate primarily from the vertical deflection resulting from vertical loading.

Gait was the only ADL that was considered in the initial package design. Although other ADL considered, the loads applied to the package correspond to ISO 14243. ISO 14243 is an established standard that is used for the durability testing of knee prostheses.

The triboelectric effect is an energy harvesting mechanism that has not been harnessed for measuring the tibiofemoral forces in knee implants. The TEGs can be used to generate usable electricity and measuring forces. However, there are limitations to this concept. The durability of the TEGs still needs to be investigated. The use of TEGs requires motion to operate optimally, but TKR designs have been developed to eliminate motion. Motion between components and within components can accelerate wear and lead to implant failure. TEGs may have a nonlinear relationship between the applied force and the measured voltage. The TEG's reliance on the frequency of the contact and separation can mean the TEGs are not usable for measuring static loads. Despite the drawbacks of using TEGs in knee implants, there are some aspects that are beneficial. The use of TEGs can be a solution for powering other load sensors which has been the main obstacle behind the development of instrumented implants. If a self-contained, selfpowering load sensing device can be successfully developed, the measurement of tibiofemoral loads will no longer be limited to a laboratory setting.

6.3 Future Directions

The package was designed to be integrated with the TEGs. With the complete triboelectric energy harvester assembly, the energy harvester's ability to generate usable electricity, accurately measure loads, and transmit load data wirelessly still needs to be investigated.

The package can undergo further iterations. The overall thickness of the package can be reduced so that more bone can be preserved in TKR. Other elastic body geometries may yield the desired design responses while decreasing the thickness of the package. Despite the elastic body geometry used, the package requires sealing with perhaps a hyperelastic biocompatible material to isolate the TEGs and the electronics from foreign biological substances.

The fatigue performance of the package was identified to be problematic. Parameters, related to AM Ti6Al4V, such as laser power, scanning strategy, alloy, heat treatment, and surface finish can be optimized to maximize the fatigue strength of the package.

6.4 Significance

This current thesis work presents a novel concept of leveraging the triboelectric effect for measuring tibiofemoral forces. The design process of creating an interpositional, compliant component can be translated to other energy harvesting components and TKR systems. Through experimental testing, the mechanical behaviour and fatigue life of AM Ti6Al4V are better understood in applications that require large deflections. The use of energy harvesting devices may serve as another approach in better understanding *in vivo* knee mechanics. Such devices can be beneficial in monitoring patient activities postoperatively, the refinement of surgical procedures and the improvement of current implant designs.

Appendices

Appendix A- Analytical Model Calculations

| E= | 130,000 | MPa |
|-----------|---------|-----|
| F-tot= | 2600 | Ν |
| delta= | 0.1 | mm |
| Perimeter | 235 | mm |

| s= | 0 mm |
|----|-------------|
| w= | 2 mm |

| | | | | b= | | | | | | | | | | | |
|----|-------------------------|----------|-------------------------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|
| L | I (based on V equation) | F/beam | F_total (8 beam config) | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | 11 | 12 |
| 1 | 7.0922E-05 | 55.31915 | 442.55 | 0.094766 | 0.075216 | 0.065707 | 0.059699 | 0.05542 | 0.052152 | 0.04954 | 0.047383 | 0.045559 | 0.043987 | 0.042611 | 0.041393 |
| 2 | 0.001134752 | 66.38298 | 531.06 | 0.238796 | 0.189533 | 0.165572 | 0.150432 | 0.139649 | 0.131415 | 0.124833 | 0.119398 | 0.114801 | 0.110839 | 0.107373 | 0.104304 |
| 3 | 0.005744681 | 77.44681 | 619.57 | 0.41003 | 0.325441 | 0.284299 | 0.258303 | 0.239787 | 0.225648 | 0.214346 | 0.205015 | 0.197122 | 0.190319 | 0.184368 | 0.179097 |
| 4 | 0.018156028 | 88.51064 | 708.09 | 0.601729 | 0.477592 | 0.417215 | 0.379065 | 0.351893 | 0.331144 | 0.314558 | 0.300864 | 0.289281 | 0.279298 | 0.270564 | 0.262829 |
| 5 | 0.044326241 | 99.57447 | 796.60 | 0.810241 | 0.643088 | 0.56179 | 0.51042 | 0.473832 | 0.445893 | 0.42356 | 0.40512 | 0.389523 | 0.37608 | 0.36432 | 0.353905 |
| 6 | 0.091914894 | 110.6383 | 885.11 | 1.033211 | 0.82006 | 0.716389 | 0.650882 | 0.604225 | 0.568598 | 0.540119 | 0.516606 | 0.496716 | 0.479574 | 0.464577 | 0.451296 |
| 7 | 0.170283688 | 121.7021 | 973.62 | 1.26897 | 1.007182 | 0.879855 | 0.799401 | 0.742098 | 0.698341 | 0.663364 | 0.634485 | 0.610057 | 0.589004 | 0.570585 | 0.554274 |
| 8 | 0.290496454 | 132.766 | 1062.13 | 1.516261 | 1.203457 | 1.051317 | 0.955185 | 0.886715 | 0.834431 | 0.792638 | 0.758131 | 0.728942 | 0.703786 | 0.681778 | 0.662288 |
| 9 | 0.465319149 | 143.8298 | 1150.64 | 1.774097 | 1.408102 | 1.23009 | 1.117611 | 1.037498 | 0.976323 | 0.927423 | 0.887049 | 0.852897 | 0.823463 | 0.797713 | 0.774908 |
| 10 | 0.709219858 | 154.8936 | 1239.15 | 2.041679 | 1.620481 | 1.415621 | 1.286177 | 1.193981 | 1.123579 | 1.067304 | 1.020839 | 0.981537 | 0.947663 | 0.918029 | 0.891785 |
| 11 | 1.038368794 | 165.9574 | 1327.66 | 2.318343 | 1.84007 | 1.607449 | 1.460464 | 1.355775 | 1.275833 | 1.211932 | 1.159171 | 1.114543 | 1.076079 | 1.04243 | 1.012629 |
| 12 | 1.470638298 | 177.0213 | 1416.17 | 2.603529 | 2.066422 | 1.805186 | 1.64012 | 1.522553 | 1.432777 | 1.361015 | 1.301764 | 1.251646 | 1.208451 | 1.170662 | 1.137196 |
| 13 | 2.025602837 | 188.0851 | 1504.68 | 2.896756 | 2.299156 | 2.008498 | 1.824842 | 1.694033 | 1.594146 | 1.514302 | 1.448378 | 1.392615 | 1.344555 | 1.30251 | 1.265275 |
| 14 | 2.724539007 | 199.1489 | 1593.19 | 3.197605 | 2.53794 | 2.217095 | 2.014365 | 1.869971 | 1.75971 | 1.671573 | 1.598802 | 1.537248 | 1.484197 | 1.437785 | 1.396682 |
| 15 | 3.590425532 | 210.2128 | 1681.70 | 3.505708 | 2.782482 | 2.430722 | 2.208458 | 2.05015 | 1.929265 | 1.832637 | 1.752854 | 1.685369 | 1.627205 | 1.576322 | 1.531259 |
| 16 | 4.647943262 | 221.2766 | 1770.21 | 3.820739 | 3.032522 | 2.649152 | 2.406915 | 2.234382 | 2.102634 | 1.997322 | 1.910369 | 1.83682 | 1.77343 | 1.717973 | 1.668861 |
| 17 | 5.923475177 | 232.3404 | 1858.72 | 4.142405 | 3.287829 | 2.872184 | 2.609552 | 2.422493 | 2.279654 | 2.165475 | 2.071203 | 1.991461 | 1.922734 | 1.862609 | 1.809362 |
| 18 | 7.445106383 | 243.4043 | 1947.23 | 4.470445 | 3.548194 | 3.099633 | 2.816204 | 2.614332 | 2.46018 | 2.33696 | 2.235222 | 2.149166 | 2.074997 | 2.01011 | 1.952647 |
| 19 | 9.242624113 | 254.4681 | 2035.74 | 4.804618 | 3.813428 | 3.331336 | 3.02672 | 2.809758 | 2.644083 | 2.511652 | 2.402309 | 2.309819 | 2.230106 | 2.160369 | 2.09861 |
| 20 | 11.34751773 | 265.5319 | 2124.26 | 5.144708 | 4.083357 | 3.567141 | 3.240963 | 3.008643 | 2.831242 | 2.689437 | 2.572354 | 2.473317 | 2.387962 | 2.313289 | 2.247158 |
| 21 | 13.79297872 | 276.5957 | 2212.77 | 5.490515 | 4.357825 | 3.806911 | 3.458808 | 3.210873 | 3.021547 | 2.870211 | 2.745258 | 2.639564 | 2.548471 | 2.468779 | 2.398203 |
| 22 | 16.61390071 | 287.6596 | 2301.28 | 5.841857 | 4.636685 | 4.050518 | 3.68014 | 3.416339 | 3.214898 | 3.053877 | 2.920929 | 2.808472 | 2.71155 | 2.626758 | 2.551666 |
| 23 | 19.84687943 | 298.7234 | 2389.79 | 6.198565 | 4.919804 | 4.297845 | 3.904851 | 3.624943 | 3.411202 | 3.240349 | 3.099282 | 2.979959 | 2.877119 | 2.787149 | 2.707473 |
| 24 | 23.53021277 | 309.7872 | 2478.30 | 6.560481 | 5.207057 | 4.548784 | 4.132844 | 3.836593 | 3.610372 | 3.429544 | 3.280241 | 3.15395 | 3.045106 | 2.949883 | 2.865554 |
| 25 | 27.70390071 | 320.8511 | 2566.81 | 6.92746 | 5.498329 | 4.803233 | 4.364027 | 4.051203 | 3.812328 | 3.621385 | 3.46373 | 3.330376 | 3.215442 | 3.114893 | 3.025847 |
| 26 | 32.40964539 | 331.9149 | 2655.32 | 7.299367 | 5.793511 | 5.061098 | 4.598313 | 4.268695 | 4.016996 | 3.815802 | 3.649683 | 3.509169 | 3.388066 | 3.282119 | 3.188292 |
| 27 | 37.69085106 | 342.9787 | 2743.83 | 7.676072 | 6.092503 | 5.322291 | 4.835623 | 4.488994 | 4.224305 | 4.012728 | 3.838036 | 3.690271 | 3.562917 | 3.451502 | 3.352833 |
| 28 | 43.59262411 | 354.0426 | 2832.34 | 8.057459 | 6.395209 | 5.58673 | 5.075881 | 4.71203 | 4.43419 | 4.212101 | 4.028729 | 3.873622 | 3.739941 | 3.62299 | 3.519419 |
| 29 | 50.16177305 | 365.1064 | 2920.85 | 8.443413 | 6.701541 | 5.854336 | 5.319017 | 4.937738 | 4.646589 | 4.413861 | 4.221707 | 4.05917 | 3.919085 | 3.796533 | 3.688 |
| 30 | 57.44680851 | 376.1702 | 3009.36 | 8.83383 | 7.011416 | 6.125036 | 5.564964 | 5.166055 | 4.861444 | 4.617955 | 4.416915 | 4.246863 | 4.100301 | 3.972082 | 3.858531 |

