



Design of a prosthetic dynamic response foot with adjustable stiffness

H van Niekerk

 **orcid.org/ 0000-0003-2514-7642**

Dissertation accepted in fulfilment of the requirements for the degree *Master of Engineering in Mechanical Engineering* at the North-West University

Supervisor: Prof AS Jonker

Co-supervisor: Dr JH Kruger

Graduation: July 2025

Acknowledgements

I would like to express my sincere gratitude to everyone who has supported me throughout the course of this dissertation.

First and foremost, I am deeply grateful to my supervisors, Prof. A.S. Jonker and Dr. J. Kruger, for their invaluable guidance, encouragement, and expertise. Their insights and constructive feedback have been instrumental in shaping this study.

I would also like to thank my colleagues from Room 213 for their stimulating discussions and unwavering support, with special thanks to university staff members Ian Thomson, J.P. van Deventer, and Mientjie “Miempie” Botha.

A special thanks to Jonker Sailplanes and their staff for their collaboration and for granting me access to their facilities and equipment to manufacture my project.

I am also grateful to the NWU finance and administrative support team for their assistance throughout this journey.

I am grateful to my family and friends; without their encouragement, this journey would have been impossible.

Finally, I acknowledge all individuals, directly or indirectly, who contributed to the completion of this dissertation. Your support has been invaluable, and I am truly grateful.

Thank you all.

Abstract

Lower limb loss due to diabetes or vehicle accidents is a harsh reality for many South Africans. Most lower limb amputations require a prosthetic foot to assist the individual in returning to their lifestyle, yet prosthetics are expensive and not affordable for the average South African.

A prosthetist fits an amputee with a prosthetic foot that accommodates their weight and activity level. Prosthetic feet that are too flexible or too stiff can have a noticeable limp effect and discomfort when the user walks. The aim of this study, therefore, was to design a prosthetic foot with an adjustable stiffness, assisting with different activities and body weights.

Multiple design options were evaluated based on their complexity and manufacturing cost, with the chosen design incorporating three composite blades and gas springs to adjust the foot stiffness. Detailed design and calculations were completed and verified through finite element analysis, for a standard adult weight of 80 kg (800 N) with a specified deflection of 23.6 mm of the forefoot and 13.6 mm of the heel.

An MTS Universal testing machine was used for experimental testing to validate the modelling and simulation methodology. The prosthetic foot was attached to a jig that would allow for forefoot testing (with and without gas springs), as well as heel testing.

The experiments were found to be repeatable, and in general the results of the measured deflections were comparable with the calculated estimations. The largest deviations in the forefoot blade deflection was 2.4% at 16 N, and for the heel blade it was 14.6% at 45.56 N.

The addition of the gas springs resulted in an increase in the reaction force to affect foot displacement. When two 150 N gas springs were attached, deflection initiated at a force of approximately 200 N, two 350 N springs required 450 N, and the last set of 500 N springs required 650 N. The measurements with the gas springs attached saw an increase in deviation between the measurements and calculations, from 3% without the gas springs and up to 8.5% when the gas springs were attached.

The results indicated that adjustment of the stiffness of the foot performed as intended. However, because the adjustments are made with discrete levels in gas force resistance, the approach might be better suited to adapting to different body weights rather than different activities.

A high level cost evaluation using realistic manufacturing and certification costs was done to compare the estimated cost of the prosthetic foot to that of commercial feet and estimated a cost of R17 000 for the prosthetic foot, with the possibility of lowering the cost when mass producing. This cost was about R4 000 lower than the lowest available compensation option from government sources.

Keywords: Dynamic response, prosthetic foot, variable stiffness, composite foot design, affordable assistive technology.

Contents

Acknowledgements	i
Abstract	ii
Contents.....	iii
List of Figures	vi
List of Tables.....	ix
List of Symbols	xi
Glossary.....	xii
List of Abbreviations.....	xiii
Chapter 1: Introduction	1
1.1. Background	1
1.1.1. Prosthetic use in South Africa.	1
1.1.2. Characteristics of Dynamic Response Feet and Their Cost Implications.....	4
1.2. Problem Statement	6
1.3. Aim and Objectives	6
1.4. Scope and Constraints	7
1.5. Dissertation Overview	7
Chapter 2: Literature Review	8
2.1. Chapter Overview	8
2.2. History of Prosthetic Limbs	9
2.3. Anatomy of a Human Foot.....	11
2.4. The Foot During the Gait Cycle.....	13
2.5. Dynamic Response Prosthetic Feet.....	16
2.5.1. Motion of a Dynamic Response Foot.....	16
2.5.2. Common Components of Dynamic Response Prosthetic Feet.....	17
2.6. Studies on Stiffness Effect	19
2.7. Studies on Prosthetic Feet with Adjustable Stiffness.....	22
2.8. Materials of a Prosthetic Foot	25
2.8.1. Adapter Materials: Stainless Steel, Aluminium, and Titanium.....	25
2.8.2. Orthotropic materials: Carbon Fibre, Kevlar, and Fibreglass	27
2.9. Chapter Summary.....	30
Chapter 3 : Concept Development and Evaluation	31
3.1. Chapter Overview	31
3.2. Foot as a Simple Cantilever Beam	32
3.3. Concept Studies.....	34
3.3.1. Concept 1: Horisontally Adjustable Blade Design.....	34
3.3.2. Concept 2: Vertically Adjustable Blade Design.....	37
3.3.3. Concept 3: Gas Spring Design	43
3.4. Concept Evaluation	45
3.5. Chapter Summary.....	46
Chapter 4: Detail Design	47
4.1. Chapter Overview	47
4.2. Detail Design Specifications	48
4.3. Calculations of Theoretical Displacement	51
4.3.1. The Forefoot Blade.....	51
4.3.2. The Heel Blade	64

4.3.3. Gas Spring Calculations	71
4.4 .Dimensional Design	73
4.4.1. Prosthetic Foot Designer Program.	73
4.4.2. Gas Spring Calculation.....	79
4.5. Layup Model Design	80
4.5.1. Initial Ply Layup	80
4.5.2. Final Layup	81
4.6. Cost Analysis of Manufacturing the Prosthetic Foot	83
4.7. Chapter Summary.....	85
Chapter 5: Experimental Procedure	86
5.1. Chapter Overview	86
5.2. Experimental Setup	86
5.3. Experimental Methodology.....	87
5.4. Data and Results Processing.....	88
5.5. Chapter Summary.....	88
Chapter 6 : Results	89
6.1. Chapter Overview	89
6.2. Validation of Calculations.....	90
6.3. Basic Prosthetic Foot Results	92
6.3.1. Validation and Verification of Forefoot Results: 0 N GSF	92
6.3.2. Validation and Verification of Forefoot Results: 300 N GSF	94
6.3.3. Validation and Verification of Forefoot Results: 700 N GSF	96
6.3.4. Validation and Verification of Forefoot Results: 1000 N GSF	98
6.3.5. Validation and Verification of Heel Results.....	100
6.4. Discussion	102
6.5. Chapter Summary.....	106
Chapter 7: Conclusion and Future Work.....	107
7.1. Chapter Overview	107
7.2. Conclusion.....	107
7.3. Future Work	110
References	113
Appendix A: Calculation Verifications	A-1
A.1. Verification of Calculation Method.....	A-1
A.2. Verification of Prosthetic Foot Designer.....	A-5
A.3. Verification of Gas Spring Calculations.....	A-30
A.4. Verification of Layup FEA results	A-32
A.5. Adjustment calculation verification.....	A-37
Appendix B : Drawings.....	B-1
B.1. Manufacturing Layup Drawings.....	B-1
B.2. Mould Drawings	B-4
Appendix C: Graphs and Tables of Results	C-1
C.1. Forefoot: No Gas Springs	C-1
C.1.1. Experiment Results	C-1
C.1.2. Comparison Plots and Tables	C-4
C.2. Forefoot: 300 N GSF	C-5
C.2.1. Experiment Results	C-5
C.2.2. Comparison Plots and Tables	C-8
C.3. Forefoot: 700 N Gas Springs	C-9

C.3.1. Experiment Results	C-9
C.3.2. Comparison Plots and Tables	C-11
C.4. Forefoot: 1000 N Gas Springs	C-12
C.4.1. Experiment Results	C-12
C.4.2. Comparison Plots and Tables	C-14
C.5. Heel.....	C-15
C.5.1. Experiment Results	C-15
C.5.2. Comparison Results	C-17
Appendix D: Cost Evaluation.....	D-1
Appendix E: Experiment Methodology.....	E-1
E.1. Equipment and Tools Used.....	E-1
E.2. Experimental Procedure.....	E-2
Appendix F: Standards, Theory and Additional Data	F-1
F.1. International Standards (ISO10328)	F-1
F.2. Ottobock Foot Size Chart.....	F-3

List of Figures

Figure 1: Amputations of the lower extremities.....	2
Figure 2: Example of a split-toe DR design (Freedom DynAdapt prosthetic foot [34]).....	4
Figure 3: Wooden toe prosthetic found in Egypt [47].....	9
Figure 4: Ambroise Paré’s design for an above-the-knee prosthesis [48].....	10
Figure 5 : Bone structure of the human foot [53].....	11
Figure 6: Movements and rotations of a human ankle plantar flexion [59].	12
Figure 7: Events in the gait cycle (adapted from Karb et al. [62])	14
Figure 8: Motion of a foot during the gait cycle [63].....	15
Figure 9: Vertical forces acting on the foot during the stance phase [63].....	15
Figure 10: Foot motion when walking.	16
Figure 11: Section view of a DR prosthetic foot [67]	17
Figure 12: Ottobock Triton foot (right) with foot shell (left) [72].	18
Figure 13: Feet used for the Klodd et al. study [74].....	19
Figure 14: Stiffness experiment [75].....	20
Figure 15: Displacement of nine prosthetic feet [75].....	21
Figure 16: Variable stiffness foot [76]	22
Figure 17: VSF beam model [76].....	22
Figure 18: Powered ankle-foot prosthesis [77]	23
Figure 19: Variable Stiffness Ankle design [78].....	24
Figure 20: The roles of the fibres and matrix of a composite material [98].....	28
Figure 21: Example of composite ply layups illustrating different layup angles [103]	29
Figure 22: A prosthetic foot considered as a cantilever beam.....	32
Figure 23: Diagrams of a simple cantilever beam.....	33
Figure 24: Cantilever beam representation of the concept of horisontal adjustment of the blade, ...	34
Figure 25: FD of horisontal adjusting blade cantilever beam.	35
Figure 26: Horisontally moving blade: fulcrum design concept.	36
Figure 27: Horisontally moving blade: horisontal adjusting design concept.	36
Figure 28: Cantilever beam simplification of vertically moving blade design	37
Figure 29: Contact instances of vertical adjusting blade.....	39
Figure 30: Vertically adjusting blades: Gear design concept.....	40
Figure 31: Vertically adjusting blades: wing design concept.....	40
Figure 32: Vertically adjusting blades: fork design concept.	41
Figure 33: Cantilever beam simplification of fork design.....	41
Figure 34: Vertically moving blades: stacked design concept.	42
Figure 35: Gas spring design concept.	43
Figure 36: Cantilever simplification and free body diagram of the gas spring concept.....	44
Figure 37: Components of the selected design concept	47
Figure 38: Foot stiffness results from Koehler-McNicholas’ study [75].	48
Figure 39: Prosthetic feet used as inspiration for prosthetic foot design.....	49
Figure 40: Simplified drawing of the proposed prosthetic foot.....	51
Figure 41: The equivalent Young’s modulus of the combined forefoot blades	53
Figure 42: The prosthetic foot with a single Young’s modulus for the forefoot blades.....	54
Figure 43: The segments and interest points of the forefoot.....	55
Figure 44: Force components of the primary blade.....	55
Figure 45: Force distribution of prosthetic forefoot segments	56
Figure 46: Conceptual representation of the initial position of the prosthetic foot segments	57
Figure 47: The new angle with the x-axis	59
Figure 48: Conceptual representation of the blade position after considering moment on AB.....	59

Figure 49: Conceptual representation of the deflection of segment BC.....	61
Figure 50: Conceptual representation of the displacement of CD.....	62
Figure 51: Conceptual representation the displacement of the blade when force is applied.....	63
Figure 52: Segments and positions of the heel blade with the forefoot blade.....	64
Figure 53: Force components of the heel blade's segments	65
Figure 54: The resulting forces and moments on the heel blade segments.	66
Figure 55: Conceptual representation of the initial position of the heel blade	67
Figure 56: Conceptual representation of the deflection of ZY segment.....	68
Figure 57: Conceptual representation of the deflection of segment YX	69
Figure 58: : Conceptual representation of the final displacement of the heel blade.....	70
Figure 59: Free body diagram of gas springs attached to the prosthetic foot.....	71
Figure 60: Resultant force breakdown on the prosthetic forefoot	72
Figure 61: Flow of the prosthetic foot Designer calculation loop.....	73
Figure 62: Flow of the forefoot function during prosthetic foot calculation.....	75
Figure 63: Flow of the heel blade function during prosthetic foot calculation	77
Figure 64: Illustration of the initial ply layup	81
Figure 65: Final layup of blade 1	82
Figure 66: The experimental setup of the forefoot tests (a) and heel tests (b)	86
Figure 67: Prosthetic foot during testing with two 500 N gas springs attached	87
Figure 68: Flow of the study's validation and verification	89
Figure 69: Comparison between intended deflection and deflection of adjusted calculations.....	91
Figure 70: Displacement vs applied load of the forefoot experiment without gas springs	92
Figure 71: Comparison of calculations and experimental results of the forefoot without gas springs..	93
Figure 72: Displacement results of the forefoot experiment of the 300 N GSF load case	95
Figure 73: Comparison of calculations and experimental results of the 300 N GSF load case.....	95
Figure 74: Displacement results of the forefoot of the 700 N GSF load case	97
Figure 75: Comparison of calculations and experimental results of the 700 N GSF load case.....	97
Figure 76: Displacement results of the forefoot of the 1000 N GSF load case	99
Figure 77: Comparison of calculations and experimental results of the 1000 N GSF load case.....	99
Figure 78: Experimental results for heel blade: load vs. displacement.	101
Figure 79: Comparison of experimental results with calculations for the heel blade	101
Figure 80: Comparison of all experiment averages and calculations.....	102
Figure 81: Illustration of the sanding on the forefoot (above) and heel blade (below)	104

Appendix A

Figure A - 1: Implementation of dimensions in CAD model.....	A-2
Figure A - 2: FEA setup of forefoot blade for calculation method verification	A-2
Figure A - 3: FEA setup of heelblade for calculation method verification	A-3
Figure A - 4: FEA results of forefoot and heel using value set 1	A-3
Figure A - 5: The forefoot segments and points.....	A-5
Figure A - 6: The force distribution of prosthetic forefoot.....	A-6
Figure A - 7: Initial position of the forefoot.....	A-10
Figure A - 8: Displacement of the forefoot blade when AB deflects.	A-12
Figure A - 9: Deflection of segment CD	A-13
Figure A - 10: Deflection of segment BC	A-15
Figure A - 11: Final forefoot displacement.....	A-17
Figure A - 12 Segments of the heel blade with the forefoot blade.....	A-18
Figure A - 13: Resulting forces and moments on the heel blade.....	A-18
Figure A - 14: Initial points of the heel blade	A-22
Figure A - 15: The deflection of segment YZ	A-23
Figure A - 16: Deflection of segment YX.....	A-26

Figure A - 17: Final heel displacement	A-27
Figure A - 18: FEA results for the verification of designer results	A-29
Figure A - 19: FEA setup for gas spring verification.....	A-30
Figure A - 20: FEA results for a forefoot with 800 N applied load and 700 N GSF.....	A-31
Figure A - 21: Display of ply layup with stagger.....	A-32
Figure A - 22: FEA setup of applied forces to the forefoot (left) and heel blade (right)	A-33
Figure A - 23: Results of FEA of the ply layup forefoot.....	A-33
Figure A - 24: Results of FEA of the ply layup heel.....	A-34

Appendix B

Figure B - 1: Layup drawing for primary forefoot blade	B-1
Figure B - 2: Layup drawing for secondary forefoot blade.....	B-2
Figure B - 3: Layup drawing for heel blade	B-3
Figure B - 4: Mould drawing for primary forefoot blade	B-4
Figure B - 5: Mould drawing for secondary forefoot blade	B-5
Figure B - 6: Mould drawing for heel blade.....	B-6

Appendix C

Figure C - 1: Results of the experiment tests: forefoot with no GSF	C-2
Figure C - 2: Calculations and experiment results: forefoot with no GSF	C-4
Figure C - 3: Results of the experiment tests: forefoot with 300 N GSF	C-6
Figure C - 4: Calculations and experiment results: forefoot with 300 N GSF	C-8
Figure C - 5: Results of the experiment tests: forefoot with 700 N GSF	C-9
Figure C - 6: Calculations and experiment results: forefoot with 700 N GSF	C-11
Figure C - 7: Results of the experiment tests: forefoot with 1000 N GSF	C-12
Figure C - 8: Calculations and experiment results: forefoot with 1000 N GSF	C-14
Figure C - 9: Results of the experiment tests: heel.....	C-15
Figure C - 10: Calculations and experiment results: heel.....	C-17

Appendix D

Figure D - 1: Bi-directional carbon fibre cost	D-1
Figure D - 2: Unidirectional carbon fibre cost	D-2
Figure D - 3: Extract from quote for resin and hardner.....	D-3
Figure D - 4: Cost of Supawood material.....	D-3
Figure D - 5: Extract of pyramid adapter quote.	D-4
Figure D - 6: Gas springs quote extract.....	D-4

Appendix F

Figure F - 1: Position of setup of the prosthetic foot during testing.....	F-1
Figure F - 2: Angles of the loading platforms during testing.	F-2

List of Tables

Table 1: Pyramid material comparison table [82].....	26
Table 2: Concept cost evaluation table.....	46
Table 3: Design Specifications Summary	50
Table 4: Input variables for forefoot design	74
Table 5: Results of the prosthetic foot Designer	78
Table 6: Designer displacement compared to target displacement	78
Table 7: Gas spring forces required for equal displacement of 800N.....	79
Table 8: Validation of calculations by FEA for a 300 N gas spring applied load.....	79
Table 9: Values for the initial layup of the three blades.....	81
Table 10: FEA Results of final layup for the forefoot and heel	82
Table 11: Total material cost.....	83
Table 12: Total manufacturing cost.....	84
Table 13: Total price estimation.....	84
Table 14: Comparison of displacement results of FEA and calculations for forefoot and heel Blade..	90
Table 15: Obtained gas spring forces	91
Table 16: Applied forces comparisons at a displacement of 6 mm.....	103
Table 17: Experiment deviations from calculations.....	105

Appendix A

Table A - 1: Input variables to test calculation method.....	A-4
Table A - 2: Results of calculation and FEA for method validation	A-4
Table A - 3: Input variables.....	A-6
Table A - 4: Input values for heel blade calculations	A-19
Table A - 5: Comparison between the Designer and hand calculation results	A-28
Table A - 6: Comparison between the Designer and FEA results for the forefoot blade.....	A-28
Table A - 7: Calculation vs FEA for gas spring verification: no gas spring	A-30
Table A - 8: Calculation vs FEA for gas spring verification: 300 N GSF	A-31
Table A - 9: Calculation vs FEA for gas spring verification: 700 N GSF	A-31
Table A - 10: Calculation vs FEA for gas spring verification: 1000 N GSF	A-31
Table A - 11: Calculation vs FEA for layup verification: heel	A-35
Table A - 12: Calculation vs FEA for layup verification: forefoot with no gas spring.....	A-35
Table A - 13: Calculation vs FEA for layup verification: forefoot with 300 N GSF	A-36
Table A - 14: Calculation vs FEA for layup verification: forefoot with 700 N GSF	A-36
Table A - 15: Calculation vs FEA for layup verification: forefoot with 1000 N GSF	A-36
Table A - 16: Calculation vs FEA for adjustment verification: heel.....	A-37
Table A - 17: Calculation vs FEA for adjustment verification: forefoot with no gas spring	A-37
Table A - 18: Calculation vs FEA for adjustment verification: forefoot with 300 N GSF	A-38
Table A - 19: Calculation vs FEA for adjustment verification: forefoot with 700 N GSF	A-38
Table A - 20: Calculation vs FEA for adjustment verification: forefoot with 1000 N GSF	A-38

Appendix C

Table C - 1: Force at displacement of experiment runs: forefoot with no GSF	C-1
Table C - 2: Deviation from average: forefoot with no GSF.....	C-2
Table C - 3: Deviation from average: forefoot with no GSF - Continued.....	C-3
Table C - 4: Calculations and experiment deviation values: forefoot with no GSF	C-4
Table C - 5: Force at displacement of experiment runs: forefoot with 300 N GSF	C-5
Table C - 6: Deviation from average: forefoot with 300 N GSF	C-6
Table C - 7: Deviation from average: forefoot with 300 N GSF - Continued.....	C-7

Table C - 8: Calculations and experiment deviation values: forefoot with 300 N GSF	C-8
Table C - 9: Force at displacement of experiment runs: forefoot with 700 N GSF	C-9
Table C - 10: Deviation from average: forefoot with 700 N GSF	C-10
Table C - 11: Calculations and experiment deviation values: forefoot with 700 N GSF	C-11
Table C - 12: Force at displacement of experiment runs: forefoot with 1000 N GSF	C-12
Table C - 13: Deviation from average: forefoot with 1000 N GSF	C-13
Table C - 14: Calculations and experiment deviation values: forefoot with 1000 N GSF	C-14
Table C - 15: Force at displacement of experiment runs: heel	C-15
Table C - 16: Deviation from average: heel	C-16
Table C - 17: Calculations and experiment deviation values: heel	C-17

Appendix D

Table D - 1: Bi-directional carbon fibre cost breakdown	D-1
Table D - 2: Unidirectional carbon fibre cost breakdown	D-2
Table D - 3: Resin cost breakdown	D-2
Table D - 4: Mould costs breakdown	D-3
Table D - 5: Manufacturing cost breakdown	D-4
Table D - 6: Gas springs cost breakdown	D-4
Table D - 7: Total cost breakdown	D-5

Appendix E

Table E - 1: Components used during experiment	E-1
--	-----

List of Symbols

Δ/δ	Displacement
A	Position of point of the forefoot A during calculations
b	Blade width
B	Position of point B of the forefoot during calculations
C	Position of point C of the forefoot during calculations
D	Position of point D of the forefoot during calculations
A	Young's modulus of material
$E_1/E_2/E_3$	Young's modulus of composite blade
E_{eq}	Equivalent Young's modulus
F/P	Force applied
G	Gas spring force
h	Blade thickness
I	Second moment of area
k	Stiffness constant
L	Length of beam/segment
M	Moment
PI	Ply failure index
R	Rotation matrix
r_h	Thickness ratio
R_x/R_y	Resulting force components
$T_1/T_2/T_{total}$	Thickness of composite layers
U	Potential energy
v_i	Volume of material
V	Volume fraction
v_t	Total volume of component
W	Position of point W of the heel during calculations
X	Position of point X of the heel during calculations
Y	Position of point Y of the heel during calculations
z	Height of the second blade
Z	Position of point Z of heel during calculations
α	Angle between segment AB and BD
γ	Angle between the BD segment and x-axis due to displacement
ε	Angle of applied gas spring force
ζ	Angle of applied load for forefoot and heel
η_e	Transformation factor
φ	Default angle between the BD segment and x-axis
θ	Slope

Glossary

Amputee	A person who has undergone surgery to remove a limb or parts of it due to injury, disease, or other medical reasons.
Bidirectional fibres	Composite fibres woven or layered in two perpendicular directions.
Composite material	A material made from two or more materials with different physical or chemical properties.
Dorsiflexion	The upward movement of the foot.
Dynamic Response Foot	A type of prosthetic foot that stores and releases energy, as a prosthetic user walks, to improve their gait cycle.
Energy Return	The ability of a prosthetic foot to store and release energy during a person's gait cycle.
Eversion	The outward movement of the foot.
Forefoot blade	Part of the prosthetic foot that mimics the forefoot.
Gait	A person's manner of walking.
Gas Spring Force	The force exerted by a gas spring.
Heel Blade	Part of the prosthetic foot that mimics the heel.
Inversion	The inward movement of the foot.
Orthotropic Materials	Materials that exhibit different material properties in three perpendicular directions.
Plantarflexion	The downward movement of the foot.
Prosthetics	Artificial limbs created to improve the lifestyle of people with limb loss.
Prosthetist	A professional that specialises in the design, fitting and customisation of prosthetic devices for amputees.
Pyramid Connector	A connecting adapter used in prosthetics to connect various components to each other.
Unidirectional fibres	Composite fibres aligned in a single direction.

List of Abbreviations

BID	Bidirectional
AKA	Above-knee amputations
AMP	Amputee mobility predictor
CAD	Computer-aided design
DR	Dynamic response
EK	Elastic keel
ET	Equivalent thickness
FD	Force diagram
GSF	Gas spring force
ISO	International Organization for Standardization
MA	Multi-axis
MTS	Material test systems
PI	Ply index
RAF	Road Accident Fund
ROM	Rule of mixtures
SA	Single-axis
SACH	Solid ankle cushioned heel
UD	Unidirectional
UTS	Ultimate tensile strength
VSA	Variable stiffness ankle
VSF	Variable stiffness foot

Chapter 1 : Introduction

1.1. Background

1.1.1. Prosthetic use in South Africa.

Losing the lower extremities is a harsh and traumatic reality for millions of people, however, at the time of this study, there was no definitive information on the number of amputees in South Africa. In 2008, the number of amputees in the world was estimated at 10 million [1]. Nine years later, in 2017, it was estimated that nearly 60 million people lived with limb amputations due to trauma or injuries [2]. This data excluded people who lost their limbs due to medical conditions such as cancer.

The most common reason for lower extremity amputation is diabetes mellitus [3], [4]. Diabetes mellitus, or type 1 and 2, is an incurable disease in which the pancreas of a patient does not produce enough insulin or the body cannot use the insulin produced [5], [6]. Almost 430 million people worldwide and an estimated 4.58 million South Africans, 52% of whom are unaware of their condition, suffer from diabetes [7], [8].

Diabetes can cause circulation problems and nerve damage that can cause ulcers. Ulcers are wounds that do not heal or return frequently. A great threat to diabetics is foot ulcers. These wounds can become infected and can lead to amputation. About 15% of diabetics develop foot ulcers that require medical attention [9], [10]. Of these admitted patients, 15% go to lower extremity amputations due to foot ulcers [9], [11].

Addington Hospital, a public hospital located in KwaZulu-Natal, South Africa, reported that over five years, more than 1,000 lower extremity amputations were performed. Most of them were due to diabetes mellitus [4], [12]. Almost 50% of all amputations performed were below-the-knee amputations (BKA), and 20% of amputations performed were above-the-knee amputations (AKA). The other 30% were amputations performed through the foot and toes [4].

Only a small number of patients (17%) were referred for physiotherapy; most of these patients were diabetic with below-the-knee amputations. Patients in therapy rarely opt for prosthetics and prefer to use crutches or walking frames [4], [12]. Although the study by Manickum et al. did not give a definitive reason for the lack of prosthetic use, they felt that ideally each patient should get a trial period with a prosthetic during rehabilitation. They, however, also stated that their findings suggest that rehabilitation doesn't seem to be widely practised within our country's setting [4], [12].

As lower limb amputations vary in extent, as illustrated in Figure 1, the need for a prosthetic limb also varies. Partial foot amputations do not necessarily require a prosthetic, but some patients may opt for a carbon fibre sole or a silicone foot shell to help protect the residual limb [13], [14]. Ankle disarticulation, which is the separation of the foot through the ankle joint, avoids cutting the bone and allows walking without a prosthetic, although a specially designed socket is often needed due to the shape of the limb [15], [16], [17].

Transtibial (below the knee) amputations require a prosthetic socket and foot for comfortable walking [18], [19], [20]. Similar to ankle disarticulation, knee disarticulation avoids cutting major bones by separating the lower leg through the knee joint, leaving the thigh muscles unaffected [19], [21]. Users with these amputations will require a socket, a prosthetic knee, and a prosthetic foot. Transfemoral (above the knee) amputations also require these components [19], [22].

Hip disarticulation, an extreme and rare procedure, involves removing the leg from the hip joint, leaving the pelvis intact [19], [23]. Prosthetics for patients with hip disarticulation are very complex, requiring a specialised socket for the pelvic residuum, a hip joint prosthetic, a knee, and a foot to allow sitting, standing, and walking [24], [25].

Thus, the prosthetic foot is required for all types of amputation above ankle disarticulation and by approximately 70% of patients seen at Addington Hospital in 5 years [4], [12].

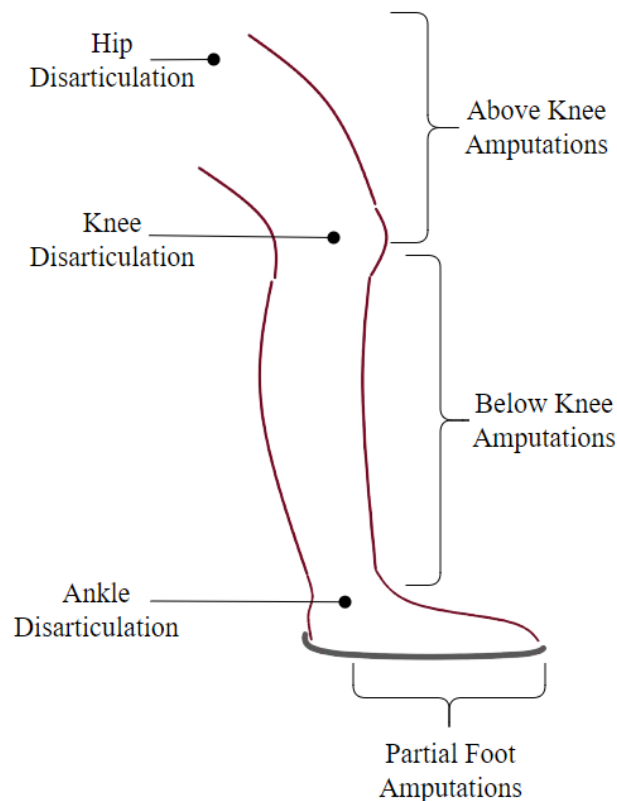


Figure 1: Amputations of the lower extremities

Prosthetic feet greatly contribute to the gait of a person, increasing their balance and allowing them to engage in activities such as running, walking, or hiking. The choice of foot is significant, as it impacts the user's performance in various activities. There are multiple types of prosthetic feet, each tailored for different activities. The prosthetist would typically classify the user with the amputee mobility predictor (AMP) test, guiding the user to choose the most suitable foot for their weight and activity level.

The everyday walking prosthetic feet fall into categories such as solid ankle cushioned heel (SACH) feet, elastic keel (EK) feet, single- or multi-axis feet, dynamic response (DR) feet, or microprocessor feet. In addition to the different types of feet recommended, there are also activity-specific feet. There are also feet designed for specific activities such as hiking, running, or swimming [26].

The SACH and the EK feet are the most basic and inexpensive feet available with no hinged parts and require minimal maintenance [20], [27]. The SACH foot consists of a solid forefoot, while the EK foot has an elastic or flexible forefoot [20], [28]. These feet are most frequently used and recommended to users with low activity levels, as they do not store and return energy as dynamic response feet [20], [28].

Single-axis feet (SA) allow for dorsiflexion and plantarflexion, providing greater stability for active users [28], [29]. However, multiaxis feet (MA) allow movement in two motion planes, allowing dorsiflexion, plantarflexion, inversion, and eversion [28], [30]. It is suitable for those with more demanding daily activities [28]. SA and MA feet are usually recommended for users with amputations between the knee and hip [27]. These prostheses are heavier, more expensive, and require more maintenance than the SACH and elastic keel foot [20].

The dynamic response foot (DR), ideal for highly active users, allows for more mobility and is ideal for varying walking speeds, rapid direction changes, and long distances [27], [28]. This prosthetic can store and release energy as the user walks, making walking with this prosthetic look and feel more natural to the user's gait. DR feet are usually made of composite materials that allow some flexion, allowing for energy build-up [31].

Microprocessor-controlled feet use sensors to adjust to the environment, supporting the gait cycle [27], [32], [33]. It is, however, much more expensive and heavier than other types and less durable in harsh conditions.

For this study, DR feet are the most suitable due to their superior energy return and robust construction, with microprocessor-controlled feet being too costly from the outset.



Figure 2: Example of a split-toe DR design (Freedom DynAdapt prosthetic foot [34]).

1.1.2. Characteristics of Dynamic Response Feet and Their Cost Implications

Dynamic response feet typically have a single fixed stiffness value linked to the foot's design and construction, which is directly proportional to the energy return of the foot. Energy return affects gait and the ease with which the user can walk [31] or perform other functions, with the amount of energy return required being influenced by the type of action. This means that users must buy different feet if they wish to participate in activities outside of an average day. For example, the foot used for everyday walking is not ideal or comfortable to use while running or walking in weighted conditions, such as hiking. The prosthetic foot for running would have a higher energy return, as its force input would be higher and its response would be faster [70]. Similarly, the prosthetic for running would not be ideal for walking. Access to multiple prosthetic feet is a luxury, but it can have a significant impact on a person's life. However, prosthetic feet are very expensive, and affording multiple prosthetic feet in order to return to your previous lifestyle is not a luxury most people have.

At the time of this study, little research is available on how often a new prosthetic is required or how often maintenance is required. Considering how much the residual limb shrinks after amputation due to decreased muscle mass or swelling, it is common to have a new socket made to adjust for the reduction in size [35], [36]. Prosthetics are already expensive devices, and prosthetic use is a continuous expense.

According to the March 2023 Prosthetics and Orthotics Gazette [37], the compensation after an accident for prosthetic feet varies significantly. For example, energy storage and return feet (DR Feet) are compensated at R9516.48 to R63921.27 depending on the user's activity level. SACH feet, a less expensive option, receive R3337.19 in compensation.

South Africa, with a population of more than 60 million, faces significant economic challenges. A third of the country is unemployed and relies on social grants of R350 per month [38], while those with applicable disabilities may receive a disability grant of R2090 per month [39]. The minimum wage is R27.58 per hour [40], which is roughly R4412.80 per month for a 40-hour workweek.

A study by Sekhampu et al. examined household expenditures, revealing an average monthly income of R1 825.25 per month and an average expenditure of R1658.41 [41]. When we weigh the average income against expenditure, it is clear that a person living on such grants could not afford a prosthetic foot when considering the prices listed in the March 2023 Prosthetics and Orthotics Gazette. A portion of their spending, 45.5%, is on food, with additional expenses on energy, insurance, household debt, and transport [41]. In particular, medical aid is absent from the list, which is reasonable since 82% of South Africans depend on public health [42]. This dependence means that prosthetic options are typically determined by what public hospitals can provide.

For those with medical aid, such as Discovery Health members, there is a payout limit of R93 550 for prostheses due to trauma [43]. Other funding options include the Road Accident Fund (RAF), which compensates road accident victims [44], and the Occupational Injuries and Diseases Compensation Fund for work-related injuries.

Given the average income and expenditure, most South Africans cannot afford prosthetics, leaving the responsibility to the public health sector. Public hospitals therefore usually prioritise cheaper options such as SACH feet to serve more patients.

A study carried out found that only about 66% of amputees returned to work after amputation [45]. From this, only 30% retained the job they had before amputation. Typically, amputees enter a new field of work that requires a higher level of education and fewer physical activities [45]. This is not a common option in South Africa, as only 7% of the population has a degree beyond a high school diploma or access to upskilling opportunities [46].

Most South Africans cannot afford a prosthetic set-up, let alone one that is advanced enough to allow them to maintain their lifestyle with multiple prosthetic feet. Currently, there was no affordable option found on the market for a prosthetic foot that can be used for multiple activities comfortably.

Although reducing the price of prosthetics could make them more accessible to some, most of the population would still struggle to afford them, and there are limited options for cost reduction of higher-functioning feet. A more pragmatic approach is therefore to attempt to broaden the functionality of a single foot design while keeping it affordable, thereby improving the value proposition for users and suppliers (be it the state or medical schemes).

1.2. Problem Statement

Currently, on-the-market feet are expensive and lack the ability to adjust stiffness and energy return, limiting the user's comfort during certain activities and therefore requiring multiple feet for optimal application. Additionally, prosthetic feet are classified into specific weight categories, meaning that significant weight gain or loss could necessitate a new foot to ensure optimal comfort and functionality. There is a need for an affordable prosthetic foot with adjustable stiffness, which could allow users to rely on a single prosthetic foot for varied activities and reduce the overall cost of assistive care. This study addresses this need by focusing on the design methodology for such a foot, rather than the development of a market-ready product.

1.3. Aim and Objectives

The aim of this study is the development of an adjustable dynamic response foot. The following objectives support the achievement of that aim:

- Investigation of how the human foot moves and how a prosthetic foot is related during the same movement.
- Determining how current prosthetic feet function, what types of feet exist, and when they would be used.
- Review on the effect the stiffness of a prosthetic foot has on the gait of the user.
- Determining what materials are typically used for prosthetic feet and if cheaper substitutes could be used.
- Evaluation of multiple design options and selection of the most suitable option, considering performance, manufacturability, and cost.
- Design and manufacture of a prototype adjustable dynamic response foot.
- Verify the prototype through comparison to the design specification and the values of commercial prosthetic feet.
- Validate the prototype by comparing the calculated data to the experimental data.

The completion of the objectives establishes the design process for an affordable and adjustable prosthetic foot.

1.4. Scope and Constraints

The study focused on the design methodology and not delivering a functional and market-ready foot. The following constraints were applied in the design process:

- The scope was limited to conceptualisation, mechanical design, and basic physical testing.
- Clinical validation, long-term durability testing, and user trials were outside the scope of this project.
- The research was constrained by the time, resources, and facilities available within a university setting.
- The prosthetic foot was designed for an 80 kg male with a size 8 shoe to simplify the design process.

1.5. Dissertation Overview

Chapter 2 provides a comprehensive review of the literature, investigating the first prosthetic limbs used, the anatomy of the human foot, and the motion that a human foot provides a person. Previous studies were explored to adopt their findings into this study.

The findings of the literature review were used for the conceptual design of Chapter 3. In this chapter, multiple concepts for the study solution are explained and reviewed. From these concepts, one concept was chosen to be taken further to the detailed design chapter, Chapter 4.

Chapter 4 provides the mathematical approach used to design the prosthetic foot. Along with the calculation method, the composite layup for the prosthetic foot was determined and documented in this chapter. The calculation method was verified by comparing the results to finite element analysis performed using Siemens NX. This chapter concludes with a cost evaluation for the prosthetic foot presented.

The testing process is briefly discussed in Chapter 5, leading to the presentation of the experiment results in Chapter 6. Here the prosthetic foot designed was verified by comparison to calculations and experimental results.

Finally, the study is concluded within Chapter 7. Additional studies and design changes are recommended, which would further improve the prosthetic foot presented by the study.

Chapter 2 : Literature Review

2.1. Chapter Overview

This chapter provides an overview of previous work from literature regarding the design of prosthetic feet. Firstly, the history of prosthetics is briefly looked at in order to understand the initial purposes of prosthetics, how this changed over the years, and what should be considered when designing a prosthetic foot.

To aid in understanding the design process of prosthetic feet, the anatomy of a human foot and the gait cycle of the foot (human and prosthetic) are investigated. This is an important consideration, as understanding where a normal foot sees its maximum load during walking or how much force a foot could experience during activities is important for the design process. Also, it is important to know how a dynamic response foot functions during the gait cycle and what components a dynamic response foot typically consists of.

With the foot stiffness and the adjustability thereof being an important part of this study, other studies were examined on the effect of forefoot stiffness on prosthetic feet, and the most important findings are discussed. Along with this, studies on how to implement variable stiffness of prosthetic feet were investigated.

Lastly, the common material options used on prosthetics were investigated. An in-depth study of composite materials was conducted. Composite materials, usually carbon fibre or glass fibre, are commonly used in prosthetic feet. Therefore, it is extremely important to understand how composite materials are used in the design of a prosthetic foot.

2.2. History of Prosthetic Limbs

In history, prosthetics have always been used to replace missing limbs lost for congenital or unnatural causes. Figure 3 shows one of the first prosthetics, a wooden toe attached with leather straps, found in Egypt, dating back to 2000 BC [47], [48]. The original prosthetics were not necessarily used for practical purposes but for cosmetic reasons [48]. The first iterations of the prosthetics were usually made of wood or metal and attached to the remaining limbs with straps and nails [48].



Figure 3: Wooden toe prosthetic found in Egypt [47]

The practicality of prosthetics changed during the Renaissance era. Iron and steel became the go-to materials, as they were more durable, and the prosthetic design was shifted toward increasing mobility and quality of life rather than appearance [47], [48]. The French surgeon Ambroise Paré made great strides in the field of prosthetic limbs in the sixteenth century. Paré saw many soldiers become amputees due to shattered arms and legs and was disturbed by the reactions to their new realities, some even taking their own lives. Paré's limbs were unique in the sense that they attempted to restore the amputee's life to what they considered normal. He created limbs that functioned to a certain extent in the same way as a biological limb would. For example, in Figure 4, his design for a mechanical knee that could lock when the patient was standing and bend according to the patient's will can be seen [48].