| м | SigmaB | b= | | | | | | | Î | | | Î | |
|----------|--------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|
| | | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | 11 | 12 |
| 5.531915 | | 3695.887 | 2933.427 | 2562.585 | 2328.263 | 2161.368 | 2033.925 | 1932.054 | 1847.943 | 1776.797 | 1715.479 | 1661.834 | 1614.327 |
| 22.12766 | | 2328.263 | 1847.943 | 1614.327 | 1466.714 | 1361.576 | 1281.292 | 1217.118 | 1164.131 | 1119.312 | 1080.684 | 1046.89 | 1016.962 |
| 49.78723 | | 1776.797 | 1410.245 | 1231.962 | 1119.312 | 1039.077 | 977.8091 | 928.8348 | 888.3985 | 854.1949 | 824.7161 | 798.9267 | 776.0876 |
| 88.51064 | | 1466.714 | 1164.131 | 1016.962 | 923.9717 | 857.7393 | 807.1636 | 766.7362 | 733.3568 | 705.1224 | 680.7882 | 659.4994 | 640.6462 |
| 138.2979 | | 1263.975 | 1003.218 | 876.3916 | 796.2547 | 739.1773 | 695.5925 | 660.7532 | 631.9877 | 607.656 | 586.6854 | 568.3394 | 552.0921 |
| 199.1489 | | 1119.312 | 888.3985 | 776.0876 | 705.1224 | 654.5776 | 615.9811 | 585.1292 | 559.656 | 538.1091 | 519.5386 | 503.2923 | 488.9045 |
| 271.0638 | | 1009.997 | 801.6349 | 700.2926 | 636.2581 | 590.6497 | 555.8226 | 527.9838 | 504.9984 | 485.5558 | 468.7989 | 454.1393 | 441.1567 |
| 354.0426 | | 923.9717 | 733.3568 | 640.6462 | 582.0657 | 540.3419 | 508.4812 | 483.0135 | 461.9858 | 444.1993 | 428.8697 | 415.4586 | 403.5818 |
| 448.0851 | | 854.1949 | 677.9749 | 592.2657 | 538.1091 | 499.5362 | 470.0816 | 446.5372 | 427.0975 | 410.6541 | 396.4822 | 384.0839 | 373.104 |
| 553.1915 | | 796.2547 | 631.9877 | 552.0921 | 501.609 | 465.6525 | 438.1958 | 416.2485 | 398.1273 | 382.7993 | 369.5887 | 358.0314 | 347.7963 |
| 669.3617 | | 747.2344 | 593.0803 | 518.1034 | 470.7282 | 436.9853 | 411.2189 | 390.6227 | 373.6172 | 359.2328 | 346.8355 | 335.9897 | 326.3847 |
| 796.5957 | | 705.1224 | 559.656 | 488.9045 | 444.1993 | 412.3581 | 388.0438 | 368.6083 | 352.5612 | 338.9875 | 327.2888 | 317.0543 | 307.9906 |
| 934.8936 | | 668.482 | 530.5746 | 463.4996 | 421.1173 | 390.9307 | 367.8798 | 349.4543 | 334.241 | 321.3726 | 310.2819 | 300.5792 | 291.9864 |
| 1084.255 | | 636.2581 | 504.9984 | 441.1567 | 400.8175 | 372.086 | 350.1463 | 332.609 | 318.129 | 305.881 | 295.3248 | 286.0898 | 277.9113 |
| 1244.681 | | 607.656 | 482.2969 | 421.3252 | 382.7993 | 355.3594 | 334.406 | 317.657 | 303.828 | 292.1305 | 282.0489 | 273.2291 | 265.4182 |
| 1416.17 | | 582.0657 | 461.9858 | 403.5818 | 366.6784 | 340.3941 | 320.3231 | 304.2795 | 291.0328 | 279.828 | 270.171 | 261.7225 | 254.2406 |
| 1598.723 | | 559.0097 | 443.6863 | 387.5957 | 352.1541 | 326.9109 | 307.6349 | 292.2268 | 279.5049 | 268.7438 | 259.4693 | 251.3555 | 244.17 |
| 1792.34 | | 538.1091 | 427.0975 | 373.104 | 338.9875 | 314.6881 | 296.1328 | 281.3008 | 269.0545 | 258.6959 | 249.7681 | 241.9577 | 235.0408 |
| 1997.021 | | 519.0584 | 411.977 | 359.895 | 326.9863 | 303.5472 | 285.6489 | 271.3419 | 259.5292 | 249.5373 | 240.9256 | 233.3917 | 226.7197 |
| 2212.766 | | 501.609 | 398.1273 | 347.7963 | 315.9939 | 293.3427 | 276.0461 | 262.2201 | 250.8045 | 241.1485 | 232.8263 | 225.5456 | 219.0979 |
| 2439.574 | | 485.5558 | 385.3859 | 336.6656 | 305.881 | 283.9547 | 267.2116 | 253.8281 | 242.7779 | 233.4309 | 225.375 | 218.3274 | 212.086 |
| 2677.447 | | 470.7282 | 373.6172 | 326.3847 | 296.5402 | 275.2835 | 259.0517 | 246.0769 | 235.3641 | 226.3025 | 218.4927 | 211.6602 | 205.6095 |
| 2926.383 | | 456.9831 | 362.7077 | 316.8544 | 287.8813 | 267.2453 | 251.4875 | 238.8915 | 228.4915 | 219.6945 | 212.1127 | 205.4798 | 199.6057 |
| 3186.383 | | 444.1993 | 352.5612 | 307.9906 | 279.828 | 259.7693 | 244.4523 | 232.2087 | 222.0996 | 213.5487 | 206.179 | 199.7317 | 194.0219 |
| 3457.447 | | 432.2735 | 343.0957 | 299.7217 | 272.3153 | 252.7951 | 237.8893 | 225.9744 | 216.1368 | 207.8154 | 200.6436 | 194.3693 | 188.8129 |
| 3739.574 | | 421.1173 | 334.241 | 291.9864 | 265.2873 | 246.2709 | 231.7498 | 220.1424 | 210.5587 | 202.4521 | 195.4653 | 189.353 | 183.9399 |
| 4032.766 | | 410.6541 | 325.9364 | 284.7316 | 258.6959 | 240.152 | 225.9916 | 214.6727 | 205.327 | 197.4219 | 190.6087 | 184.6483 | 179.3697 |
| 4337.021 | | 400.8175 | 318.129 | 277.9113 | 252.4992 | 234.3995 | 220.5783 | 209.5305 | 200.4087 | 192.6929 | 186.043 | 180.2253 | 175.0732 |
| 4652.34 | | 391.5495 | 310.773 | 271.4852 | 246.6607 | 228.9795 | 215.478 | 204.6856 | 195.7747 | 188.2374 | 181.7412 | 176.058 | 171.025 |
| 4978.723 | | 382.7993 | 303.828 | 265.4182 | 241.1485 | 223.8624 | 210.6626 | 200.1114 | 191.3997 | 184.0307 | 177.6797 | 172.1235 | 167.203 |

| Sigma_she | ear | | | | | | | | | | |
|-----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|
| 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | 11 | 12 |
| 116.7485 | 73.54697 | 56.12684 | 46.33169 | 39.92744 | 35.35769 | 31.90456 | 29.18713 | 26.98297 | 25.15271 | 23.60422 | 22.27395 |
| 92.66337 | 58.37427 | 44.5479 | 36.77348 | 31.69043 | 28.06342 | 25.32267 | 23.16584 | 21.4164 | 19.96372 | 18.73468 | 17.67885 |
| 80.94891 | 50.99462 | 38.91618 | 32.1246 | 27.68414 | 24.51566 | 22.12139 | 20.23723 | 18.70895 | 17.43991 | 16.36625 | 15.4439 |
| 73.54697 | 46.33169 | 35.35769 | 29.18713 | 25.15271 | 22.27395 | 20.09861 | 18.38674 | 16.99821 | 15.84521 | 14.86973 | 14.03171 |
| 68.27496 | 43.01053 | 32.82318 | 27.09493 | 23.34971 | 20.67731 | 18.6579 | 17.06874 | 15.77974 | 14.70939 | 13.80383 | 13.02589 |
| 64.2492 | 40.47446 | 30.88779 | 25.49731 | 21.97292 | 19.45809 | 17.55776 | 16.0623 | 14.8493 | 13.84207 | 12.9899 | 12.25783 |
| 61.03123 | 38.44726 | 29.34075 | 24.22026 | 20.87239 | 18.48352 | 16.67836 | 15.25781 | 14.10556 | 13.14878 | 12.33929 | 11.64389 |
| 58.37427 | 36.77348 | 28.06342 | 23.16584 | 19.96372 | 17.67885 | 15.95228 | 14.59357 | 13.49149 | 12.57635 | 11.80211 | 11.13698 |
| 56.12684 | 35.35769 | 26.98297 | 22.27395 | 19.19511 | 16.99821 | 15.33811 | 14.03171 | 12.97206 | 12.09216 | 11.34773 | 10.7082 |
| 54.18987 | 34.13748 | 26.05177 | 21.50526 | 18.53267 | 16.41159 | 14.80878 | 13.54747 | 12.52439 | 11.67485 | 10.95611 | 10.33865 |
| 52.49532 | 33.06998 | 25.23712 | 20.83278 | 17.95315 | 15.89839 | 14.3457 | 13.12383 | 12.13274 | 11.30977 | 10.6135 | 10.01536 |
| 50.99462 | 32.1246 | 24.51566 | 20.23723 | 17.43991 | 15.4439 | 13.9356 | 12.74865 | 11.7859 | 10.98646 | 10.31009 | 9.729045 |
| 49.65203 | 31.27882 | 23.87021 | 19.70442 | 16.98075 | 15.03729 | 13.5687 | 12.41301 | 11.4756 | 10.69721 | 10.03865 | 9.472897 |
| 48.44052 | 30.51561 | 23.28777 | 19.22363 | 16.56642 | 14.67038 | 13.23762 | 12.11013 | 11.19559 | 10.43619 | 9.793704 | 9.241758 |
| 47.33921 | 29.82183 | 22.75832 | 18.78658 | 16.18978 | 14.33684 | 12.93666 | 11.8348 | 10.94106 | 10.19892 | 9.571042 | 9.031645 |
| 46.33169 | 29.18713 | 22.27395 | 18.38674 | 15.84521 | 14.03171 | 12.66133 | 11.58292 | 10.7082 | 9.981859 | 9.367341 | 8.839424 |
| 45.4048 | 28.60323 | 21.82835 | 18.01891 | 15.52822 | 13.751 | 12.40804 | 11.3512 | 10.49398 | 9.782168 | 9.179944 | 8.662588 |
| 44.5479 | 28.06342 | 21.4164 | 17.67885 | 15.23517 | 13.49149 | 12.17387 | 11.13698 | 10.29593 | 9.597555 | 9.006696 | 8.499103 |
| 43.75223 | 27.56218 | 21.03388 | 17.36309 | 14.96305 | 13.25051 | 11.95643 | 10.93806 | 10.11204 | 9.426133 | 8.845827 | 8.347301 |
| 43.01053 | 27.09493 | 20.67731 | 17.06874 | 14.70939 | 13.02589 | 11.75374 | 10.75263 | 9.940612 | 9.266337 | 8.695869 | 8.205794 |
| 42.31669 | 26.65784 | 20.34374 | 16.79339 | 14.4721 | 12.81575 | 11.56413 | 10.57917 | 9.780251 | 9.116854 | 8.555588 | 8.073419 |
| 41.66556 | 26.24766 | 20.03071 | 16.53499 | 14.24942 | 12.61856 | 11.38619 | 10.41639 | 9.629762 | 8.976573 | 8.423943 | 7.949193 |
| 41.05274 | 25.86161 | 19.7361 | 16.29179 | 14.03984 | 12.43296 | 11.21872 | 10.26319 | 9.488127 | 8.844545 | 8.300044 | 7.832276 |
| 40.47446 | 25.49731 | 19.45809 | 16.0623 | 13.84207 | 12.25783 | 11.06069 | 10.11861 | 9.354474 | 8.719957 | 8.183126 | 7.721948 |
| 39.92744 | 25.15271 | 19.19511 | 15.84521 | 13.65499 | 12.09216 | 10.91121 | 9.981859 | 9.228046 | 8.602106 | 8.07253 | 7.617584 |
| 39.40884 | 24.82601 | 18.94579 | 15.63941 | 13.47763 | 11.9351 | 10.76949 | 9.85221 | 9.108188 | 8.490377 | 7.96768 | 7.518644 |
| 38.91618 | 24.51566 | 18.70895 | 15.4439 | 13.30915 | 11.7859 | 10.63485 | 9.729045 | 8.994324 | 8.384236 | 7.868073 | 7.424651 |
| 38.44726 | 24.22026 | 18.48352 | 15.25781 | 13.14878 | 11.64389 | 10.50671 | 9.611816 | 8.885948 | 8.283212 | 7.773268 | 7.335188 |
| 38.00016 | 23.9386 | 18.26857 | 15.08037 | 12.99587 | 11.50848 | 10.38453 | 9.50004 | 8.782613 | 8.186887 | 7.682873 | 7.249888 |
| 37.57316 | 23.66961 | 18.06329 | 14.91092 | 12.84984 | 11.37916 | 10.26784 | 9.393289 | 8.683924 | 8.094891 | 7.596541 | 7.168421 |

| Bending a | nd Shear St | tresses | | | | | | | | | |
|-----------|-------------|----------|----------|----------|----------|----------|----------|----------|----------|----------|----------|
| 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | 11 | 12 |
| 3812.635 | 3006.974 | 2618.712 | 2374.594 | 2201.295 | 2069.283 | 1963.959 | 1877.13 | 1803.78 | 1740.631 | 1685.439 | 1636.601 |
| 2420.926 | 1906.318 | 1658.875 | 1503.487 | 1393.267 | 1309.356 | 1242.441 | 1187.297 | 1140.728 | 1100.648 | 1065.625 | 1034.641 |
| 1857.746 | 1461.239 | 1270.878 | 1151.437 | 1066.761 | 1002.325 | 950.9562 | 908.6357 | 872.9039 | 842.156 | 815.2929 | 791.5315 |
| 1540.261 | 1210.463 | 1052.32 | 953.1588 | 882.892 | 829.4376 | 786.8348 | 751.7436 | 722.1206 | 696.6334 | 674.3692 | 654.6779 |
| 1332.25 | 1046.229 | 909.2148 | 823.3496 | 762.5271 | 716.2698 | 679.4111 | 649.0565 | 623.4358 | 601.3948 | 582.1432 | 565.118 |
| 1183.561 | 928.873 | 806.9754 | 730.6197 | 676.5505 | 635.4392 | 602.687 | 575.7183 | 552.9584 | 533.3807 | 516.2822 | 501.1624 |
| 1071.028 | 840.0822 | 729.6334 | 660.4783 | 611.522 | 574.3061 | 544.6622 | 520.2562 | 499.6613 | 481.9477 | 466.4786 | 452.8006 |
| 982.3459 | 770.1303 | 668.7096 | 605.2315 | 560.3056 | 526.1601 | 498.9658 | 476.5794 | 457.6907 | 441.446 | 427.2607 | 414.7188 |
| 910.3218 | 713.3326 | 619.2486 | 560.383 | 518.7313 | 487.0798 | 461.8753 | 441.1292 | 423.6261 | 408.5743 | 395.4316 | 383.8122 |
| 850.4445 | 666.1252 | 578.1439 | 523.1143 | 484.1852 | 454.6074 | 431.0572 | 411.6748 | 395.3237 | 381.2635 | 368.9875 | 358.1349 |
| 799.7297 | 626.1503 | 543.3405 | 491.561 | 454.9385 | 427.1173 | 404.9684 | 386.741 | 371.3656 | 358.1453 | 346.6032 | 336.4 |
| 756.117 | 591.7806 | 513.4202 | 464.4365 | 429.798 | 403.4877 | 382.5439 | 365.3098 | 350.7734 | 338.2753 | 327.3644 | 317.7196 |
| 718.1341 | 561.8534 | 487.3698 | 440.8217 | 407.9114 | 382.9171 | 363.023 | 346.654 | 332.8482 | 320.9791 | 310.6178 | 301.4593 |
| 684.6986 | 535.514 | 464.4445 | 420.0411 | 388.6524 | 364.8167 | 345.8466 | 330.2392 | 317.0766 | 305.761 | 295.8835 | 287.1531 |
| 654.9952 | 512.1187 | 444.0835 | 401.5859 | 371.5492 | 348.7428 | 330.5937 | 315.6628 | 303.0716 | 292.2479 | 282.8001 | 274.4499 |
| 628.3974 | 491.173 | 425.8558 | 385.0651 | 356.2393 | 334.3548 | 316.9408 | 302.6158 | 290.5362 | 280.1528 | 271.0899 | 263.08 |
| 604.4145 | 472.2896 | 409.4241 | 370.173 | 342.4391 | 321.3859 | 304.6348 | 290.8561 | 279.2378 | 269.2515 | 260.5355 | 252.8326 |
| 582.657 | 455.1609 | 394.5204 | 356.6663 | 329.9233 | 309.6243 | 293.4747 | 280.1915 | 268.9918 | 259.3657 | 250.9644 | 243.5399 |
| 562.8107 | 439.5391 | 380.9289 | 344.3494 | 318.5103 | 298.8994 | 283.2984 | 270.4673 | 259.6493 | 250.3517 | 242.2375 | 235.067 |
| 544.6195 | 425.2223 | 368.4736 | 333.0626 | 308.0521 | 289.072 | 273.9738 | 261.5571 | 251.0891 | 242.0926 | 234.2415 | 227.3037 |
| 527.8725 | 412.0437 | 357.0093 | 322.6744 | 298.4268 | 280.0274 | 265.3923 | 253.3571 | 243.2111 | 234.4919 | 226.883 | 220.1594 |
| 512.3937 | 399.8649 | 346.4154 | 313.0752 | 289.5329 | 271.6703 | 257.4631 | 245.7805 | 235.9323 | 227.4692 | 220.0842 | 213.5587 |
| 498.0358 | 388.5693 | 336.5905 | 304.1731 | 281.2852 | 263.9204 | 250.1103 | 238.7547 | 229.1827 | 220.9573 | 213.7799 | 207.438 |
| 484.6737 | 378.0585 | 327.4486 | 295.8903 | 273.6114 | 256.7101 | 243.2694 | 232.2182 | 222.9032 | 214.899 | 207.9148 | 201.7438 |
| 472.201 | 368.2484 | 318.9168 | 288.1605 | 266.4501 | 249.9815 | 236.8856 | 226.1186 | 217.0435 | 209.2457 | 202.4419 | 196.4304 |
| 460.5261 | 359.067 | 310.9322 | 280.9267 | 259.7485 | 243.6849 | 230.9119 | 220.4109 | 211.5603 | 203.9557 | 197.3207 | 191.4586 |
| 449.5703 | 350.452 | 303.4406 | 274.1398 | 253.4611 | 237.7775 | 225.3075 | 215.0561 | 206.4162 | 198.993 | 192.5163 | 186.7943 |
| 439.2647 | 342.3493 | 296.3948 | 267.757 | 247.5483 | 232.2222 | 220.0372 | 210.0205 | 201.5789 | 194.3262 | 187.9986 | 182.4083 |
| 429.5496 | 334.7116 | 289.7538 | 261.7411 | 241.9754 | 226.9865 | 215.0701 | 205.2748 | 197.02 | 189.9281 | 183.7409 | 178.2749 |
| 420.3725 | 327.4976 | 283.4815 | 256.0594 | 236.7122 | 222.0417 | 210.3792 | 200.7929 | 192.7146 | 185.7746 | 179.7201 | 174.3714 |