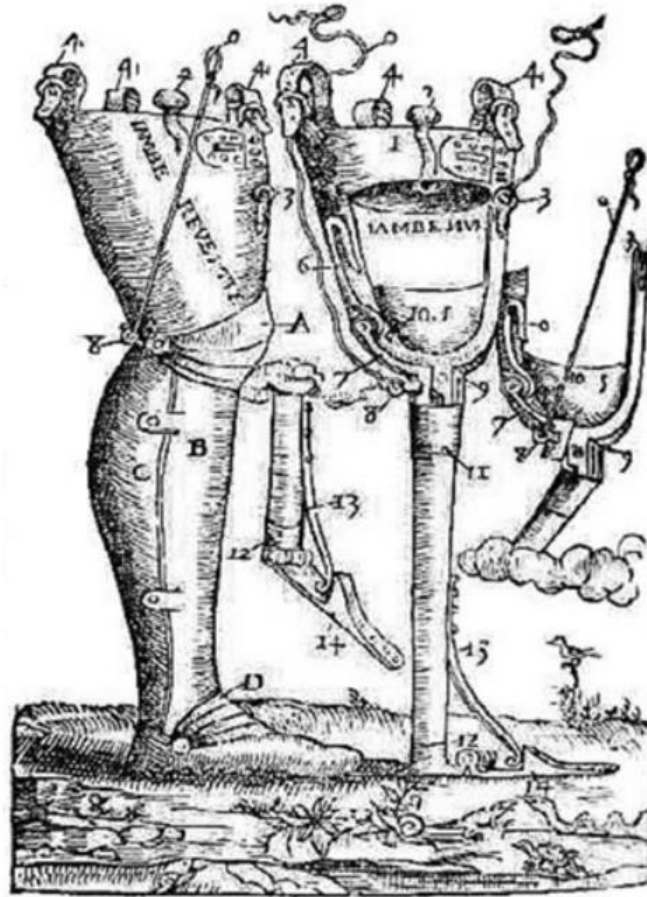


Figure 4: Ambroise Paré's design for an above-the-knee prosthesis [48].

Greater strides were made in the twentieth century, as material and manufacturing advances made it possible for more flexible and responsive feet, allowing for more natural walking motions. The first dynamic response foot was developed by Van Phillips, who also invented the flex foot brand in the 1960s. He later also invented the first running blade in 1989 [50], [51]. Van Phillips was an amputee himself and was able to test many models himself, helping him to redefine the final design. The foot was made of carbon fibre and had an L-shape with a heel. This allowed for natural responsive movement and energy return as the foot acted like a spring [51], [52]. His work and discoveries had a massive impact on how feet are designed today. Currently, there are many different types of prosthetic feet, each designed for a specific need of the wearer, helping with a natural gait, different activities, and some are even able to adjust to different environments.

Similar to Paré's design philosophy, modern prosthetic legs and feet are functional before they are cosmetic. Understanding the movements of a real human foot aids in understanding what functionalities are required for a prosthetic foot and what movements can and should be mimicked by the prosthetic foot.

2.3. Anatomy of a Human Foot

The human foot is a complex structure consisting of 26 bones, 33 joints, and more than 100 muscles [53], [54]. The foot is divided into three parts: the forefoot, the midfoot, and the hindfoot, as illustrated in Figure 5. The forefoot is made up of the toes and bones that connect the toes to the midfoot. The forefoot provides balance and stability during activities such as walking and running. The force required to jump off the ground is supplied by the forefoot. The forefoot is also subject to the greatest stress and impact compared to the rest of the foot [55].

The midfoot is made up of the arch bones that are supported by various muscles and tendons. The midfoot is important, as it is responsible for the transfer of forces from the heel to the forefoot during walking [56], [57]. The hind foot consists of two large bones: the heel bone (calcaneus) and the ankle bone (talus) [58]. The heel bone is the largest foot bone and therefore serves as the foot's foundation. The ankle bone is attached to the heel bone by the tendons and ligaments, and the joint is responsible for the dorsiflexion and plantarflexion of the foot [54], [56].

Dorsiflexion is the upward motion, and plantar flexion is the downward motion of the foot, or, as illustrated in Figure 6, the rotation around the y-axis [59]. A normal ankle can have a dorsiflexion rotation between 10° and 20° and a plantarflexion between 40° and 55° [59].

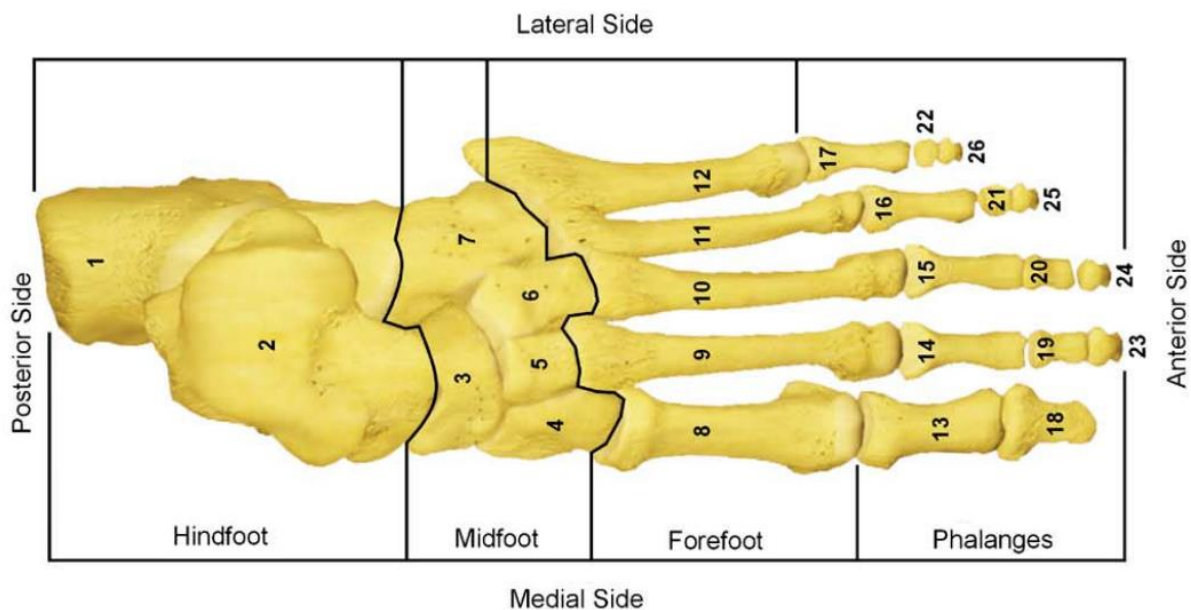


Figure 5 : Bone structure of the human foot [53]

In addition to dorsiflexion and plantarflexion, the human foot is also capable of inversion, eversion, external axial rotation, and internal axial rotation [59], [60]. Inversion and eversion describe the ability to rotate around the x-axis. Inversion is the inward rotation of the foot, and eversion is the outward rotation, as illustrated in Figure 6 [59], [60]. The average human foot has an inversion angle of 23° and an eversion angle of 12° [59]. The external axial rotation and internal axial rotation are the rotations about the z-axis of the foot, away and towards the body. The external axial rotation is the outward rotation, and the internal axial rotation is the inward rotation [59], [60], as illustrated in Figure 6.

Every foot is extremely important for balance, prevention of injuries, adaptation to different surfaces, obstacles, and terrains, and uniform distribution of weight evenly during activities such as walking or running. For the purpose of this study, it is therefore required to understand the motions of the foot and to know which motions are the most crucial to address in the design of the foot. Next, the gait cycle is studied to understand the movements and conditions of the foot while walking.

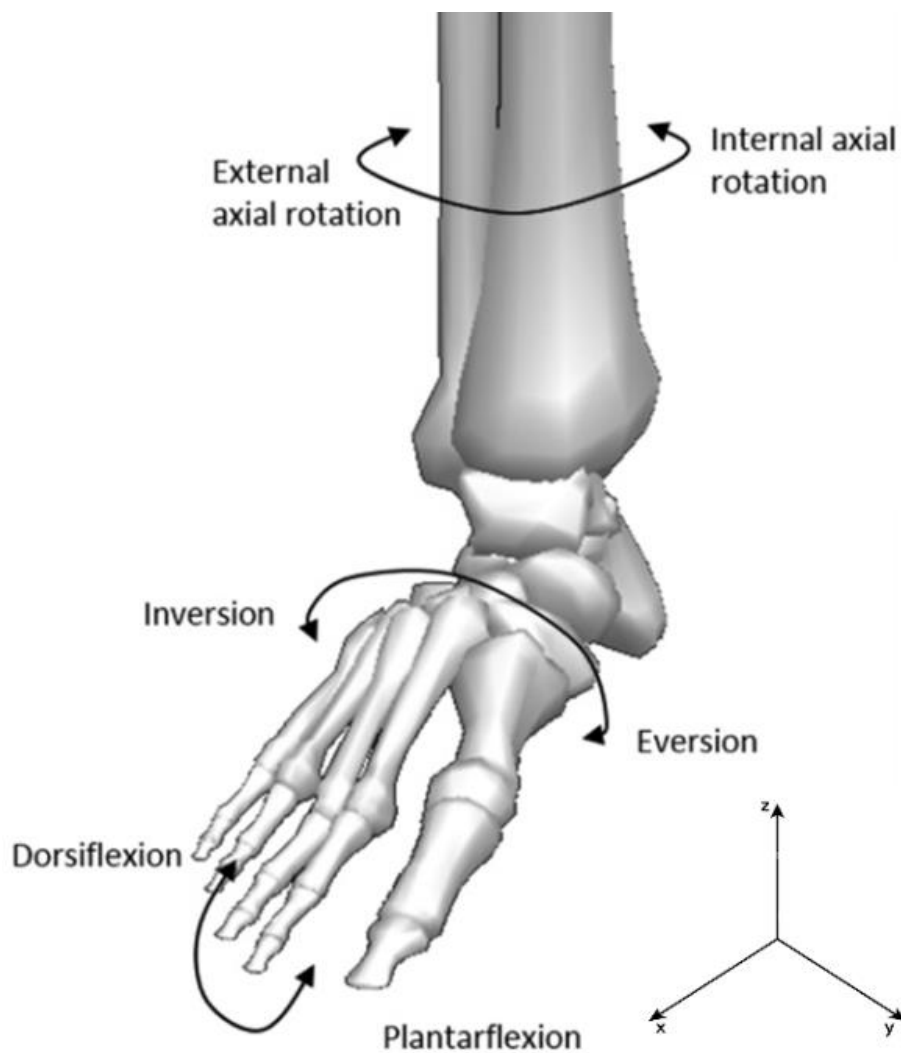


Figure 6: Movements and rotations of a human ankle plantar flexion [59].

2.4. The Foot During the Gait Cycle

Gait refers to the manner in which a person walks and the phases through which a person transitions while walking. The bending and movement of the feet, knees, and hip joints play an important role in the study of the gait cycle, as seen in Abbass and Abdulrahman [61]. Their study showed that the angles of the knees and hips were different in each of the three gait cycles during walking [61]. They also found that the step lengths on the right side were larger than those on the left.

There are two phases of the gait cycle through which the hip joint and knee bend, namely the stance phase and the swing phase [54], [62]. Within these two phases, there are sub-phases that better explain how the foot acts during the gait cycle.

Stance Phase

The first phase is the stance phase. The stance phase occupies about 60% of the gait cycle [54], [62], [63]. The largest actions that occur within this phase are (i) opposite toe-off, (ii) heel rise, and (iii) opposite initial contact. These actions are illustrated in Figure 7 on the outside of the diagram. Within the stance phase, the subphases are (i) loading response, (ii) mid-stance, (iii) terminal stance, and (iv) pre-swing [62], [64]. The subphases are represented in the centre of the diagram in Figure 7.

The loading response sub-phase is where body weight is distributed to the leg, which is the stance phase. Here, the leg is placed in front of the body with the heel on the ground. The heel acts as a point of rotation until lifted as the foot moves through the stance phase [62], [63].

The ankle allows for dorsiflexion, and the foot rotates around the heel to a flat position in the midstance subphase. During the mid-stance phase, only one leg is on the ground [54], [62], [64].

The terminal stance phase is also done with only one leg grounded. In this phase, the heel is lifted, and the rotation now occurs on the forefoot. This phase ends with the opposite side leg contacting the ground [62], [64]. The final subphase of the stance phase is the pre-swing phase. Here, the body weight is transferred to the other leg, which is now experiencing its loading phase [62].

Swing Phase

The second phase is the swing phase, occupying 40% of the gait cycle [62], [63]. The two noticeable actions within the swing phase are (i) feet adjacent and (ii) tibia vertical. These actions are illustrated outside of the diagram in Figure 7. The subphases within the swing phase are (i) initial swing, (ii) mid-swing, and (iii) terminal swing.

In the initial swing phase, the foot is completely off the ground and is moved forward by hip and knee rotation [62], [64]. The initial swing phase is followed by the mid-swing phase. This phase starts adjacent to the opposite leg and ends when the leg is in front of the other [62], [64]. The final phase is the terminal swing phase. This happened after the tibia was vertical and ended with the heel touching the floor [62], [64].

The (i) initial contact action and (ii) toe-off actions are what split the two phases. Initial contact is the start of the stance phase and the end of the swing phase, and the toe-off action is the end of the stance phase and the start of the swing phase. It is also important to note that the gait cycles of the two legs overlap. The gait cycle of the opposite leg would start at ‘opposite initial contact’, as indicated in Figure 7.

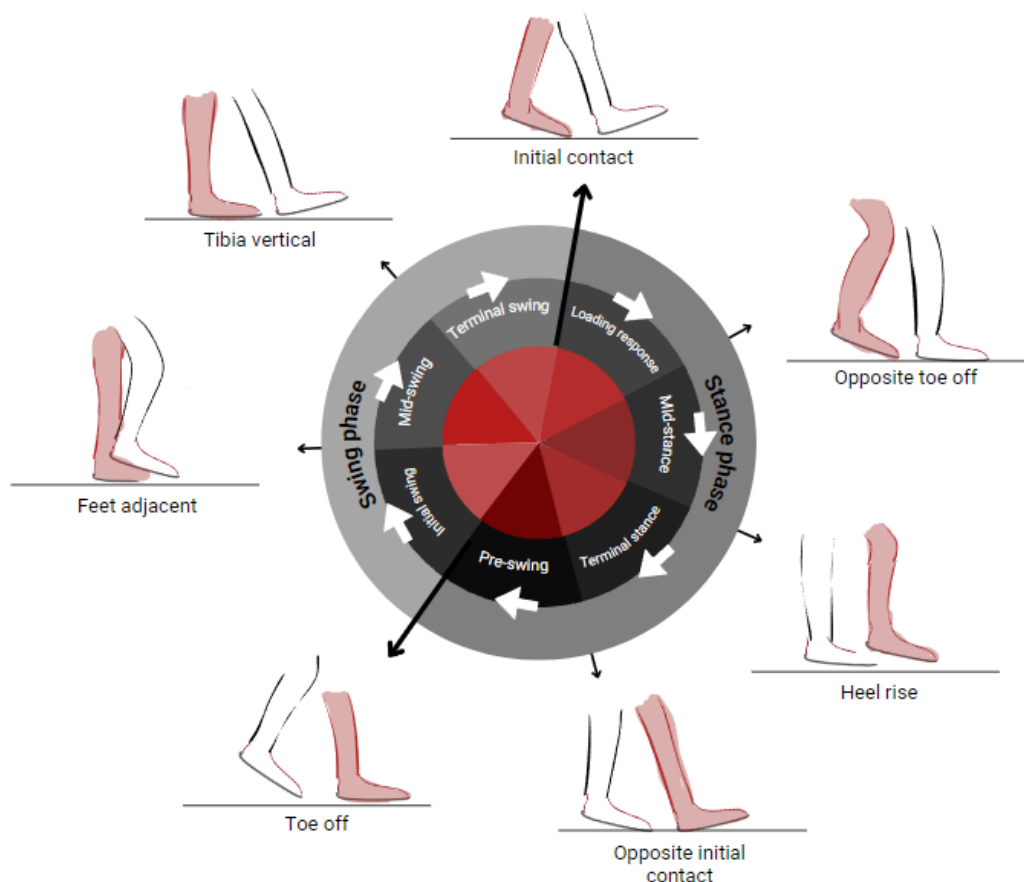


Figure 7: Events in the gait cycle (adapted from Karb et al. [62])

Ankle During the Gait Cycle

When considering what the foot and ankle do during the gait cycle, the stance phase is the most important. Figure 8 illustrates the ankle during the stance phase, starting with the initial contact action illustrated in Figure 8A and moving on to Figure 8B by plantarflexing and finding itself in the mid-stance subphase [54], [63]. From here, the shin moves over the ankle, as seen in Figure 8C, allowing the toe to lift, as illustrated in Figure 8D [63]. Figure 9 shows how the foot is loaded with vertical forces during the stance phase. As Figure 9 illustrates, the maximum load is observed during the loading response and the push-off. The forces experienced here are almost 1.2 times the body weight of a person [63], [65]. The reason why the maximum load is seen at these two points is that only one foot is on the ground.

The fitting of a user with the proper prosthetic is crucial to maintaining a balanced gait cycle. What prosthetics a person will require depends on the level of amputation of the person, and the desired level of activity. Next, the motion of a dynamic response foot during the gait cycle is investigated along with the components present in a dynamic response prosthetic foot.

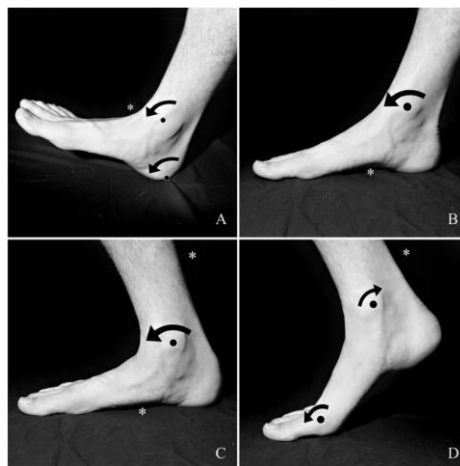


Figure 8: Motion of a foot during the gait cycle [63]

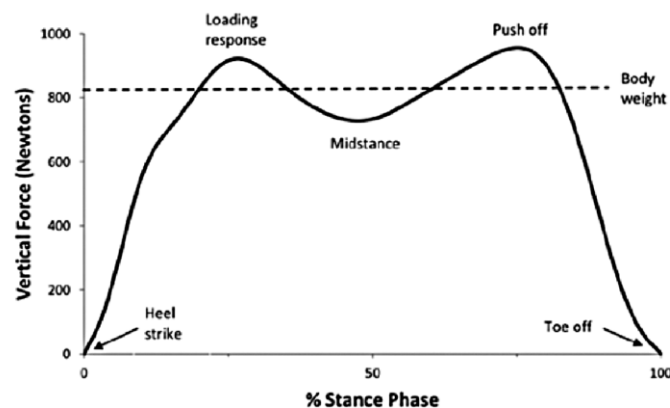


Figure 9: Vertical forces acting on the foot during the stance phase [63]

2.5. Dynamic Response Prosthetic Feet

2.5.1. Motion of a Dynamic Response Foot

The dynamic response foot (DR), or the energy storage and return foot, seen in Figure 2, is commonly reserved for the most active users. It allows more mobility and is ideal for varying walking speeds, rapid direction changes, and long distances [27], [28]. This prosthetic can store and release energy as the user walks, making walking with this prosthetic look and feel more natural to the user's gait. DR feet are usually made of composite materials that allow some flexion, allowing for energy build-up [31].

When the prosthetic foot meets the ground, as shown in Figure 10(a), the heel compresses, as it is the first action in the stance phase. The heel is designed to compress and dampen the initial shock. The heel compression releases enough force to propel the foot to the mid-stance phase, illustrated in Figure 10(b), and allows the foot to roll over to the heel release action, as seen in Figure 10(c). As the heel lifts, the toe compresses, and energy is stored in the prosthetic foot. This allows for the final part of the stance phase as the foot rises from the ground, as seen in Figure 10(d).

The split toe dynamic response foot (Figure 2) allows inversion and eversion movements, improving stability, and mimics the movement of a real ankle [27], [33]. Dynamic response feet tend to be relatively lightweight since they are made of composites. They are also on the more expensive side of the prosthetic foot market [66].

Next, the different components of a DR foot were investigated to understand which components should be considered for this study.

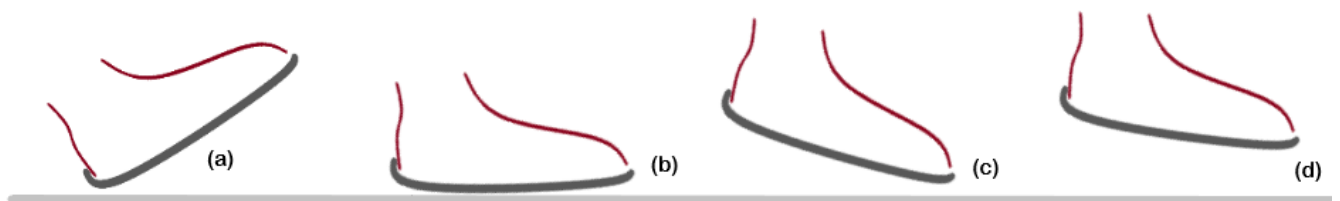


Figure 10: Foot motion when walking.

Foot Shell or Cover

The foot shell is a cover that covers the prosthetic foot, usually with the appearance of a normal foot, as illustrated in Figure 11 and Figure 12. In addition to aesthetic function, the shell helps fill the user's shoe, preventing the prosthetic foot from sliding around in the shoe. It is important that the foot shells are made of a soft and flexible material so that they can be easily removed from the prosthetic if cleaning or maintenance is needed. Polyurethane foam is an ideal material for the shell, as it is lightweight, flexible, and easily mouldable [71].

A final component that is not present on all DR feet is the sole plate. The sole plate keeps the composite blades in place and helps fill out the foot shell. An example of a foot with a sole plate is Ottobock's Trion foot, shown in Figure 12.

From this section, it is clear that the following components will be the most important to consider in this study: the male pyramid, the toe blade, and the heel blade. The foot shell is an important part of the prosthetic foot; however, for the purpose of this study, the cosmetic aspect of the foot and fitting the foot inside a shoe are not a priority.

As mentioned throughout the chapter, one of the most important aspects of the DR foot is its ability to store energy. The energy stored is due to the deflection of the foot. Therefore, the stiffness of the prosthetic foot is crucial due to its effect on bending and thus energy storage. Next, studies that focus on the stiffness of prosthetic feet and the effect of stiffness are looked at.



Figure 12: Ottobock Trion foot (right) with foot shell (left) [72].

2.6. Studies on Stiffness Effect

The effect that foot stiffness has on the gait cycle is very important. An adjustable prosthetic foot will offer variable levels of stiffness, so it is important to review the effects that different prosthetic feet with different stiffness levels have on the user.

Houdijk et al. investigated how increasing the push force of a dynamic response foot affects centre of mass velocity and step length symmetry without compromising stability. The study involved 15 transtibial prosthetic users, with an average age of 55.8 years and a weight of 86 kg, all experienced with dynamic-response feet [73].

Over two days, participants walked with their everyday dynamic response feet and a SACH foot, which is typically recommended for less active users due to its low energy response [73]. The dynamic response foot generated greater energy return, leading to higher centre-of-mass velocity, improved symmetry, and preserved stability. This showed a positive impact on gait [73].

Klodd et al. investigated the impact of forefoot stiffness on gait using five prosthetic feet with varying flexibility and 14 participants [74], all with at least one year of experience using dynamic response feet. The feet were identical in weight and shape, with different flexibilities achieved by modifying the forefoot. The feet used can be seen in Figure 13.

The participants had to walk on a flat 10-metre walkway at their normal walking speeds with each foot in random order. The study found that the more flexible feet did not allow a push force that would propel the prosthetic sufficiently. This had a chain reaction, as it caused a reduction in the movement of the prosthetic ankle and the length and caused a noticeable limp [74]. The study found that there was a wide range in forefoot flexibility where no effects could be observed on the gait [74]. Large differences were seen in the gait cycle between the three most flexible feet (F1, F2, and F3) and small differences between the three least flexible feet (F3, F4, F5). The results supported the idea that overly flexible feet contribute to a noticeable limp, which could impact the sound limb.

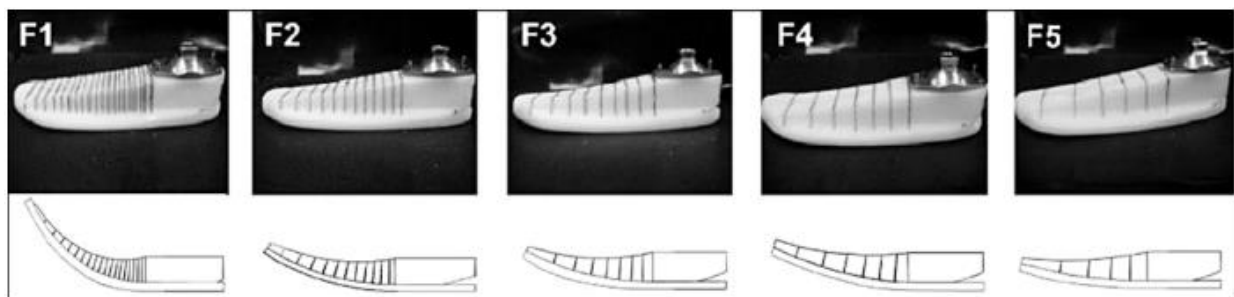


Figure 13: Feet used for the Klodd et al. study [74]

Koehler-McNicholas SR et al. [75] investigated how prosthetic feet, which are currently prescribed to people, perform under weighted walking conditions. Three participants, with an average age of 39 years, younger than that of the Klodd and Houdjik studies, were used in the study. Each participant had at least two and a half years of experience walking with prosthetics.

The study involved testing nine available prosthetic feet by measuring their stiffness. Each foot was placed in a load frame, as seen in Figure 14, and subjected to increasing force at 100 N/s intervals and held at maximum value for one second before decreasing at 100 N/s intervals and held at 0 N for one second. The maximum value was the value of the simulated body weight of a subject plus a weighted vest with a value of 22 kg [75]. Each cycle was repeated nine times on each foot, and the applied force and deflection of the prosthetic were recorded. The results of one of the participants can be seen in Figure 15, which shows displacements ranging from 25 to 50 mm under an 800 N load.



Figure 14: Stiffness experiment [75]

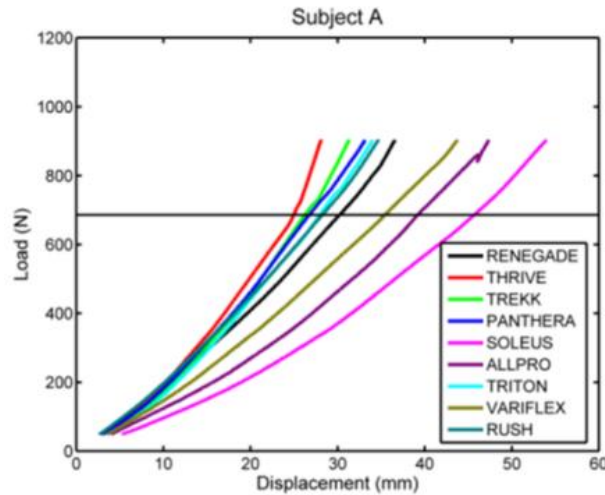


Figure 15: Displacement of nine prosthetic feet [75]

Participants in the Koehler-McNicholas study were fitted with all nine feet in random order and walked with and without a weighted vest while being monitored. The study found that less stiff feet provided a better energy return between unweighted and weighted walking [75].

These studies highlight the critical role of prosthetic foot stiffness. Both excessive flexibility, as seen in the Klodd et al. study, and stiffer feet, as seen in the Koehler-McNicholas study, can lead to discomfort and energy loss. Adjustable foot stiffness would allow for better customisation to the user, improving comfort and efficiency. Next, studies on adjustable stiffness were reviewed to inform future design improvements.

2.7. Studies on Prosthetic Feet with Adjustable Stiffness

A study by Glanzer and Adamczyk highlighted the challenge that current prosthetic feet cannot adapt to daily activities, with fully actuated robotic ankle units (microprocessing feet) being the best solution [76]. Another option is to make prosthetic feet with hydraulic elements, but the hydraulic elements cause power loss through the movement of the ankle.

The study offers the variable stiffness foot (VSF), illustrated in Figure 16, which uses a movable fulcrum to adjust the stiffness of the foot. The fulcrum allows the blade to deflect at different points, changing the keel length. The fulcrum is moved by a motor and a drive belt during the swing phase of the gait cycle to minimise power consumption. This design, which uses a composite blade, provides adjustable stiffness for improved adaptability.

The beam model in Figure 17 illustrates how the fulcrum (B) moves along the length of the beam, changing the overhang (a). The predicted stiffness range for the VSF is between 9 and 40 N/mm, allowing for adaptable stiffness depending on the position of the fulcrum.

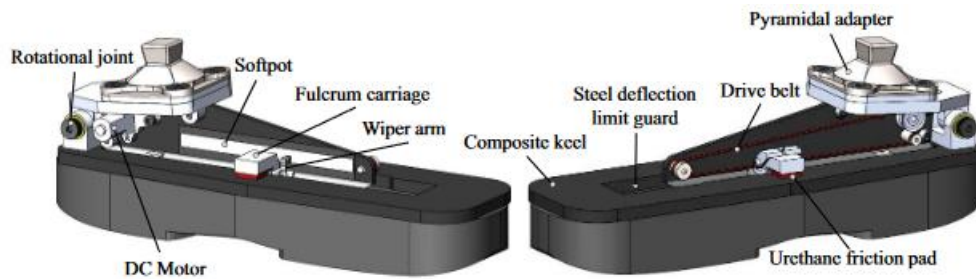


Figure 16: Variable stiffness foot [76]

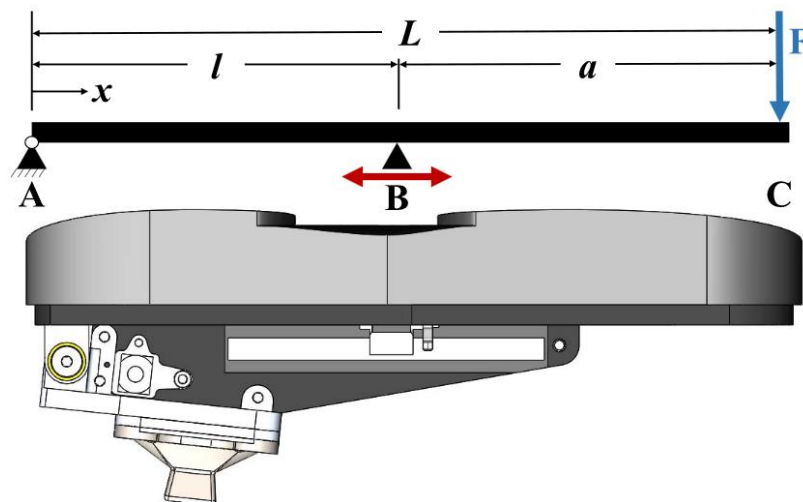


Figure 17: VSF beam model [76]

The VSF model was tested both mechanically and with human participants. While the machine could only apply 700 N (despite the foot design for 1200 N), the stiffness range was measured between 10 and 32 N/mm with the additional supporting foam. Three participants tested the foot and reported positive differences when climbing ramps, however, the energy return decreased as forefoot stiffness increased. The drawbacks of the VSF were a one-stride delay when adjusting the fulcrum and an overly large foot size, which made participants' gait unnatural.

A study by Dong et al. [77] addressed the high power consumption of powered ankle prostheses and highlighted the importance of variable stiffness to adjust to walking speeds. The study introduced an adjustable stiffness foot using a spring and ball screw mechanism, which performed well in low-speed walking. The concept is illustrated in Figure 18.

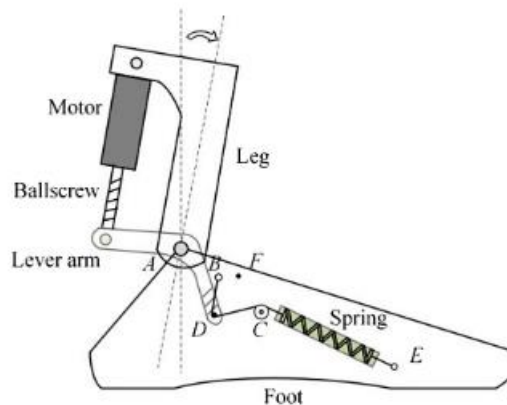


Figure 18: Powered ankle-foot prosthesis [77]

Christophe Lecomte et al. aimed to design a prosthetic ankle, the Variable Stiffness Ankle (VSA), which would allow a stiffness range of 20% for plantar and dorsiflexion movements and be tested by machine [78].

The VSA, shown in Figure 19, used two leaf springs and a commercial dynamic response foot. Under load, the blades will deflect and rotate around the pivot, and adjusting the leaf spring position changes the foot stiffness by limiting pivot rotation [78].

Some of their design constraints were to keep the foot under 1.5 kg, not exceed a height of 180 mm, and use a size 27 prosthetic foot. The Soleus leaf spring was made of pre-impregnated glass fibres, while rigid components were aerospace-grade aluminium [78]. Their calculations relied on deflection and stiffness formulas, using Young's moduli of 97 GPa for the carbon fibre blade and 50 GPa for the glass fibres. Machine testing was applied at 1250 N at 15° and 20° to the heel and toe, showing that the VSA could achieve a 50% change in stiffness.

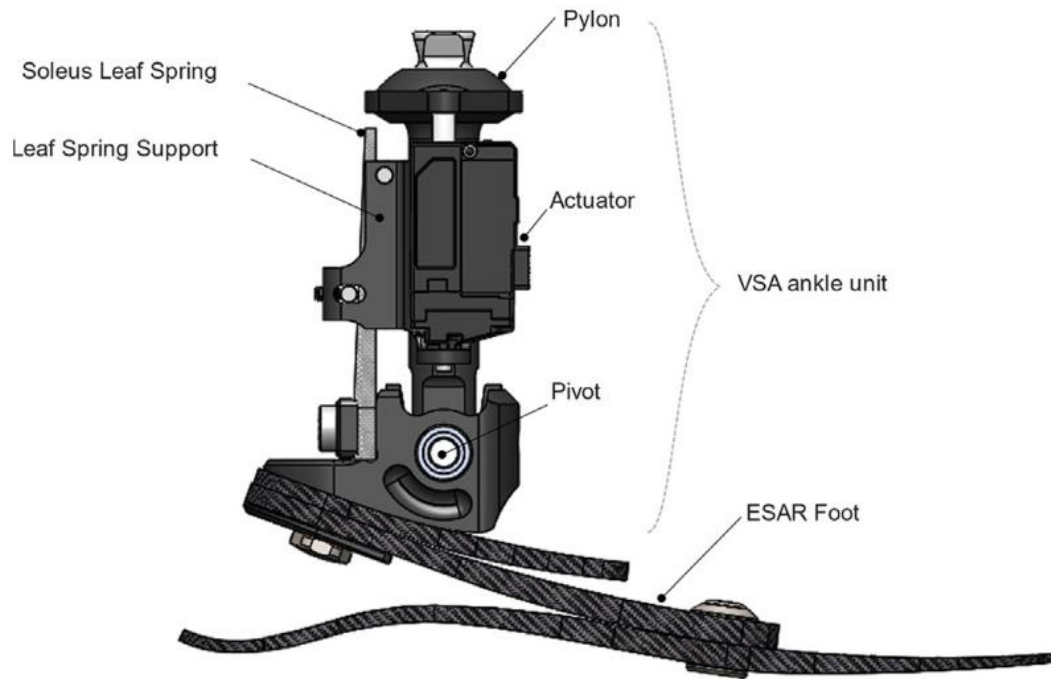


Figure 19: Variable Stiffness Ankle design [78]

Section 2.4 highlighted that prosthetic feet can experience nearly 1.2 times the user's body weight while walking. The VSF model was tested with static loads of 700 N, with a designed capacity of 1200 N, providing a 1.7 safety factor. Similarly, the VSA foot withstood 1250 N during testing. According to ISO 10328, a prosthetic foot designed for a 70 kg user (~ 700 N) should withstand 1610 N, a load that has not been tested in any of the studies discussed here. However, it is likely that these studies did not aim to develop market-ready prosthetic feet.

Although meeting market standards was not the objective, understanding these standards remains critical. Each study underscored the importance of adjustable stiffness in prosthetic feet, focusing on how stiffness impacts energy storage and user energy expenditure. The findings of these studies align with those on forefoot stiffness in Section 2.6, where participants reported greater comfort and adaptability with adjustable foot stiffness. In general, these studies confirmed the need for adjustable prosthetic feet, which enhance user comfort and provide increased freedom of movement.

2.8. Materials of a Prosthetic Foot

2.8.1. Adapter Materials: Stainless Steel, Aluminium, and Titanium

The materials used for the pyramid connector play a crucial role in the strength, weight (density), corrosion resistance, machinability, and cost of a prosthetic foot. The pyramid is a central part of the prosthetic foot that helps connect to the prosthetic knee or socket. This section takes a closer look at the selection of materials for the pyramid connector based on a combination of material properties and structural requirements, with the goal of ensuring strength, durability, and manufacturability.

Strength: The connector is required to withstand forces and torques encountered during the gait cycle, which can reach up to 1.2 times the user's body weight during walking [63], [65]. Strength is a structural outcome which is influenced by the material's yield strength and ultimate tensile strength. A suitable material should therefore possess mechanical properties that allow adequate structural performance with a sufficient safety factor. The required loading conditions and test procedures for prosthetic components are specified in ISO 10328, which outlines the standards for structural testing. A summary of these requirements is provided in Appendix F.

Density: The weight of the connector contributes to the overall weight of the prosthetic foot. A heavier foot leads to an increase in energy expenditure during walking [79], [80]. Reducing the weight of the foot contributes to comfort and gait efficiency.

Corrosion Resistance: The material must be resistant to corrosion to ensure durability, especially since the prosthesis may be exposed to moisture or particles.

Machinability: Materials with good machinability can be manufactured more efficiently and cost-effectively. Machinability must, however, be balanced against strength, as more machinable materials may have lower strength [81].

Material Costs: Both raw material cost and processing cost should be considered in the selection. The material must meet performance requirements while remaining within cost constraints for the intended affordability of the design.

The material properties of the most-used alloys of titanium, aluminium and stainless steel and a comparison of the properties for the three materials can be seen in Table 1.

Table 1: Pyramid material comparison table [82].

Material	Yield Strength (MPa)	Ultimate Tensile Strength (MPa)	Density (kg/m ³)
Titanium (Ti-6Al-4V)	924	1000	4430
Aluminium (6061-T6)	255	290	2710
Stainless steel (304)	207	517	7860

Among the three materials considered for the pyramid connector in Table 1, titanium stands out as the strongest, with a yield strength of 924 MPa and an ultimate tensile strength (UTS) of 1000 MPa. High-strength materials like titanium enable the design of smaller or thinner components while still maintaining structural integrity, which can contribute to weight reduction. Stainless steel follows with a UTS of 517 MPa, offering a good balance of strength and affordability. Aluminium, while the lightest of the three, has a lower UTS of 290 MPa. This means that, to achieve equivalent structural performance, aluminium components may need to be bulkier, which could contradict some of its weight advantages. These strength properties are crucial for ensuring that the connector safely withstands the loading conditions defined in ISO 10328 while also meeting the dimensional and weight constraints of a prosthetic foot design.

Although aluminium has lower strength, it offers a relatively high strength relative to its weight due to its low density, less than half that of stainless steel, which makes it attractive for applications where weight reduction is a priority. However, to compensate for its lower strength, more material may be needed, potentially offsetting some of the weight advantage. Titanium is also an attractive option in terms of density, being almost 60% as dense as stainless steel [83].

Both titanium and stainless steel excel at corrosion resistance, making them ideal for prosthetics that could encounter moisture [83], [84], [85]. Stainless steel may be vulnerable in chloride-rich environments where its protective layer can degrade [86]. Aluminium also has corrosion-resistant properties but may lack them when submerged in water for extended periods of time [87]. The corrosion resistance depends on the pH levels to which it is subjected, being ideal only in a certain range [87], [88].

The machinability rating uses 160 Brinell B1112 steel as the reference point (100%) to compare other materials. Values higher than 100% are found in soft materials and more easily machined, whereas values lower than 100% are found in harder materials [89]. Of the three materials considered, aluminium is the material that is easiest to machine, with a machinability rating of 190% [90]. Stainless steel has a machinability rating of 40% [90]. This means that stainless steel requires more machining time and thus also has a higher machining cost. Titanium is a hard material but easily bendable, which means that the material takes a toll on the machining tools and must be cut slowly, as the material can bend under cutting forces [91]. This contributes to the high cost and long machining time.

Aluminium is the most affordable and widely available material among the three. However, due to its lower strength, more aluminium may be required to achieve the same performance as stainless steel. Despite this, stainless steel remains more expensive overall. [92]. Titanium stands out as the most expensive material of the three [93].

2.8.2. Orthotropic materials: Carbon Fibre, Kevlar, and Fibreglass

Composite materials are frequently used in prosthetic applications due to their high strength-to-density ratio, shock resistance, and energy storage. The properties of prosthetic components, such as feet, are influenced by factors such as the type of composite material used, the number of layers, the orientation of the layers, and the composition of the resin. Composite materials such as carbon fibre, Kevlar, and fibreglass are the most widely used composites for prosthetic feet. These materials are classified as orthotropic materials, characterised by their different material properties along three perpendicular axes (x, y, and z axes) [94], [95]. Properties such as modulus of elasticity, yield strength, Poisson ratio, shear modulus, ultimate tensile strength, and thermal coefficients can vary in different directions [94]. This divergence separates orthotropic materials from isotropic materials, such as metals or polymers, where material properties are consistent in all directions [94], [95].

To achieve these properties, various tests are conducted on orthotropic specimens. Tensile, compression, and shear tests are performed in various directions to determine the unique properties of the material in each direction. For example, the ASTM D695 test method is ideal for determining composite compression properties, while the D3039 method is ideal for testing composite tensile properties [96], [97].

The unique characteristic of orthotropic materials comes from their microstructure. A composite material consists of two key elements: reinforcement fibre and the matrix material, where the fibre materials (such as carbon fibre, fibreglass, or Kevlar) are complemented by the matrix material (usually resin or epoxy), which combine to form the composite [98]. The reinforcement fibres contribute to the strength, stiffness, and density properties of the composite, and the matrix material provides shear properties. Figure 20 illustrates how fibres and the matrix work together.