Appendix B- Simplified Design Iterations using OFAT

| Factors | min | inc | max |
|---------|------|------|-----|
| h | 0.5 | 0.5 | 2 |
| L | 10 | 2 | 26 |
| r | 0.25 | 0.25 | 0.5 |

| Responses | Target Value |
|----------------|--------------|
| sigma_vm [Mpa] | 550 |
| d [mm] | 0.2 |

| | Factors | | | Repsonses | | |
|----------|---------|----|-------|-----------|---------|----------------------------|
| Test Run | h | L | r | sigma_vm | d | Notes |
| 1 | 1 | 6 | 0.25 | 1113 | 0.1948 | |
| 2 | 1 | 8 | 0.25 | 1774 | 0.4918 | |
| 3 | 1 | 10 | 0.25 | 2200 | 1.001 | beams come into contact; |
| 4 | 1 | 12 | 0.25 | | | |
| 5 | 1 | 14 | 0.25 | | | |
| 6 | 1 | 16 | 0.25 | | | |
| 7 | 1 | 18 | 0.25 | | | |
| 8 | 1 | 20 | 0.25 | | | |
| 9 | 1 | 22 | 0.25 | | | |
| 10 | 1 | 24 | 0.25 | | | |
| 11 | 1 | 26 | 0.25 | | | |
| 12 | 1.5 | 6 | 0.25 | 632.4 | 0.079 | |
| 13 | 1.5 | 8 | 0.25 | 984.4 | 0.1867 | |
| 14 | 1.5 | 10 | 0.25 | 1414 | 0.38 | |
| 15 | 1.5 | 12 | 0.25 | 1896 | 0.6879 | |
| 16 | 1.5 | 14 | 0.25 | | | |
| 17 | 1.5 | 16 | 0.25 | | | |
| 18 | 1.5 | 18 | 0.25 | | | |
| 19 | 1.5 | 20 | 0.25 | | | |
| 20 | 1.5 | 22 | 0.25 | | | |
| | | | | | | |
| 21 | 4 5 | 24 | 0.25 | 1.025.00 | 60.2F | beams come into contact; |
| 21 | 1.5 | 24 | 0.25 | 1.02E+06 | 00.25 | noninear problem |
| | | | | | | hooms como into contact: |
| 22 | 1 5 | 26 | 0.25 | 7 725+05 | 202 | pealins come into contact, |
| 22 | 1.5 | 20 | 0.25 | 7.72E+05 | 203 | noniniear problem |
| | | | | | | herause of stress |
| 23 | 2 | 6 | 0.25 | 1 00F+02 | 0.04426 | singuarity |
| | 2 | 0 | 0.25 | 4.002102 | 0.04420 | nrohed stresses at hends |
| | | | | | | hecause of stress |
| 24 | 2 | 8 | 0.25 | 6 08F+02 | 0 09934 | singuarity |
| 25 | 2 | 10 | 0.25 | 815 | 0 196 | Singulary |
| 26 | 2 | 12 | 0.25 | 594 | 0.165 | |
| | | | 0.120 | | 0.100 | |
| | | | | | | beams come into contact: |
| 27 | 2 | 14 | 0.25 | 27060 | 0.3095 | nonlinear problem |
| | | | | | | |
| | | | | | | beams come into contact: |
| 28 | 2 | 16 | 0.25 | 347200 | 0.3115 | nonlinear problem |
| 29 | 2 | 18 | 0.25 | | | • |
| 30 | 2 | 20 | 0.25 | | | |

| 31 | 2 | 22 | 0.25 | | | |
|----------|-----|----|------|----------|---------|------------------------------|
| 32 | 2 | 24 | 0.25 | | | |
| 33 | 2 | 26 | 0.25 | | | |
| 34 | 1 | 6 | 0.5 | 839.2 | 0.1744 | |
| 35 | 1 | 8 | 0.5 | 1358 | 0.4497 | |
| 36 | 1 | 10 | 0.5 | 1999 | 0.97 | |
| | | | | | | |
| | | | | | | contact; mesh adaptation |
| 37 | 1 | 12 | 0.5 | 1591 | 1.004 | failed at 2nd iteration |
| | | | | | | |
| | | | | | | beams come into contact; |
| 38 | 1 | 14 | 0.5 | 3.13E+06 | 41.69 | nonlinear problem |
| 39 | 1 | 16 | 0.5 | | | |
| 40 | 1 | 18 | 0.5 | | | |
| 41 | 1 | 20 | 0.5 | | | |
| 42 | 1 | 22 | 0.5 | | | |
| 43 | 1 | 24 | 0.5 | | | |
| 44 | 1 | 26 | 0.5 | | | |
| | | | | | | Stress singularity at fillet |
| | | | | | | locations; probed |
| 45 | 1.5 | 6 | 0.5 | 388.9 | 0.07167 | stresses |
| | | | | | | Stress singularity at fillet |
| | | | | | | locations; probed |
| 46 | 1.5 | 8 | 0.5 | 631.9 | 0.1722 | stresses |
| | | | | | | stress singularity at fillet |
| 47 | 1.5 | 10 | 0.5 | 1919 | 0.3547 | locations |
| | | | | | | stress singularity at fillet |
| | | | | | | locations; probed stress |
| 48 | 1.5 | 12 | 0.5 | 1300 | 0.6478 | result |
| 49 | 1.5 | 14 | 0.5 | | | |
| 50 | 1.5 | 16 | 0.5 | | | |
| 51 | 1.5 | 18 | 0.5 | | | |
| 52 | 1.5 | 20 | 0.5 | | | |
| 53 | 1.5 | 22 | 0.5 | | | |
| 54 | 1.5 | 24 | 0.5 | | | |
| 55 | 1.5 | 26 | 0.5 | | | |
| | | _ | 0.5 | | | not possible with this h |
| 56 | 2 | 6 | 0.5 | | | and H value |
| 5/ | 2 | 8 | 0.5 | | | |
| 58 | 2 | 10 | 0.5 | | | |
| 59 | 2 | 12 | 0.5 | | | |
| 60 | 2 | 14 | 0.5 | | | |
| 10 | 2 | 10 | 0.5 | | | |
| 62 62 | 2 | 18 | 0.5 | | | |
| 60 | 2 | 20 | 0.5 | | | |
| 65 | 2 | 22 | 0.5 | | | |
| 33 | 2 | 24 | 0.5 | | | |
| 00 | | 20 | 0.0 | | | |
| Whittled | Results | | | | | |
|----------|---------|----|------|----------|---------|---|
| | Factors | | | Repsonse | S | |
| Test Run | h | L | r | sigma_vm | d | Notes |
| 1 | 1 | 6 | 0.25 | 1113 | 0.1948 | |
| 2 | 1 | 8 | 0.25 | 1774 | 0.4918 | |
| 3 | 1 | 10 | 0.25 | 2200 | 1.001 | beams come into contact; nonlinear problem; probed stresses at bends because of stress singuarity |
| 4 | 1.5 | 6 | 0.25 | 632.4 | 0.079 | |
| 5 | 1.5 | 8 | 0.25 | 984.4 | 0.1867 | |
| 6 | 1.5 | 10 | 0.25 | 1414 | 0.38 | |
| 7 | 1.5 | 12 | 0.25 | 1896 | 0.6879 | |
| 8 | 2 | 6 | 0.25 | 400 | 0.04426 | probed stresses at bends because of stress singuarity |
| 9 | 2 | 8 | 0.25 | 608.1 | 0.09934 | probed stresses at bends because of stress singuarity |
| 10 | 2 | 10 | 0.25 | 815 | 0.196 | |
| 11 | 2 | 12 | 0.25 | 594 | 0.165 | |
| 12 | 1 | 6 | 0.5 | 839.2 | 0.1744 | |
| 13 | 1 | 8 | 0.5 | 1358 | 0.4497 | |
| 14 | 1 | 10 | 0.5 | 1999 | 0.97 | |
| 15 | 1.5 | 6 | 0.5 | 388.9 | 0.07167 | |
| 16 | 1.5 | 8 | 0.5 | 631.9 | 0.1722 | |
| 17 | 1.5 | 10 | 0.5 | 1919 | 0.3547 | |
| 18 | 1.5 | 12 | 0.5 | 1300 | 0.6478 | |

| Adjusting | L (H=6mm) | | | | | | |
|-----------|-----------|-----|------|------|------|----------|------|
| | Factors | | | | | Repsonse | s |
| Test Run | h_s | h_l | L | r_s | r_1 | sigma_vm | F_R |
| | | | | | | | |
| 1 | 0.5 | 4 | 8 | 0.25 | 0.75 | 1128 | 1200 |
| 2 | 0.75 | 3.5 | 8 | 0.5 | 0.75 | 956.2 | 1293 |
| 3 | 0.75 | 3.5 | 8.25 | 0.5 | 0.75 | 741 | 1205 |
| 3 | 0.75 | 3.5 | 8.5 | 0.5 | 0.75 | 733.7 | 1116 |
| | 0.75 | 2 5 | 0 75 | 0.5 | 0.75 | 605.7 | 1025 |
| 4 | 0.75 | 3.5 | 8.75 | 0.5 | 0.75 | 695.7 | 1035 |

Appendix C: Axial Loading of Rectangular Models

Methods

Computational Model

Rectangular models with S-shaped beams were created using CAD software (Solidworks). S-shaped beams, positioned along the edges of the model, provide deflection in the entire model through bending. Three configurations with identical beam geometries were tested. However, the three models differ in thickness and in the number of beams along the periphery. Two of the configurations consist of having the S-beams located along two opposing sides, but the thickness of the beams vary where one configuration had a beam thickness of 2 mm and the other configuration had a beam thickness of 3mm. These parts will be referred to as D2 and D3, respectively. The third configuration has the same beam geometry as D3, but the beams are located on all four sides of the part. This configuration will be referred to as Q3 hereinafter. The three rectangular models are presented in the table below.

| | Side View | Isometric View |
|---|--------------------------|--------------------|
| 2 mm Thick Beams Along Two Opposing Sides (D2) | | |
| 3 mm Thick Beams Along | | |
| Two Opposing Sides (D3) | | |
| 3 mm Thick Beams Along | | |
| | | |
| Beam Dimensions | [mm] | |
| | | |
| | | |
| | | |
| a = | 80.62°, b = 1.25. c = 2. | .55, d = 15, e = 3 |

Rectangular models with S-beam design.

These models were used to develop computational models that calculated the stiffness of the parts during axial compression. Solidworks Simulation was used to perform static analyses on the parts. A fixed restraint was applied to one of the corners on the bottom surface of the part and a roller/slider was applied to the bottom surface. A normal force applied to the top surface of the part. The default tetrahedral element mesh with a size of 1.6 mm was used. Local mesh refinement was achieved by using the h-adaptive mesh refinement where target accuracy was set to 99% and was set to run five iterations. The models accounted for large displacements so that changes in the stiffness would be updated as the model deformed during the analyses. Because the stiffness varies between the different configurations, the maximum applied loads were 600 N, 1200 N and 2400 N for the thin beam, thick beam and eight beam configurations, respectively. These loads correspond to the load that resulted in von Mises stresses equal to the yield strength of Ti6Al4V.



Locations of boundary conditions and load application: (a) a fixed restraint was defined to a corner on the bottom surface of the part, (b) a roller/slider was defined to the bottom surface and (c) the compressive load was applied to the top surface of the rectangular models.

Experimental Model

Experimental testing was performed on 3D printed versions of the CAD generated parts. The 3D printing parameters were identical to what was used on the package prototype (Table 3 1). These 3D printed parts were compressed using a TTD Series mechanical testing machine (Adelaide Testing Machine (ATM), Inc., Toronto, ON). Compression plates were secured to the machine's fixture adapters. The part being tested was placed in the centre of the bottom compression plate. Once in position, by manually controlling the position of the crosshead using the dial responsible for actuation, the crosshead was lowered at a loading rate of 6 mm/min until contact was achieved. The initial force and crosshead position were recorded. Again, by manually controlling the crosshead position,

the force and position were recorded in approximately 100 N increments. As in the computational models, the maximum applied loads were 600 N, 1200 N and 2400 N for the thin beam, thick beam and eight beam configurations, respectively. The experimental stiffness of the three configurations, calculated from the measured loads and crosshead displacements, were compared to the computational models' stiffness.

Results

Axial Loading of Rectangular Models

The stiffness in the 3D printed rectangular models and their respective computational models had a similar trend as what was observed with the prototype. The results are summarized in the table below. D2, D3 and Q3 had stiffness values 61%, 59% and 67% less than what was calculated in the computational models, respectively.

| | k [N/mn | k [N/mm] | |
|---------------|---------|----------|------|
| | D2 | D3 | Q3 |
| SW Simulation | 2222 | 3243 | 6154 |
| ATM | 863 | 1328 | 2027 |
| Percent Error | 61 | 59 | 67 |

Tabulated stiffness from computational and experimental models. The manufactured parts have shown to be more compliant than what was determined in the computational models.

Discussion

In addition, 3D printed rectangular models were subjected to axial loading. The axial compression testing of additional rectangular models yielded similar results as the prototype— the stiffness of the as-built rectangular parts were about 60% less than the computational model's stiffness. Overall, from the axial testing performed on these 3D

printed specimens, the compliance of as-built 3D printed parts were consistently lower than what was expected based on the computational models.

Appendix D: Stiffness Calculation of Single Silicone Spacer

w = 15 mm, L = 6 mm, h = 18 mm

 $A = 15 \text{ x } 18 = 270 \text{ mm}^2$

Shore A Hardness = 25

 $E = 0.949 \text{ N/mm}^2$

 $\mathbf{k}_{silicone} = EA/L = (0.949)(270)/6 = 42.7 \text{ N/mm}$

Appendix E: AP Translation and IE Rotation Testing

Methods

The addition of an internal-external (IE) rotation of $\pm 10^{\circ}$, and an anteroposterior (AP) translation of ± 5 mm were tested in separate subsequent loading scenarios.

Results

The Flexiforce sensors measured changes in the load distribution when a cyclic AP translation was applied. Loads reached their highest magnitude as the tibia translated anteriorly with respect to the femur. As the tibia translated posteriorly, the measured compartment loads decreased. During IE rotation testing, the measured medial compartmental load had a larger range than the lateral compartment. The medial condyle was more mobile along the AP axis; the joint seemed to exhibit a lateral pivot.



Flexiforce measurements during anteroposterior translation.



Flexiforce measurements during cyclic internal/external rotation.