Composite materials are available in different fibre orientations, primarily unidirectional and bidirectional. Unidirectional (UD) composite fibres run in a single direction and are typically held together by thin perpendicular fibres. This structure results in high strength and stiffness along the primary fibre direction (0° layup) while being significantly weaker in the perpendicular (90°) direction [99].

Bi-directional (BID) composite fibres, on the other hand, have fibres running in both the 0° and 90° directions. This woven or cross-structured arrangement provides improved mechanical performance along those directions compared to unidirectional layups, which are optimised for strength in only one direction [99].

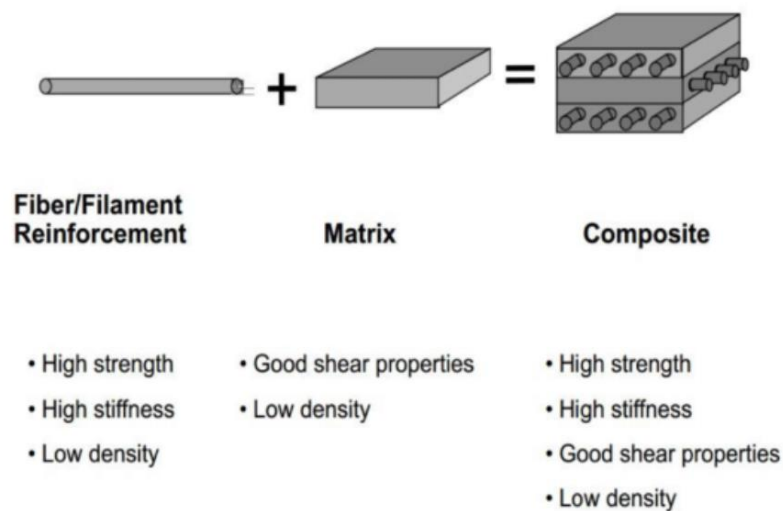


Figure 20: The roles of the fibres and matrix of a composite material [98]

The strategic orientation of the fibre layers significantly influences the performance of the composite. Given the orthotropic nature, rotating the layers means that the directional properties change depending on the reference point. A 0° ply layup runs parallel to the reference datum line and is useful for handling parallel bending stresses. Plies laid at 45° are ideal for handling shear stresses, and 90° layups are great for handling perpendicular stresses [70]. Figure 21 illustrates how the angles of layers could differ within a composite ply layup.

Combining orthotropic materials can create a composite layup with the ideal material properties for a prosthetic foot. Carbon fibre has a high stiffness, which makes it more resistant to deformation [70], [100]. Although this does not directly align with prosthetic applications that require displacement and energy storage, carbon fibre excels with a weight-to-strength ratio, an ideal quality for prosthetics [100]. The main drawback of carbon fibre is that it is a very brittle material, making it crack or fracture under too much stress [101]. It is also an expensive material [102].

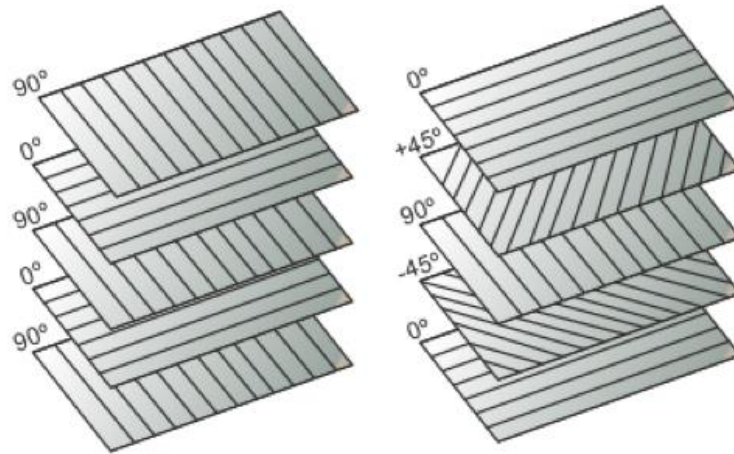


Figure 21: Example of composite ply layups illustrating different layup angles [103]

Unlike carbon fibre, fibreglass is a cost-effective and easy-to-manufacture alternative to carbon fibre [70], [101]. Its enhanced flexibility, although fibreglass has a lower strength than carbon fibre and Kevlar, remains an attractive quality for a prosthetic solution. However, caution is warranted for overflexibility of prosthetic feet, which can lead to energy loss on the toe during the gait cycle [70].

Kevlar exceeds carbon fibre's strength and is recognised for its application in bulletproof vests and aircraft parts [104]. Being the lightest material of the three and more flexible than carbon fibre [70], [101], Kevlar has ideal attributes for prosthetic use. Nevertheless, vigilance for over-flexibility must be kept in mind. Kevlar is, however, the most expensive of the three fibre types, which can negatively impact the affordability of components using it in construction [70], [101].

2.9. Chapter Summary

In this chapter the history of prosthetic limbs were briefly described, followed by the anatomy of the foot. The focus of the bone structure was on the three parts of the foot (the forefoot, midfoot, and heel) and how each of these sections allowed movement of the foot. The most important movements were the upward and downward movement (dorsiflexion and plantarflexion) and the inward and outward rotation (inversion and eversion) of the foot. Understanding the movements of the foot was required when further investigation of the foot's motion during the gait cycle was performed.

With the importance of the gait cycle established, the motion of dynamic response feet during the gait cycle was investigated. Multiple studies with a focus on the stiffness of a prosthetic foot and what effect it has on a person's gait were studied, and it was clear that a foot that was too stiff and not stiff enough could negatively affect a person's gait and comfort while walking.

Studies that investigated prosthetic feet with adjustable stiffness reported participants were able to walk comfortably at different walking speeds and complete obstacles such as ascending and descending ramps with more ease than a prosthetic foot with constant stiffness. These studies highlighted the comfort of walking with an adjustable prosthetic foot.

Lastly, the materials used typically for prosthetic feet were looked at. Typically carbon fibre and fibreglass are used for the composite blades of a prosthetic foot, as they have a high strength-to-weight ratio and can be quite flexible. Titanium or stainless steel is commonly used for the machined parts.

In the next chapter, concept designs for a more cost-effective and adjustable stiffness prosthetic foot will be discussed and considered.

Chapter 3 : Concept Development and Evaluation

3.1. Chapter Overview

Within this chapter, the multiple possible design solutions that were evaluated are discussed. For clarity, the chapter starts with a brief explanation of the required theoretical knowledge needed to explain how each concept functions. Thereafter, three different concepts are presented.

Concept 1 and Concept 2 present a prosthetic foot with a secondary moving blade, whereas Concept 3 presents the addition of gas springs to the prosthetic foot. The manner in which the stiffness adjusts within the three chapters is explained within the chapter.

The two main criteria by which concepts were evaluated are overall functionality and affordability. Functionality determines if a specific concept can perform as required, that is, provide the required levels of stiffness while being adjustable in some manner. The amount of composite blades, the finishing of composite blades, the number of parts requiring machining, the complexity of the machining, and whether or not additional parts were required were all compared when considering the high-level costing of various concepts.

3.2. Foot as a Simple Cantilever Beam

When pressure is applied to the forefoot, the foot bends, storing energy. As the toe lifts off the ground, the foot returns to its original shape, only to be bent again throughout the gait cycle. On the most basic level, a prosthetic foot functions as a cantilever beam with an applied force. Figure 22 illustrates this concept by simplifying the prosthetic foot as a cantilever beam. For further clarity, Figure 23 presents a basic cantilever beam with a perpendicular applied load. Assuming the force is perpendicular simplifies the mathematical equations, making the design concepts easier to understand and evaluate.

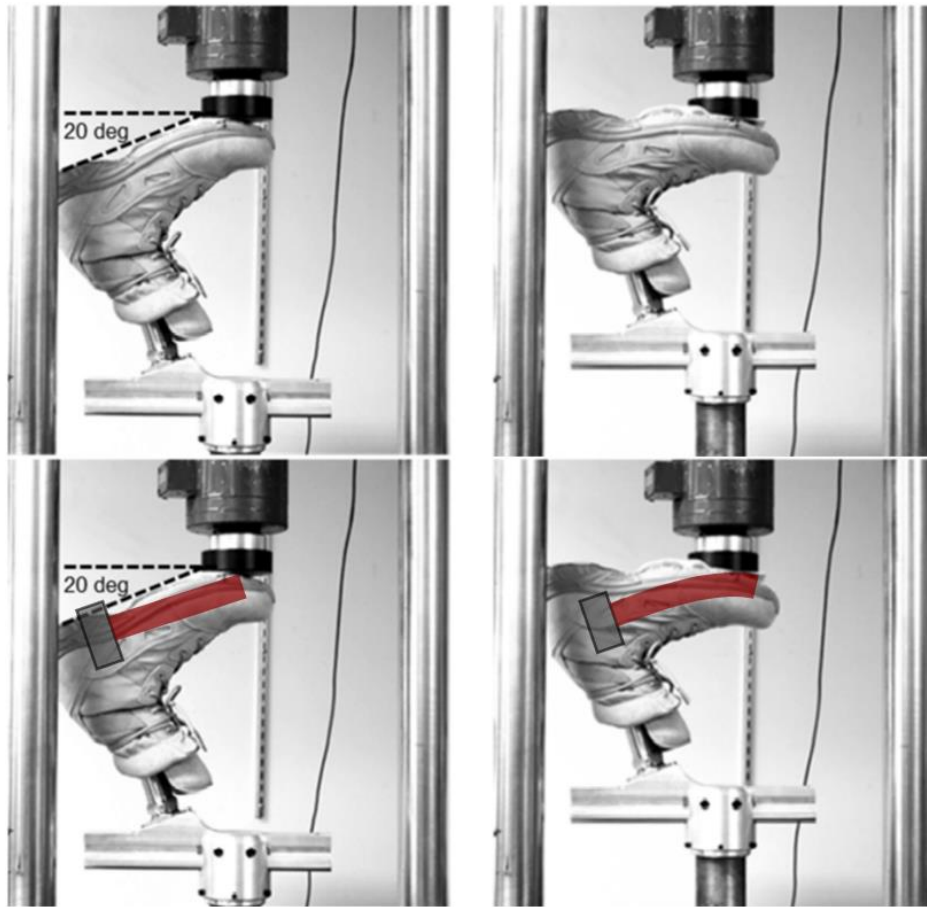


Figure 22: A prosthetic foot considered as a cantilever beam

The prosthetic foot tested in the figure is from the study of Koehler-McNicholas SR et al. [75]. The figures on top show the foot as it bends. And illustrates how a prosthetic foot can be simplified as a cantilever beam. This simplification of the foot is used for concept evaluation.

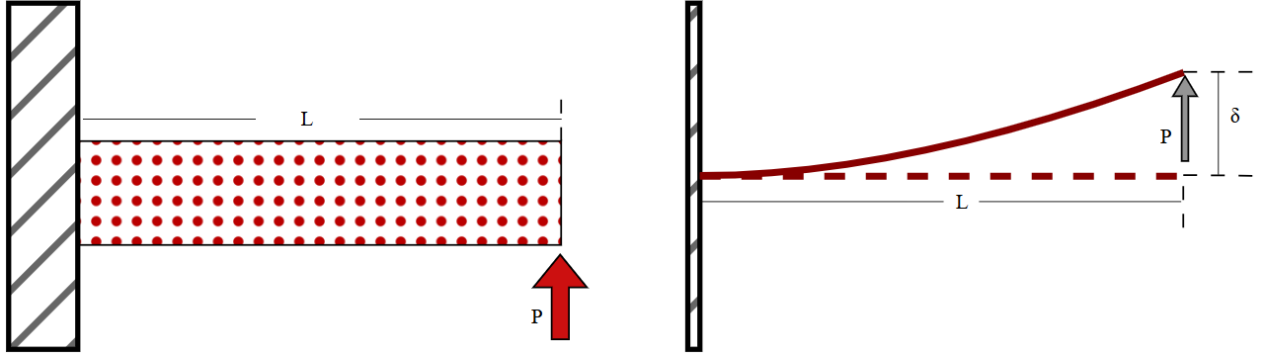


Figure 23: Diagrams of a simple cantilever beam.

On the left side of Figure 23, a cantilever beam with an applied load is illustrated. A free body diagram of this cantilever beam is presented by the figure on the left. The dotted line presents the original position of the beam, while the solid line illustrates the exaggerated deflection of the beam.

The displacement of the cantilever beam can be calculated using equation 1 [82]. The displacement (δ) is directly proportional to the force inflicted on the beam (P) and the length of the beam (L). Young's modulus of the beam (E) and the second area moment (I) are indirectly proportional to displacement. The stiffness constant (k) of the beam, given by equation 2 [105], is determined by dividing the inflected force by the maximum deflection. Thus, to have an effect on the stiffness, either the force applied or the displacement would have to be adjusted. Finally, using both the displacement and the stiffness constant, the potential energy (U) is determined with equation 3 [105].

$$\delta = \frac{PL^3}{3EI} \quad \dots (1)$$

$$k = \frac{P}{\delta_{max}} = \frac{3EI}{L^3} \quad \dots (2)$$

$$U = \frac{1}{2} k \delta_{max}^2 \quad \dots (3)$$

3.3. Concept Studies

3.3.1. Concept 1: Horizontally Adjustable Blade Design

The variable stiffness foot in Figure 16 of the Glanzer and Adamczyk study [76] works with a horizontally moving fulcrum. The same concept can be investigated as a concept for this study. By adding a horizontally adjusting blade, the stiffness of the prosthetic foot could be changed.

An additional composite blade that moves horizontally to the primary blade is the main idea for concept 1. The primary blade would be allowed to deflect normally until it encountered the secondary blade. From here the two blades would deflect together. The adjustment comes from the horizontal movement of the blade. As the additional blade moves, the point where the two blades connect changes, thus changing the amount of deflection allowed for the primary blade. The movement of the concept is represented in Figure 24 as a simple cantilever beam.

The configuration on the left of Figure 24 is less rigid than on the right due to the shorter grey beam illustration. As the grey beam lengthens, as illustrated on the right of Figure 24, the stiffness increases due to the change of the fulcrum location.

The change of fulcrum location effectively changes the length used to calculate the displacement. Where equation 1 would generally be used to calculate the displacement of the beam, equation 4 is now used along with the adjusted beam length (L_i). The force diagram (FD) in Figure 25 illustrates the displacement experienced due to the change in beam length.

$$\delta = \frac{PL_i^3}{3EI} \quad \dots (4)$$

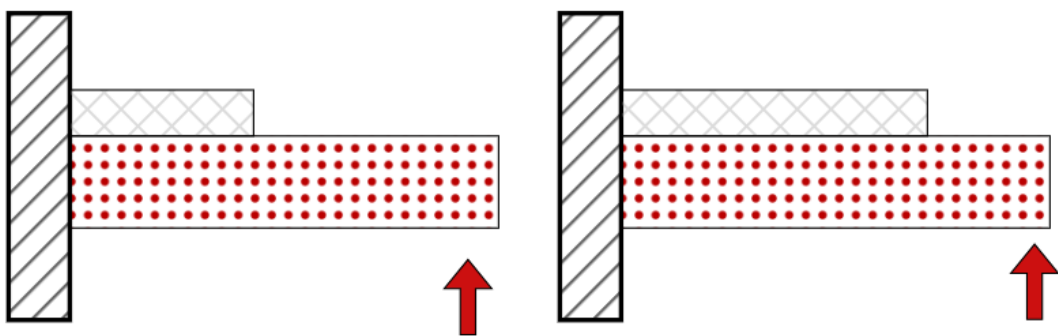


Figure 24: Cantilever beam representation of the concept of horizontal adjustment of the blade, The blade on the left more flexible due to a shorter fulcrum length. Once the blade moves horizontally to the right, as seen on the right of Figure 24, the blade becomes stiffer due to the moved fulcrum.

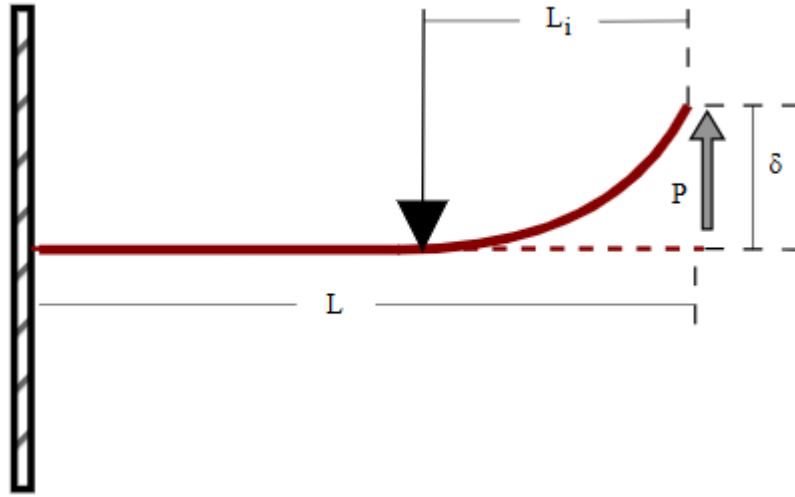


Figure 25: FD of horizontal adjusting blade cantilever beam.

Assuming the force remains constant, the adjustment of the beam length would adjust the displacement of the beam through equation 4, in turn, the stiffness would be adjusted using equation 2.

However, this is somewhat of a simplification of the concept. In reality, the shorter secondary blade would also start to deflect along with the primary blade. However, for the purpose of this chapter and the concept evaluation, it is assumed that the secondary blade is of high enough stiffness so that the displacement, as described in equation 1 is sufficient.

Possible designs using this concept could be to have a fulcrum piece, which consists of a composite material and moves over the primary blade, simply illustrated in Figure 26. With this design, multiple interchangeable fulcrum blades could be made to further add to the adjustability of the foot stiffness. This design would require a system that would keep the fulcrum blade in place at the desired location.

Another possible design using this concept could be to have a full-length additional blade, with the adjustment made by using a unique pyramid adapter. The pyramid adapter connects the prosthetic blades of the foot and allows connection of the pylon or socket (see Chapter 2). The full-length primary blade would deflect along with the secondary blade as the two blades interact with each other. Figure 27 is an illustration of this possible concept solution.

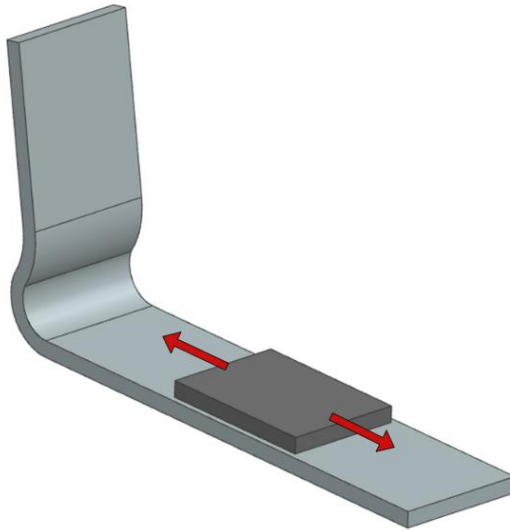


Figure 26: Horizontally moving blade: fulcrum design concept.

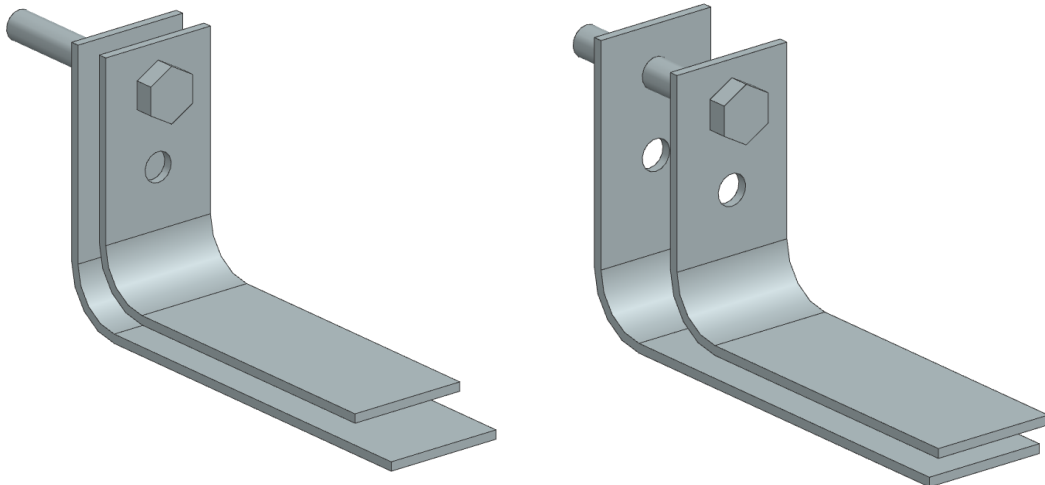


Figure 27: Horizontally moving blade: horizontal adjusting design concept.

3.3.2. Concept 2: Vertically Adjustable Blade Design

Similarly to concept 1, concept 2 also has an additional composite blade. The difference between the two is that the additional blade for concept 2 would adjust vertically instead of horizontally like in concept 1.

The primary blade would deflect as usual. When the primary blade comes into contact with the additional blade, the two blades will deflect together. A vertically adjustable blade would allow the user to change the height at which the two blades interact with each other. The vertical movement of the concept is illustrated in Figure 28 as a cantilever beam (grey crosshatched beam). When a load is applied to the primary blade (red dotted beam), the blade will start to deflect, first on its own. When deflected enough, the primary blade will make contact with the additional blade, and this will become a fulcrum for the primary blade. Figure 29(b) illustrates the displacement when the two blades meet. There are two points of rotation for the beam; the first is the fixed point on the left edge of the beam, and the second is the grey circle, which represents the point of contact between the two blades.

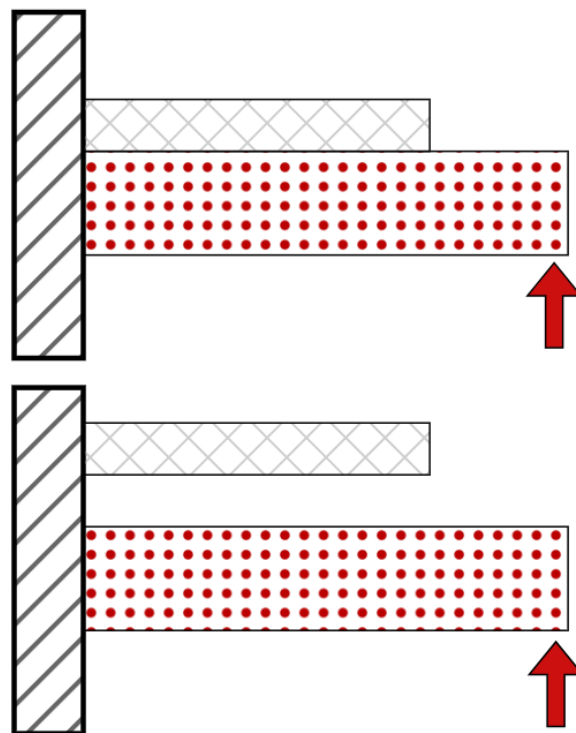


Figure 28: Cantilever beam simplification of vertically moving blade design

In the upper figure, the additional blade deflects as the primary blade deflects. This is stiffer than the instance at the bottom, where the primary blade deflects alone until contact is made with the additional blade.

Before contact is made between the two blades (Figure 29(a)), the primary blade would be calculated with equation 1. At the point of contact, the displacement of the primary blade (δ_B) would depend on the height of the second blade, as can be seen when studying equations 6 and Figure 29(b). Here, z represents the height of the second blade and L_{1x} represents the x-axis length of the primary blade. The slope of the beam (θ) is determined as described by Rao & Fah [105]. When the blades are in contact and the displacement continues, illustrated in Figure 29(c), the displacement of the primary blade can be calculated by equation 7. This is dependent on the resultant force of the primary blade. The resultant force exerted by the primary blade is the difference between the applied force P and the spring force of the primary blade.

The equations for the three instances, illustrated in Figure 29, can be estimated with equations 5 to 7. Since the displacement can be adjusted, the potential energy can be adjusted with equation 3.

$$\delta_A = \frac{PL_1^3}{3E_1I_1} \quad \text{Before contact} \quad \dots (5)$$

$$\delta_B = (L_{1x} - L_2) \times \tan \theta + z \quad \text{At contact} \quad \dots (6)$$

$$\delta_c = \frac{(P - \delta_B \times k_1)L_2^3}{3E_2I_2} \quad \text{After contact} \quad \dots (7)$$

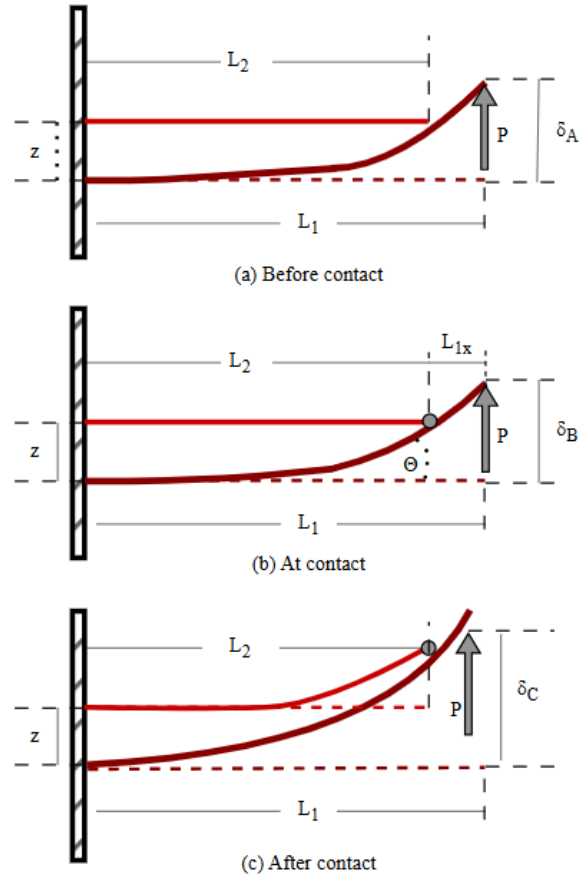


Figure 29: Contact instances of vertical adjusting blade.

Figure 29(a) illustrates the vertical adjusting concepts before the two blades are in contact. At this point only the primary blade deflects. Figure 29(b) shows the instance when the two blades make contact; after this point, the additional blade deflects along with the primary blade, illustrated by Figure 29(c). The primary blade's deflection is affected by the contact of the additional blade.

Various options are available to implement horizontal adjustment, one of which is to have a gear system (Figure 30) within the pyramid adapter that would allow the vertical movement of the blade and the adjustment to be made on the side of the prosthetic foot. Alternatively, a screw mechanism could be used to allow adjustment to be made at the top, eliminating the complexity of the gear system. A guiding system, as illustrated in Figure 31, would need to be present in both cases to ensure that the secondary blade is kept aligned with the primary blade.

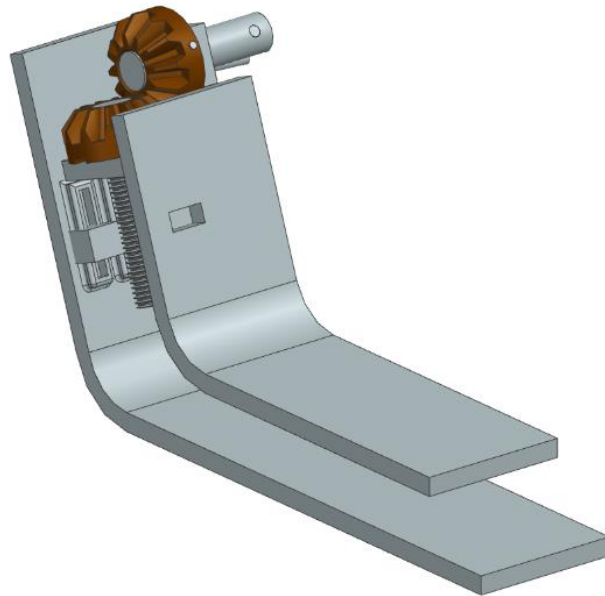


Figure 30: Vertically adjusting blades: Gear design concept

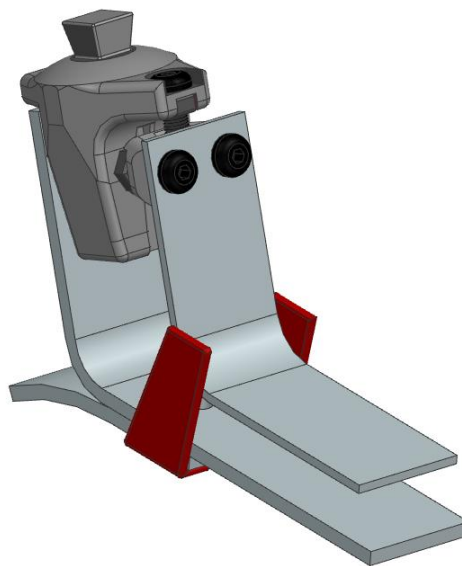


Figure 31: Vertically adjusting blades: wing design concept

Alternatively, the additional blade could be fixed to a base plate or sole plate along with the primary blade. With this solution, the problem of blade alignment is solved. Slight changes in the mathematical model would be present with this concept. If the additional blade is split into two to promote inversion and eversion, as illustrated by the concept in Figure 32, the force applied to the system when the blade is moved vertically would act as an initial applied force. The idea is illustrated as a cantilever beam in Figure 33.



Figure 32: Vertically adjusting blades: fork design concept.

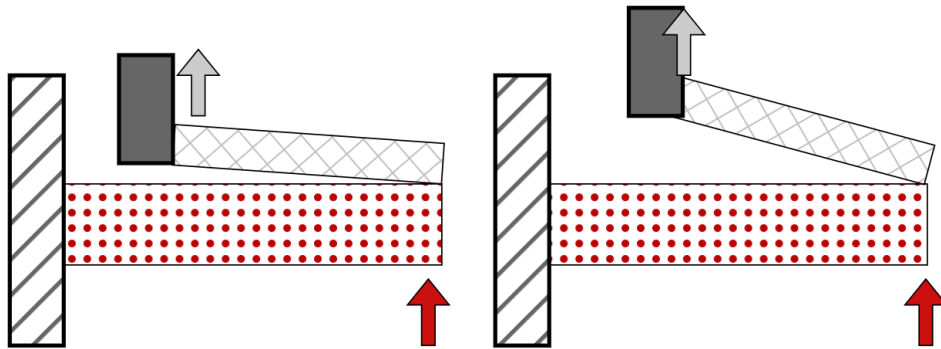


Figure 33: Cantilever beam simplification of fork design

The tensioning of the additional blade adds to the applied force of the foot. This means that a foot could deflect at lower applied loads due to additional blade tension force.

Equation 8 calculates how the displacement of this concept is estimated. The force inflicted by the adjustment is represented by F_a , and the other variables are applicable to the additional blade. As the displacement is adjusted, the beam stiffness, calculated with equation 2, energy stored, calculated with equation 3, will also be adjusted.

$$\delta = \frac{(P + F_a)L^3}{3EI} \quad \dots (8)$$

The concern of the split blades being next to the primary blade is the possible abrasion between the blades. If the foot experiences inversion or eversion rotation, the blades could wear due to the additional fork blades sliding against the primary blade. Alternatively, the additional split blade could be stacked above the primary blade to eliminate the risk of this wear. This concept is illustrated in Figure 34.

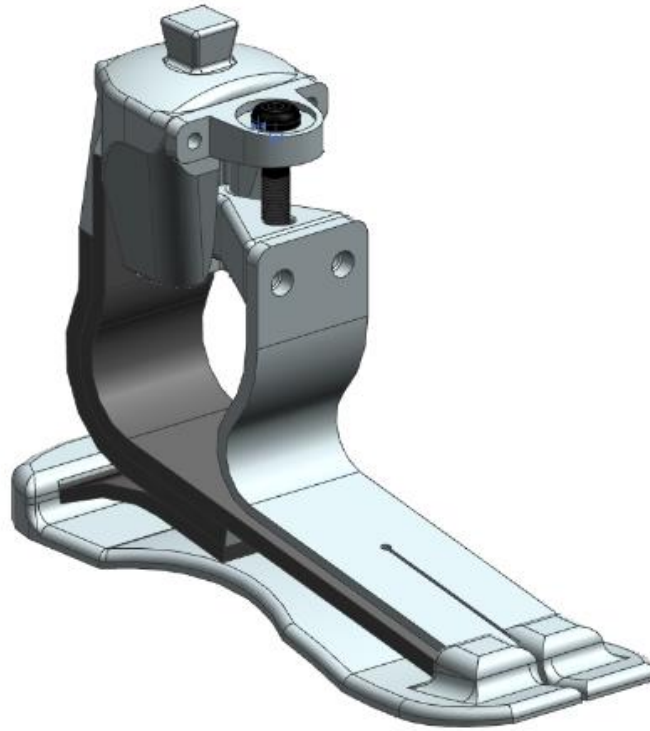


Figure 34: Vertically moving blades: stacked design concept.

3.3.3. Concept 3: Gas Spring Design

The gas spring design, shown in Figure 35, considers adding commercially available gas springs. The gas springs would be fitted to both sides of the pyramid connector and the toe part of the composite blade. The adjustment would be to swap the gas spring for stronger or weaker springs. The compression of the spring would mean more force pushing back as the foot deflects.

The addition of gas springs makes it possible to adjust the inversion and eversion independently from each other, by attaching different-strength gas springs on each side of the foot. Gas springs usually have upper force limits but can be pumped to any force below the limit, significantly expanding customisation potential. A prosthetic foot could then be adjusted to a person's specific body weight or comfort of walking, ensuring support and gait adaptation.

Figure 36 illustrates the cantilever setup for this concept, where the additional arrow represents the push-back forces of the gas spring. The y component of these forces is represented by the arrow associated with F_y in the force diagram (FD), illustrated in the bottom figure in Figure 36.

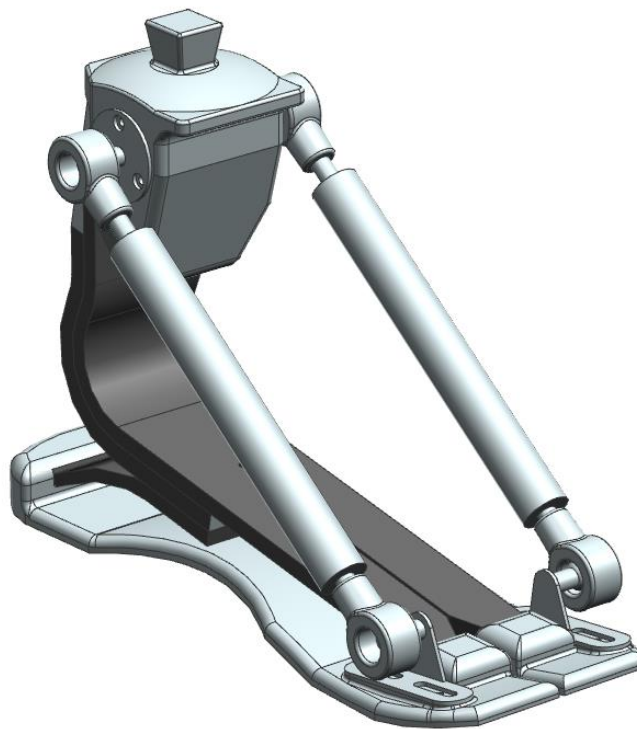


Figure 35: Gas spring design concept.

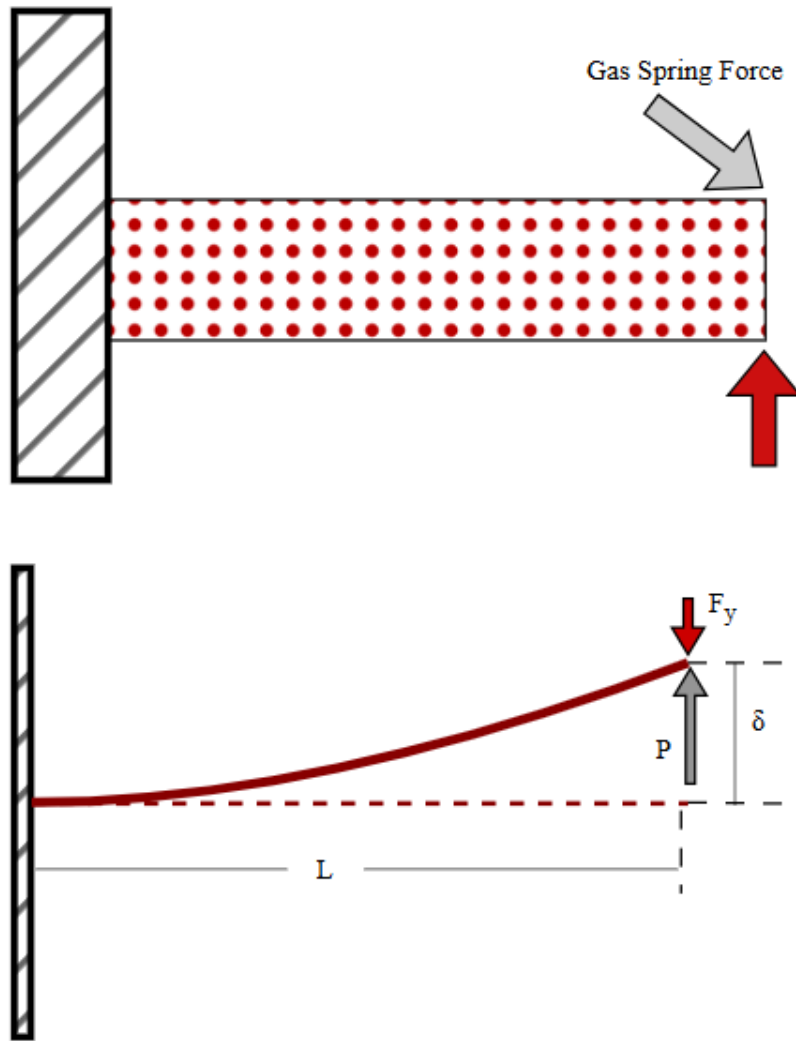


Figure 36: Cantilever simplification and free body diagram of the gas spring concept

The cantilever illustration of the gas spring design is illustrated in the above figure of Figure 36. This illustration includes the gas spring force at an angle due to the way it would have to be mounted. The y component is then seen acting against the applied load in the free body diagram shown in the bottom figure of Figure 36.

The displacement of this design can be determined with equation 9, where the force aspect is now the resultant between the applied force and the force supplied by the gas springs. By changing the force supplied by the gas springs, the displacement can be changed. By changing the displacement, beam stiffness and the energy output can be determined by equations 2 and 3 respectively.

$$\delta = \frac{(P - F_y)L^3}{3EI} \quad \dots (9)$$

The biggest concern for this design is that the gas spring setup would have to be designed in such a way that the foot fits in a foot shell and in a shoe. The foot shell would need to be specifically designed for this foot.

3.4. Concept Evaluation

All three of the concepts discussed would prove to be a solution for an adjustable stiffness foot; however, another goal of the study is to make the foot more affordable. Therefore, the evaluation of the concepts discussed in this section will be done based on a high-level cost estimation of the concept's manufacturing. Each concept would be scored on a component count, with the argument that the more composite blades, machined parts, and additional components required by the concept, the more the manufacturing of the component would cost. The scores are kept simple. If the design requires the same number of parts as a commercial prosthetic, a score of 1 is allocated; if the concept has more than a commercial foot, a score of 2 is assigned.

The composite score refers to the number of composite blades that would have to be manufactured for the concept design. The standard parts that would achieve a composite score of 1 are a heel blade, a primary forefoot blade, and a supporting blade for the primary blade. The types of composite materials used are not considered at this point in the study.

The machining score is given on the number of components that will have to be machined. A standard prosthetic foot typically comes with only one machined component, the pyramid adapter.

The additional component score is allocated for the amount of additional components that will have to be added to the prosthetic foot to ensure that the stiffness change can be applied. A score of one in this category would mean that no additional components are required to adjust the stiffness.

Concept 1, the concept of horizontally moving blades, consists of the standard composite blades with the addition of another blade that moves horizontally and allows for stiffness adjustment. In addition to the composite blades, additional machined parts must have a pyramid adapter that allows for the adjustment of stiffness. Depending on the solution, the concept would require either a system to restrain the fulcrum blade or a barrier component to keep the alignment correct. Concept 1 therefore has a score of 2 for all three criteria.

Concept 2 also has a score of 2 in all three criteria, requiring additional composite blades for stiffness adjustment, additional machined parts for vertical blade movement, and additional components that keep the blades aligned or fixed to the base sole plate of the prosthetic foot.

Concept 3 only has the standard composite blades found on commercial prosthetic DR feet. Along with these, this concept works with only one machined component if off-the-shelf gas springs are used, as they are relatively inexpensive. The machined pyramid adapter would include the attachment points for the gas springs, therefore scoring 2 for its machining. The gas springs are additional components required for this concept.

Table 2: Concept cost evaluation table

Concept	Composite Score	Machining Score	Component Score	Total Score
Horisontally Adjustable Blade Design	2	2	2	6
Vertically Adjustable Blade Design	2	2	2	6
Gas Spring Design	1	2	2	5

Although the score difference is marginal, as seen in Table 2, Concept 3 has the lowest score and requires the lowest number of manufactured components. Commercially available gas springs can be used, reducing manufacturing time and cost. Concept 3 presents the possibility of refined adjustment, as gas springs can be pumped to any force within the spring's limit. The addition of gas springs can also present the opportunity of adjusting the inversion and eversion of the foot, by adding different-strength gas springs to the respective sides of the foot. Concept 3 was therefore chosen for further development during this study.

3.5. Chapter Summary

In this chapter, three concepts are presented as a possible solution for the study. The horizontal adjustable blade concept proposed an additional composite blade that would move horizontally over the prosthetic foot. The second concept proposed also presented the addition of a moving composite blade, the movement, however, being vertical. The final concept proposed the addition of gas springs to the prosthetic foot.

The fundamental beam theory models for each of the concepts are explained within the chapter, and the three concepts were evaluated based on the number of composite blades that would be required, the amount of components that would need to be machined, and if any other additional components would be required, the argument being that more components lead to a higher cost.

The concept with the gas springs was chosen in the evaluation, and the next chapter describes the detailed model development and design thereof.

Chapter 4 : Detail Design

4.1. Chapter Overview

The previous chapter explored multiple concept designs for an adjustable-stiffness prosthetic foot, along with a high-level consideration of the cost implications to manufacture each concept. Of these concepts, the gas spring concept was selected, and a detailed design is described in this chapter.

Firstly, the specifications are discussed that would be required for the detailed design, which is then followed by the calculation method. The calculation method presents how the displacement of both the forefoot and heel is calculated. These calculations are used to determine the dimensions required for the prosthetic foot. These dimensions were validated using finite element analysis (FEA) performed in Siemens NX (version 1984), which utilises the Nastran solver for structural simulations [106].

Furthermore, a composite layup analysis of the blade was done, exploring multiple materials and alternating composite layup angles in simulations.

Lastly, an elementary cost evaluation was done on the prosthetic foot, estimating material costs, manufacturing costs, and what the end-user could possibly pay.

Figure 37 illustrates the different parts of the foot design, as may be referenced in this chapter.

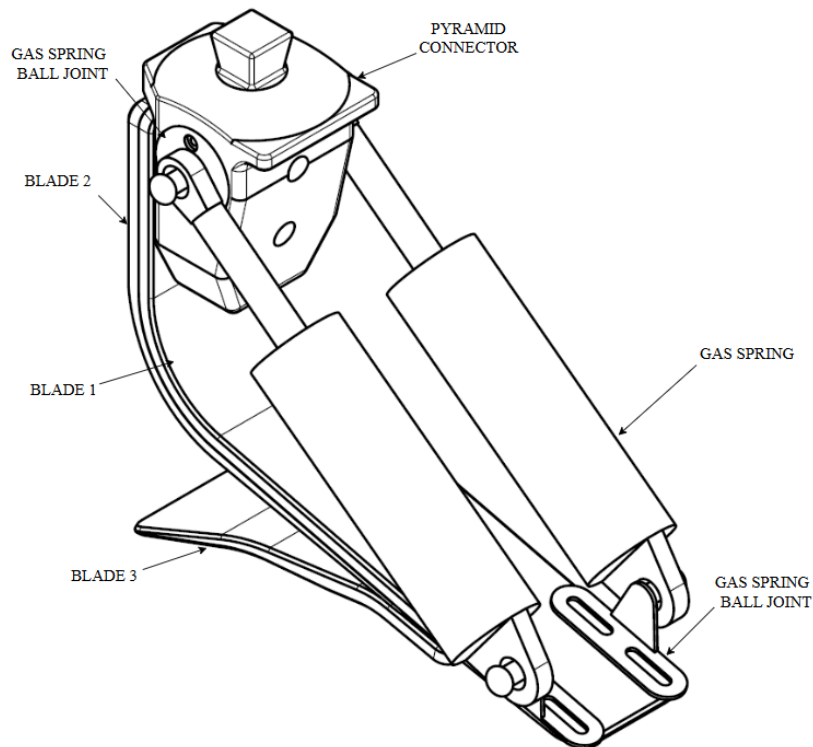


*Figure 37: Components of the selected design concept
(Some mechanical assembly details omitted for clarity.)*

4.2. Detail Design Specifications

There were some important aspects and specifications that had to be considered in order to design the prosthetic foot. These aspects included the weight for which the foot was designed, the size of the foot, and the stiffness the foot had to achieve.

Weight

For this study, the prosthetic foot was designed to handle a person of 80 kg and a size 27 prosthetic foot. This design mass was simplified to 800 N for a manageable comparison to static tests as prescribed by ISO 10328. From Chapter 2, it was learnt that a foot can see up to 1.2 times a person's body weight. This means that the foot should be able to withstand about 960 N. Designing the foot with an additional safety factor of 1.25 would allow the foot to endure a force of 1200 N before failure. This is a safety factor of 1.5 on the foot when loaded with 800 N.

Design Stiffness

The study by Koehler-McNicholas et al. [75] tested multiple feet with different participants, measuring stiffness in terms of % body mass per millimetre displacement. These results are presented in Figure 38. As discussed in the literature review in Chapter 2, a prosthetic foot that is too stiff or too flexible can cause discomfort and a noticeable limp.

Although the foot is on the stiffer side, the *Triton Heavy Duty*, presented as “Triton” in Figure 38, serves as a useful benchmark. Marketed as a prosthetic foot for all amputation levels and active lifestyles [107], the *Triton* also features higher pyramid placement. This is ideal when considering the attachment of gas springs. This made the *Triton* an ideal foot to use as inspiration for the prosthetic foot of this study.

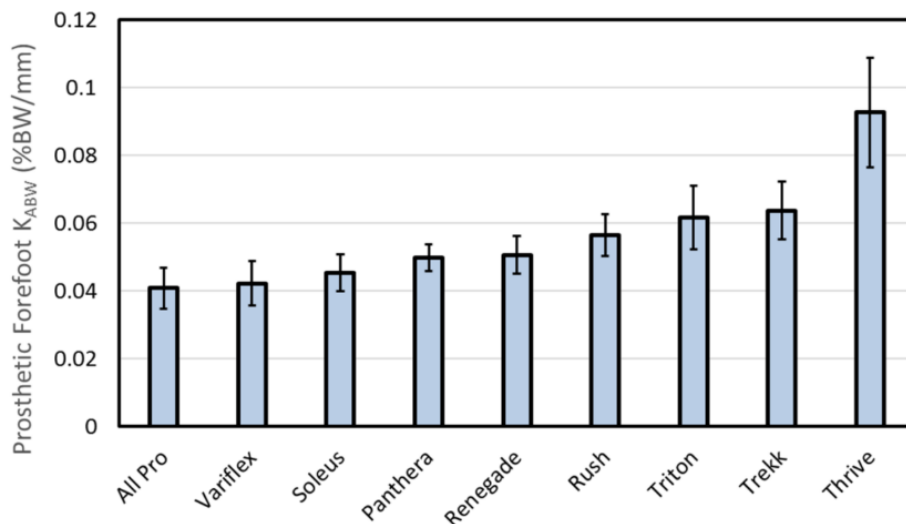


Figure 38: Foot stiffness results from Koehler-McNicholas' study [75].

Koehler-McNicholas's study did not provide the exact stiffness values for the prosthetic feet but instead reported stiffness as a percentage of body mass per millimetre value. Although the foot stiffness could be calculated, a study by Turner et al. provided the stiffness of the *Rush HiPro* foot as 34.8 N/mm for the forefoot stiffness and 53.4 N/mm for the heel stiffness [108]. These values were obtained through static testing as prescribed by ISO 10328.

[108].

The *Rush HiPro*, presented as “Rush” in Figure 38, is positioned next to the *Triton* and exhibits a similar stiffness. The two feet have similar designs, as seen in Figure 39. Given these similarities, the stiffness values of the *Rush HiPro* foot were used as target values for the design. Additionally, a *Triton* foot was available for physical measurements, which was used to make assumptions for the design measurements.



Figure 39: Prosthetic feet used as inspiration for prosthetic foot design

The *Triton Heavy Duty* [107] (left) and the *Rush High Pro* [109] (right) are prosthetic feet made for high activity levels.

Foot Dimensions

Some dimensions were kept constant during the design process, including the foot length, height and width, allowing the focus to remain on optimising segment lengths, blade angles, and blade thicknesses.

The prosthetic foot was designed as a right-foot model to fit a size 8 shoe, which, according to Ottobock's conversion chart in Appendix F, corresponds to a foot length of 27 cm. The blade height was set at 140 mm, corresponding to that of the *Triton Heavy Duty*. This height would allow for a gas spring to be mounted at an angle such that its vertical force component is greater than its horizontal component.

The width of the blade used for the design is 45 mm, 5 mm wider than the *Triton*, to accommodate the attachment of the gas springs. The contact length of the pyramid was selected as 45mm; this is comparable to that of the *Triton*.

Specification Summary

Table 3 presents a summary of the design specifications that were discussed in this section.

Table 3: Design Specifications Summary

Specification	Value	Unit
Design mass for an 80 kg person	800	[N]
Safety Factor	1.5	[-]
Design Stiffness: Forefoot	34.8	[N/mm]
Design Stiffness Heel	53.4	[N/mm]
Foot Size	270	[mm]
Foot Height	130	[mm]
Foot Width	45	[mm]
Pyramid Length	45	[mm]

4.3. Calculations of Theoretical Displacement

4.3.1. The Forefoot Blade

Design overview

Figure 40 presents a simplified illustration of the proposed design for the study's prosthetic foot. To determine what the dimensions of the prosthetic foot blade should adhere to achieve the desired displacement and applied force, calculations were performed. In Chapter 3, the design concepts were shown as linear cantilever beams. This was sufficient for the conceptual evaluation, but for the detailed design, more realistic shapes were required for the prosthetic foot blade.

To avoid the possibility that the forefoot blade became too thick for manufacturing, it was split into two blades of different lengths, as can be seen in Figure 40. When a composite layup is too thick over a bend, wrinkling can occur. Wrinkling in composites is caused by the accumulation of excess material, typically around a bend or complex shape. The excess material causes the composite layers to wrinkle, resulting in creases along the structure. The wrinkling effect is due to excessive thickness, insufficient resin flow, inadequate compaction, or a lack of moulding support. In the case of this component, wrinkling would most likely be present due to the thickness (number of plies) of the composite material [110], [111].

The two-part split would thus help prevent wrinkling, as it would have had undesirable consequences for the component, like cosmetic defects, structural integrity, dimensional inaccuracy, and increased weight. All these factors were unwanted, as they would have led to inaccurate testing and inconsistent outcomes when compared to the simulations [110], [111].

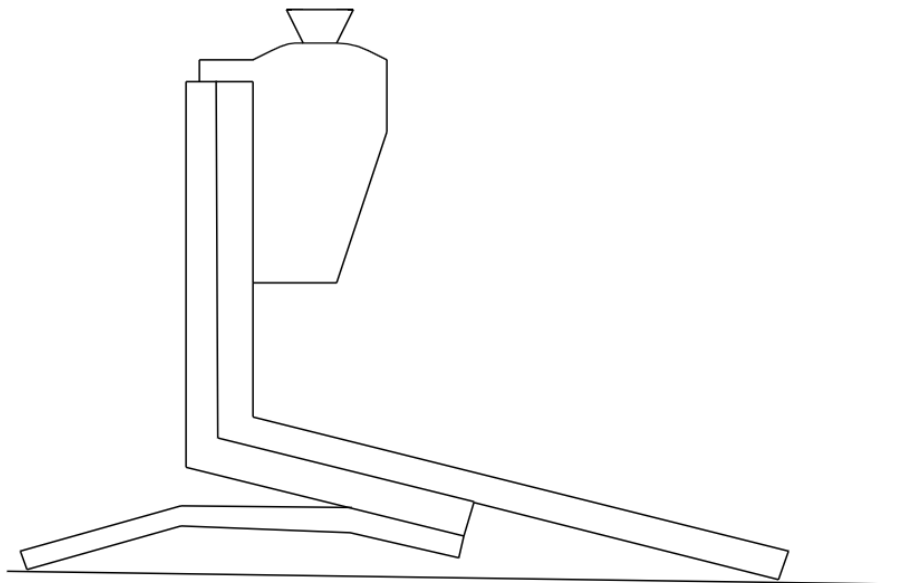


Figure 40: Simplified drawing of the proposed prosthetic foot

Equivalent Young's Modulus for the Forefoot Blades

The Rule of Mixtures (ROM) serves as a valuable rule for approximating the material properties within a composite mixture, following a similar approach as Talla HK et al. in their study on a prosthetic running blade [112]. The ROM provides estimated values for various properties of elasticity, mass density, and maximum tensile strength [95]. The ROM can accurately predict the longitudinal Young's modulus but lacks when predicting the transverse [113]. Classical laminate theory would offer a more accurate prediction, accounting for ply orientation and stacking sequence [114]. ROM was selected for its simplicity in the early-stage design phase, as well as its ability to accurately predict the longitudinal Young's modulus, which was the primary property calculated with the ROM [113].

Equation 10 illustrates the use of the ROM to estimate the longitudinal modulus of elasticity, E_{eq} , of the composite element of a unidirectional lamina [95], [115]. The product of the volume fraction and the modulus of elasticity of each material in the layup, represented by V_i and E_i respectively in equation 10 are summed together to estimate the Young's modulus in the longitudinal direction [95], [115].

The volume fraction, V_i , represents the ratio between the volume of a material, v_i , and the total value of the component. The volume fraction is calculated with equation 11

$$E_{eq} = \sum_{i=1}^n V_i E_i \quad \dots (10)$$

$$V_i = \frac{v_i}{v_t} \quad \dots (11)$$

The two blades that contribute to the forefoot, as illustrated in Figure 40, each had their own Young's modulus dependent on their composite layup. In the red hatched area displayed in Figure 41, the two blades are considered homogeneous. The ROM, equation 11, was used to estimate the equivalent Young's modulus for the homogeneous section.

This approach did, however, assume the following:

- That the two forefoot blades were perfectly bonded together.
- The homogenous section was assumed to behave in a linear elastic manner when loads were applied.
- Uniform material properties within each blade.
- The homogeneous section was treated as unidirectional orthotropic; only the longitudinal Young's was calculated and assumed constant through the section.

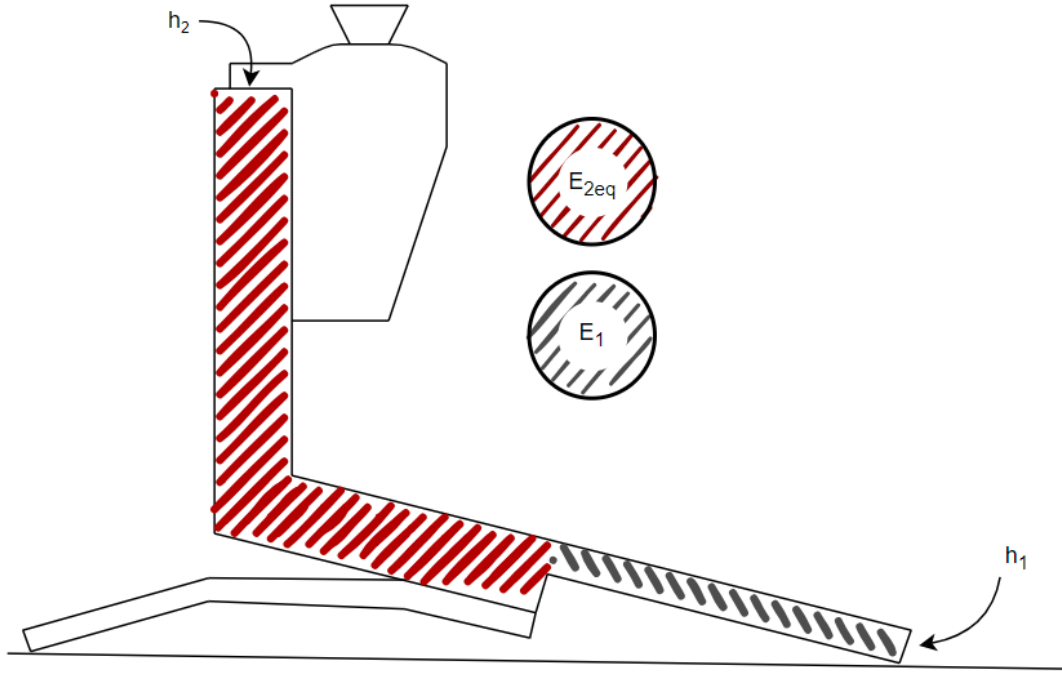


Figure 41: The equivalent Young's modulus of the combined forefoot blades

Using equation 10 and 11 an equivalent Young's modulus was calculated with equation 15, using the values collected from equations 12-14, as well as the blade width, b , and thicknesses h_1 and h_2 , illustrated in Figure 41 [95]. The length L is the length that the two beams overlap. This is, however, along with the width b , cancelled out when calculating the volume fractions V_1 and V_2

$$v_t = h_2 \times b \times L \quad \dots (12)$$

$$V_1 = \frac{h_1 \times b \times L}{v_t} \quad \dots (13)$$

$$V_2 = \frac{(h_2 - h_1) \times b \times L}{v_t} \quad \dots (14)$$

$$E_{2eq} = E_1 V_1 + E_2 V_2 \quad \dots (15)$$

To simplify the calculations further, an approach was taken using a calculated equivalent thickness (ET) for the two forefoot blades. This approach used the transformation factor, n_E , for composite beams [82]. The ET approach considered the two blades of the forefoot as one, as illustrated in Figure 42. With this method, a singular Young's modulus was selected for the two-blade system, and the appropriate thickness, h_{eq} , was calculated by multiplying the original thickness by the transformation factor n_E . The transformation factor n_E , calculated with equation 16 is the cube root of the transformation factor between the modulus of the first and second blades. The equivalent thickness was calculated with equation 17.

$$n_E = \sqrt[3]{\frac{E_2}{E_1}} \quad \dots (16)$$

$$h_{eq} = n_E h_2 \quad \dots (17)$$

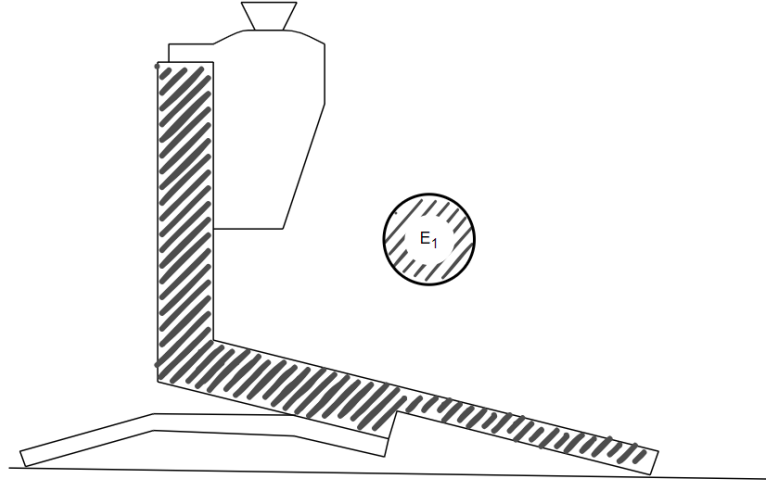


Figure 42: The prosthetic foot with a single Young's modulus for the forefoot blades

Although this method might seem redundant when the ROM method could rather be used, it proved useful during the design phase later in this chapter. During the design, multiple dimensions and variables were looped through a MATLAB script. Using the ET approach allowed for a more efficient functioning script.

Forces and dimensions

During the gait cycle, as described in Chapter 2, the forefoot and heel are only on the ground together for a short while. In this instance the full body weight is split between the forefoot and heel and thus not the maximum that these components will see. For this reason, calculations for the forefoot and heel are considered apart when fully loaded.

To calculate the estimated displacement of the forefoot blade, the blade was split into sections, each considering a different effect on the blade system. Figure 43(a) is an illustration of a simplified prosthetic foot. The markers, points A to D, were points of interest during the calculations for the forefoot. Point A was fixed during the calculations, and the applied force was located at point D. Point B represented the bending point of the blade, and point C represented the point where the second forefoot blade stopped.

The symbols used for dimensions that are shown in Figure 43(b) correlate to the dimensions used in the calculations and in the equations listed further in this chapter.

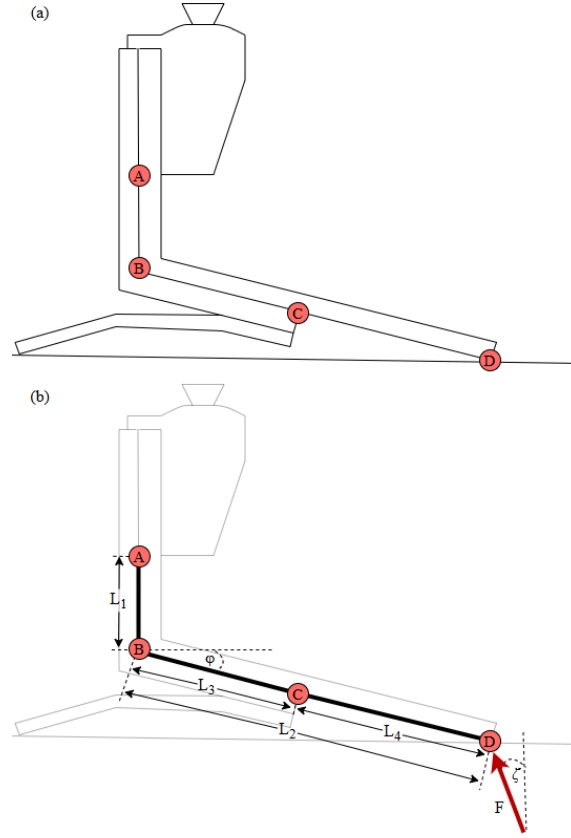


Figure 43: The segments and interest points of the forefoot

The applied force, F , was split into x and y components to be used in calculations. The first split, illustrated in Figure 44(a), was used to calculate the deflection of the AB segment. The force F_{y1} is parallel to the AB segment. The second split, illustrated in Figure 44(b), was done to fit the Cartesian plane of segments BC and CD, meaning the force F_{y2} was perpendicular to segments BC and CD. Forces F_{x1} , F_{y1} , F_{x2} , and F_{y2} , were calculated with equations 18, 19, 20, and 21 respectively.

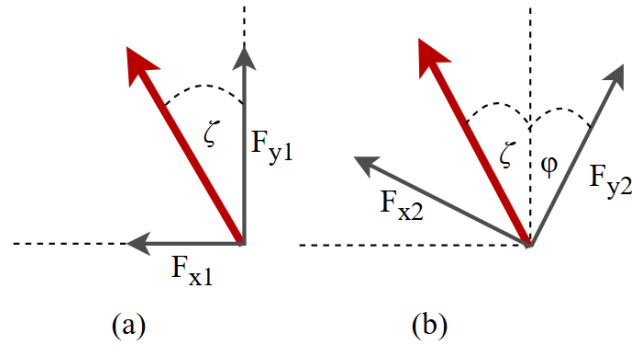


Figure 44: Force components of the primary blade

Figure 44(a) illustrates the components of the applied load on the conventional cartesian plane. Figure 44(b) shows the applied load split into components to fit the blade segments BC and CD.

$$F_{x1} = -F * \sin(\zeta) \quad \dots (18)$$

$$F_{y1} = F * \cos(\zeta) \quad \dots (19)$$

$$F_{x2} = -F * \sin(\zeta + \varphi) \quad \dots (20)$$

$$F_{y2} = F * \cos(\zeta + \varphi) \quad \dots (21)$$

Figure 45 illustrates the segments from Figure 43(b) separated with the applicable force components. The moments M_2 and M_1 were calculated with 22, and 23 respectively.

$$M_2 = F_{y2}L_4 \quad \dots (22)$$

$$M_1 = M_2 + F_{y2}L_3 \quad \dots (23)$$

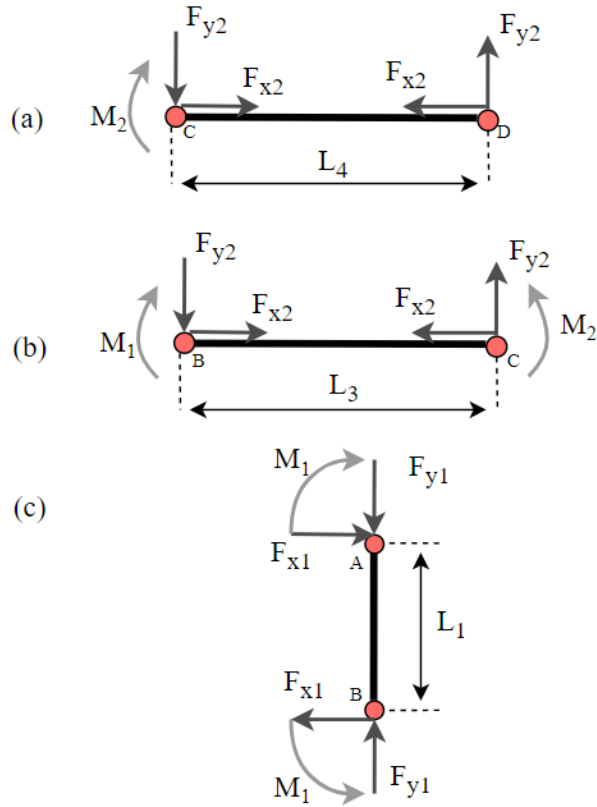


Figure 45: Force distribution of prosthetic forefoot segments

Figure 45(a) illustrated the segment CD with the load components acting upon the segment and the movement that results from the applied load. Figure 45(b) shows the next segment, BC, with the applicable acting loads and moments as well as the resulting loads and moments. The final segment, AB, is illustrated in Figure 45(c) along with the applicable moments and forces.

Initial position

Figure 43(b) illustrates the primary blade with the segment points, lengths, and applicable angles. The angles and lengths in this illustration represent the initial position of the blade. The initial position is used as a reference to investigate and illustrate the deflection of the blade in each section. To determine the displacement seen by the blade, superposition was used on the segments AB, BC, and CD to determine each segment's displacement first individually and then add it together to find the entire system's displacement. The calculation of initial points, A_0 , B_0 , C_0 , and D_0 , was done by equations 24-27. Figure 46 illustrates the initial points of the forefoot blade.

$$A_0 = [0, 0] \quad \dots (24)$$

$$B_0 = [0, -L_1] \quad \dots (25)$$

$$C_0 = [L_3 \cos(-\varphi) + B_{0x}, L_3 \sin(-\varphi) + B_{0y}] \quad \dots (26)$$

$$D_0 = [L_2 \cos(-\varphi) + B_{0x}, L_2 \sin(-\varphi) + B_{0y}] \quad \dots (27)$$

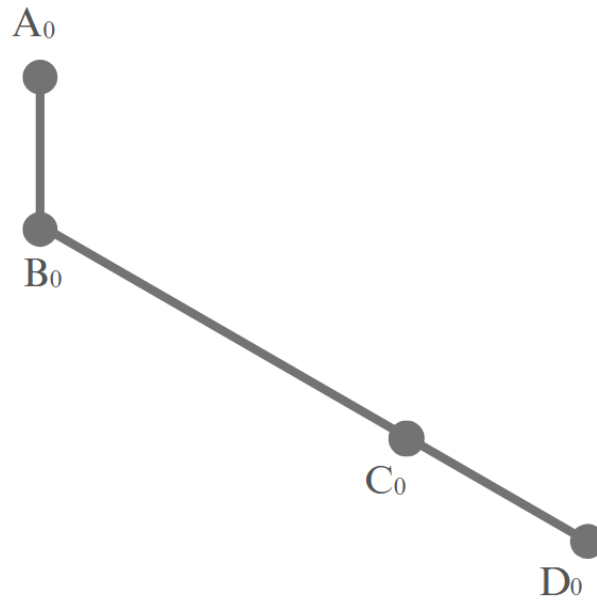


Figure 46: Conceptual representation of the initial position of the prosthetic foot segments

Next, the deflection of the segment AB due to the moment caused by the applied force is examined. This is a very important segment since point C and point D move when point B deflects.

Displacement of Segment AB

For segment AB, the moment M_2 and the force F_{x1} contributed to the displacement. Typically, the displacement of a cantilever beam due to a force is determined with equation 28, and the displacement due to a moment is determined with equation 29 [82]. The slopes of the displacement under an acting force or moment are given by equations 30 and 31 [82].

$$\Delta_F = \frac{FL^3}{3EI} \quad \dots (28)$$

$$\Delta_M = \frac{ML^2}{2EI} \quad \dots (29)$$

$$\theta_F = \frac{FL^2}{2EI} \quad \dots (30)$$

$$\theta_M = \frac{ML}{EI} \quad \dots (31)$$

The displacement of segment AB was calculated with equation 32, being the sum of the equations 28 and 29. This displacement was also the x value of position B. The new point B, point B_1 , had the same y-value as point B_0 , as the change in y was deemed negligible. The slope of the deflected segment AB was determined with equation 33, where equations 30 and 31 are summed together. The variable names match those of Figure 45(c).

$$\Delta_{AB} = \frac{M_2 L_1^2}{2E_{2eq} I_2} + \frac{F_{x1} L_1^3}{3E_{2eq} I_2} \quad \dots (32)$$

$$\theta_{AB} = \frac{M_2 L_1}{E_{2eq} I_2} + \frac{F_{x1} L_1^2}{2E_{2eq} I_2} \quad \dots (33)$$

During the displacement of segment AB, the segments BC and CD did not undergo individual changes. They did, however, displace, as a complete segment, segment BD. During AB's displacement, a new angle, γ_1 , was established between the x-axis and the segment BD. This angle is illustrated by Figure 47 and determined by equation 34. With this angle, the new positions for C and D, points C_1 and D_1 , were determined. The equations for points B_1 , C_1 , and D_1 are given by equations 35-37. The exaggerated deflection of AB is illustrated in Figure 48. Next, the deflection of the BC segment is investigated when a force is applied to the blade.

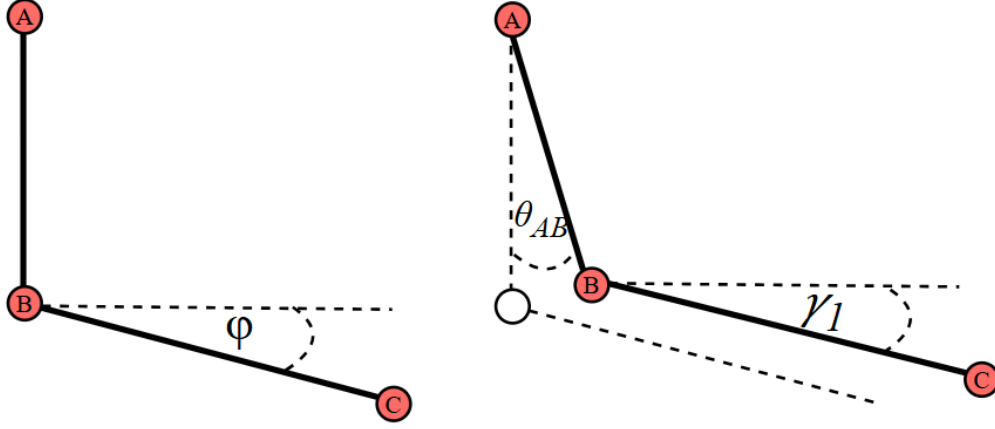


Figure 47: The new angle with the x-axis

This shows how the location of B changes the location of the segments connected to it. The angle γ_1 represent the new angle of segment BC with the x-axis.

$$\gamma_1 = \varphi + \theta_{AB} \quad \dots (34)$$

$$B_1 = [\Delta_{AB}, \quad -L_1] \quad \dots (35)$$

$$C_1 = [L_3 \cos(\gamma_1) + \Delta_{AB}, \quad -L_3 \sin(\gamma_1) - L_1] \quad \dots (36)$$

$$D_1 = [L_2 \cos(\gamma_1) + \Delta_{AB}, \quad -L_2 \sin(\gamma_1) - L_1] \quad \dots (37)$$

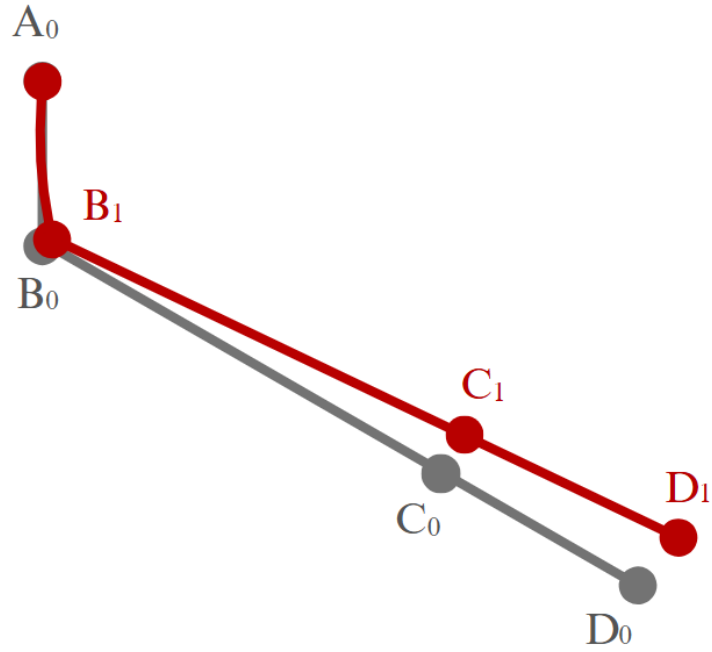


Figure 48: Conceptual representation of the blade position after considering moment on AB

Displacement of Segment BC

Segment BC deflected both due to the applied force and to the moment, M_1 . In this case, the displacement was calculated considering the theoretical equations of an acting force (equation 28) and an acting moment (equation 29). The sum of the two equations gave the displacement of the segment. The same was done when calculating the slope of the deflected beam. The sum of equations 31 and 30 gave the slope of the beam. The equation for Δ_{BC} and θ_{BC} with the dimension as illustrated in Figure 43 and Figure 45 is given in equation 38 and 39.

$$\Delta_{BC} = \frac{F_{y2}(L_3)^3}{3E_{2eq}I_2} + \frac{M_1(L_3)^2}{2E_{2eq}I_2} \quad \dots (38)$$

$$\theta_{BC} = -\frac{M_1(L_3)}{E_{2eq}I_2} + \frac{F_{y2}(L_3)^2}{2E_{2eq}I_2} \quad \dots (39)$$

To illustrate what this deflection looks like on the blade, the results had to be rotated. A rotation matrix, R , as calculated in equation 41, was used to turn the point C_{2a} to get C_{2b} as calculated in equation 42. After the rotation, the new point C_2 was calculated with equation 43, where C_{2b} was adjusted to meet the point B_0 .

$$C_{2a} = [L_3, \Delta_{BC}] \quad \dots (40)$$

$$R = \begin{bmatrix} \cos(-\varphi) & \sin(-\varphi) \\ -\sin(-\varphi) & \cos(-\varphi) \end{bmatrix} \quad \dots (41)$$

$$C_{2b} = R \times C_{2a} \quad \dots (42)$$

$$C_2 = [C_{2bx} + B_{0x}, C_{2by} + B_{0y}] \quad \dots (43)$$

Segment CD displaced as point C deflected. The displacement of CD was determined by calculating the slope of the deflected BC. The slope, θ_{BC} , was used in equation 44 to calculate the new angle between segment CD and the x-axis, γ_3 . With this, the new location of D was determined with equation 45.

$$\gamma_2 = -\varphi + \theta_3 \quad \dots (44)$$

$$D_2 = [L_4 \cos(\gamma_2) + C_{2x}, L_4 \sin(\gamma_2) + C_{2y}] \quad \dots (45)$$

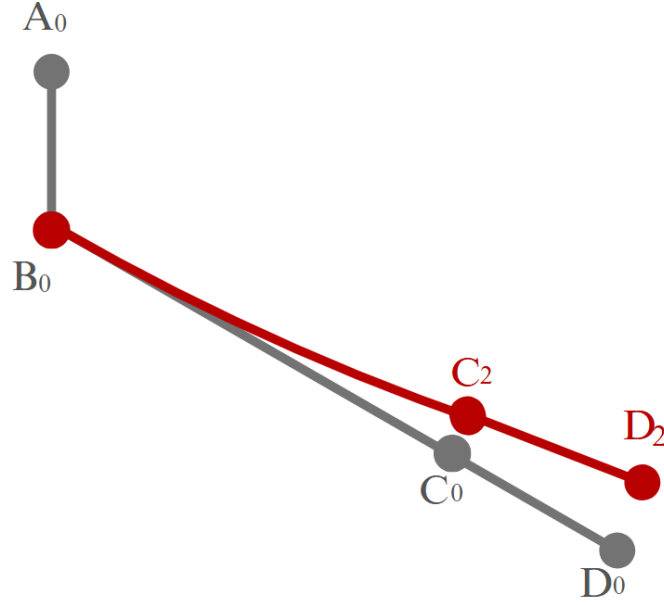


Figure 49: Conceptual representation of the deflection of segment BC

Displacement of Segment CD

The CD segment deflected only due to the applied force. Thus, the simple cantilever equation, equation 28 [82], was used. The slope of a cantilever beam under this condition was calculated with equation 30 [82]. The equation for displacement is rewritten in equation 46 as Δ_{CD} with the dimensions as illustrated in Figure 43 and Figure 45.

$$\Delta_{CD} = \frac{F_{y2}L_4^3}{3E_1I_1} \quad \dots (46)$$

Similarly to the calculation of C_2 , to calculate D_3 the results had to be rotated with the rotation matrix in equation 48. Then the results were adjusted to meet the point C_0 . The calculations were done with equations 47-50.

$$D_{3a} = [L_4, \Delta_{CD}] \quad \dots (47)$$

$$R = \begin{bmatrix} \cos(-\varphi) & \sin(-\varphi) \\ -\sin(-\varphi) & \cos(-\varphi) \end{bmatrix} \quad \dots (48)$$

$$D_{3b} = R \times D_{3a} \quad \dots (49)$$

$$D_3 = [D_{3bx} + C_{0x}, D_{3by} + C_{0y}] \quad \dots (50)$$

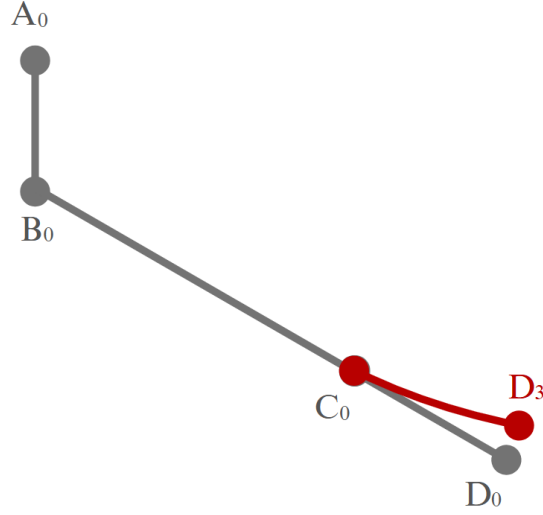


Figure 50: Conceptual representation of the displacement of CD

Final Displacement

Finally, the displacements from the BC and DC were merged with the displacement of the AB segment. First, the deflection of point C was rotated to fit the BD slope after AB's displacement. Equations 51-53 were used to calculate the final location of C.

$$R = \begin{bmatrix} \cos(\gamma_1) & \sin(\gamma_1) \\ -\sin(\gamma_1) & \cos(\gamma_1) \end{bmatrix} \quad \dots (51)$$

$$C_{4a} = R \times C_{2a} \quad \dots (52)$$

$$C_4 = [C_{4ax} + B_{1x}, C_{4ay} + B_{1y}] \quad \dots (53)$$

After this, the displacement of point B was determined. The angle between the x-axis and the slope of BC, γ_4 , was first determined with equation 54, as the value of the D deflections was rotated around it, with the rotation matrix from equation 55. The flow of the calculations is the same as the calculations of points D_3 and C_2 , and was calculated with 56 and 57.

$$\gamma_4 = -\varphi + \theta_2 - \theta_1 \quad \dots (54)$$

$$R = \begin{bmatrix} \cos(\gamma_4) & \sin(\gamma_4) \\ -\sin(\gamma_4) & \cos(\gamma_4) \end{bmatrix} \quad \dots (55)$$

$$D_{4a} = R \times D_{3a} \quad \dots (56)$$

$$D_4 = [D_{4ax} + C_{4x}, D_{4ay} + C_{4y}] \quad \dots (57)$$

With all the displacements considered, the total displacement of the forefoot blade was given by the difference between the final and the first D-y values, as seen in equation 58.

$$\Delta_y = D_{4y} - D_{0y} \quad \dots (58)$$

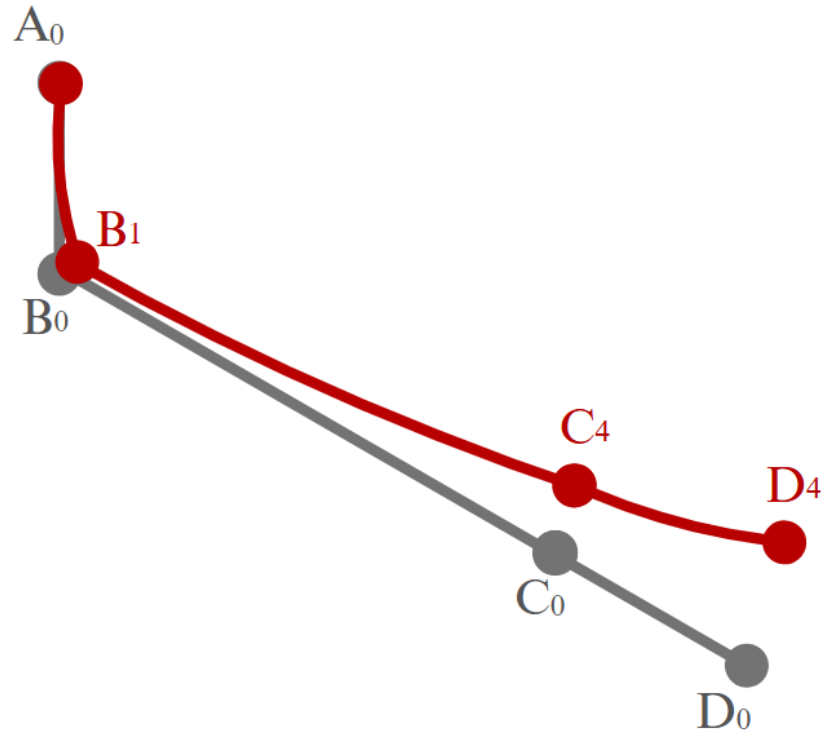


Figure 51: Conceptual representation the displacement of the blade when force is applied

4.3.2. The Heel Blade

The proposed simplified look of the heel blade is illustrated in Figure 40. The heel blade calculation method was similar to the forefoot calculation.