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Appendix G- Mechanical Testing Machine Technical Data Sheets

Instron 8874 Mechanical Testing Machine



| FRAME SPECIFICATIONS | | | 1 |
|--|-------|--|--|
| Daylight Opening (Maximum Between Load Cell and Actuator at | mm | 1001 | |
| Mid-stroke, with Largest Capacity Actuator) | IN IN | 30,41 | |
| Dynamic Load Capacity | lbf | +5820 | |
| | Nm | 100 | |
| Torque Load Capacity | inlb | 880 | |
| Actuator Stroke | mm | 100 | |
| (Total) | in | 4 | |
| Actuator Rotation | kN | ±130° | |
| Configuration | | Twin-Column High-Stiffness Load Frame with Actuator in Upper Crosshead and T-Slot Base | |
| Lift and Locks | | Hydraulically-Powered Lifts and Manual Locks | - 2025 |
| Load Cell | | Petented: Dynacell TM Fatigue-RatedLoad Cell with Capacity to Suit Actuator | |
| Load Weighing Accuracy | | ±0.002% of Load Cell Capacity or 0.5% of Indicated Load, Whichever is Greater - Down to 1/250th of Full Scale | я Н |
| Hydraulic Pressure Supply | bar | 207 | WWX |
| (Required) | psi | 3000 | 8 |
| Electrical Supply | | Single-Phase Mains 80-132 or 180-284 V 45/85 Hz with Power Consumption 800 VA Max | 455 |
| Operating Environment | | +10 to +38°C (+50 to +100°F) with 10 to 80% Humidity Non-Condensing | HI: 202 |
| Frame Stiffness | kN/mm | 260 | |
| | kg | 287 | |
| Frame Weight | lb | 834 | + |
| MECHANICAL ACCESSO | ORIES | | |
| Load Cell | | 8 × M8 on 75 PCD | |
| Actuator | | 6 × M8 on 75 mm PCD 6 × 9 mm Diameter Through Holes on 75 mm PCD | |
| Table and Crosshead | | 4 × M10 Holee on a 280 mm × 90 mm for Accessory Mounting 6 × M10 x 20 Deep on 100 mm PCD (Table) | |
| | | with 40 mm Location Diameter 4 × M10 T-Slota Running Front to Back, Spaced 80 and 100 mm from Centerline | |
| Table and Crosshead | | 4 x M10 Holes on a 280 mm x 80 mm for Accessory Mounting 8 x M10 x 20 deep pn 100 mm PCD (Table) with 40 mm Location Diameter | 667 |
| ACCESSORIES | | | Instron®8874 Dimensions (All Dimensions in mm) |
| 8260C | | ±25 kN / ±100 Nm Fatigue Rated | - |
| 1) US Patent Number 6508132 | | Hyangulic Wedge Gripe | |
| | | | www.instro |
| | | | |

Adapted from: https://www.instron.us/-/media/literature-library/products/2012/10/8874-servohydraulic-fatigue-testing-system.pdf?la=en

Instron Load Cells



| Catalog Number | Force | Capacity | Tor Cap | que acity | Mechanical Fitting (Frame) | Mechanical Fitting (Load String) | Effe Ler | ctive ngth | Diar | neter | We | ight |
|-------------------|-------|----------|------------|--------------|--|---|-------------|---------------|------|-------|--------|--------|
| | kN | lbf | Nm | in-lb | | | mm | in | mm | in | kg | lb |
| 2527-131 | ±0.25 | 56 | - | - | Sx M6 on 57 mm PCD | Central M6 x 1 | 42 | 1.65 | 75 | 2.95 | 1 | 2.2 |
| 2527-130 | ±1 | 225 | - | - | 3x M6 on 57 mm PCD | Central M6 x 1 | 42 | 1.65 | 75 | 2.95 | 1 | 2.2 |
| 2527-302 | ±1 | 225 | ±25 | ±225 | 3x M6 on 75 mm PCD | 3x M6 on 75 mm PCD or 3x M6 on 57 mm PCD | 68 | 2.68 | 94 | 3.7 | 1.3 | 2.87 |
| 2527-129 | ±2 | 450 | - | - | 3x M8 on 57 mm PCD | Central M6 x 1 | 42 | 1.65 | 75 | 2.95 | 1 | 2.2 |
| 2527-153 | ±5 | 1,125 | - | - | 3x M6 on 75 mm PCD | 3x M6 on 75 mm PCD or 3x M6 on 57 mm PCD | 68 | 2.68 | 94 | 3.7 | 2 | 4.4 |
| 2527-303 | ±5 | 1,125 | ±25 | ±225 | 3x M6 on 75 mm PCD | 3x M8 on 75 mm PCD or 3x M8 on 57 mm PCD | 68 | 2.68 | 94 | 3.7 | 2 | 4.4 |
| 2527-102 | ±10 | 2,250 | - | - | 6x M6 on 75 mm PCD | Central M20x1.5 | 71 | 2.8 | 107 | 4.2 | 4 | 8.8 |
| 2527-202 | ±10 | 2,250 | ±100 | ±880 | 6x M8 on 75 mm PCD | 6x M8 on 75 mm PCD | 86 | 3.38 | 94 | 3.7 | 2.5 | 5.5 |
| 2527-101 | ±25 | 5,620 | - | - | Central M20x1.5 or 6 x M10 on 75 mm PCD | Central M20x1.5 | 71 | 2.8 | 107 | 4.2 | 4 | 8.8 |
| 2527-201 | ±25 | 5,620 | ±100 | ±880 | 6x M8 on 75 mm PCD | 6x M8 on 75 mm PCD | 86 | 3.38 | 94 | 3.7 | 2.5 | 5.5 |
| 2527-100 | ±50 | 11,250 | - | - | Central M30x2 or 6x M10 on 100 mm PCD | Central MS0x2 | 99 | 3.9 | 154 | 6.06 | 10 | 22 |
| 2527-111 | ±100 | 22,500 | - | - | Central M30x2 or 6x M10 on 100 mm PCD | Central M30x2 | 99 | 3.9 | 154 | 6.06 | 10 | 22 |
| 2527-115 | ±250 | 56,250 | - | - | Central M48x2 | Central M48x2 | 131 | 5.15 | 203 | 8 | 25.5 | 56.2 |
| 2527-125 | ±500 | 112,500 | - | - | Central M72 | Central M72x2 6x M20 on 150 mm PCD 6x M30 on 225 mm PCD | 232 | 9.13 | 305 | 12 | 95 | 209.4 |
| 2527-120 | ±1000 | 225,000 | - | - | Central M100 | Central M100 6x M20 on 150 mm PCD 12x M30 on 225 mm PCD | 360 | 12.17 | 305 | 12 | 140 | 308.0 |
| 2527-140 | ±2500 | 562,500 | - | - | Central M150 | Central M150 | 450 | 17.71 | 428 | 16.85 | 322 | 710 |
| | | | | | | | | | | ww | w.inst | ron.co |

(Adapted from: https://www.instron.us/-/media/literaturelibrary/products/2005/06/dynacell-fatigue-rated-load-cells.pdf?la=enus&hash=35B791E3BD0C0B1AA5142F5E4408B4066304A5C6)

8800 MiniTower Control Electronics



8800 MiniTower Control Electronics

The 8800 controller is a class-leading, fully digital dynamic controller that utilizes Instrome core technologies and is capable of running static and high-frequency dynamic tests. Found at the heart of Instrom's servohydraulic testing systems, the 8800 controller provides full system control, machine safety, transducer conditioning, and data acquisition, as well as acts as the foundation for the user interface to the testing machine.

Features

- Dedicated materials testing hardware and firmware-based controller developed through decades of research, development, and continuous use
- Continuous synchronous data acquisition and loop closure rates of 5 kHz
- Up to 24-bit data resolution across the entire span of each transducer provides maximum data quality
- Automatic identification and calibration of all compatible transducers prevents configuration errors and simplifies setup
- Specimen Protect function helps to avoid damage of specimen and fixtures during test setup and end of test
- Continuous update of PID control terms with Adaptive Control - optimizes the control parameters throughout a test to suit the changing stiffness characteristics of the specimen
- Expandable architecture; extensive analog and digital channel capabilities

Console Software





Handset and Frame Controls

The handset, frame controls, and emergency stop button make up the hardware interface that is rigidly mounted to the testing machine. Their functionality includes switching the machine into low power or high power mode; offering fast or fine positioning of the actuator; and where fitted, opening and closing of hydraulic grips. Uniquely, the 8800MT offers additional protection by locking out the actuator and grip controls when a waveform is running, or when in load/strain control.



Console Software is the main user interface to the 8800 controller. Running on a PC, it allows all controller functions to be viewed and configured including control-loop optimization, setting of operational limits, and running of simple cyclic tests. Console provides the foundation for running more demanding tests in application software such as WaveMatrix*, Bluehill®, or specific software, such as the Low Cycle Fatigue or Fracture Mechanics suite.