Forces and dimensions

Figure 52(a) is an illustration of the simplified prosthetic foot's heel blade segment points. Figure 52(b) illustrates the segment lengths and angles of the heel blade as well as the force acting on the blade with a angle ζ .

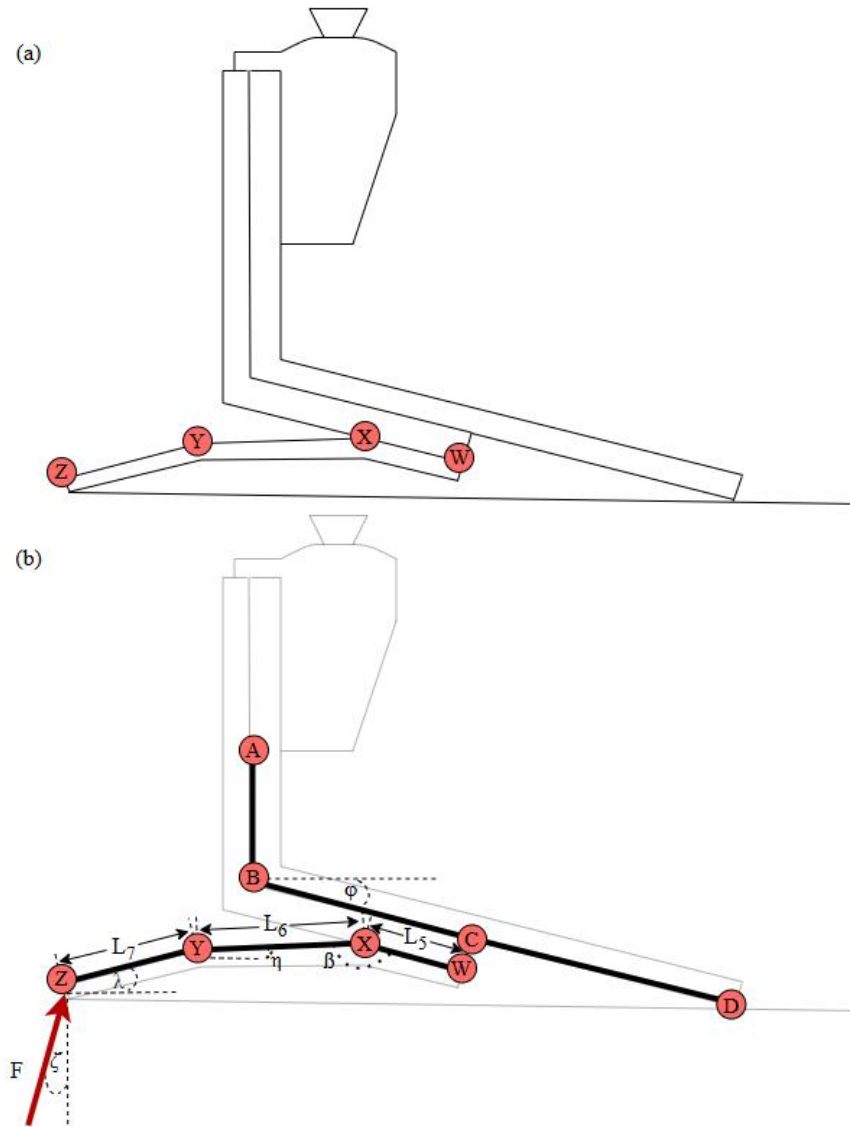


Figure 52: Segments and positions of the heel blade with the forefoot blade.

The forces in each segment in regard to the x-y plane are illustrated in Figure 53. Figure 53(a) illustrates the x and y force components, F_{x7} and F_{y7} , where F_{y7} is perpendicular to segment ZY. Forces F_{x7} and F_{y7} were calculated with equation 59 and 60 respectively. Similarly, the x and y force components of the force acting on segment YX, where F_{y6} is perpendicular to segment YX, as illustrated by Figure 53(b), were calculated with equation 62 and 61 respectively.

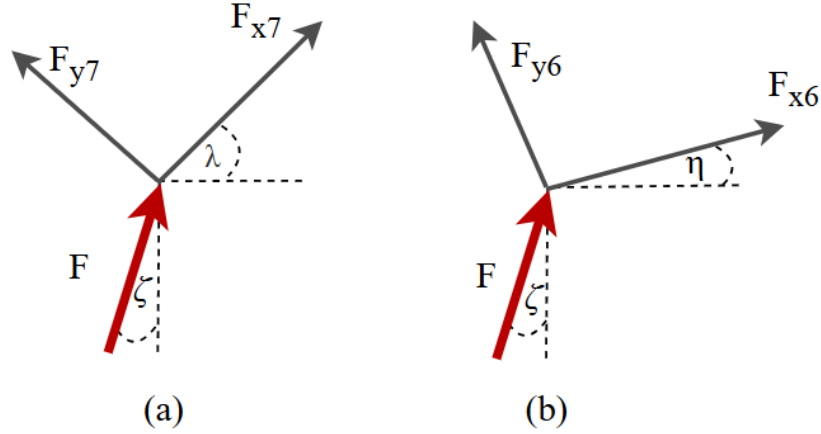


Figure 53: Force components of the heel blade's segments

$$F_{x7} = F \times \cos (90 - \lambda - \zeta) \quad \dots (59)$$

$$F_{y7} = F \times \sin (90 - \lambda - \zeta) \quad \dots (60)$$

$$F_{x6} = F \times \cos(90 - \zeta - \eta) \quad \dots (61)$$

$$F_{y6} = F \times \sin (90 - \zeta - \eta) \quad \dots (62)$$

The segments of the heel blade are shown as separate sections in Figure 54 to illustrate the forces and moments required to determine the total displacement of the heel blade. The moments M_7 and M_6 were calculated with equations 63 and 64.

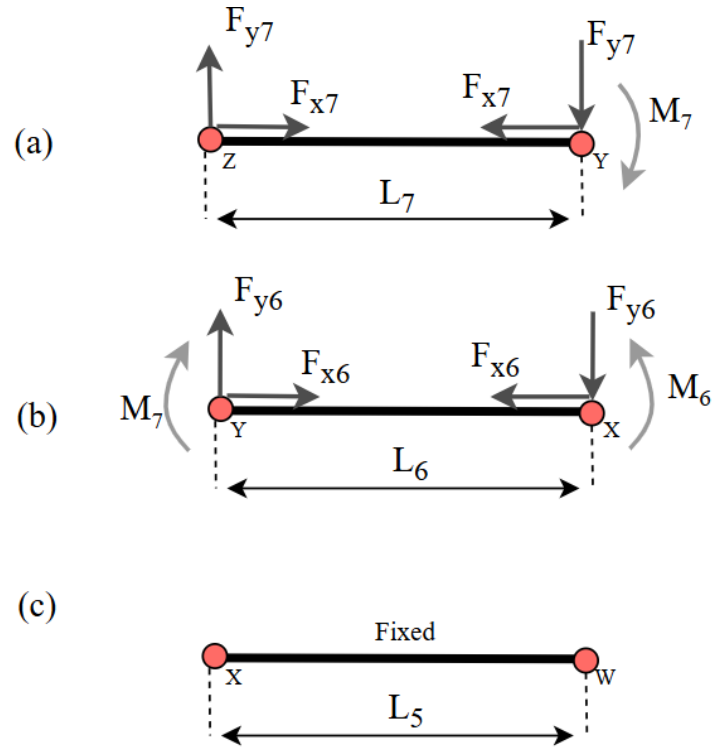


Figure 54: The resulting forces and moments on the heel blade segments.

Figure 54(a) illustrates the segment ZY with the applied load components acting upon the segment and the moment that results from the applied load. Segment XY is shown in Figure 54(b) with the applicable loads and moments. Segment WX in Figure 54(c) is fixed to the forefoot blade.

$$M_7 = F_{y7}L_7 \quad \dots (63)$$

$$M_6 = F_{y6}L_6 + M_7 \quad \dots (64)$$

Initial position

The heel blade configuration shown in Figure 55 was used to determine the displacement of the heel blade under the applied load. First, the initial points were required. Since the heel blade is aligned with the forefoot blade at point C, the dimensions of the forefoot blade and the location of C_0 were required as geometric constraints. The initial locations of heel nodes were calculated with equations 65-68.

$$W_0 = [C_{0x} - h_2 \cos(\varphi), C_{0y} - h_2 \sin(\varphi)] \quad \dots (65)$$

$$X_0 = [-L_5 \cos(\varphi) + W_{0x}, L_5 \sin(\varphi) + W_{0y}] \quad \dots (66)$$

$$Y_0 = [-L_6 \cos(\eta) + X_{0x}, L_6 \sin(\eta) + X_{0y}] \quad \dots (67)$$

$$Z_0 = [-L_7 \cos(\lambda) + Y_{0x}, -L_7 \sin(\lambda) + Y_{0y}] \quad \dots (68)$$

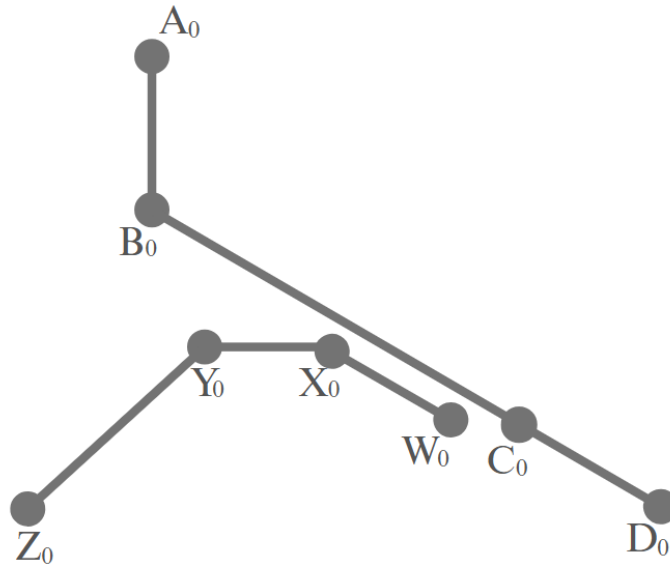


Figure 55: Conceptual representation of the initial position of the heel blade

Displacement of Segment ZY

Similar to the forefoot blade calculations, the heel blade calculations were split into parts. First, the YZ deflection of the segment was considered. In this segment, the applied force was used to calculate the deflection; see equation 69.

Just like the forefoot calculations, the deflection results of the cantilever formula were rotated and adjusted to align with the curvature. The coordinates of the deflected Z point were determined by equations 70-73.

$$\Delta_{ZY} = \frac{F_{y7} L_7^3}{3 E_3 I_3} \quad \dots (69)$$

$$Z_{6a} = [-L_7, \Delta_{ZY}] \quad \dots (70)$$

$$R = \begin{bmatrix} \cos(\lambda) & \sin(\lambda) \\ -\sin(\lambda) & \cos(\lambda) \end{bmatrix} \quad \dots (71)$$

$$Z_{6b} = R \times Z_{6a} \quad \dots (72)$$

$$Z_6 = [Z_{6bx} + Y_{0x}, Z_{6by} + Y_{0y}] \quad \dots (73)$$

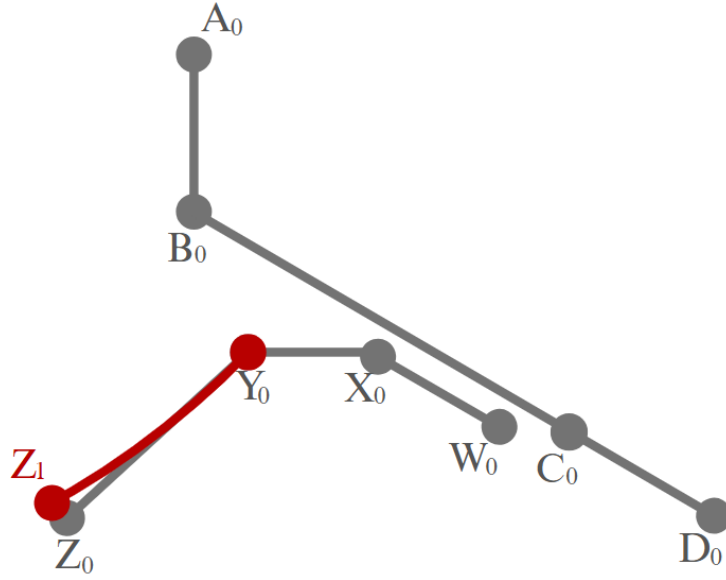


Figure 56: Conceptual representation of the deflection of ZY segment

Displacement of Segment YX

Next, the displacement of segment YX was considered. The displacement of the segment was caused by the force F_{y6} , and the moment M_7 , as illustrated in Figure 54. For segment YX, the displacement Δ_{YX} was calculated with equation 74 and the slope of the segment θ_7 with equation 75. The results of the deflection calculation had to be rotated with equation 77 to fit the slope of the initial YX segment. The equations used (76-79) are found below.

$$\Delta_{YX} = \frac{M_7 L_6^2}{2E_3 I_3} + \frac{F_{y6} L_6^3}{3E_3 I_3} \quad \dots (74)$$

$$\theta_7 = \frac{M_7 L_6}{E_3 I_3} + \frac{F_{y6} L_6^2}{2E_3 I_3} \quad \dots (75)$$

$$Y_{7a} = [-L_6, \Delta_{YX}] \quad \dots (76)$$

$$R = \begin{bmatrix} \cos(-\varepsilon) & \sin(-\varepsilon) \\ -\sin(-\varepsilon) & \cos(-\varepsilon) \end{bmatrix} \quad \dots (77)$$

$$Y_{7b} = R \times Y_{7a} \quad \dots (78)$$

$$Y_7 = [Y_{7bx} + X_{0x}, Y_{7by} + X_{0y}] \quad \dots (79)$$

The displacement of Y caused Z to displace as well. The new angle with the x-axis, γ_7 was calculated with equation 80. The Z-point was calculated with equation 81.

$$\gamma_7 = \lambda + \theta_7 \quad \dots (80)$$

$$Z_7 = [-L_7 \cos(\gamma_7) + Y_{7x}, -L_7 \sin(\gamma_7) + Y_{7y}] \quad \dots (81)$$

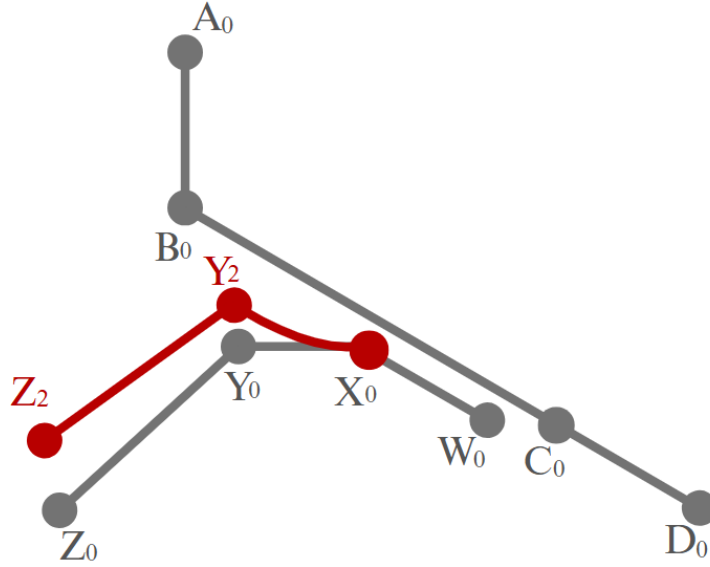


Figure 57: Conceptual representation of the deflection of segment YX

Final Displacement

Finally, the two cases were brought together to obtain the final displacement. The YX segment due to the moment and force, M_7 and F_{y6} , will be the final displacement for this section. The calculations to find the final Z point were done by rotating the point Z_{6a} with the rotation matrix, equation 83, and making the necessary adjustments towards the point Y_7 , equation 84.

$$R = \begin{bmatrix} \cos(\gamma_7) & \sin(\gamma_7) \\ -\sin(\gamma_7) & \cos(\gamma_7) \end{bmatrix} \quad \dots (82)$$

$$Z_{8a} = R \times Z_{6a} \quad \dots (83)$$

$$Z_8 = [Z_{8ax} + Y_{7x}, Z_{8ay} + Y_{7y}] \quad \dots (84)$$

The final displacement was calculated by subtracting the final Z point from the initial location of the Z node, as seen in equation 85.

$$\Delta_Z = Z_{8y} - Z_{0y} \quad \dots (85)$$

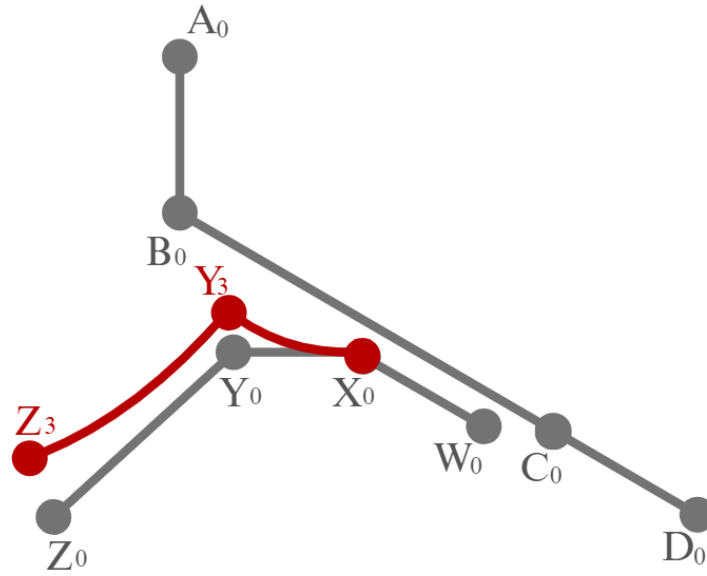


Figure 58: : Conceptual representation of the final displacement of the heel blade

The calculations were all verified with FEA in Siemens NX, and the results can be found in Appendix A.1. Of the tested cases, the deviation between the calculations and the FEA was below 1 mm. The largest absolute deviation was 0.77 mm (2.3%), observed during a forefoot test with a calculated displacement of 33 mm. The highest percentage deviation was 6.53 %, seen in a heel test with a calculated displacement of 10.29 mm and an absolute deviation of 0.6 mm. Despite the higher percentage, the small absolute deviation made this deviation acceptable. These results confirmed that the calculation method used was sound.

The calculations were all verified with FEA in Siemens NX. Of the tested cases, the deviation between the calculations and the FEA was below 1 mm. This was a small deviation when considering the displacement of each case. The percentages of these deviations varied between 0.9% and 6.6%, verifying that the calculation method was sound. The verification results for calculations can be found in Appendix A.1.

4.3.3. Gas Spring Calculations

The gas spring calculations followed the same steps as the forefoot blade in order to calculate the displacement of the forefoot when the gas springs were attached to the foot. Figure 59(a) is a simple illustration of where the gas spring attaches to the foot. Figure 59(b) illustrates how the gas spring force (GSF) acts on the forefoot along with the applied load.

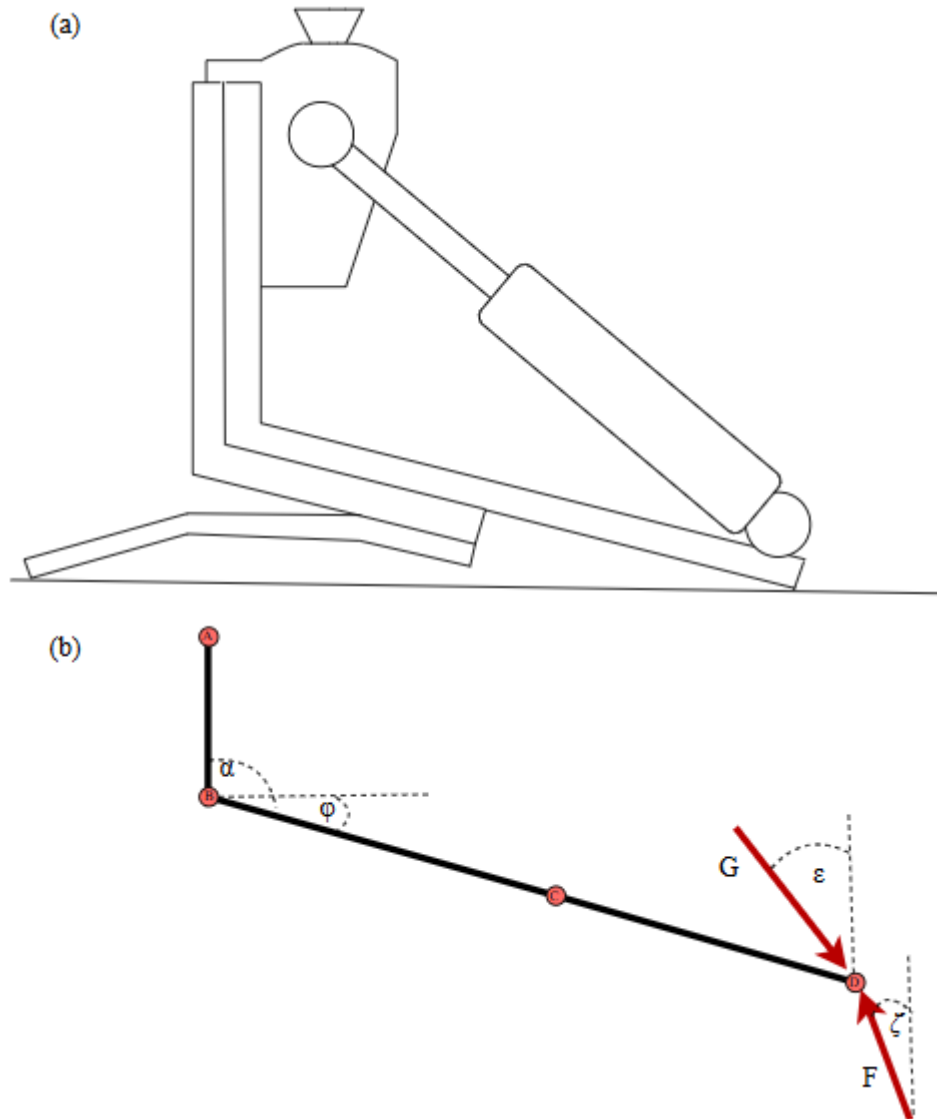


Figure 59: Free body diagram of gas springs attached to the prosthetic foot

Figure 60 illustrates how the applied load, F , and the gas spring force, G , were split into their respective x and y components. The components were then summed together to achieve the resultant force component, R_x and R_y . Equations 86-91 were used to calculate the resultant components. From here the resultant components were used as the acting force in the displacement calculation, and the same methodology was used as for the forefoot calculations.

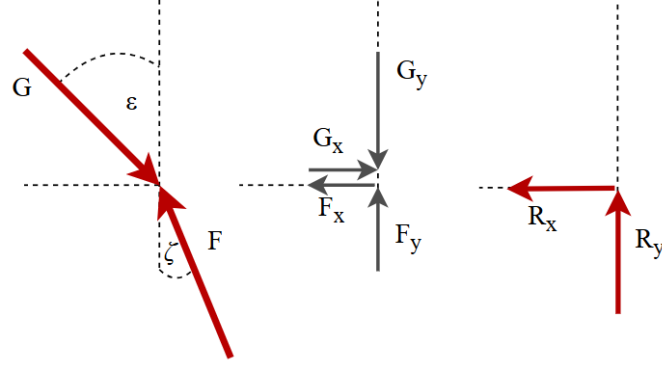


Figure 60: Resultant force breakdown on the prosthetic forefoot

$$G_x = G \times \sin (\varepsilon) \quad \dots (86)$$

$$G_y = G \times \cos (\varepsilon) \quad \dots (87)$$

$$F_x = F \times \sin (\varepsilon) \quad \dots (88)$$

$$F_y = F \times \cos (\varepsilon) \quad \dots (89)$$

$$R_x = -F_x + G_x \quad \dots (90)$$

$$R_y = F_y - G_y \quad \dots (91)$$

4.4. Dimensional Design

4.4.1. Prosthetic Foot Designer Program.

Input Variables

In section 4.2, the design specifications were discussed and summarised in Table 3. These specifications were used for the input values of the program. Some additional constraints were added to improve the efficiency of the program.

Figure 61 illustrates the flow of the MATLAB calculation program. This program is referred to as the “Designer” in this study, as it provided the dimensions with which to produce the prosthetic foot. The designer followed a stepwise approach in order to calculate the ideal dimensions for the input variables given to the program.

The available materials, unidirectional carbon fibre, allowed for a compression stress of 500 MPa, which was used as the maximum allowed stress. The Designer calculated an appropriate equivalent Young’s modulus of the three blades, restricted between 20 GPa and 90 GPa.

The length of L_1 had a minimum limit of 35 mm. This ensured that the angle of the blade allowed the foot to fit inside a shoe.

The program made use of two functions: the forefoot function and the heel blade function. The input variables described above were given to the main program. From here the input variables were sent to each function as required.

The forefoot function provided the solution for the forefoot dimensions, some of which were required to achieve a solution for the heel blade. The forefoot function is described in the next section.

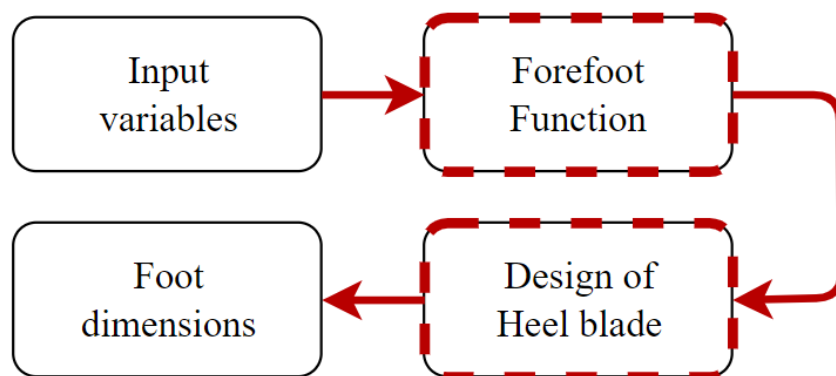


Figure 61: Flow of the prosthetic foot Designer calculation loop

Forefoot design function.

A programmed loop, illustrated in Figure 62, found the suitable values for the forefoot. To do this, input variables, minimum and maximum range values, as well as initial variable values, were given to the program. These values are recorded in Table 4.

Table 4: Input variables for forefoot design

Input Variable	Value	Unit
Mass	800	[N]
Safety Factor	1.5	[-]
Forefoot Stiffness	34.8	[N/mm]
Foot Size	270	[mm]
Foot Height	130	[mm]
Foot Width	45	[mm]
Pyramid Length	45	[mm]
Minimum L_1 value	35	[mm]
Allowed Compression Stress	500	[MPa]
Minimum Young's Modulus	20	[GPa]
Minimum Young's Modulus	90	[GPa]
Applied Force Angle ζ ,	20	[°]
Initial Angle α Value	115	[°]
Initial Thickness Ratio r_h Value	1.5	[-]
Initial Transformation Factor n_E Value	0.9	[-]

With the constant and initial values presented in Table 4, the function started with the initial value for the angle α , this starts Loop 1, as illustrated in Figure 62. Then the values for L_3 and L_4 were calculated using the angle α , the foot length, and the blade height. With these lengths known, the forces and moments were calculated with equations 18-23, and then used to calculate h_1 .

The value for h_2 was calculated by using the initial value of the thickness ratio r_h . The thickness ratio is the ratio between h_2 and h_1 . This started the second loop. Loop 3 initiated with the initial value of the transformation factor n_E . The flow of these loops is shown in Figure 62. The equivalent thickness was calculated as described, using equation 16 and 17. Using the equivalent height and the design displacement, Young's modulus, E_1 , was calculated. With this, along with the value of n_E , the value of E_{2eq} was calculated, and then the final E_2 was calculated.

The results delivered by the function were then checked. The results were only saved if they conformed to the limit of L_1 , were in the allowed composite Young's modulus range and did not exceed the allowed stress.

After considering the results, the loops started over. First, Loop 3 ran through all the possible iterations, increasing n_E by 0.05 until the transformation factor equalled 1.3. Once all the possible variables were run through, the value for r_h was increased by 0.05. and Loop 2 ran again as well, as the entirety of Loop 3 with the new h_2 value. This process was repeated until r_h equals 2.5. Once this was the case, Loop 1 restarted and α decreased by one degree. The entire process ran again, completing Loop 3 for each new Loop 2 run, and once this was completed, Loop 1 restarted. This process continued until α equals 95 degrees. These limits were gathered by running the code multiple times and investigating the results. The limits made the code more efficient by removing the cases where no results were gathered.

When all loops were complete, the saved results were sorted in order of ascending blade thickness, h_1 , angle α . This was done to ensure that the lowest values of each of these were used.

The captured values were the solution to the forefoot blade. Next, the heel blade function is explained.

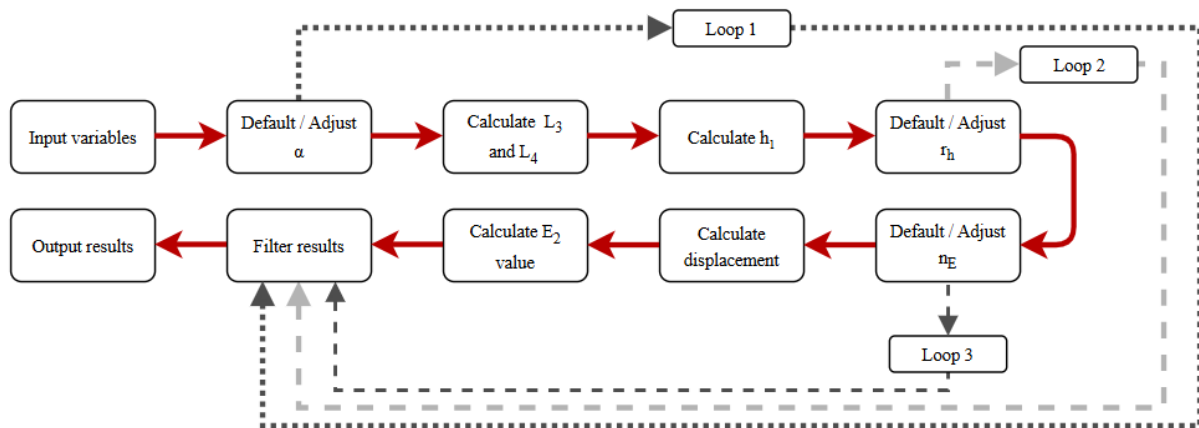


Figure 62: Flow of the forefoot function during prosthetic foot calculation

Design of the heel blade

Similarly to the forefoot function, the heel blade was designed using a MATLAB function. The heel blade function, illustrated in Figure 63, was given the input variables named below:

Mass	800	[N]
Safety Factor	1.5	[-]
Design Stiffness Heel	53.4	[N/mm]
Foot Size	270	[mm]
Foot Width	45	[mm]
Lenght L_1 Value from Results	L_1	[mm]
Lenght L_2 Value from Results	L_2	[mm]
Lenght L_3 Value from Results	L_3	[mm]
Lenght L_4 Value from Results	L_4	[mm]
Allowed Compression Stress	500	[MPa]
Minimum Young's Modulus	20	[GPa]
Minimum Young's Modulus	90	[GPa]
Applied Force Angle ζ ,	15	[°]
Angle α From Results	115	[°]
Thickness h_2 From Results	1.5	[-]

The heel blade function consists of four loops. Continuing the numbering from the forefoot function, Loop 4 started by initialising L_5 as 30. L_6 , has an initial value of 50 mm, used for Loop 5, and angle β an initial value of 155°, used for Loop 6. The final loop started with an initial value of 4 for h_3 . With all the variables set to their initial values, the initial positions of the heelblade points and the applicable forces were calculated. The Young's modulus was calculated, and the results were filtered.

The results had to adhere to the following conditions:

- The angle λ is less than 30 °. This design choice ensures that the heel blade does not have a large angle with the ground.
- E_3 is within the allowed range
- L_7 is within 10 mm of the L_6 value. This is done to keep angles from being too sharp.
- The maximum stress is within the allowed range.

After the results were considered, the loops were repeated with the next set of input variables. The loops are nested within each other, as illustrated in Figure 63. Loop 7 increased the value of h_3 0.1 mm until the value reached 8 mm. Once Loop 7 was complete, Loop 6 ran through its possible values, increasing the angle β by 5° until the value reached 180° . The value of L_6 was increased to 120 mm in increments of 10 mm within Loop 5 and L_5 increased in 5 mm increments until a value of 50 mm was reached, completing Loop 4.

The results were sorted to have λ and the thickness h_3 ascending. The top value, the value with the lowest λ angle and then the lowest thickness, was then chosen as the blade's solution.

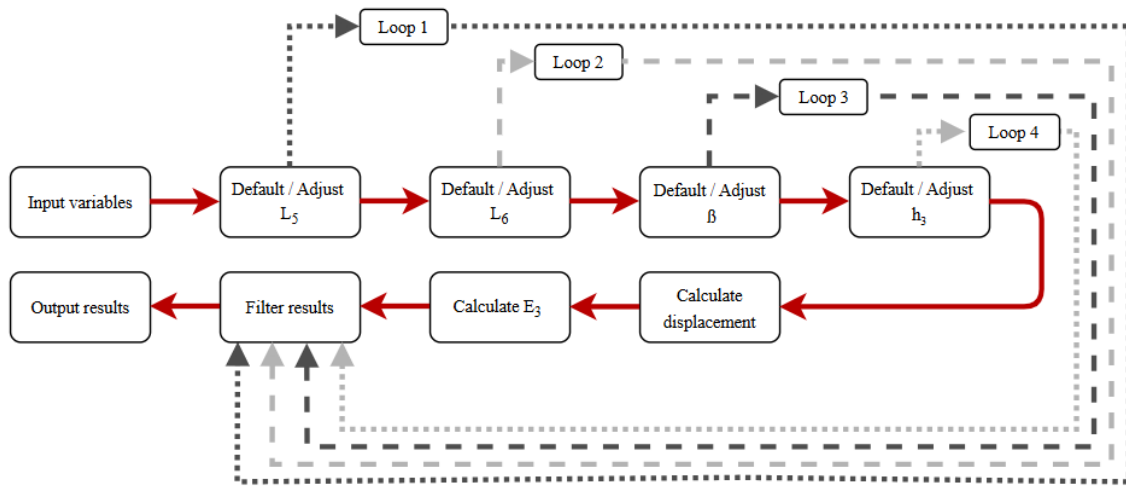


Figure 63: Flow of the heel blade function during prosthetic foot calculation

Designer results

The results of the MATLAB Designer are summarised in Table 5. These results were rounded to the nearest whole number, which was a more practical choice. However, the blade heights (h_1 , h_2 , and h_3) and the Young's moduli (E_1 , E_2 , and E_3) were rounded to one decimal place. This level of precision was retained since Young's moduli are usually large values, and ply layers composing the blade heights are often thinner than a millimetre. Therefore, the blade can be built to a height with one decimal point of accuracy.

These results are used in the hand calculations in Appendix A.2 to verify the output of the Designer. After the Designer results were verified, the output values were further verified using FEA simulations.

As can be seen in Table 5, the values of h_2 and h_{eq} are the same, as well as the Young's moduli E_2 and E_{eq} . This means that the thickness and elasticity of the blade would not have to change when using the equivalent thickness approach to validate the results from the Designer.

Table 5: Results of the prosthetic foot Designer

Dimension	Value	Unit
α	104	[°]
β	160	[°]
λ	21	[°]
η	-6	[°]
L_1	44	[mm]
L_2	175	[mm]
L_3	105	[mm]
L_4	70	[mm]
L_5	30	[mm]
L_6	50	[mm]
L_7	49	[mm]
h_1	4.3	[mm]
h_2	7.4	[mm]
h_{eq}	7.4	[mm]
h_3	5.6	[mm]
E_1	63.7	[GPa]
E_{2eq}	63.7	[GPa]
E_2	63.7	[GPa]
E_3	24.2	[GPa]
$\Delta_{Forefoot}$	24.73	[mm]
Δ_{Heel}	13.31	[mm]

Table 6 compares the displacement obtained by using the rounded dimensions with the target displacement values that were used in the Designer to calculate the dimension. The results show that the rounding of the Designer values had a large effect on the outcome of the displacements. The forefoot had nearly an 8% difference between the target and calculated displacements. Similarly, the rounded heel blade dimensions lead to an 11% difference. While these differences were relatively large, the results from the Designer were considered acceptable since the target displacements only served as parameters to inform the choice of input parameters for the Designer algorithm.

Table 6: Designer displacement compared to target displacement

Forefoot	Designer Displacement	24.73	[mm]
	Target Displacement	22.99	[mm]
	Difference	1.74	[mm]
	Difference	7.55	[%]
Heel	Designer Displacement	13.31	[mm]
	Target Displacement	14.94	[mm]
	Difference	1.631	[mm]
	Difference	10.92	[%]

4.4.2. Gas Spring Calculation

Gas springs are commercially available in standard sizes and can be pressurised for the force output required. Based on the values from the designer, an angle of 54° for angle ε was predicted, as illustrated in Figure 59(b). This angle affected the x and y force components of the gas springs. With the angle known, the target spring forces were calculated.

As discussed during the design concept evaluation, the usage of gas springs could adjust the stiffness of the prosthetic foot to fit a person's bodyweight. For this study, three gas spring cases will be studied. The suggested gas spring forces were calculated in 150 N increments of the design mass of 800 N.

The target applied loads along with the suggested gas spring force (GSF) are recorded in Table 7. These gas spring forces, when applied, were expected to result in the same displacement as an applied load of 800 N with no gas springs attached.

Table 7: Gas spring forces required for equal displacement of 800N

Target Applied Load [N]	Calculated Suggested GSF [N]
950	307.2
1100	629.5
1250	961.5

The suggested gas spring force values, along with the applicable target applied load values from Table 7, were used to acquire the displacement values from FEA. The setup of the FEA is discussed in Appendix A.3.

Table 8 presents the recoded displacement values from FEA and compares the displacements to the displacement of the foot with no gas springs attached under an applied load of 800 N. The results are within 1 mm of each other and were therefore considered acceptable.

Table 8: Validation of calculations by FEA for a 300 N gas spring applied load

Target Applied Load [N] /	Displacement from FEA [mm]	Deviation from displacement at 800 N [mm]	Deviation from displacement at 800 N [%]
Suggested GSF [N]	24.73	N/A	N/A
800 / 0	24.73	0.05	0.20
950 / 307.2	24.78	0.5	2.02
1100 / 629.5	25.23	0.83	3.36

4.5. Layup Model Design

In this section the dimensions calculated in the previous section were used to calculate the layup schedule for each of the composite components of the blade. Composite materials allow the layer thickness, material type and fibre angle to be modified to obtain the required strength and stiffness of each component.

As mentioned, the foot is designed for a person of about 80 kg. As discussed in Section 4.2, the foot can experience 1.2 times the body weight, and an additional safety factor of 1.25 was applied to the design.

There are multiple failure criteria to be used when considering composite design. Tsai- Wu, for example, considers how stress works together and treats tensile and compressive strength differently and therefore requires more material properties in order to be more accurate [116]. Tsai-Hill also considers how stresses work together but does not differentiate between compression and tensile strength and does not differentiate between the fibre and the matrix [116]. Hashin's criteria examines the fibre and the matrix separately for failures and account for the effects of shear stresses [117].

The Tsai-Hill criterion was used for the evaluation of the simulations, as done by Talla HK et al. in their study on a prosthetic running blade [112]. This criterion is typically used for anisotropic composites with varying material properties to predict the failure of the composite material [118]. The criterion predicts failure as the failure index (PI) reaches one, as can be seen when studying equation 92 . From this equation, it was learnt that a PI of 0.45 must not be exceeded to comply with the safety factor.

$$Safety\ factor = \frac{1}{\sqrt{PI}} \quad \dots (92)$$

The layup analysis was done in Siemens NX, and FEA gave an estimated displacement and PI. The displacement was aimed at being within 1 mm of the calculated displacement of the forefoot and heel, 24.73 mm and 13.31 mm, respectively. Along with the aimed displacement, the PI shouldn't have exceeded the aforementioned 0.45.

4.5.1. Initial Ply Layup

Several layup orders may be suitable for the study's prosthetic foot. Consequently, an initial layup served as a starting point for developing the final configuration. The rule of mixtures was used to predict the layup.

Using two composite materials with different moduli, C_1 and C_2 , a layup that results in Young's modulus equivalent to E_1 , E_2 , and E_3 , from the Designer outputs in Table 5, was calculated. Figure 64 illustrates how the plies were laid for the initial layup. The stiffer and stronger composite, C_1 , was stacked at the outsides of the blade where the highest bending stresses were encountered.

The less stiff material, C_2 , was used closer to the neutral axis where lower stresses are found, contributing to weight saving and potential improvements in flexibility. Using C_2 helped tailor the overall flexibility of the foot without significantly compromising the strength.

The combination of the two composites delivered an equivalent modulus for the blade comparable to those from the Designer outputs in Table 5. The values for the initial layup can be found in Table 9. Here the thicknesses, T_1 , T_2 , and T_{total} , are given as well.