| Aves of control | | | |
|--|---|--|------------------------------------|
| All a control | • | 1-2 | |
| Sensor conditioning channels | • | Up to 8 | |
| Channels as Standard | • | Position and Load Strain 1 and Strain 2 or any other compatible sensors | 10 |
| apare channel slocio | | Type PID (Proportional, Integral, Derivative), Lag, Feed Forward (2 Term), | |
| Control Loop Type | • | Notch (4 Term) and External Compensation Input (e.g. Acceleration or Pressure Feedback) | |
| Control Loop Update Rate | • | 5 kHz | |
| Auto Loop Shaping Adentive Loop Shaping | • | Position, Load, and Strain | Ö |
| Low Power "Specimen Loading Mode Feature | e" . | Meximum Actuetor Velocity Limited by Control System | 479 |
| External Inputs and Outputs | _ | | |
| Analog Input | - | 1 off Per Axis, +/-10V Scalable | |
| Analog Outputs | | 4 off Per Axis, +/-10V with 20% Over-Range, Zero Suppressed and Scalable Selectable from Eachbork Science Demond Series of | 00 |
| Bidial Inc. do | | 4 off, Programmable, Low Level Opto Isolated | QU) |
| Digital Inputs | | Optional: 4 off 24V Inputs | 425 |
| Digital Outputs | | 4 off, Programmable, Low Level Opto Isolated for High-Speed Switching Optional: 4 off 24V, 1A Outputs for High Power Switching | |
| Waveform Generation | | | 198 |
| Frequency Range | - | 0.00001 to 1,000 Hz | SD View (All Measurements in mm) |
| Resolution Waveforms | • | 32-bit Sine, Triangle, Square, Haversine, Havertriangle, Havesquare, Remp, Dual Remo. Trapszoidel. and Random | op view (vir weasshements in mini) |
| Signal Conditioning | | Carry, Captornary, and Carry and | |
| Compatible Transducer Types | | Resistive Bridges (e.g. Strain Gauged Load Cells and Extensometers), AC Devices (e.g. LVDT) and DC (e.g. Pre-Conditioned Devices) | |
| Transducer Recognition / Calibration | - | Automatic with Instron+ Devices, Manual with Others | |
| Data Acquisition Rate | - | 5 kHz | |
| Resolution | | 19-bit (1k Hz Bendwidth) 24-bit (1 Hz Bendwidth vie e Digitel Reedout) | |
| System Measurement Accuracies (wi | ith Instron | Transducers) | |
| Position | - | ±0.2% of Transducer Full Travel Under Normal Operating Conditions | |
| | | ±0.002% of Load Cell Capacity or 0.5% of Indicated Load, Whichever is Greater - Meets or Surpasses 1807500-1 Class 0.5, A8TM E 4, EN10002- 2 Class 0.5, JIS (87721, 87733) Down to 1/250th of Full Scale. | |
| | | ±0.005% of Transducer Capacity or ±0.25% of Reading ±Transducer Accuracy, Whichever is Greater. Meets or Surpasses 1808513 (lose 0.5, 1, 2, ASTM E 83 Class B1, B2, C, D, EN 10002-4 Class 0.5, 1, 2 and 187741. Crede 0.5, 12 Depending on the Extensioneter Used. | |
| | | | |
| Strain General Specifications | | | |
| Strain General Specifications Weizht (Fully Populated) | kć | 14 | |
| Strain General Specifications Weight (Fully Populated) | kg Ib mm | 14 31 450 | |
| Strain General Specifications Weight (Fully Populated) Height | kg Ib mm in | 14 31 450 17.7 199 | |
| Strain General Specifications Weight (Fully Populated) Height Width | kg Ib mm in mm in | 14 31 450 17.7 188 7.8 | |
| Strain General Specifications Weight (Fully Populated) Height Width Depth | kg Ib mm in mm in in | 14 31 450 17.7 188 7.8 475 18.7 | |
| Strain General Specifications Weight (Fully Populated) Height Width Depth Electrical Supply | kg Ib mm in mm in in | 14 31 450 17.7 188 7.8 475 18.7 80-132 and 180-284V 45-85 Hz Single Phase (Auto Switching) | |
| Strain General Specifications Weight (Fully Populated) Height Width Depth Electrical Supply Power Consumption | kg Ib mm in mm in in - | 14 31 450 17.7 188 7.8 475 18.7 80-132 and 180-284V 45-85 Hz Single Phase (Auto Switching) 800 VA Maximum | |

(Adapted from: https://www.instron.us/-/media/literature-library/products/2012/10/8800minitower-control-electronics.pdf?la=en-US)

AMTI VIVO Joint Simulator



SPECIFICATIONS CONTINUED

| ACTUATOR TYPE | SPECIFICATION | | |
|------------------------------------|--|--------------------------------|--|
| All Degrees of Freedom | Servo-hydraulic | | |
| D.O.F. / AXIS | SENSORS | AVAILABLE CONTROL MODES | |
| Vertical Actuator | Fz force & position | Force or displacement control | |
| Flexion | Mx torque & position | Torque or displacement control | |
| IE Rotation | Mz torque & position Torque or displacement of | | |
| Y (AP) Translation | Fy force & position | Force or displacement control | |
| X (ML) Translation | Fx force & position | Force or displacement control | |
| Abduction/Varus | My torque & position Torque or displacement | | |
| LOAD CELLS | SPECIFICATION | MEASURED FORCES & MOMENTS | |
| Six-Axis | Removable | Fx, Fy, Fz, Mx, My, Mz | |
| PHYSICAL SPECIFICATIONS | SPECIFICATION | | |
| Height | 1900 mm | | |
| Width | 635 mm | | |
| Depth | 965 mm | | |
| Weight | 272 kg (800 lbs) | | |
| HYDRAULIC SYSTEM [1] SPECIFICATION | | COMMENT | |
| External hydraulic power supply | | Required, quoted separately | |
| Pressure | 5.5 MPa (800 psi) | Required | |
| Required Flow | 30 LPM (8 GPM) Typical | Required | |
| | Gravity drain return | | |
| Oil Temperature | 38°C | Recommended at outlet | |
| Water Cooling | 3kW Typical | 26°C Max cooling water | |
| | 2 LPM @ 15°C | temperature | |
| POWER [2] | SPECIFICATION | COMMENT | |
| Electric | 115/230 VAC, 1500 watts | 1 phase, 50/60 Hz | |
| CE Compliant | | | |

DYNAMIC PERFORMANCE

| ITEM | MAXIMUM REPETITION RATE (REPETITIONS/SECOND) | | | | |
|--------------------|---|--------------------------------------|-------------------------|--|--|
| Controller | 30 Hz | | | | |
| D.O.F | TYPICAL REPETITION RATE (REPETITIONS/SECOND) [3] | MAXIMUM SPECTRAL CONTENT (HZ) [4] | RMS ERROR (% FS) [5] | | |
| Axial Load | 2.0 Hz | 10 Hz | < 1 % | | |
| Flexion Extension | 2.0 Hz | 10 Hz | < 1 % | | |
| IE Rotation | 2.0 Hz | 10 Hz | < 1 % | | |
| Y (AP) Translation | 2.0 Hz | 10 Hz | < 1 % | | |
| X (ML) Translation | 2.0 Hz | 10 Hz | < 1 % | | |
| Abduction/Varus | 2.0 Hz | 10 Hz | < 1 % | | |



ENVIRONMENTAL CONDITIONING FOR SPECIMENS

| SPECIMEN FLUID RECIRCULATION | SPECIFICATION | COMMENT | |
|------------------------------|---|---|--|
| Pump | 100 ml/min | 60 RPM peristaltic with #25 silicone tubing | |
| Reservoir | 500 ml | Stainless steel tank | |
| Fluid Level | High/Low | Magnetic sensor/float | |
| SPECIMEN FLUID TEMPERATURE | SPECIFICATION | COMMENT | |
| Temperature Controller | 50 watts heating/cooling Thermoelectric | | |
| SPECIMEN FLUID | SPECIFICATION | | |
| Suitable fluids | Bovine serum, saline solution, | , water | |

MEASUREMENT INSTRUMENTATION

| DATA ACQUISITION | CHANNEL | RANGE | COMMENT | |
|-----------------------|-------------------------------|-----------------------|-----------------|--|
| Data Rate | All channels | 10-2000 samples/sec | User selectable | |
| ADC Resolution | All channels | 14 bit | | |
| Digital Filters | All channels | 10-500Hz | User selectable | |
| Anti-Aliasing Filters | All channels | 600 Hz | | |
| MULTI-AXIS LOAD CELL | CHANNEL | RANGE | | |
| Axial Load | Fz | ±4500 N | | |
| AP Force | Fy | ±2200 N | | |
| ML Force | Fx | ±2200 N | | |
| Flexion Moment | Mx | ±200 N-m | | |
| Valgus Moment | My | ±200 N-m | | |
| Axial Moment | Mz | ±100 N-m | | |
| ANGLE AND POSITION | RANGE | NOMINAL RESOLUTION | N | |
| Axial Position | ±25 mm | 0.1 mm | | |
| Flexion | ±100° | 0.1° | | |
| IE Rotation | ±40° | 0.1° | | |
| Y (AP) Translation | ±25 mm | 0.1 mm | | |
| X (ML) Translation | ±25 mm | 0.1 mm | | |
| Abduction/Varus | ±30° | 0.1° | | |
| OTHER SENSORS | COMMENT | | | |
| Serum Temperature | Dual sensors | Dual sensors | | |
| Oil Temperature | Table top temperat | Table top temperature | | |
| Serum Fluid Level | High/low safety shutoff | | | |
| Hydraulic Pressure | Dual gauges and supply sensor | | | |



SPECIFICATIONS CONTINUED

CONTROL SYSTEM NETCONTROL INTERFACE SPECIFICATION COMMENTS Included Computer Hardware Windows based PC and accessories Interface Ethernet 1Gb REAL-TIME CONTROLLER SPECIFICATIONS COMMENTS Intel Core 2 Controller 2.8 GHz Double precision floating point math Update Rate 2000 Hz CONTROL MODES CHANNELS State Space Control 6 Gain Scheduling 6 Adaptive Control 6 Virtual Soft Tissue 6 WAVEFORM GENERATOR CHANNELS RANGE COMMENTS Reference Waveforms 6 Repetition Rate All channels 0.01 to 30 Hz Programmable 1024 points Interpolated EVENT MONITOR CHANNELS SPECIFICATION COMMENTS Threshold Trigger 5 per station Rising or falling edge Response Time All channels 0.0005 seconds Programmable Response Soft stop, hold, shut down All channels DATA ACQUISITION CHANNELS Force/Torque 6 Displacement/Rotation 6 Temperature 4 Reference Waveforms 6 Servo Error 6 Soft Tissue Constraint 6 Sum Signals 6 Oil Pressure 1

[1] The external hydraulia system requires adoling water for operation, usually available from your laboratory's infrastruature. If not available, contact AMT for information about ohillers. [2] Contact AMT for other power configurations. [3] The typical repetition rate corresponds to the maximum rate at which satisfactory performance will be achieved running the ISO standard gait availe watersms for knee testing. This is a somewhat subjective indication of dynamic performance. Overall tracking performance is reduced with higher frequency. (4] The ISO waveforms contain spectral content in considerable excess of the fundamental driving frequency. Analysis of these waveforms indicates that tracking performance at 1 fix repetition rate is excellent up to the indicated frequency. (5) The RMS error provides a measure of the simulator's tracking performance of 1 fix repetition rate is excellent up to the indicated frequency. These values are typical for festing at a 1 harepetition rate while running the ISO waveforms and represent standard results while evaluating conventional proteins and represent standard results while evaluating conventional proteins and represent standard results while evaluating according a conventional proteins and represent standard results while evaluating conventional proteins are sing AMT at Adaptive Control Technology (iterative learning control algorithm). Different prosthetic devices or conditions may result in an increased or decreased tracking error.