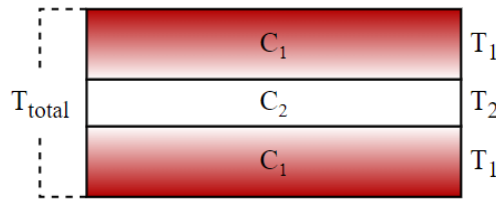


Figure 64: Illustration of the initial ply layup

Table 9: Values for the initial layup of the three blades

	E of C_1 [GPA]	E of C_2 [GPA]	T_1 [mm]	T_2 [mm]	T_total [mm]
Blade 1 (Primary Forefoot Blade)	96.0	55.0	0.4	3.5	4.3
Blade 2 (Primary Forefoot Blade)	96.0	55.0	0.4	2.3	3.1
Blade 3 (Heel Blade)	96.0	17.5	0.2	5.2	5.6

4.5.2. Final Layup

After multiple iterations of ply layups, a solution was achieved where the FEA displacement was close to the calculated displacement and the PI conforms to the desired safety factor. Overall, no changes were made to the dimensions from the Designer. The only difference was that the forefoot blade was a bit thicker than what the calculations suggest. The thickness of the primary blade, h_1 , was changed from 4.3 mm to 4.7 mm to improve the PI value in the foot. The layers of the blades were staggered to decrease the thickness in the areas where less load was seen. The stagger of these layers increased the deflection again to accommodate the loss in displacement observed due to the increased thickness.

Figure 65 illustrates the final layup schedule of the primary blade of the forefoot. For the most part, the layup schedule was precisely as the initial layup predicted. The second composite material used, bidirectional carbon fibre, was rotated 45° at the centre of the blade. This change, along with the sanding of the staggered edge as seen in Figure 65, increased the displacement of the forefoot and also eliminated the stress concentrations that were caused on the blade. Multiple iterations were considered for the ply-layups, but the end layup structures fit best in regard to displacement and PI. Illustrations of the final layups of all the blades can be found in Appendix B.

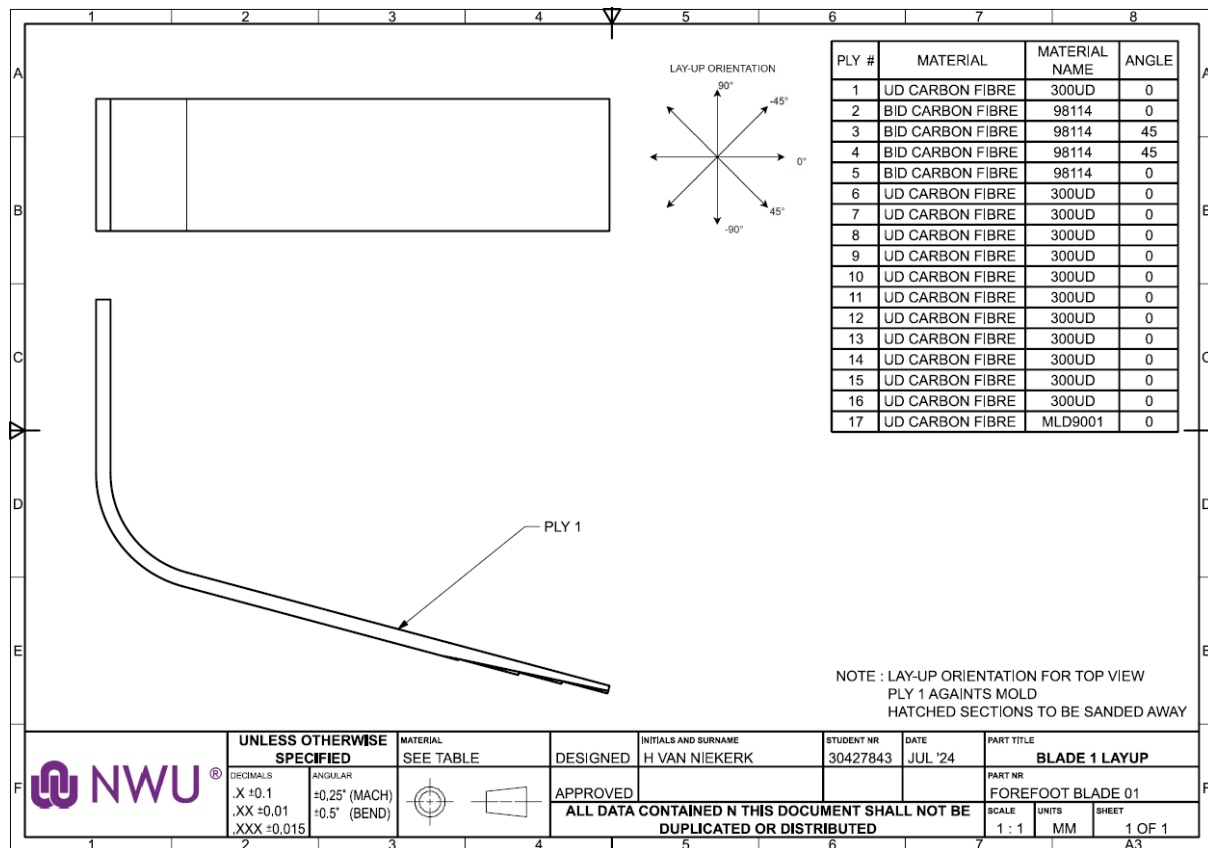


Figure 65: Final layup of blade 1

The recorded FEA results of the final layup are found in Table 10 along with the displacement calculated with the Designer results. As can be seen, both the forefoot difference and the heel difference were below 5% from the Designer value, corresponding to a difference of approximately 1 mm. This comparison verifies that the layup solution based on the Designers dimensions can produce displacement results that remain within an acceptable range. Full comparison tables for the heel blade and all instances of the forefoot blade can be found in Appendix A.4.

Table 10: FEA Results of final layup for the forefoot and heel

Forefoot	Designer Displacement	24.73	[mm]
	Layup FEA Displacement	23.61	[mm]
	Difference	1.12	[mm]
	Difference	4.73	[%]
Heel	Designer Displacement	13.31	[mm]
	Layup FEA Displacement	13.58	[mm]
	Difference	0.27	[mm]
	Difference	2.00	[%]

4.6. Cost Analysis of Manufacturing the Prosthetic Foot

This section describes the costs of the prosthetic foot suggested by this study to see how it compared to the prices found in the Gazette 2023 [37]. The cost analysis was divided into material costs, manufacturing costs, labour costs, and additional costs. Additionally, a profit margin was considered to determine whether a prosthetic foot can be provided at an affordable price. Since this analysis was an estimation, approximate prices were used from online stores, and no formal quotes were necessarily obtained. To ensure conservative estimates, the highest possible costs were assumed.

Material Costs

The primary materials used for the prosthetic foot are the composite materials for the blades, namely bi-directional and unidirectional carbon fibre sheets, as well as a resin-hardener mixture. Supawood is used for the moulds on which the blades are built. Assuming the foot is able to provide three different stiffness levels, one with no gas springs attached and then two with different sizes of gas springs, two sets of springs would be required along with eight end fittings. A summary of the cost estimation is in Table 11 given below. The full breakdown can be found in Appendix D.

Table 11: Total material cost

Material	Cost per unit (R/u)	Units required	Cost
Bi-directional material	526.70	0.41	214.27
Unidirectional material	1720.40	0.59	1012.23
Resin	238.44	0.28	67.52
Hardner	289.94	0.08	22.17
Mould material	178.64	0.33	59.68
Gas springs	122.76	4.00	491.04
End Fittings	25.41	8.00	203.28
Total material Cost			2070.19

Manufacturing costs

The cost to manufacture the mould, out of a high-density fibre board, on which the blades will be manufactured, as well as the manufacturing cost of the composite blades, to sand and finish them, is summarised below. The estimated cost of the pyramid adapter is also captured below. This cost is estimated from a quote on a similar type of component. The full breakdown of the costs is captured in Appendix D.

Table 12: Total manufacturing cost

Component	Rate (R/h)	Estimated time	Cost
Mould manufacturing	500.00	1.00	500.00
Composite blades (layup)	500.00	4.00	2000.00
Composite blades (finishing)	500.00	4.00	2000.00
Pyramid connector			1000.00
Total manufacturing Cost			5500.00

Total Cost Estimate

Combining all material, manufacturing, and labour costs, a total of R7570.19. Additionally, another 50% is added to the total cost to cover any additional costs not mentioned in this evaluation, such as costs associated with medical or ISO standard testing, overheads, and logistic costs. These testing expenses would be distributed across the total number of prosthetic feet produced; this would depend on the market size. A study would be required on the South African market size, as very little information is available on this topic. Furthermore, 50% is added to the total cost of the prosthetic foot as a profit margin. This is only an estimate, as it would require an in-depth cost evaluation to see if the profit margin would be acceptable.

Table 13: Total price estimation

Cost	Value (R)
Total material cost	2070.19
Total manufacturing cost	5500.00
Total material and manufacturing	7570.19
Additional Cost (50 %)	3785.09
Total Cost	11355.28
Profit (50%)	5677.64
Estimated price	17032.93

As outlined in Chapter 1, the compensation figures in the *Gazette 2023* [37] indicate that the most affordable dynamic response (DR) foot costs R21 032, and the most expensive costs R63 921. In comparison, the cost evaluation for the prosthetic foot concept in this study is more affordable by about R 4 000. Not only is it more affordable, but it may also remove the need for multiple prosthetic feet for different activities or weight classes due to its adjustable stiffness. Detailed cost analysis and further studies focusing on market readiness will need to be done to get a more accurate understanding of the true cost of supplying an affordable prosthetic foot.

The cost estimation for the prosthetic foot was done quite conservatively, assuming the higher costs of materials and manufacturing rates. It is possible that the cost of the prosthetic foot could be decreased when economies of scale are considered, which would make the foot even more affordable.

Cost reduction

Increasing the number of prosthetic feet that are manufactured would decrease the cost per prosthetic foot.

- **Mould Costs:** The moulds are reusable after each layup. The mould costs can therefore be divided between the total number of manufactured prosthetic feet. The cost of moulds can be distributed across multiple units.
- **Material Costs:** On the sides of the manufactured blades, there are certain amounts of excess material that are removed. By building more than one foot at a time on a larger mould, less material would be wasted.

4.7. Chapter Summary

In this chapter, the development of the detailed design of the chosen concept for an adjustable foot was described. The chapter started off with the design specifications for the prosthetic foot, followed by determining the best values for the blade dimensions by using the Designer program that followed the calculation method presented. The dimensions were verified by hand calculations.

There was a difference between the displacement calculated and the displacement that was originally designed for. This deviation was due to the rounding of the input dimension values, and since the target displacement values were not intended to be rigidly fixed goals, the calculated dimensions were accepted.

The force output of the gas springs was calculated and rounded. The effects of the gas spring on the prosthetic foot were calculated and determined by FEA, and the results were compared.

With these dimensions, a composite layup model was developed. The FEA of the layup results was validated by comparison to the Designer displacements.

Finally, the high-level cost analysis presented a breakdown of the cost of labour and materials used to manufacture the prosthetic foot. The evaluation included a profit margin and insurance for missed expenses. The cost was estimated to be R 4 000 lower than the lowest available compensation option from the *Gazette 2023* [37].

Chapter 5 : Experimental Procedure

5.1. Chapter Overview

To validate that the prosthetic foot presented in this study performed as the calculations intended, the calculation results were compared to experimental results. This chapter briefly explains the experimental procedure used to test the prosthetic foot. A full list of the equipment used is found in Appendix E.1, and the step-by-step methodology is in Appendix E.2.

5.2. Experimental Setup

An MTS machine performed a compression test on the prosthetic foot. The machine has hydraulic clamps at the top and bottom which were used during testing to position the foot at the correct angles for the applied load lines. The bottom clamp was used to hold the testing jig in place, and the top clamp was used to press down on the prosthetic foot. Figure 66 illustrates the setup for the forefoot with the gas springs connected (Figure 66 a) and the heel test (Figure 66 b).

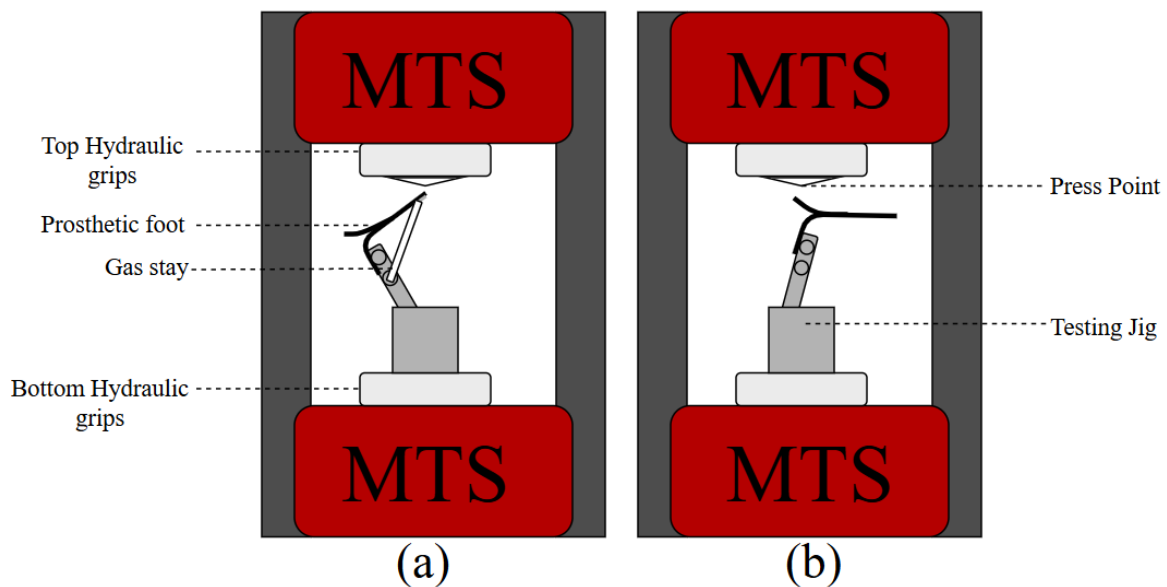


Figure 66: The experimental setup of the forefoot tests (a) and heel tests (b)

The test jig allowed for the prosthetic foot to be tested at the angles provided by ISO 10328, summarised in Appendix F. The jig has two positions, one for forefoot testing and the second for heel testing. The test positions were easily changed using a wrench and an allen key to move the bolt and nut to the appropriate hole.

The prosthetic foot was attached to the test jig using the holes where the pyramid adapter would have fit. The foot was attached with nuts and bolts, using allen keys to tighten the attachment. The gas springs were attached to the connector fittings found on the forefoot of the prosthetic foot and on the connector fitting found on the testing jig.

5.3. Experimental Methodology

The prosthetic foot was attached to the testing jig and clamped in the MTS machine, as seen in Figure 67. The jig allowed for the forefoot to be tested at 20° and the heel at 15° as recommended by ISO 10328. The MTS machine pressed down on the prosthetic foot at 1.5 mm/s, and the displacement along with the applied force was recorded.

Three tests were performed on both the forefoot and heel to demonstrate repeatability and to obtain consistent and accurate data. The tests were stopped at 800 N, which was the design load.

After the forefoot and heels were tested in the basic configuration, the gas springs were attached to the foot. The test was stopped at the appropriate force or if the gas spring was fully compressed.



Figure 67: Prosthetic foot during testing with two 500 N gas springs attached

5.4. Data and Results Processing

A program on the MTS machine was used to press down on the test subject at 1.5 mm/s. However, tests conducted on a prosthetic foot are typically done at a Newtons-per-second rate, and at the time of testing, such a program was not available for use.

The data collected from the MTS machine required clean-up and analysis before it could be reliably compared to calculations. This process involved identifying and removing abnormalities such as sudden drops or spikes in the data, which are common during experimental testing, particularly when using sensors as the MTS machine does. Sudden drops in force readings may result from the shifting or settling of carbon strands within the test model. These shifts could temporarily affect the force measurement, causing the machine to register a lower force at a single point before continuing with accurate force readings. Since the MTS machine moved at millimetres per second, such faulty force measurements resulted in errors in the displacement data. To ensure accurate analysis and comparison, these data points had to be carefully identified and filtered.

Some deviation was expected to occur between experimental runs and between the experimental and calculated results. The data comparisons were done at every 1 mm of deflection. The data for all the experimental runs can be found in Appendix C.

5.5. Chapter Summary

This chapter described the experimental setup and method used to gather the experimental data for the next chapter.

An MTS machine was used to gather the experimental data. The prosthetic foot was placed in a jig in the MTS machine, and the jig allowed the foot to be pressed at the required angles and the attachment of the gas springs.

The data from the MTS machine was processed before it could be compared to the calculated values. This comparison is presented in the following chapter.

The full methodology can be found in Appendix E along with a list of the equipment used.

Chapter 6 : Results

6.1. Chapter Overview

In this chapter, the results from the experimental procedure are presented and compared to the calculated results. Firstly, the dimensions of the prosthetic blades were verified to confirm if the manufactured blades meet the original specifications. After this verification, any dimensional discrepancies between the designed foot and the manufactured foot were accounted for in the calculations.

The newly calculated values were then verified against FEA to determine whether the calculations done with the adjusted dimensions are correct. Verification in this context confirmed the alignment of the calculation method and the finite element analysis.

The results of the experimental runs were verified against each other to confirm the repeatability of the experiments. Once the results were verified, the calculation results could be validated against the experimental results. This validation step confirmed if the prosthetic foot performed as expected.

This verification and validation process, as illustrated in Figure 68, was done for the heel blade, the forefoot, and three gas spring force configurations on the forefoot. The results between the different gas spring load cases were compared to determine if the addition of gas springs did affect the prosthetic foot stiffness.

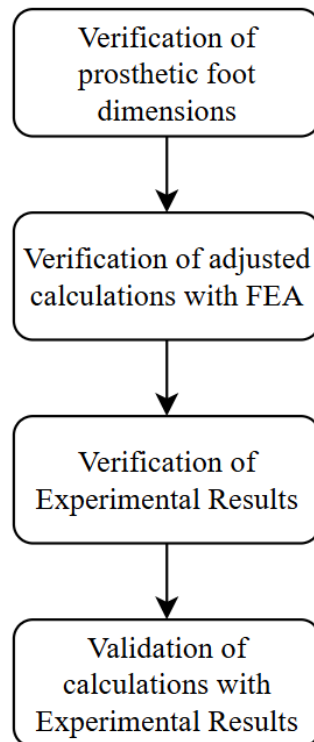


Figure 68: Flow of the study's validation and verification

6.2. Validation of Calculations

The prosthetic foot was fabricated according to the layup structures specified in the detailed drawings in Appendix B. However, the process did not achieve the exact thickness specified in the design. Specifically, the combined thickness of the two forefoot blades, h_2 in Figure 41, increased from 7.4 mm to 8.5 mm, approximately 14.9%. Similarly, the heel blade thickness, h_3 , increased from 5.6 mm to 6.7 mm, representing a 19.6% increase.

This is likely due to an excess amount of resin retained in the parts during the manufacturing process. The measured thickness values were updated in the FEA model and the calculations. The corresponding displacement results were gathered as explained in Chapter 4 and are presented in Table 14.

As shown in Table 14, the FEA and calculation values for the forefoot blade were within 2% (approximately 0.2 mm) of each other. This correlation was considered satisfactory, allowing the calculations to be used as a baseline for verifying whether the experimental results align with expectations. A similar comparison for the heel blade, shown in Table 14, indicated that the FEA and calculated values were within 3% of each other, which was also acceptable.

Table 14: Comparison of displacement results of FEA and calculations for forefoot and heel Blade

Forefoot	Calculation Displacement	15.35	[mm]
	FEA Displacement	15.18	[mm]
	Difference	0.17	[mm]
	Difference	1.13	[%]
Heel	Calculation Displacement	7.74	[mm]
	FEA Displacement	7.56	[mm]
	Difference	0.18	[mm]
	Difference	2.33	[%]

Despite the strong agreement between the FEA and calculations, there was a noticeable difference in deflection values between the designed specifications and predictions from the calculations, as can be seen in Figure 69. The results of the adjusted calculations were stiffer than those of the intended calculations for the target design specifications due to the additional thickness.

The increased stiffness led to an increase in force to achieve the intended displacement. This increased stiffness would likely not be ideal for user experience, as the studies from Chapter 2 found that feet which were too stiff led to discomfort. and a noticeable limp when walking.

Since this study did not require human participation and the primary objective was the design of a variable stiffness prosthetic foot, the slightly thicker foot was still used for testing to prove that the method of characterisation and adjustment of the stiffness was a valid approach. The verified calculations, listed in Table 14, provided a reliable basis for evaluating the experimental data.

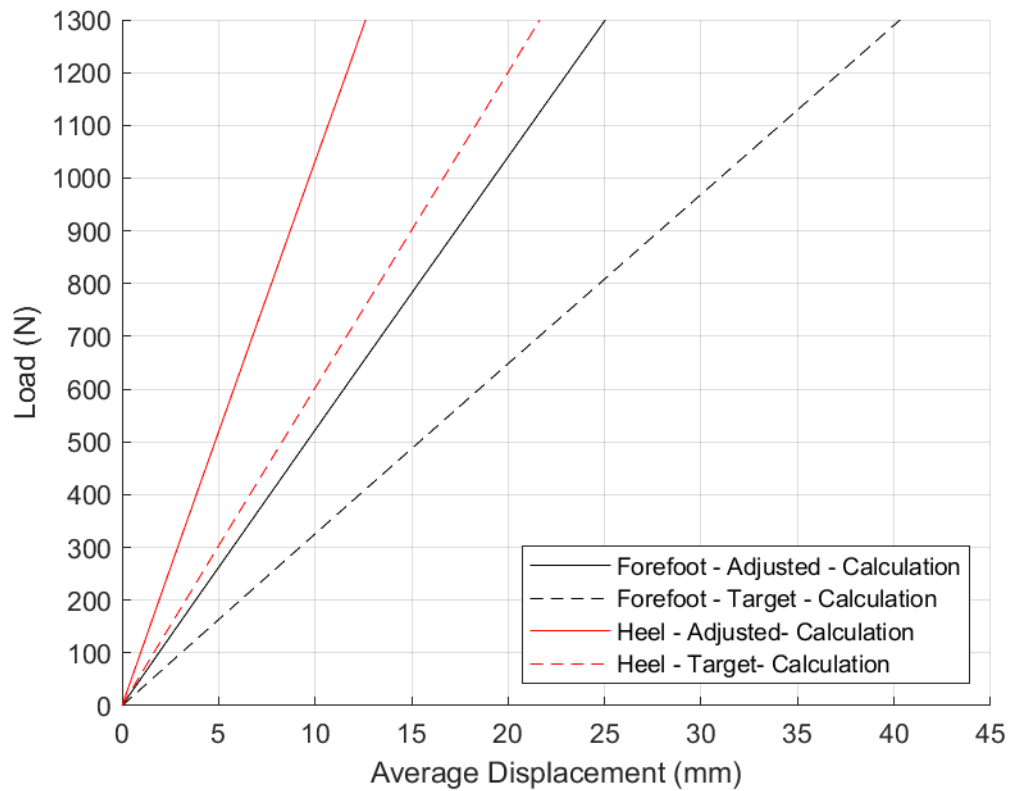


Figure 69: Comparison between intended deflection and deflection of adjusted calculations

Three gas springs were acquired from a commercial source for the experiment, based on the suggested values from Chapter 4. The closest available alternatives were obtained, with their values recorded in Table 15. Since the gas springs were mounted onto the jig, the angle ε was adjusted to 46° . This angle increased the affected component of the gas spring's force, making its effects apparent. All these modifications were incorporated into the calculations, and the results were verified through FEA. The verification results are provided in Appendix 0.

Table 15: Obtained gas spring forces

GSF of a single gas spring [N]	Total GSF for two gas springs [N]
150	300
350	700
500	1000

Next, the verification and consistency check of the experimental data is described, followed by the validation process of the design calculations.

6.3. Basic Prosthetic Foot Results

6.3.1. Validation and Verification of Forefoot Results: 0 N GSF

The repeatability of the forefoot measurements without any applied gas spring force (GSF) to it is examined first, with the test results presented in Figure 70. The results of the different runs were highly comparable, confirming the repeatability of the experiment.

Slight deviations between the runs were expected. The largest observed standard deviation, 8.51 N, occurred around 9 mm, which can be seen in Figure 70. The largest deviation of a run from the average plotted data was from Run 2, where the force deviation was about 15.03 N at a displacement of 11.7 mm.

These deviations observed between runs were minimal and acceptable within the scope of the results and confirmed the repeatability of the experimental method. A table presenting the deviation of each run from the experimental average can be found in Appendix C.1.

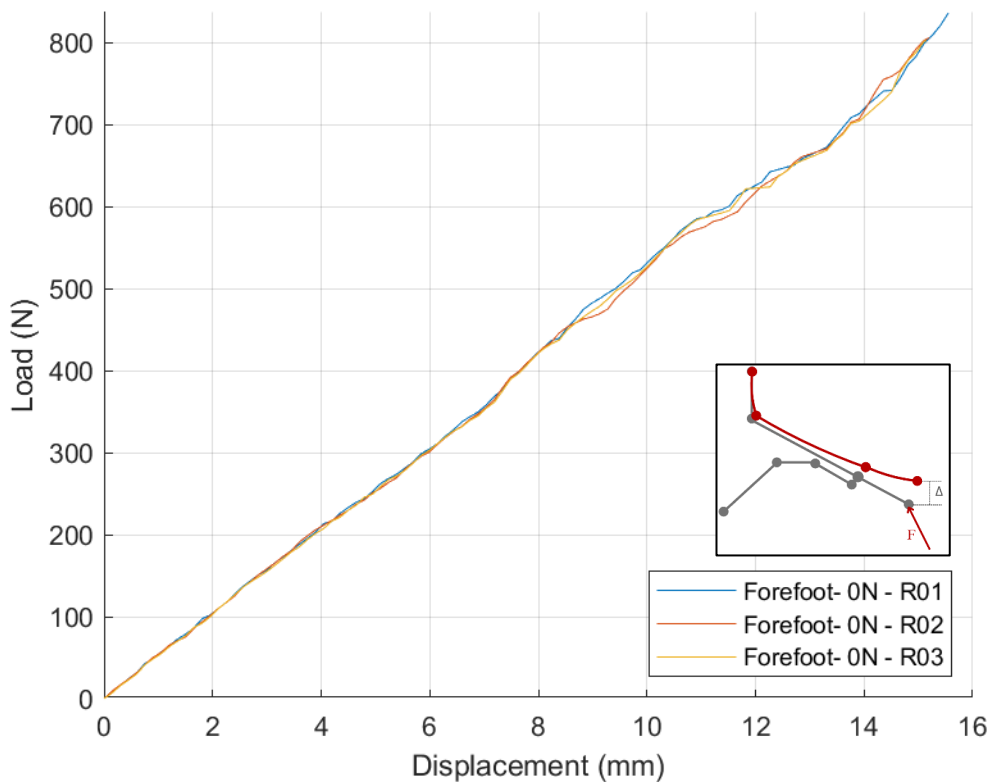


Figure 70: Displacement vs applied load of the forefoot experiment without gas springs

Subsequently, the numerical modelling approach and calculations were validated through comparison with the experimental results. Figure 71 presents a plot of the calculation results, including a 10% deviation reference margin above and below the calculated data. The averaged values of the recorded force data are also included in this plot.

The average data points followed the trendline established by the calculation results, with the largest deviations from calculations occurring around displacements of 13 mm and 14 mm, with deviations of about 16 N, as can be seen in the table found in Appendix C.1. Overall, the comparison between the calculations and the experimental results of the forefoot without gas springs demonstrated a strong correlation, confirming the validity of the design methodology against real-world performance.

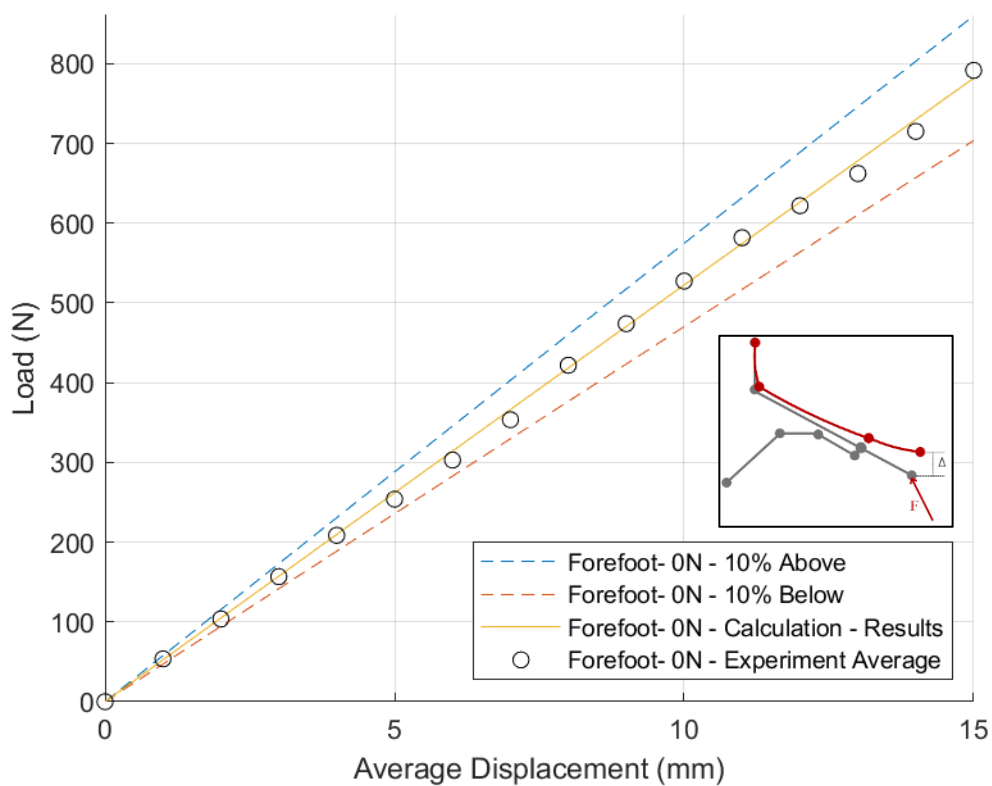


Figure 71: Comparison of calculations and experimental results of the forefoot without gas springs

6.3.2. Validation and Verification of Forefoot Results: 300 N GSF

When gas springs were attached to the foot, the force exerted by the gas springs must first be overcome before the foot could start to deflect. With the two 150 N gas springs attached to the forefoot, adding to the 300 N GSF, approximately 200 N of force was required to initiate deflection, as can be seen in Figure 72. Run 3 deviated more from the other two, presenting a slightly stiffer response than Runs 1 and 2.

When analysing the displacement data in millimetre intervals, the largest standard deviation was 13.6 N at a displacement of 13 mm. Run 3 deviated most from the average data at 12.9 mm with a deviation of 18.96 N. These deviations are, however, minimal when considering the recorded load, thus confirming the repeatability of the experimental approach (for a table of detailed measurements, see Appendix C.2).

A comparison of the calculation results with the experimental data is shown in Figure 73. Noticeable deviations occurred as the applied load increased. The largest deviation of 79.45 N (9.15%) can be seen at 13 mm.

From the plot in Figure 73 it seems that the experimental results are less stiff than the calculations predicted. Since the results of the prosthetic foot without gas springs did not indicate a significant difference between the results of the calculations and the experiment, it can be deduced that the source of deviations does not appear to be related to the blades of the foot. Rather, it seems plausible that some aspects related to the addition of the gas springs to the foot may contribute to the differences.

The gas springs were attached to ball joints on steel brackets on the foot, which in turn were in contact with the MTS press. This steel-on-steel contact possibly caused slipping due to the friction between the two steel parts, which caused deviations in the measured data; it is also plausible that the gas springs were not exerting exactly the theoretical force as expected, meaning the gas springs were filled to within an unknown tolerance of the specified pressure by the commercial suppliers.

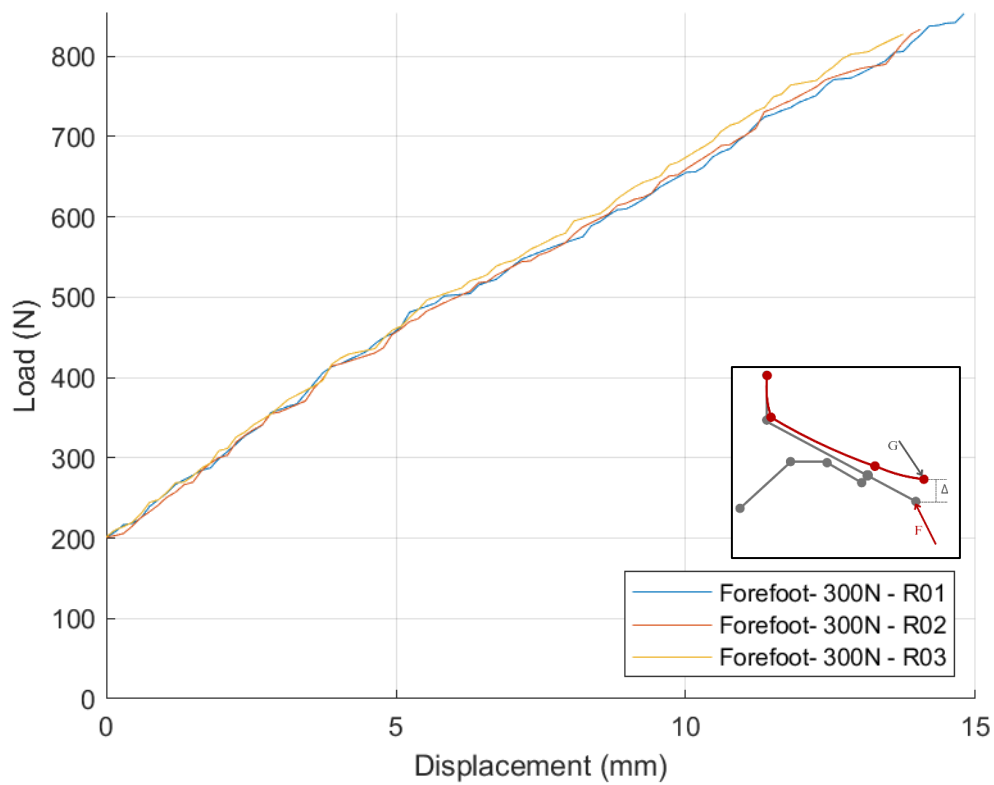


Figure 72: Displacement results of the forefoot experiment of the 300 N GSF load case

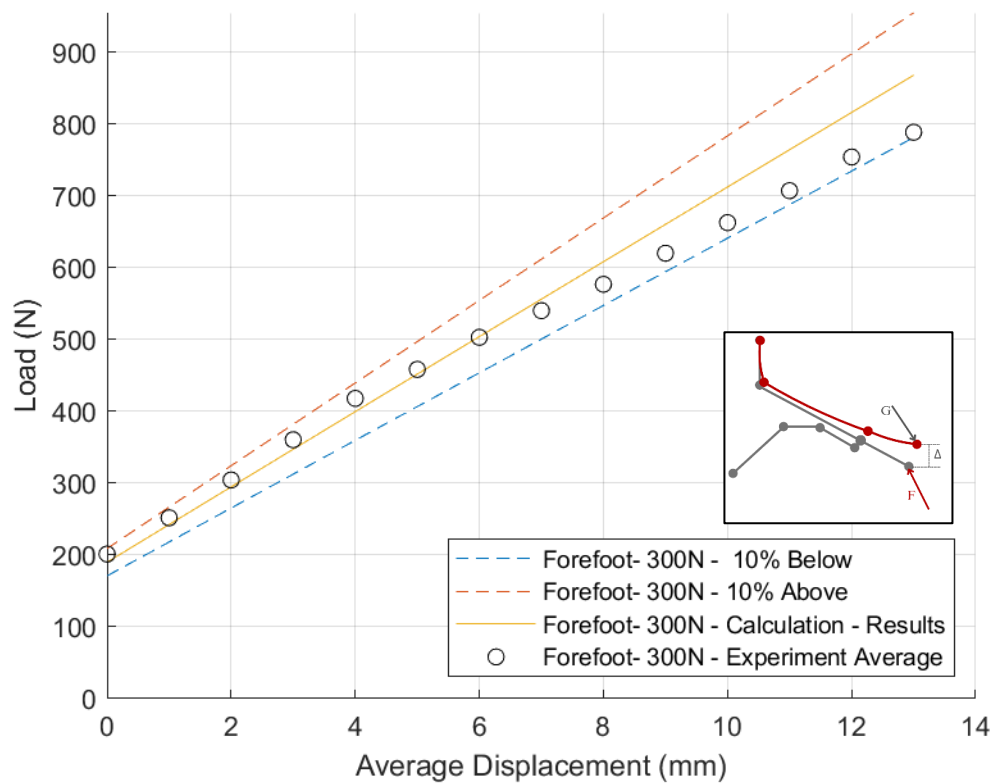


Figure 73: Comparison of calculations and experimental results of the 300 N GSF load case

6.3.3. Validation and Verification of Forefoot Results: 700 N GSF

When the two 350 N gas springs, adding up to a GSF of 700 N, were attached to the foot, a force of approximately 450 N was required to initiate deflection. The results of these experiments are shown in Figure 74. Despite the deviations between experimental runs, the data followed a noticeable linear trend.

The larger standard deviations of around 8 N occurred at displacements of 1 mm, 3 mm, and 4 mm. Run 2 had the furthest deviation from the average data, with 17.9 N at 1.35 mm. The maximum deviations between each millimetre point of the three runs are recorded in a table in Appendix C.3. This was a small deviation (3.3%) when considering the recorded applied load of 530 N. The experimental runs with the applied 700 N GSF were considered repeatable due to the highly comparable results of the three runs.

Unlike the results comparison between the calculations and the 300 N GSF experiment, the comparison of the 700 GSF case presented the experimental average results stiffer than what the calculations predicted. The largest deviation from the calculations, 64.5 N, was found at 6 mm. Not only was this deviation smaller than the largest deviation from the 300 N GSF experiment, but it was also closer compared to the experimental average, deviating only 8.5% from the predicted value.

Although the experimental data shows some variability, the comparison with the calculated results still demonstrated acceptable alignment. This indicates that the experimental data provides sufficient validation of the calculations for the 700 N GSF load case on the prosthetic foot.

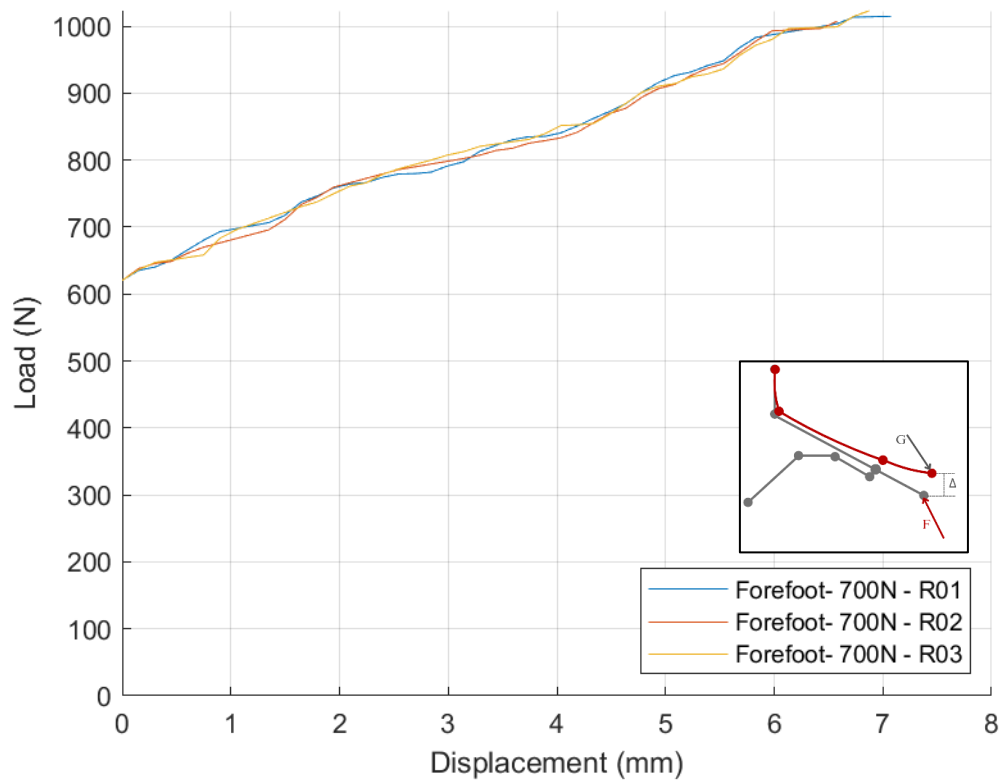


Figure 74: Displacement results of the forefoot of the 700 N GSF load case

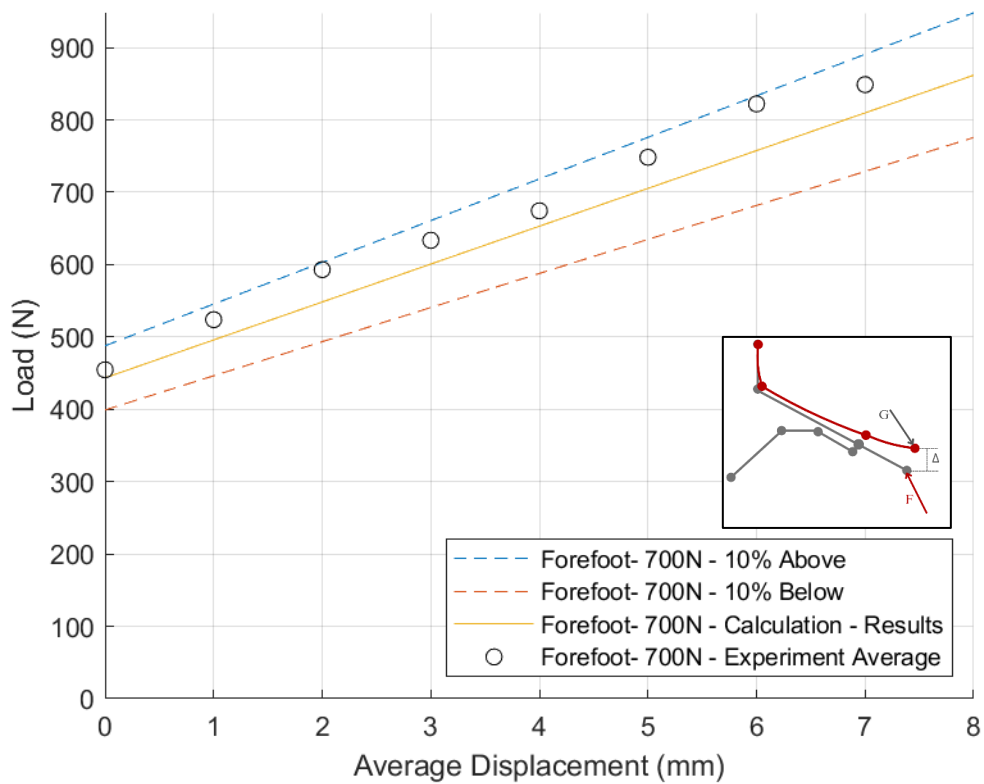


Figure 75: Comparison of calculations and experimental results of the 700 N GSF load case

6.3.4. Validation and Verification of Forefoot Results: 1000 N GSF

To displace the foot with an applied GSF of 1000 N (two 500 N gas springs), a force of approximately 650 N had to be exceeded. The results of the experiments, illustrated in Figure 76, showed a correlation between the experimental runs; however, the correlation deviated more between runs than that of previous experiments with gas springs applied.