(Adapted from: https://www.amti.jp/AMTI-VIVO-Brochure-Rev2-HiRes.pdf)

AMTI Boston



AMTI-Boston 6-Station Knee Simulator Model KS2-6-1000

The AMTI-Boston Knee Simulator, developed with the guidance of the medical profession and manufacturers of implants, simulates the loading and multi-axis motion of the knee in a physiologic environment. At each station the implant components are mounted in the proper fashion and the contact area is immersed in a scaled, temperature controlled fluid bath. The microprocessor controlled system allows programming of an endless combination of loads and motions in order to simulate walking, stair climbing etc. The simulator provides a system for evaluating materials, implant component designs and implant life testing.

The knee simulator has six stations arranged in two banks of three. Each bank may be operated at a different set of conditions. The system has four controlled degrees of freedom (load and three motions) on each bank. The femoral component rotates about the horizontal axis to a maximum of 134° in flexion (±67° of flexion-extension); the femoral component translates (anterior/posterior translation) a maximum of ±25 mm (±1 in.) relative to the tibial tray. The tibial component has a maximum possible rotation about the vertical axis of ±20°. Two other motions are allowed but not controlled; the tibial component is allowed to translate along the horizontal plane and rotate about the horizontal axis to provide "nesting" of the femoral and tibial components (a maximum of ±10 mm (0.5 in.) of translation, and ±3° of rotation). A programmable axial load up to 4500 N (1000 lb) is applied at each station, where typical walking patterns may contain four load peaks during each step. Prior to the initiation of a test, adjustments may be made to provide for different condyle loading; for example, the distribution of the load applied may be set at 50% on each condyle, or up to 80% loading on either condyle.

Available as an option is medial/lateral force control on the tibial component. This feature results in a five degree-of-freedom simulator with ± 7.5 mm. of allowable translation and $\pm 5^{\circ}$ of tibial tray rotation.

The operating speed of the machine is variable up to speeds of 2 Hz - the speed is programmed through the computer. Each joint is immersed in its own bath of boxine serum, saline or water temperature-controlled fluid. Separate reservoirs (including level sensors) are provided for each station and each station includes its own pump to circulate the fluid to the test station. The knee components are enclosed in a flexible plastic bag (LV, bag) to limit fluid evaporation and prevent foreign materials from influencing the test results.

At each station is a force/torque (Fx, Fy, Fz, Mx, My, Mz) load cell allowing any or all of the load components to be monitored continuously or at periodic intervals. In addition, motions and temperatures are available for monitoring at each station. These parameters are available for evaluation of the implant, wear, changing friction etc. as the testing proceeds. The simulator runs unattended 24 hours per day, allowing completion of 172 000 cycles per day when

testing at 2 Hz. Each fluid reservoir contains level sensors, which shut off the machine if a leak in the system occurs or certain other malfunctions occur. Polycarbonate doors provide environmental and safety protection.



Specifications

6-Station Knee Simulator Model No. KS2-6-1000

- Six Station Knee Simulator arranged in 2 banks of 3.
 4 Separate actuator axes for each
- 4 Separate actuator axes for each bank; total of 8 servohydraulic control valves.
- Fz loads up to 4500 N (1000 lb) at each station; each bank of three under separate control.
- Mz loads up to 45 N·m (400 in-lb) at each station; each bank of three under separate control (±20° rotation).
- Flexion of 134° total rotation with 56 Nm (500 in-lb) of torque per station; each bank of three under separate control.
- ±25 mm of anterior/posterior motion at up to 1800 N (400 lb) per station; each bank of three under separate control.
- All motions servohydraulically generated with laminar seal actuators for long life (>10^e cycles).
- Separate six component load cells at each station (Fx, Fy, Fz, Mx, My, Mz).
- Separate peristaltic pump and reservoir unit with temperature and level sensors at each station. Automatic shut-down due to test fluid containment system leakage.
- Temperature control of test fluid at 37°C (±1K) or other set point, provided by a closed loop recirculating heater/chiller unit.

- A total of 8 separate software programmable drive signals for the axes of motion.
- The four actuators on each bank can be "manually" controlled in addition to their full automatic operation.
- Separate cycle counter for each bank in addition to computer monitor of cycles for machine shut down.
- Up to 2 Hz speed adjustable by computer.
- Separate unbreakable polycarbonate doors over each bank.
- Rated for continuous 24 hour operation.
- High pressure hydraulic filter: 6 micron (6 μm) with indicator light.
- Separate hydraulic power supply required but not included -5 MPa @ 1 L/s. (800 psi @ 15 gpm), gravity drain return line with no restrictions.
- Included is software (SIMMAC[®]) to operate the simulator, prepare "load" and "motion" templates for simulation of different conditions (walking, stair climbing etc.) and safety /low fluid shutdown of system. Also, included is a software program which is used in calibration of the simulator load cells.

Knee Simulator Options

Knee Simulator Vertical Position Sensor (model VPS-6)

This option provides the six stations with a measurement of the vertical position of the tibial tray specimen holder. The sensors are of a non-contacting magnetic design and provide an output of 10 V/in. (0.4 V/mm). The measurement resolution is less than .001 inches $(25\mu m)$. All electrical outputs are wired in and sampled as part of the data acquisition routine for the stations.

Knee Simulator Tibial Torque Actuators (model KTTA)

This option provides an independent servo controlled medial/lateral force on each tibial component, with \pm 7.5 mm of allowable medial/lateral translation and \pm 5' of allowable rocking about the A/P axis. Sensors are provided for the translation and rocking motions. The controlled output is \pm 900 N (200 lb) for \pm 10 V of command. There are individual pneumatic servo valves with a closed loop controller on the top of each load cell. All six actuators are controlled by a common command signal. With this option, the left and right banks of the machine are not provided with separate command signals. A continuous source of oilless air is required and recommendations for a suitable compressor will be provided.

Knee Machine Load Cell Calibration Fixtures (model KLCC) Hardware and software to calibrate the six-component load cell is provided in this package. A single axis calibration cell is used to calibrate all channels at levels simulating operating loads.



(Adapted from: https://www.yumpu.com/en/document/read/42207110/amti-boston)

Curriculum Vitae

| Name: | Geofrey Yamomo |
|---|---|
| Post-secondary Education and Degrees: | The University of Western Ontario London, Ontario, Canada 2012-2017 BESc |
| Related Work Experience | Teaching Assistant– FEM in Mechanical Engineering The University of Western Ontario Jan 2019-Apr 2019 |
| | Teaching Assistant– Advanced CAE: Reverse Engineering The University of Western Ontario Jan 2018-Apr 2018, Sept 2018-Dec 2018 |
| | Teaching Assistant– Advanced CAE: Simulation The University of Western Ontario Sept 2017-Dec 2017 |

Publications:

Willing R, Moslemian A, Yamomo G, Wood T, Howard J, Lanting B. Condylar-Stabilized TKR May Not Fully Compensate for PCL-Deficiency: An In Vitro Cadaver Study. Journal of Orthopaedic Research, 37:2172–2181, 2019.

Ibrahim A, Yamomo G, Willing R, Towfighian S. Analysis of a triboelectric energy harvester for total knee replacements under gait loading. Proc. SPIE 10967, Active and Passive Smart Structures and Integrated Systems XIII, 109671D (21 March 2019).

Conference Presentations and Posters:

Yamomo G, Ibrahim A, Towfighian S, Willing R. (2019) Tuning of 3D Printed Energy Harvester Package Design. *International Society for Technology in Arthoplasty 32nd Annual Congress, October 2-5, Toronto, Ontario*

Yamomo G, Moslemian A, Wood T, Lanting B, Willing R. (2019) Kinematic and Laxity Comparison between Cruciate Retaining and Cruciate Sacrificing Total Knee Replacements. *International Combined Orthopaedic Research Societies* 2nd Annual *Meeting, June 19-22, Montreal, Quebec*

Yamomo G, Moslemian A, Wood T, Lanting B, Willing R. (2019) Kinematic Comparison between Cruciate Retaining and Cruciate Sacrificing Total Knee Replacements during Gait and Stair Ascent. International Combined Orthopaedic Research Societies 2nd Annual Meeting, June 19-22, Montreal, Quebec Yamomo G, Moslemian A, Wood T, Lanting B, Willing R. (2019) Post-Operative Laxity Outcomes of Cruciate Retaining and Condylar Stabilizing Total Knee Replacements. London Health Research Day, April 30, London, Ontario

Yamomo G, Moslemian A, Wood T, Lanting B, Willing R. (2019) Post-Operative Laxity Differences Between Cruciate Retaining and Condylar Stabilizing Total Knee Replacements. Ontario Biomechanics Conference, March 8-10, Alliston, Ontario