The larger deviations were seen at the start of the experimental runs, following a linear trendline as the experiment continued. The largest standard deviation out of all the experiments, 17.72 N, was present at 2 mm. The largest run-average deviation came from Run 2 at 1.65 mm with a deviation of 24.35 N. A table of displacement values at every millimetre displacement interval can be found in Appendix C.4.

Additional differences were evident in the final values of the three runs.

- Run 1 was stopped before reaching 1000 N, likely due to the MTS machine's starting position being set higher than intended.
- Run 3 exhibits a sudden incline just after 1000 N, likely caused by the metal brackets slipping on the testing machine's metal press.
- Run 2 continues linearly beyond 1000 N without the irregularities seen in Runs 1 and 3. Although the limited number of tests beyond this point restricts statistical confidence, Run 2 was considered as the representation of the experiment's results due to its alignment with the expected behaviour.

The experimental data was compared to the calculations in Figure 77. The results indicated that the experiment results and the calculations were comparable, with the largest deviation of 79.37 N at the 6 mm mark. This was no surprise, as Run 1 did not exceed the 6 mm line, and Run 3 inclined just before the 6 mm mark. This caused the average calculated at the point to be higher. When excluding the 6 mm data, all the remaining data points were very close to the calculation line, the largest deviation being 17.5 N (5.22%) at 3 mm.

Overall, the experimental data provided a reasonable validation of the calculated response for the 1000 N GSF load case. Despite the deviations, the data followed a trend that matched that predicted by the calculations, with deviations less than 5.5% across the displacement range.

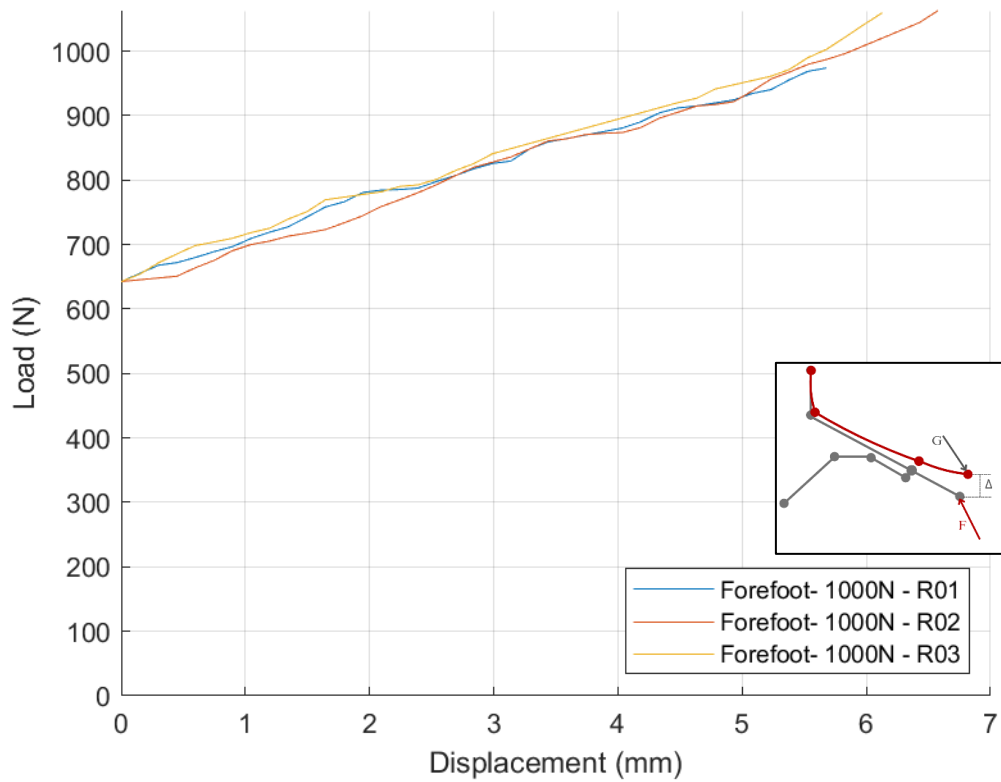


Figure 76: Displacement results of the forefoot of the 1000 N GSF load case

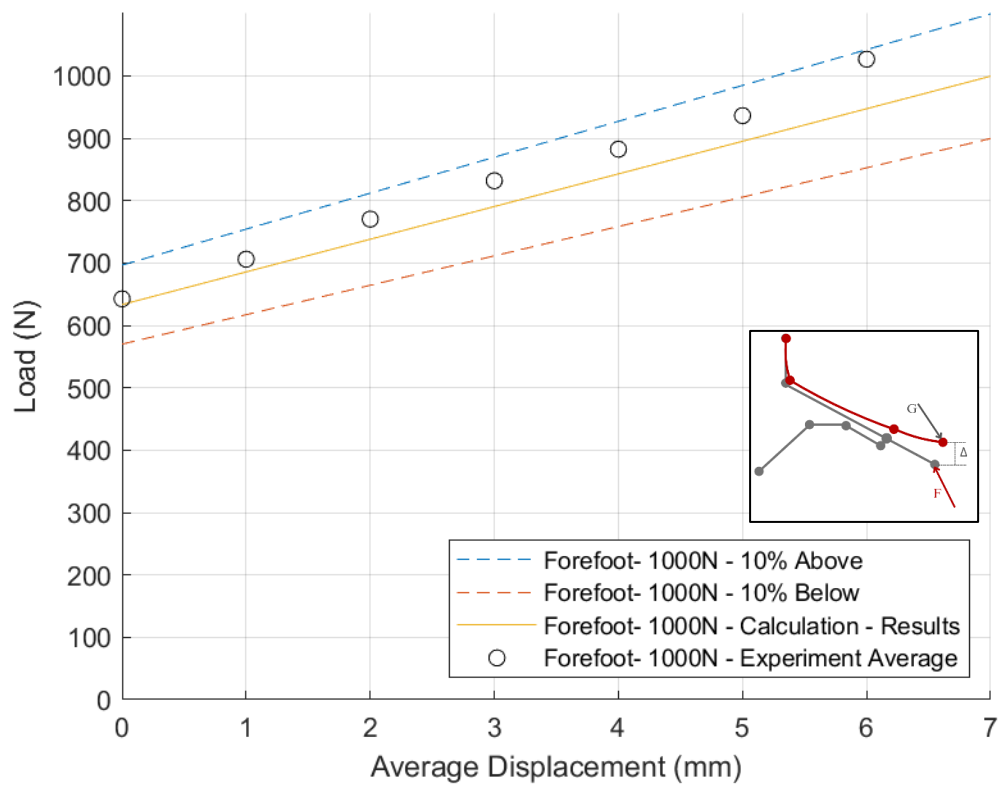


Figure 77: Comparison of calculations and experimental results of the 1000 N GSF load case

6.3.5. Validation and Verification of Heel Results

The results of the experiments conducted on the heel blade, illustrated in Figure 78, showed a clear correspondence among the three runs, with the data closely following a linear trend. However, a deviation could be seen in runs 2 and 3 between 400N and 500N. This did, however, self-correct further along the graph.

Appendix C.5 includes a table summarising the displacements at every millimetre interval and the standard deviation between the three runs. Both Runs 2 and 3 had their maximum deviations from the experimental average at 3.9 mm displacement. Run 2 had a deviation of 17.42 N, and Run 3, the largest deviation between the three runs, had a deviation of 20.68 N. These deviations were small when the applied load at that instance was considered and thus complemented the experiment's repeatability.

The comparison results in Figure 79, between the calculations and the experiment, reveal that the heel blade was stiffer than the calculations predicted. This was likely due to the blade not being sanded as intended or errors in the ply layup angles during the manufacturing process.

It was difficult to sand the heel blade exactly to the size and dimensions as the design intended due to its shape. Its curves created the risk of oversanding or unintentionally creating grooves in the part, leading to insufficient sanding of the blade. Insufficient sanding led to increased stiffness due to the extra material. An excess of sanding would have led to a less stiff foot and possibly weak spots.

The heelblade also had multiple plies that had to be laid up at 45°. Errors in cutting the fabric to the exact angle or the plies moving while the blades were manufactured could also likely result in the blade being stiffer than predicted.

Overall, the calculations for the heelblade were validated by the experiment results, as the results demonstrated a strong correlation between the calculations and experimental data. Although a stronger correlation between the experimental results and the 10% deviation reference margin was seen, indicating that the heel is stiffer than the calculation model.

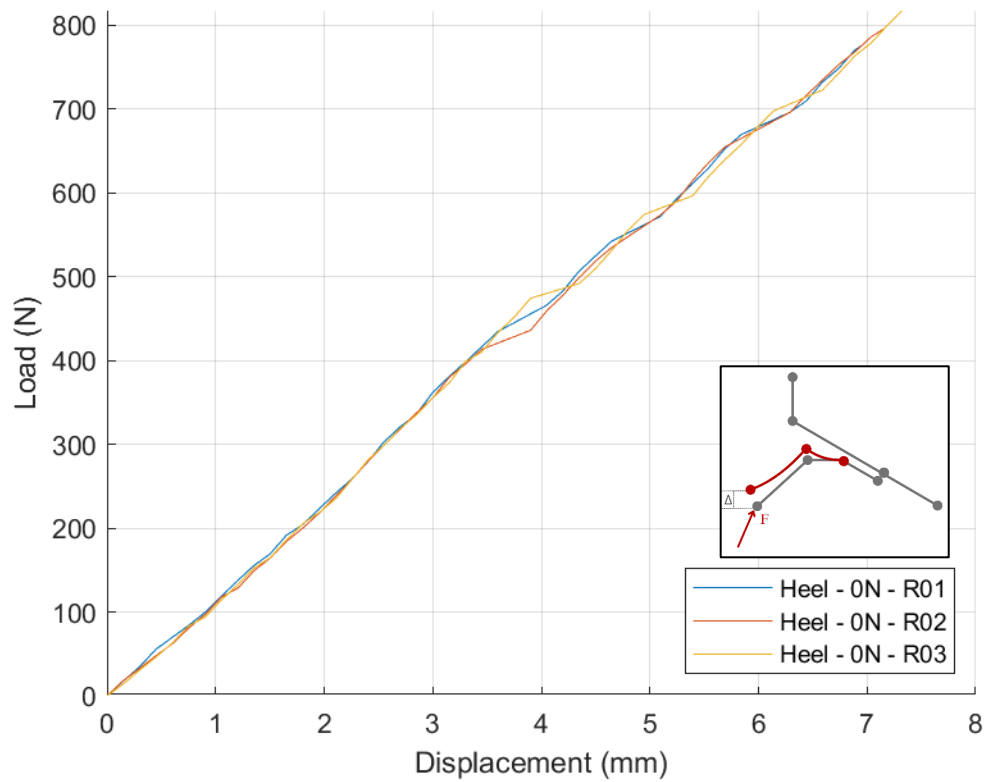


Figure 78: Experimental results for heel blade: load vs. displacement.

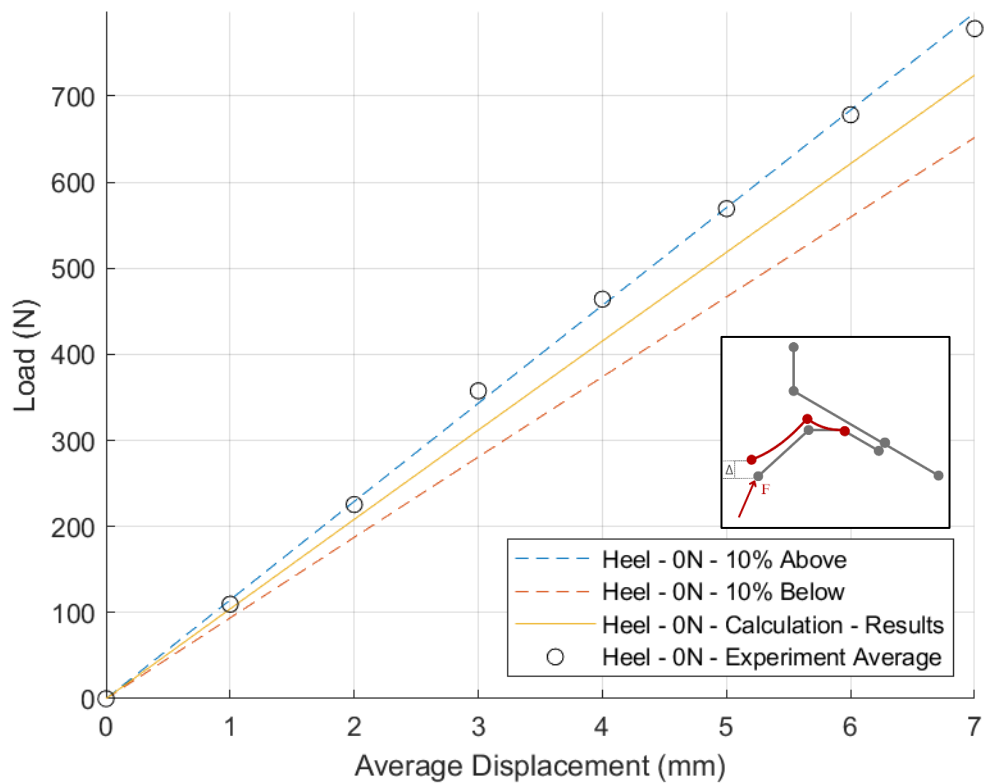


Figure 79: Comparison of experimental results with calculations for the heel blade

6.4. Discussion

When considering all the averages of the measured values alongside the calculations, as illustrated in Figure 80, the variation in stiffness between various configurations became evident. The addition of a stronger gas spring increased the force required to initiate deflection, while the gradient of the force-displacement curve remained similar to the original foot configuration (without the gas springs).

The plot in Figure 80 validates the design choice that the addition of gas springs does adjust the stiffness of the prosthetic foot. It is, however, clear that the deviations increased with the addition of gas springs. The deviations were, however, not excessively large. The largest deviation came from the 300 N GSF experiment, with a deviation of 79.45 N (9.15%).

The possible range of the stiffness adjustment suggests that the prosthetic foot design would be suitable to implement adjustability for varying activity levels or body weights. Meaning that a basic, blade-only foot designed for a specific weight could be strengthened by adding gas springs so that the foot can support the additional loads. Further research would help clarify whether this adjustment is sufficient for the intended activities and dynamic movement.

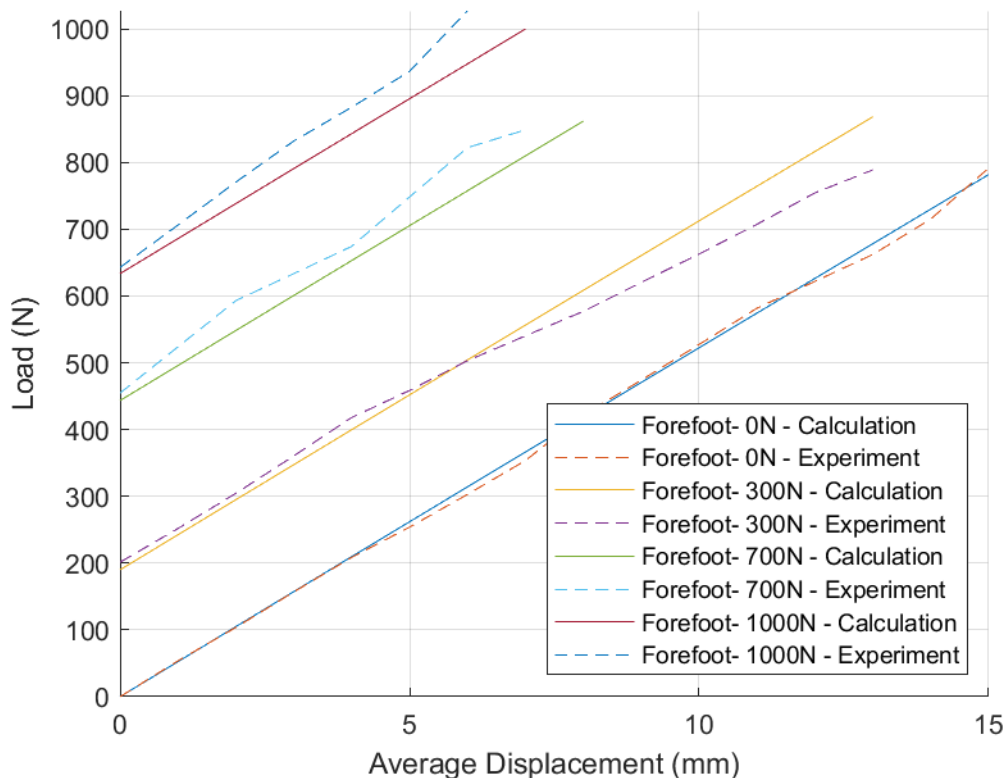


Figure 80: Comparison of all experiment averages and calculations

Table 16: Applied forces comparisons at a displacement of 6 mm

GSF [N]	Calculated Applied Load at 6 mm displacement[N]	Experimental Load at 6 mm Displacement [N]	Deviation (N)	Deviation (%)
0	314.21	302.96	11.25	3.58
300	504.22	503.24	0.98	0.19
700	757.54	822.08	64.54	8.52
1000	947.54	1026.91	79.37	8.38

Table 16 presents the force results at a 6 mm displacement for the forefoot, both with and without gas springs. The prosthetic foot, originally designed without springs for an 800 N applied load, required approximately 300 N to achieve a 6 mm displacement during experimental testing. With the gas springs, the force required for the same displacement increased.

Considering the 300 N GSF, closest to the gas spring values from the Chapter 4 calculations, it is clear that the gas spring adjusted the foot within the predicted calculation range. The same can be said for the 700 N and 1000 N gas spring cases, although the error percentage was larger in these cases.

Even though the results from the experiments served as validation that the prosthetic foot performed as the calculations predicted, there were still deviations from the expected results. Possible causes of these deviations could be dimensional inaccuracies; manufacturing errors, equipment issues, or discrepancies during the testing process.

As mentioned, a possible cause for the deviations was due to dimensional inaccuracies. Both the manufactured heel blade and the forefoot blade were thicker than the design intended. The calculations were, however, adjusted to account for this change in thickness for both the heel and forefoot. The second problem was sanding the blades to the intended position.

Figure 81 illustrates where the heel blade and the forefoot blade should be sanded. Sanding to the exact dimensions of the design posed a problem during the manufacturing process, especially on the heel blade. The curve in the heel blade's shape made it more difficult to sand compared to the forefoot that had to be sanded to a flat edge.

An insufficient level of sanding of the heel blade was the most probable cause leading to the experimental results being stiffer than the calculations, as it can be seen in Figure 79.

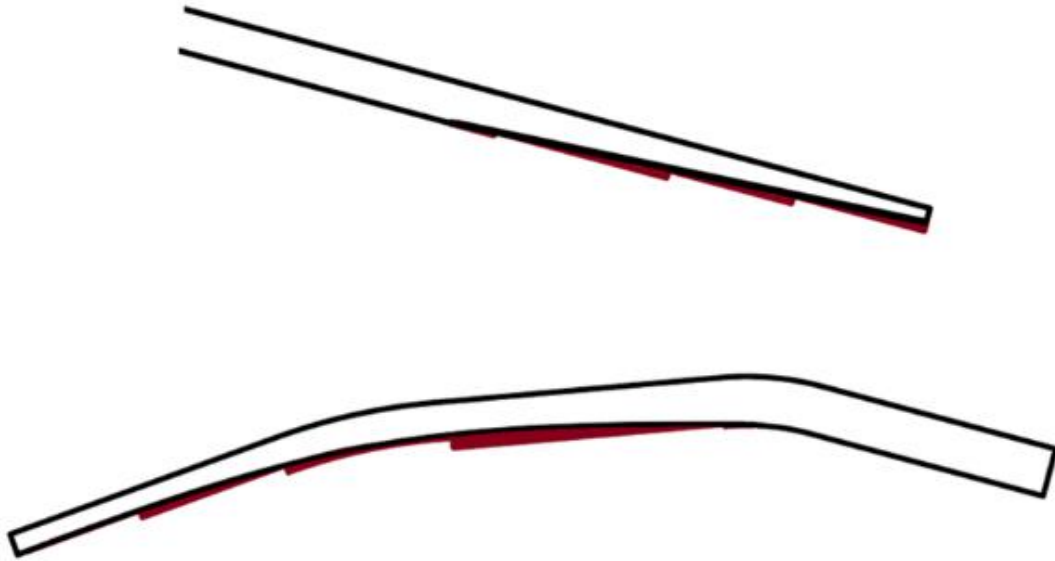


Figure 81: Illustration of the sanding on the forefoot (above) and heel blade (below)

Dimensional inaccuracies were less likely to be the main cause of deviation in the forefoot tests due to its simpler geometry, which allowed for more controlled sanding. The experimental results closely correlated with the calculations, and the slight acceptable deviations were likely caused by minor discrepancies in ply layup angles during manufacturing. Future studies should incorporate a formal uncertainty analysis to quantify the influence of manufacturing variability on the experimental results.

In most cases, the largest deviation occurred at higher applied loads when the gas springs were attached. This contrasts with results from the experiment without gas springs, which displayed a highly linear trend and closely aligned with the calculations. This suggested a correlation between the observed deviations and the attachment of the gas springs. The large deflections could have influenced the angle of the gas struts during loading, affecting the applied force direction and magnitude. This factor was not considered in the current model and should be investigated in future analyses.

The percentage error of the largest deviations of each experiment can be seen in Table 17. When the gas springs were attached, the maximum deviation increased. Without the gas springs, a maximum deviation of 16 N (2.4%) was observed between the calculated applied load and the experimental applied load. With gas springs attached, a deviation of approximately 80 N was observed between the calculated and experimental loads for both the 300 N GSF load case, which saw a 9.2% error, and the 1000 N GSF load cases, which saw an 8.4 % error.

Table 17: Experiment deviations from calculations

GSF Case	Calculated Applied Load [N]	Experiment Load [N]	Deviation (N)	Deviation (%)
No Gas Spring	678.22	662.20	16.02	2.36
300 N	868.24	788.79	79.45	9.15
700 N	757.54	822.08	64.54	8.52
1000 N	947.54	1026.91	79.37	8.38

This issue could potentially be mitigated by configuring the MTS machine to operate based on force per second rather than displacement per second. A force-controlled approach is typically used for the ISO testing for prosthetics. This approach was, however, not used for this study, as the program did not exist on the MTS at the time of the study. A force-controlled approach could allow for more precise determination of the initial loading conditions for each experimental run and could yield more consistent and representative graphs.

Another possible cause for the increase in deviation when the gas springs were attached was slipping between the foot and the press area. At this contact point, the steel press and the steel brackets are in contact, and the gas springs add an additional force, increasing the friction between the press and the foot.

To mitigate the slippage, two different materials with a lower friction coefficient could be used. This would reduce the slipping due to the friction between the two parts. Using different materials would then prove whether the deviations were truly caused by the slipping between the two parts or if something else was at fault for the deviations. The addition of gas springs may have increased the friction between these surfaces, causing deviations.

6.5. Chapter Summary

In this chapter the experimental and calculated displacement values for each experiment case were compared.

Firstly, the dimensions of the manufactured blades were verified against the target dimensions. It was found that the manufactured blades were thicker than intended. This had a noticeable effect on the stiffness of the foot. The adjusted dimensions were used to calculate new displacement results, and the calculations were verified against the results of an FEA model. The thicker manufactured blades were subsequently used during the experiments.

The results of different experimental runs under the same load cases were verified against each other to ensure that the data captured was repeatable between different experiment runs. Overall, the experimental results were highly comparable with the calculations, validating that the prosthetic foot performed as the calculations intended.

The heel blade experiment was found to be stiffer than the calculations predicted. This was concluded to be likely due to how the heel blade was sanded after manufacturing.

The comparison of all the gas spring experiments validated that the stiffness of the foot was adjustable. It was clear from the graphed results that the foot stiffness changed, as with each gas spring addition, a higher initial force had to be overcome in order for the foot to experience displacement.

The results gathered from the experiments were satisfactory when compared to the calculations, and the results proved that the stiffness of the prosthetic foot can ultimately be adjusted.

Chapter 7 : Conclusion and Future Work

7.1. Chapter Overview

This chapter concludes the study by revisiting its motivation, the design concept evaluation, and the manufacturing of the prosthetic foot. It reviews the experimental results to consider whether the study presents a valid solution for an adjustable prosthetic foot.

Finally, the chapter proposes several future research and design options that would improve upon the work done in this study.

7.2. Conclusion

The motivation for this study stemmed from how expensive prosthetic setups are and how few people are able to afford them in low-income countries like South Africa. The prosthetic foot was identified as the most commonly used component, as most lower limb prosthetic setups require them. Additionally, it was found that people who would wish to participate in various activities would require more than one prosthetic foot.

The literature review provided answers to four of the objectives stated in Chapter 1.

- *Investigation of how the human foot moves and how a prosthetic foot is related during the same movement.*
 - The gait cycle was investigated, and its different stages were explained. It was found that during the gait cycle the foot can experience 1.2 times the user's body weight, with peak forces occurring at heel impact and toe push-off.
- *Determining how current prosthetic feet function, what types of feet exist, and when they would be used.*
 - Various types of prosthetic feet were examined, leading to the selection of a dynamic response foot as the basis for this study. These feet provide more movement and energy return, making them ideal. Analysis of dynamic response foot movement showed that heel and forefoot displacements propel the user through the gait cycle.
- *Review on the effect the stiffness of a prosthetic foot has on the gait of the user.*
 - Stiffness was found to be crucial in the users' gait cycle. If a foot was experienced as too stiff or flexible, a noticeable limp was observed and discomfort was experienced.

- *Determining what materials are typically used for prosthetic feet and if cheaper substitutes could be used.*
 - It was found that carbon fibre, fibreglass, and Kevlar were commonly used materials for prosthetic feet due to their desirable characteristics of strength and light weight. Carbon fibre was selected as the material for the blades of the prosthetic foot for this study.

Following the literature review, three concepts were presented, each offering a different method for adjusting stiffness. This led to a concept evaluation, completing another objective from Chapter 1:

- *Evaluation of multiple design options and selection of the most suitable option, considering performance, manufacturability, and cost.*
 - The selected concept integrated gas springs into the prosthetic foot, allowing users to swap gas springs of different strengths based on comfort, activity level, or body weight.

The chosen concept was refined through calculations, using the stiffness of a commercial foot and maintaining constant dimensions as a reference. A target displacement of 23.0 mm and 14.9 mm for the forefoot and heel was set in the Designer program. The Designer produced target dimensions for the foot. For practicality, these dimensions were rounded, resulting in final displacements which led to a change of 24.7 mm and 13.3 mm in the displacements of the forefoot and heel, respectively.

A CAD model based on these dimensions was developed, and finite element analysis (FEA) performed on the model verified the Designer's dimensions, confirming that the calculation methodology was sound. A layup model was then designed for manufacturing, completing another study objective:

- *Verify the prototype through comparison to the design specification and the values of commercial prosthetic feet.*
 - The stiffness of a commercial prosthetic foot was used as a reference. Calculations were verified through FEA models, ensuring consistency with the intended design.

The design of the composite blades and the manufacturing thereof was an objective of this study.

- *Design and manufacture of a prototype adjustable dynamic response foot.*
 - The composite blades consisted of unidirectional and bidirectional carbon fibre layers. The FEA for the composite model aligned closely with the Designer calculations.

The manufacturing model deviated from the initial design in blade thickness, with the combined thickness of the two forefoot blades and the heel blade being approximately 1mm thicker. The calculations and the FEA models were adjusted to fit the dimensions of the manufactured foot. The calculations were once again verified through FEA and showed a strong correlation, with deviations within 3%. Although these differences affected stiffness, the primary objective—demonstrating the feasibility of stiffness adjustment using gas springs—remained unchanged.

- *Validate the prototype by comparing the calculated data to the experimental data.*
 - Experiments were conducted on the manufactured prosthetic foot, measuring the displacement of the heel, the forefoot without any gas springs, and the forefoot with the 300 N GSF, 700 N GSF, and 1000 N GSF. The results aligned with calculated data, confirming that the addition of gas springs effectively increased the stiffness of the prosthetic foot.

The experiments without the gas springs applied showed a highly linear correlation, with the largest deviations from the calculations from the forefoot being 2.4% at 16 N and 14.6% at 45.56 N for the heel. However, when gas springs were applied, there was an increase in deviations, typically falling within a range of 3% to 8.5% of the calculated results.

The addition of a gas spring increased the force required to initiate deflection. With a 300 N GSF, a force of 200 N initiated deflection; a 700 N GSF required 450 N, and the 1000 N GSF required 650 N. This verified that the stiffness of the prosthetic foot could be adjusted. Overall, most experiments aligned closely with the predicted linear behaviour from the calculations.

The deviations observed during testing were likely due to slipping or excessive friction between the steel bracket and the machine's steel press. Manufacturing challenges, especially inconsistencies in sanding the components, impacted the blade thickness and consequently their stiffness.

Multiple design options were evaluated based on their complexity and manufacturing cost, with the chosen design incorporating three composite blades and gas springs to adjust the foot stiffness. Detailed design and calculations were completed and verified through finite element analysis for a standard adult weight of 80 kg (800 N) with a specified deflection of 23.6 mm of the forefoot and 13.6 mm of the heel.

The study demonstrates the potential of an affordable variable stiffness prosthetic foot. When comparing the estimated costs of the foot with the values presented in *Gazette 2023* [37], it appears that the prosthetic foot presented in this study could be a more cost-effective option for South Africans. While a detailed cost analysis remains necessary; preliminary calculations estimate that the design may be approximately R4 000 less than the most affordable compensation amount of R21 032 in the *Gazette 2023* [37].

However, a prosthetic foot equipped with gas springs may not be ideal for all activities. As indicated in the experimental results (see the plot in Figure 80), the variability in stiffness primarily influences the force required to initiate displacement. This suggested that the design could be better suited to adjust the foot to different body weights rather than activities. Future refinements and additional testing will be necessary to fully assess its practical applications and limitations.

7.3. Future Work

Improved gas spring attachment and materials

- For this study, steel brackets were used to test the feasibility of the concept. However, steel brackets are heavy, and during testing, the steel-on-steel contact between the brackets and the machine steel press caused excessive friction. This friction would not be present when the prosthetic foot is worn, as it would fit into a foot shell and likely in a shoe. However, the steel could still be abrasive within the footshell.
- A lighter, softer material such as nylon, which is both strong and less damaging to the foot shell, would be more beneficial. Composite materials could also be considered, though careful attention to the orientation of the layer would be required to support the reaction forces from the ball joint fitting in the bracket.
- For this study carbon fibre was used for the blades. Further investigation can be done on the ideal composite material for a foot in terms of weight and cost.
- This study did not characterise the behavior of the gas springs independently. Future work should test the springs on their own to better understand their force-displacement characteristic. These results could then be incorporated in the calculation model to improve the accuracy of the predicted displacement.

Improved manufacturing processes

- To improve the accuracy between manufactured models and calculated models, a better finishing processes for the blades should be developed. A jig could be designed to hold the prefinished parts, aiding in consistent sanding of the blades.
- Alternatively, an additional layer could be applied over the staggered blade layers to reinforce the areas where the layers end, eliminating the need for sanding.

Improved design process

- Designing the foot for a baseline user weight of 60 kg, rather than the 80 kg used in this study, might also be beneficial. The gas springs could be used to adjust the foot for higher weights, offering a wider range of adjustability.
- The Designer could also include an input for the standard foot sizes. This would allow the Designer to calculate the appropriate dimensions for the different foot sizes.
- A classical laminate theory approach can be used to calculate instead of the rule of mixtures.
- The change in angle of the gas springs should be brought into consideration along with the effect this would have on the applied gas spring force.

User studies and customisation of stiffness:

- A user study could be conducted in which participants test the prosthetic foot with different configurations of gas springs in various obstacles and activities. Ideally, participants would vary in weight to assess the practicality of adjusting one foot to suit different weights using gas springs.
- This study would not only determine whether changes in stiffness enhance comfort during specific activities but also verify whether gas springs effectively adjust a standard prosthetic foot to accommodate individual weights.
- Since gas springs can be adjusted to specific forces, a future study could explore their customisation based on individual users. Adjusting the force of the gas spring to complement the user's gait could improve walking dynamics and comfort. A gait analysis could be performed to evaluate the impact of these adjustments.

Integration of compression springs and split toe forefoot design:

- Compression springs could be integrated into the design, combined with threading on the gas spring body and a screw to allow fine-tuning for specific activities. The gas springs would then adjust the foot to the user's weight, while the compression springs would handle activity-specific requirements.
- A potential design enhancement is the inclusion of a split toe forefoot. This would enable better eversion and inversion of the foot, which would potentially improve comfort and performance.
- A study could investigate this feature with the addition of gas springs on each side of the foot. By attaching different-sized gas springs to either side of the foot, the inversion and eversion of the foot could be adjusted. Participant feedback would determine the effectiveness and comfort of this feature in daily activities.

Heel gas spring and repositioning of gas springs:

- Adding a gas spring to the heel could increase the adjustability and suspension of the prosthetic foot during heel strikes, further aiding in the adjustability to certain activities or weights.
- The gas springs should be repositioned to fit within a foot shell and a shoe. Lowering the gas springs on the pyramid fitting side would cause higher force requirements to reduce the displacement. This is, however, a crucial change that needs to be made if the foot is to be market-ready.

Implementing the improvements mentioned, such as specifically adjusted gas spring forces, a split toe design, and a gas spring on the heel of the foot, significantly reinforces the ideas brought forward during this study. By adding more movement through the split toe and adjustability through the heel gas spring and specific gas spring forces, the user would have a foot that the user can adjust according to their comfort. A foot that, in essence, is unique to each person.

References

- [1] A. Calado, F. Soares, and D. Matos, "A Review on Commercially Available Anthropomorphic Myoelectric Prosthetic Hands, Pattern-Recognition-Based Microcontrollers and sEMG Sensors used for Prosthetic Control," in *2019 IEEE International Conference on Autonomous Robot Systems and Competitions (ICARSC)*, Apr. 2019, pp. 1–6. doi: 10.1109/ICARSC.2019.8733629.
- [2] C. L. McDonald, S. Westcott-McCoy, M. R. Weaver, J. Haagsma, and D. Kartin, "Global prevalence of traumatic non-fatal limb amputation," *Prosthet Orthot Int*, p. 0309364620972258, Dec. 2020, doi: 10.1177/0309364620972258.
- [3] P. A. Lazzarini, D. Clark, and P. H. Derhy, "What are the major causes of lower limb amputations in a major Australian teaching hospital? The Queensland Diabetic Foot Innovation Project, 2006 – 2007," *J Foot Ankle Res*, vol. 4, no. Suppl 1, p. O24, May 2011, doi: 10.1186/1757-1146-4-S1-O24.
- [4] P. Manickum, S. S. Ramklass, and T. E. Madiba, "Diabetes and lower extremity amputation - rehabilitation pathways and outcomes at a regional hospital," *S Afr J Surg*, vol. 59, no. 3, pp. 128a–128g, Sep. 2021.
- [5] A. M. Egan and S. F. Dinneen, "What is diabetes?," *Medicine*, vol. 47, no. 1, pp. 1–4, Jan. 2019, doi: 10.1016/j.mpmed.2018.10.002.
- [6] E. Peschke, "Melatonin, endocrine pancreas and diabetes," *Journal of Pineal Research*, vol. 44, no. 1, pp. 26–40, 2008, doi: 10.1111/j.1600-079X.2007.00519.x.
- [7] World Health Organization, *Global report on diabetes*. Geneva: World Health Organization, 2016. Accessed: Jul. 10, 2023. [Online]. Available: <https://apps.who.int/iris/handle/10665/204871>
- [8] N. Grundlingh, T. T. Zewotir, D. J. Roberts, and S. Manda, "Assessment of prevalence and risk factors of diabetes and pre-diabetes in South Africa," *Journal of Health, Population and Nutrition*, vol. 41, no. 1, p. 7, Mar. 2022, doi: 10.1186/s41043-022-00281-2.
- [9] R. G. Frykberg, "Diabetic Foot Ulcers: Pathogenesis and Management," *afp*, vol. 66, no. 9, pp. 1655–1663, Nov. 2002.
- [10] s.d., "Frequently Asked Questions: Diabetic Foot Ulcers," Michigan Medicine. Accessed: Jan. 10, 2023. [Online]. Available: <https://www.uofmhealth.org/conditions-treatments/podiatry-foot-care/frequently-asked-questions-diabetic-foot-ulcers>
- [11] D. G. Armstrong, N. D.p.m, and L. A. Lavery, "Diabetic Foot Ulcers: Prevention, Diagnosis and Classification," *afp*, vol. 57, no. 6, pp. 1325–1332, Mar. 1998.
- [12] P. Manickum, S. S. Ramklass, and T. E. Madiba, "A five-year audit of lower limb amputations below the knee and rehabilitation outcomes: The Durban experience," *Journal of Endocrinology, Metabolism and Diabetes of South Africa*, vol. 24, no. 2, Art. no. 2, 2019.
- [13] S. M. Abbas, K. Resan, A. Muhammad, and M. Al-Waily, "Mechanical and Fatigue Behaviors of Prosthetic for Partial Foot Amputation with Various Composite Materials Types Effect," vol. 09, pp. 383–394, Sep. 2018.
- [14] s.a., "Living with a partial foot amputation." Accessed: Mar. 09, 2023. [Online]. Available: <https://www.ottobock.com/en-us/resources/living-with-an-amputation/partial-foot-amputation>
- [15] S. M. Abbas, A. M. Takhakh, M. A. Al-Shammari, and M. Al-Waily, "MANUFACTURING AND ANALYSIS OF ANKLE DISARTICULATION PROSTHETIC SOCKET (SYMES)," *International Journal of Mechanical Engineering and Technology (IJMET)*, vol. 9, no. 7, pp. 560–569, Jul. 2018.
- [16] R. Braaksma, P. U. Dijkstra, and J. H. B. Geertzen, "Syme Amputation: A Systematic Review," *Foot Ankle Int.*, vol. 39, no. 3, pp. 284–291, Mar. 2018, doi: 10.1177/1071100717745313.
- [17] R. I. Harris, "Syme's amputation; the technical details essential for success," *J Bone Joint Surg Br*, vol. 38-B, no. 3, pp. 614–632, Aug. 1956, doi: 10.1302/0301-620X.38B3.614.

- [18] D. H. K. Chow, A. D. Holmes, C. K. L. Lee, and S. W. Sin, "The Effect of Prosthesis Alignment on the Symmetry of Gait in Subjects with Unilateral Transtibial Amputation," *Prosthet Orthot Int*, vol. 30, no. 2, pp. 114–128, Aug. 2006, doi: 10.1080/03093640600568617.
- [19] s.a., "Lower Extremity Amputation Types & Levels - MossRehab." Accessed: Mar. 09, 2023. [Online]. Available: <https://www.mossrehab.com/amputation-and-prosthetics/lower-extremity-amputation>
- [20] R. Versluys *et al.*, "From conventional prosthetic feet to bionic feet: A review study," in *2008 2nd IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics*, Scottsdale, AZ, USA: IEEE, Oct. 2008, pp. 49–54. doi: 10.1109/BIOROB.2008.4762839.
- [21] M. S. Pinzur and J. H. Bowker, "Knee Disarticulation," *Clinical Orthopaedics and Related Research®*, vol. 361, p. 23, Apr. 1999.
- [22] M. D. Muller, "Transfemoral Amputation: Prosthetic Management," 2016.
- [23] P. H. Sugarbaker and P. B. Chretien, "A surgical technique for hip disarticulation," *Surgery*, vol. 90, no. 3, pp. 546–553, Sep. 1981.
- [24] G. Stark, "Overview of Hip Disarticulation Prostheses," *JPO: Journal of Prosthetics and Orthotics*, vol. 13, no. 2, p. 50, Jun. 2001.
- [25] S. M. Zaffer, R. L. Braddom, A. Conti, J. Goff, and D. Bokma, "TOTAL HIP DISARTICULATION PROSTHESIS WITH SUCTION SOCKET: Report of Two Cases: 1," *American Journal of Physical Medicine & Rehabilitation*, vol. 78, no. 2, p. 160, Apr. 1999.
- [26] J. J. Stokosa, "Limb Prosthesis Preparation - Special Subjects," MSD Manual Consumer Version. Accessed: Oct. 11, 2023. [Online]. Available: <https://www.msdmanuals.com/home/special-subjects/limb-prosthetics/limb-prosthesis-preparation>
- [27] s.a., "Prosthetic Feet," Amputee Coalition. Accessed: Jan. 10, 2023. [Online]. Available: <https://www.amputee-coalition.org/resources/prosthetic-feet/>
- [28] P. M. Stevens, J. Rheinstein, and S. R. Wurdeman, "Prosthetic Foot Selection for Individuals with Lower-Limb Amputation: A Clinical Practice Guideline," *J Prosthet Orthot*, vol. 30, no. 4, pp. 175–180, Oct. 2018, doi: 10.1097/JPO.000000000000181.
- [29] N. E. Doane and L. E. Holt, "A comparison of the SACH and single axis foot in the gait of unilateral below-knee amputees," *Prosthet Orthot Int*, vol. 7, no. 1, pp. 33–36, Apr. 1983, doi: 10.3109/03093648309146713.
- [30] M. I. S. Fadul, F. G. A. Elsayed, M. E. A. S. Aldeen, H. G. A. Abdalla, S. E. M. Hamad, and A. M. A. Babeker, "Design of a Multi-axis Prosthetic foot for the amputees in the developing world based on human's foot bones structure for Sudanese environment," in *2021 World Engineering Education Forum/Global Engineering Deans Council (WEEF/GEDC)*, Nov. 2021, pp. 190–195. doi: 10.1109/WEEF/GEDC53299.2021.9657273.
- [31] S. L. Shanfield, "Below-knee amputee gait with dynamic elastic response prosthetic foot: A pilot study," *JRRD*, vol. 27, no. 4, p. 369, 1990, doi: 10.1682/JRRD.1990.10.0369.
- [32] V. Agrawal, R. Gailey, I. A. Gaunaud, C. O'Toole, and A. A. Finnieston, "Comparison between microprocessor-controlled ankle/foot and conventional prosthetic feet during stair negotiation in people with unilateral transtibial amputation," *J Rehabil Res Dev*, vol. 50, no. 7, pp. 941–950, 2013, doi: 10.1682/JRRD.2012.05.0093.
- [33] M. Ernst, B. Altenburg, and T. Schmalz, "Characterizing adaptations of prosthetic feet in the frontal plane," *Prosthet Orthot Int*, vol. 44, no. 4, pp. 225–233, Aug. 2020, doi: 10.1177/0309364620917838.
- [34] s.a., "Steeper Group - Steeper Group - DynAdapt Foot," Steeper Group. Accessed: Mar. 22, 2023. [Online]. Available: <https://www.steepergroup.com/prosthetics/lower-limb-prosthetics/feet/dynadapt/>
- [35] F. L. Golbranson, "Volume changes occurring in postoperative below-knee residual limbs," 1988.
- [36] A. T. Tantua, J. H. B. Geertzen, J. J. A. M. van den Dungen, J.-K. C. Breek, and P. U. Dijkstra, "Reduction of residual limb volume in people with transtibial amputation," *Journal of Rehabilitation Research and Development*, vol. 51, no. 7, pp. 1119–1126, 2014, doi: 10.1682/JRRD.2013.11.0243.

- [37] “Prosthetics and Orthotics Gazette 2023.” Accessed: Mar. 27, 2024. [Online]. Available: <https://archive.gazettes.africa/archive/za/2023/za-government-gazette-dated-2023-03-27-no-48299.pdf>
- [38] s.a., “The Rapid Assessment of the Implementation and Utilisation of the Special Covid-19 Social Relief of Distress Grant,” Department of Social Development. Accessed: Oct. 10, 2023. [Online]. Available: <https://www.dsd.gov.za/index.php/latest-news/21-latest-news/410-the-rapid-assessment-of-the-implementation-and-utilisation-of-the-special-covid-19-social-relief-of-distress-grant>
- [39] s.a., “Disability grant | South African Government.” Accessed: Oct. 10, 2023. [Online]. Available: <https://www.gov.za/services/social-benefits/disability-grant>
- [40] South African Government, “National Minimum Wage for 2024.” Accessed: Mar. 27, 2024. [Online]. Available: <https://www.gov.za/news/media-statements/government-activities/minister-thulas-nxesi-announces-national-minimum-wage>
- [41] T. J. Sekhampu and F. Niyimbaniira, “Analysis Of The Factors Influencing Household Expenditure In A South African Township,” *International Business & Economics Research Journal (IBER)*, vol. 12, no. 3, Art. no. 3, Feb. 2013, doi: 10.19030/iber.v12i3.7671.
- [42] s.a., “Public healthcare: How much per person? | Statistics South Africa,” Statistics South Africa. Accessed: Jul. 10, 2023. [Online]. Available: <https://www.statssa.gov.za/?p=10548>
- [43] M. Muller, “Trauma recovery extender benefit,” no. 1125.
- [44] “Home | Road Accident Fund.” Accessed: Mar. 27, 2024. [Online]. Available: <https://www.raf.co.za/Pages/Default.aspx>
- [45] H. Burger and Č. Marinček, “Return to work after lower limb amputation,” *Disability and Rehabilitation*, vol. 29, no. 17, pp. 1323–1329, Jan. 2007, doi: 10.1080/09638280701320797.
- [46] “Fact Sheet -High Level Education Attainment in South Africa- April 2024.pdf.” Accessed: Feb. 07, 2025. [Online]. Available: <https://www.dhet.gov.za/Planning%20Monitoring%20and%20Evaluation%20Coordination/Fact%20Sheet%20-High%20Level%20Education%20Attainment%20in%20South%20Africa-%20April%20%202024.pdf>
- [47] B. Daniele, “Chapter 69 - Evolution of prosthetic feet and design based on gait analysis data,” in *Clinical Engineering Handbook (Second Edition)*, E. Iadanza, Ed., Academic Press, 2020, pp. 458–468. doi: 10.1016/B978-0-12-813467-2.00070-5.
- [48] P. Hernigou, “Ambroise Paré IV: The early history of artificial limbs (from robotic to prostheses),” *Int Orthop*, vol. 37, no. 6, pp. 1195–1197, Jun. 2013, doi: 10.1007/s00264-013-1884-7.
- [49] P. Hernigou, “Ambroise Paré IV: The early history of artificial limbs (from robotic to prostheses),” *Int Orthop*, vol. 37, no. 6, pp. 1195–1197, Jun. 2013, doi: 10.1007/s00264-013-1884-7.
- [50] Y. Alizadeh, *Design and Structural Analysis of Composite Prosthetic Running Blades for Athletes: A case of dynamic explicit analysis using Abaqus CAE*. 2020. doi: 10.13140/RG.2.2.20693.04327.
- [51] s.a., “Van Phillips | Lemelson.” Accessed: Feb. 10, 2023. [Online]. Available: <https://lemelson.mit.edu/resources/van-phillips>
- [52] S. Kapil, “Advanced Prosthetics: Changing the world of sports.” Accessed: Sep. 18, 2023. [Online]. Available: <https://nopr.niscpr.res.in/bitstream/123456789/44974/1/SR%2055%289%29%2030-32.pdf>
- [53] R. J. Abboud, “(i) Relevant foot biomechanics,” *Current Orthopaedics*, vol. 16, no. 3, pp. 165–179, Jun. 2002, doi: 10.1054/cuor.2002.0268.
- [54] A. Das, “A Review on the Anatomy and Biomechanics of the Foot-Ankle Complex,” *Asian Journal For Convergence In Technology (AJCT) ISSN -2350-1146*, Apr. 2018, Accessed: Jan. 19, 2023. [Online]. Available: <https://asianssr.org/index.php/ajct/article/view/418>
- [55] F. Bojsen-Møller, “Anatomy of the Forefoot, Normal and Pathologic,” *Clinical Orthopaedics and Related Research®*, no. 142, p. 10, Aug. 1979.
- [56] J. J. Fraser, M. A. Feger, and J. Hertel, “MIDFOOT AND FOREFOOT INVOLVEMENT IN LATERAL ANKLE SPRAINS AND CHRONIC ANKLE INSTABILITY. PART 1:

- ANATOMY AND BIOMECHANICS,” *Int J Sports Phys Ther*, vol. 11, no. 6, pp. 992–1005, Dec. 2016.
- [57] H. Kurup and N. Vasukutty, “Midfoot arthritis- current concepts review,” *J Clin Orthop Trauma*, vol. 11, no. 3, pp. 399–405, 2020, doi: 10.1016/j.jcot.2020.03.002.
 - [58] H. B. Menz, “Clinical hindfoot measurement: a critical review of the literature,” *The Foot*, vol. 5, no. 2, pp. 57–64, Jun. 1995, doi: 10.1016/0958-2592(95)90012-8.
 - [59] C. L. Brockett and G. J. Chapman, “Biomechanics of the ankle,” *Orthop Trauma*, vol. 30, no. 3, pp. 232–238, Jun. 2016, doi: 10.1016/j.morth.2016.04.015.
 - [60] P. Wolf *et al.*, “Functional units of the human foot,” *Gait & Posture*, vol. 28, no. 3, pp. 434–441, Oct. 2008, doi: 10.1016/j.gaitpost.2008.02.004.
 - [61] S. J. Abbass and G. Abdulrahman, “Kinematic analysis of human gait cycle,” 2014.
 - [62] A. Kharb, V. Saini, Y. Jain, S. Dhiman, M. Tech, and Scholar, “A review of gait cycle and its parameters,” *IJCEM Int J Comput Eng Manag*, vol. 13, Jan. 2011.
 - [63] D. J. Mayich, A. Novak, D. Vena, T. R. Daniels, and J. W. Brodsky, “Gait Analysis in Orthopedic Foot and Ankle Surgery—Topical Review, Part 1: Principles and Uses of Gait Analysis,” *Foot Ankle Int.*, vol. 35, no. 1, pp. 80–90, Jan. 2014, doi: 10.1177/1071100713508394.
 - [64] W. Tao, T. Liu, R. Zheng, and H. Feng, “Gait Analysis Using Wearable Sensors,” *Sensors*, vol. 12, no. 2, Art. no. 2, Feb. 2012, doi: 10.3390/s120202255.
 - [65] H. A. C. Jacob, “Forces acting in the forefoot during normal gait – an estimate,” *Clinical Biomechanics*, vol. 16, no. 9, pp. 783–792, Nov. 2001, doi: 10.1016/S0268-0033(01)00070-5.
 - [66] D. C. Wing and D. A. Hittenberger, “Energy-storing prosthetic feet,” *Archives of Physical Medicine and Rehabilitation*, vol. 70, no. 4, pp. 330–335, Apr. 1989, doi: 10.5555/uri:pii:000399938990155X.
 - [67] M. Omasta, D. Paloušek, T. Návrát, and J. Rosický, “Finite element analysis for the evaluation of the structural behaviour, of a prosthesis for trans-tibial amputees,” *Medical Engineering & Physics*, vol. 34, no. 1, pp. 38–45, Jan. 2012, doi: 10.1016/j.medengphy.2011.06.014.
 - [68] Z. Tao, H.-J. Ahn, C. Lian, K.-H. Lee, and C.-H. Lee, “Design and optimization of prosthetic foot by using polylactic acid 3D printing,” *J Mech Sci Technol*, vol. 31, no. 5, pp. 2393–2398, May 2017, doi: 10.1007/s12206-017-0436-2.
 - [69] J. K. Oleiwi and A. N. Hadi, “Properties of Materials and Models of Prosthetic Feet: A Review,” *IOP Conf. Ser.: Mater. Sci. Eng.*, vol. 1094, no. 1, p. 012151, Feb. 2021, doi: 10.1088/1757-899X/1094/1/012151.
 - [70] K. M. Walke, “Mechanical Properties of Materials Used For Prosthetic Foot: A Review,” *IOSR JMCE*, vol. 17, no. 01, pp. 61–65, Mar. 2017, doi: 10.9790/1684-17010026165.
 - [71] S. Suleman, S. M. Khan, N. Gull, W. Aleem, M. Shafiq, and T. Jamil, “A Comprehensive Short Review on Polyurethane Foam,” vol. 12, no. 1, 2014.
 - [72] “About Feet | Ottobock US Shop.” Accessed: Mar. 17, 2023. [Online]. Available: <https://shop.ottobock.us/About-Feet>
 - [73] H. Houdijk, D. Wezenberg, L. Hak, and A. G. Cutti, “Energy storing and return prosthetic feet improve step length symmetry while preserving margins of stability in persons with transtibial amputation,” *Journal of NeuroEngineering and Rehabilitation*, vol. 15, no. 1, p. 76, Sep. 2018, doi: 10.1186/s12984-018-0404-9.
 - [74] E. Klodd, A. Hansen, S. Fatone, and M. Edwards, “Effects of prosthetic foot forefoot flexibility on gait of unilateral transtibial prosthesis users,” *JRRD*, vol. 47, no. 9, p. 899, 2010, doi: 10.1682/JRRD.2009.10.0166.
 - [75] S. R. Koehler-McNicholas *et al.*, “Mechanical and dynamic characterization of prosthetic feet for high activity users during weighted and unweighted walking,” *PLOS ONE*, vol. 13, no. 9, p. e0202884, Sep. 2018, doi: 10.1371/journal.pone.0202884.
 - [76] E. M. Glanzer and P. G. Adamczyk, “Design and Validation of a Semi-Active Variable Stiffness Foot Prosthesis,” *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 26, no. 12, pp. 2351–2359, Dec. 2018, doi: 10.1109/TNSRE.2018.2877962.
 - [77] D. Dong, W. Ge, J. Wang, Y. Sun, and D. Zhao, “Optimal Design and Analysis of a Powered Ankle-Foot Prosthesis with Adjustable Actuation Stiffness: Proceedings of International

- Conference on Mechatronics and Intelligent Robotics (ICMIR2018),” 2019, pp. 578–587. doi: 10.1007/978-3-030-00214-5_73.
- [78] C. Lecomte, A. L. Ármannsdóttir, F. Starker, H. Tryggvason, K. Briem, and S. Brynjolfsson, “Variable stiffness foot design and validation,” *Journal of Biomechanics*, vol. 122, p. 110440, Jun. 2021, doi: 10.1016/j.jbiomech.2021.110440.
 - [79] “15: The Energy Expenditure of Amputee Gait | O&P Virtual Library.” Accessed: May 21, 2025. [Online]. Available: <http://www.oandplibrary.org/alp/chap15-01.asp>
 - [80] A. Mutlu, M. D. Kharooty, and Y. Yakut, “The effect of segmental weight of prosthesis on hemodynamic responses and energy expenditure of lower extremity amputees,” *J Phys Ther Sci*, vol. 29, no. 4, pp. 629–634, Apr. 2017, doi: 10.1589/jpts.29.629.
 - [81] Z. Liao *et al.*, “Review of current best-practices in machinability evaluation and understanding for improving machining performance,” *CIRP Journal of Manufacturing Science and Technology*, vol. 50, pp. 151–184, Jun. 2024, doi: 10.1016/j.cirpj.2024.02.008.
 - [82] R. C. Hibbeler and R. C. Hibbeler, *Mechanics of materials*, 7. ed. Upper Saddle River, NJ: Pearson Prentice Hall [u.a.], 2008.
 - [83] J. Yang, Y. Song, K. Dong, and E.-H. Han, “Research progress on the corrosion behavior of titanium alloys,” *Corrosion Reviews*, vol. 41, no. 1, pp. 5–20, Feb. 2023, doi: 10.1515/corrrev-2022-0031.
 - [84] M. Ishii, T. Oda, and M. Kaneko, “Titanium and Its Alloys As Key Materials for Corrosion Protection Engineering,” no. 87, 2003.
 - [85] T. O. Olugbade, “Corrosion Resistance, Evaluation Methods, and Surface Treatments of Stainless Steels,” in *Stainless Steels*, IntechOpen, 2022. doi: 10.5772/intechopen.101430.
 - [86] M. P. Ryan, D. E. Williams, R. J. Chater, B. M. Hutton, and D. S. McPhail, “Why stainless steel corrodes,” *Nature*, vol. 415, no. 6873, pp. 770–774, Feb. 2002, doi: 10.1038/415770a.
 - [87] W. S. Tait, “28 - Corrosion Prevention and Control of Chemical Processing Equipment,” in *Handbook of Environmental Degradation of Materials (Second Edition)*, M. Kutz, Ed., Oxford: William Andrew Publishing, 2012, pp. 863–886. doi: 10.1016/B978-1-4377-3455-3.00028-6.
 - [88] Q. Li, J. O. Jensen, and N. J. Bjerrum, “CHEMISTRY, ELECTROCHEMISTRY, AND ELECTROCHEMICAL APPLICATIONS | Aluminum,” in *Encyclopedia of Electrochemical Power Sources*, J. Garche, Ed., Amsterdam: Elsevier, 2009, pp. 695–708. doi: 10.1016/B978-044452745-5.00951-5.
 - [89] “MACHINABILITY INDEX OF AIRFRAME MATERIALS.” Accessed: Oct. 19, 2023. [Online]. Available: <https://apps.dtic.mil/sti/citations/AD0663696>
 - [90] S. Genculu, “Factors Affecting Machinability of Metals”.
 - [91] P.-J. Arrazola, A. Garay, L.-M. Iriarte, M. Armendia, S. Marya, and F. Le Maître, “Machinability of titanium alloys (Ti6Al4V and Ti555.3),” *Journal of Materials Processing Technology*, vol. 209, no. 5, pp. 2223–2230, Mar. 2009, doi: 10.1016/j.jmatprotec.2008.06.020.
 - [92] S. Gupta, D. Singh, A. Yadav, S. Jain, and B. Pratap, “A comparative study of 5083 aluminium alloy and 316L stainless steel for shipbuilding material,” *Materials Today: Proceedings*, vol. 28, pp. 2358–2363, Jan. 2020, doi: 10.1016/j.matpr.2020.04.641.
 - [93] F. H. Froes, H. Friedrich, J. Kiese, and D. Bergoint, “Titanium in the family automobile: The cost challenge,” *JOM*, vol. 56, no. 2, pp. 40–44, Feb. 2004, doi: 10.1007/s11837-004-0144-0.
 - [94] Y. Li and J. Barbic, “Stable Orthotropic Materials,” *Eurographics/ ACM SIGGRAPH Symposium on Computer Animation*, p. 6 pages, 2014, doi: 10.2312/SCA.20141121.
 - [95] A. K. Kaw, *Mechanics of composite materials*, 2. ed. in Mechanical engineering series, no. 29. Boca Raton, FL: CRC Taylor & Francis, 2006.
 - [96] “ASTM D3039 Tensile Testing of Composite Laminates.” Accessed: Aug. 14, 2023. [Online]. Available: <https://www.instron.com/en/testing-solutions/astm-standards/astm-d3039>
 - [97] “ASTM D695 Compression Testing Rigid Plastics.” Accessed: Aug. 14, 2023. [Online]. Available: <https://www.instron.com/en/testing-solutions/astm-standards/astm-d695-plastics>
 - [98] Y. K. Kumar and D. D. S. Lohchab, “Influence of Aviation Fuel on Mechanical properties of Glass Fiber-Reinforced Plastic Composite,” vol. 3, no. 4, 2016.
 - [99] A. K. Tanwer, “Mechanical Properties Testing of Uni-directional and Bi-directional Glass Fibre Reinforced Epoxy Based Composites,” 2014.

- [100] M. Sharma, S. Gao, E. Mäder, H. Sharma, L. Y. Wei, and J. Bijwe, "Carbon fiber surfaces and composite interphases," *Composites Science and Technology*, vol. 102, pp. 35–50, Oct. 2014, doi: 10.1016/j.compscitech.2014.07.005.
- [101] D. A. Berry, "Composite Materials for Orthotics and Prosthetics," *Orthotics and Prosthetics*, vol. 40, pp. 35-43., 1987.
- [102] M. Holmes, "Lowering the cost of carbon fiber," *Reinforced Plastics*, vol. 61, no. 5, pp. 279–283, Sep. 2017, doi: 10.1016/j.repl.2017.02.001.
- [103] "Carbon Fiber sheets," Oxyblack. Accessed: Oct. 19, 2023. [Online]. Available: <https://www.oxyblack.com/index.php/en/products-services/sheets-tubes>
- [104] Engr. Reashad Bin Kabir and Engr. Nasrin Ferdous, "Kevlar-The Super Tough Fiber," *TEXTILE*, vol. 1, no. 6, pp. 78–83, Jan. 2013, doi: 10.5923/j.textile.20120106.04.
- [105] S. S. Rao and Y. F. Fah, *Mechanical vibrations*, 5. ed. in SI units. in Always learning. Singapore: Prentice Hall/Pearson, 2011.
- [106] "NX software | Siemens Software," Siemens Digital Industries Software. Accessed: Feb. 28, 2025. [Online]. Available: <https://plm.sw.siemens.com/en-US/nx/>
- [107] "Triton Heavy Duty | Your will. Your way.," Ottobock. Accessed: Feb. 28, 2025. [Online]. Available: <https://www.ottobock.com/en-za/product/1C64>
- [108] A. T. Turner *et al.*, "Prosthetic forefoot and heel stiffness across consecutive foot stiffness categories and sizes," *PLoS One*, vol. 17, no. 5, p. e0268136, May 2022, doi: 10.1371/journal.pone.0268136.
- [109] "Steeper Group - Steeper Group - RUSH HiPro®," Steeper Group. Accessed: Feb. 28, 2025. [Online]. Available: <https://www.steepergroup.com/prosthetics/lower-limb-prosthetics/feet/rush-hipro/>
- [110] P. Boisse, J. Colmars, N. Hamila, N. Naouar, and Q. Steer, "Bending and wrinkling of composite fiber preforms and prepregs. A review and new developments in the draping simulations," *Composites Part B: Engineering*, vol. 141, pp. 234–249, May 2018, doi: 10.1016/j.compositesb.2017.12.061.
- [111] J. Huang, P. Boisse, N. Hamila, and Y. Zhu, "Simulation of Wrinkling during Bending of Composite Reinforcement Laminates," *Materials (Basel)*, vol. 13, no. 10, p. 2374, May 2020, doi: 10.3390/ma13102374.
- [112] H. Talla, A. Hassan, and J. Oleiwi, "An Investigation into the Effect of Adding Carbon and Glass Fibres to UHMWPE Fibres on the Mechanical Characteristics of a Sports Prosthetic Foot," *Revue des composites et des matériaux avancés*, vol. 32, pp. 69–76, Apr. 2022, doi: 10.18280/rcma.320203.
- [113] B. Samir, M. Belkacem, and G. Brahim, "Numerical Modeling of the Effects of Fiber Packing and Reinforcement Volume Ratio on the Transverse Elasticity Modulus of a Unidirectional Composite Material Glass / Epoxy.," *Journal of Composite & Advanced Materials / Revue des Composites et des Matériaux Avancés*, vol. 30, no. 5/6, pp. 203–210, Dec. 2020, doi: 10.18280/rcma.305-602.
- [114] Y. Chen, Z. Wang, Y. Lin, H. Wang, and L. Hua, "Theoretical Modeling and Experimental Verification of the Bending Deformation of Fiber Metal Laminates," *Materials (Basel)*, vol. 16, no. 9, p. 3486, Apr. 2023, doi: 10.3390/ma16093486.
- [115] A. J. Rapoff, D. M. Heisey, and R. Vanderby, "A probabilistic rule of mixtures for elastic moduli," *Journal of Biomechanics*, vol. 32, no. 2, pp. 189–193, Feb. 1999, doi: 10.1016/S0021-9290(98)00155-9.
- [116] N. T. Mascia and R. A. Simoni, "Analysis of failure criteria applied to wood," *Engineering Failure Analysis*, vol. 35, pp. 703–712, Dec. 2013, doi: 10.1016/j.engfailanal.2013.07.001.
- [117] A. P. C. Duarte, A. Díaz Sáez, and N. Silvestre, "Comparative study between XFEM and Hashin damage criterion applied to failure of composites," *Thin-Walled Structures*, vol. 115, pp. 277–288, Jun. 2017, doi: 10.1016/j.tws.2017.02.020.
- [118] A. A. Groenwold and R. T. Haftka, "Optimization with non-homogeneous failure criteria like Tsai–Wu for composite laminates," *Struct Multidisc Optim*, vol. 32, no. 3, pp. 183–190, Sep. 2006, doi: 10.1007/s00158-006-0020-3.

Appendix A : Calculation Verifications

A.1. Verification of Calculation Method.

To verify the calculation method explained in Chapter 4, the vertical displacement results from 3 different input variable sets were compared to those from FEA in NX. Two NX models were used to verify the calculations.

Finite element analysis (FEA) was performed on models matching the dimensions used in Table A - 1 using Siemens NX with Nastran as the finite element solver. The implementation of the values on the models can be seen in Figure A - 1. The purpose of the FEA was to verify the displacements obtained from the calculation method.

The material used in the FEA was Aluminium 2014 , as defined by the NX material library. The Young's modulus was, however, adjusted according to the values in Table A - 1, since it was the only material property influencing the displacement in the calculations.

A Ctetra(10) mesh (10-node tetrahedral elements) was used. The automatically suggested element size was 10.4mm refined to 5mm to improve the accuracy without excessive computational cost.

Fixed supports were applied as described in the calculation method:

- The top edges of the forefoot blades, as seen in Figure A - 2. The two forefoot blades were bonded together by a glue constraint.
- The face of the heel blade that would be glued to the forefoot blade, as seen in Figure A - 3.

The loads were applied to the opposite free edge of the forefoot and heel blades, consistent with the calculation method, as seen in Figure A - 2 and Figure A - 3. The loads were resolved into their respective horizontal (F_{x1}) and vertical (F_{y1}) componets, as explained with equations 18 and 19 respectively. NX supports parametric inputs, allowing the use of trigonometric expressions like $\sin()$ and $\cos()$ directly in the load definitions, which simplified variation and control of applied loads.

The FEA results of the value set run for both the forefoot and heel can be seen in Figure A - 4. The results of all the value sets are recorded in Table A - 2.

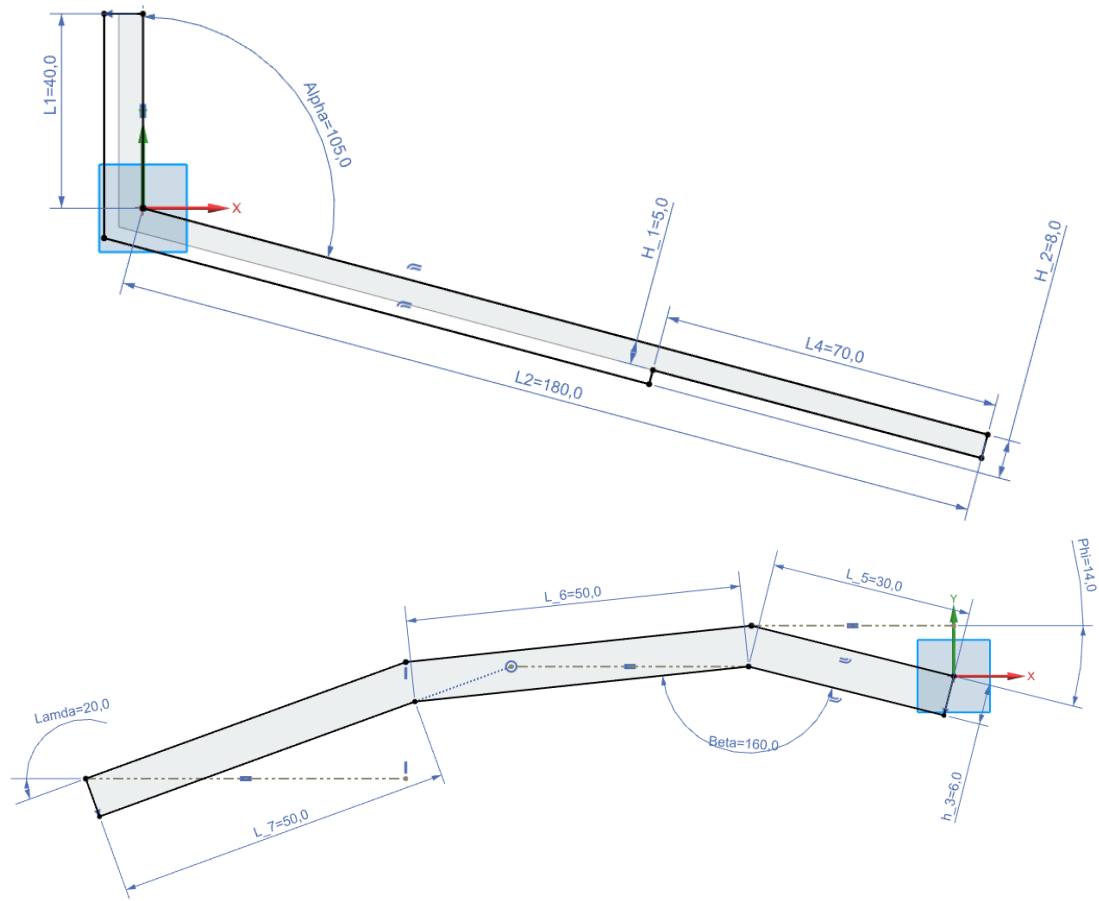


Figure A - 1: Implementation of dimensions in CAD model

The dimensions from the value sets from Table A - 1 are implemented to the CAD model. This figure illustrates how value set 1 was implemented on the forefoot (above) and heel (below).

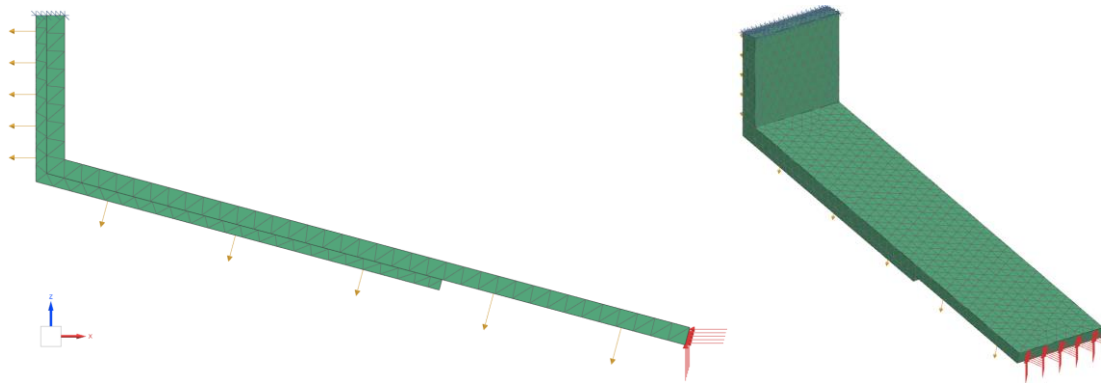


Figure A - 2: FEA setup of forefoot blade for calculation method verification

The FEA of a model with value set 1 from Table A - 1. The blue hatching represents the fixed support on the top of the forefoot blade segment. The yellow arrows on the body represent the glue bonding of the two parts. The red arrows at the free edge of the blade, represent the force components F_{x1} and F_{y1} .

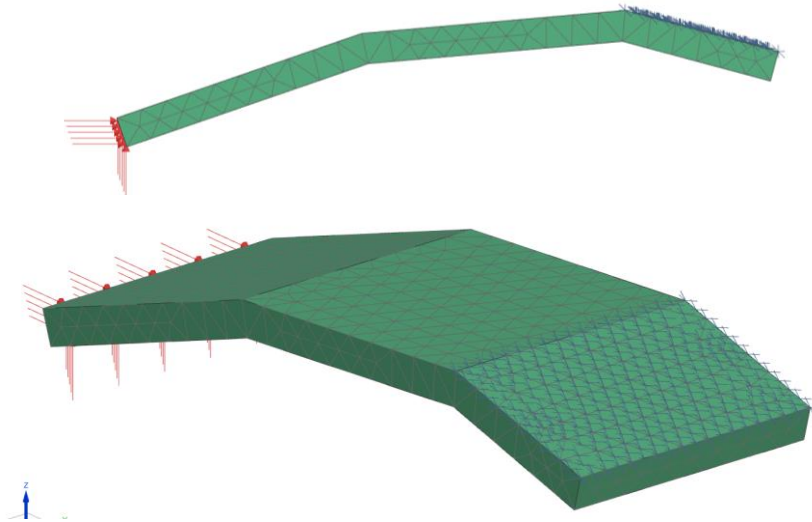


Figure A - 3: FEA setup of heelblade for calculation method verification

The FEA of a model with value set 1 from Table A - 1. The blue hatching represents the fixed support on the top of the forefoot blade segment. The yellow arrows on the body represent the glue bonding of the two parts. The red arrows at the free edge of the blade represent the force components F_{x1} and F_{y1} .

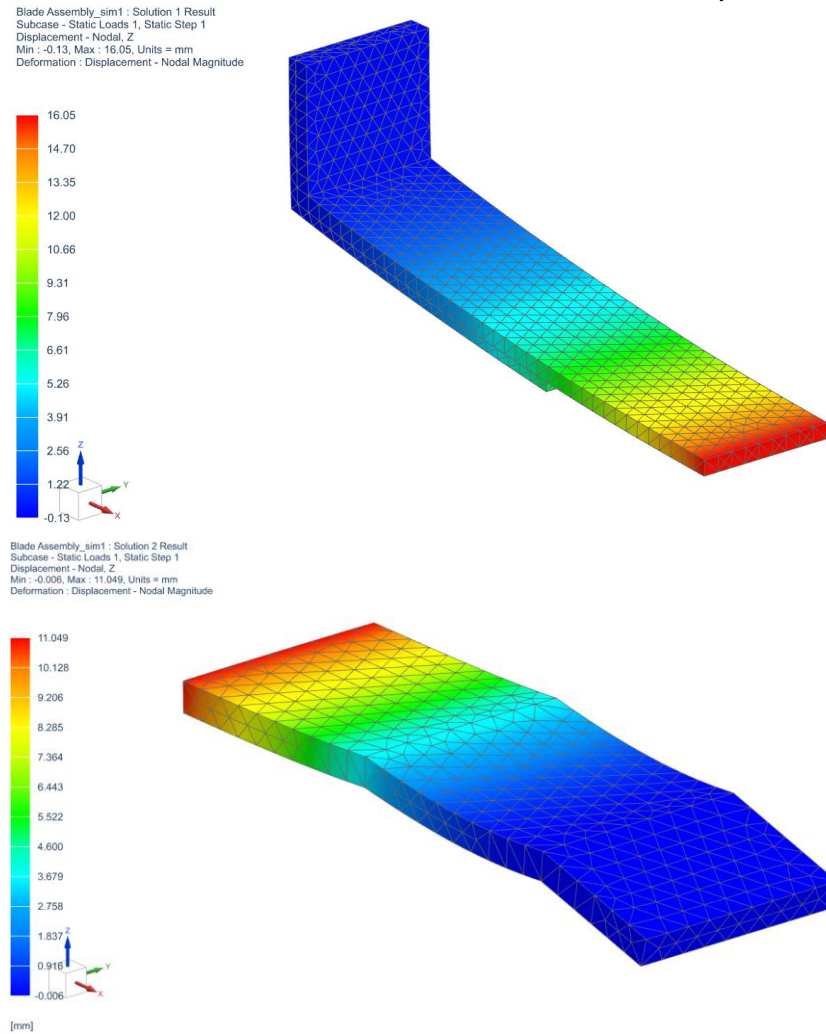


Figure A - 4: FEA results of forefoot and heel using value set 1

The figure presents the results from the FEA for value set 1. The results of all the FEA for the value sets presented in Table A - 1 can be found in Table A - 2.

Table A - 1 summarises the three sets of input variables. The calculations and finite element analysis used these variables to return a displacement result for both the forefoot and the heel. The results for the forefoot and heel calculations and FEA are found in Table A - 2.

Table A - 1: Input variables to test calculation method

Variable symbol	Value			Unit
	Set 1	Set 2	Set 3	
F	800	500	1200	[N]
b	45	50	30	[mm]
α	105	110	100	[°]
ζ	20	20	20	[°]
λ	20	20	18	[°]
β	160	160	165	[°]
L ₁	40	40	50	[mm]
L ₂	180	180	160	[mm]
L ₃	110	90	80	[mm]
L ₄	70	90	80	[mm]
L ₅	30	20	20	[mm]
L ₆	50	50	45	[mm]
L ₇	50	30	20	[mm]
h ₁	5	6	4	[mm]
h ₂	8	8	10	[mm]
h ₃	6	3	5	[mm]
E ₁	75	80	60	[GPa]
E ₂	75	80	80	[GPa]
E ₃	25	50	30	[GPa]

Table A - 2: Results of calculation and FEA for method validation

		Set 1	Set 2	Set 3
Forefoot	Calculation (mm)	16.22	7.73	33.03
	FEA (mm)	16.05	7.60	33.80
	Difference (mm)	0.17	0.13	0.77
	Difference (%)	1.07	1.74	2.27
Heel	Calculation (mm)	10.94	13.16	10.29
	FEA (mm)	11.05	12.59	9.66
	Difference (mm)	0.11	0.57	0.63
	Difference (%)	0.97	4.50	6.53

As can be seen in Table A - 2, the results between the calculations and the FEA are highly comparable. The results for the forefoot calculations were found to deviate roughly 2% from the FEA results. The highest percentage deviation is 2.27% at a 0.77 mm difference between the displacement values.

The results between the calculations and FEA of the heel are also comparable; the percentage error, however, was found to be higher when compared to the forefoot results. The highest percentage deviation, 6.53%, was from the heel comparison of Set 3. The difference in millimetres was only 0.63 mm, this was considered acceptable.

A.2. Verification of Prosthetic Foot Designer

Forefoot

Figure A - 5 shows the prosthetic foot as base for the calculations described in Chapter 4. Figure A - 6 illustrates how the different sections were affected by the applied force.

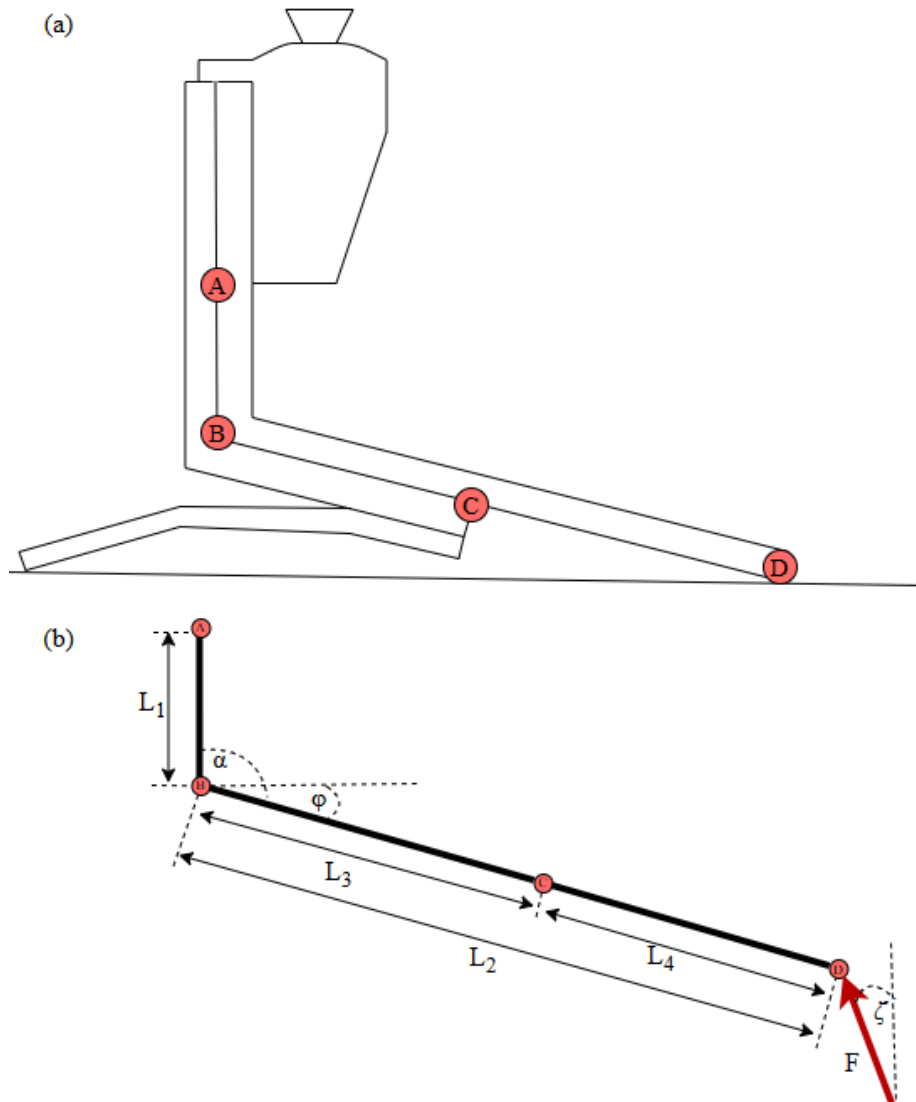


Figure A - 5: The forefoot segments and points

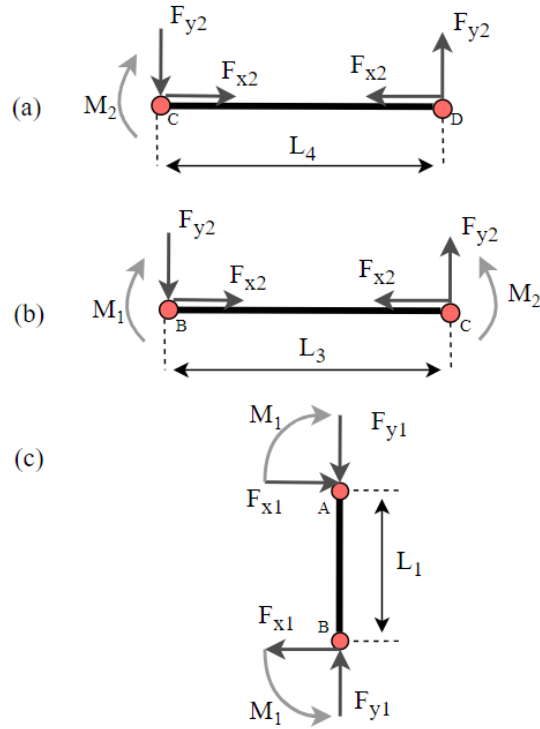


Figure A - 6: The force distribution of prosthetic forefoot

Constants and Assumptions

The input variables for the calculation, delivered by the Designer, are captured in Table A - 3. The symbols of the variables correspond to those illustrated in Figure A - 5 and Figure A - 6.

E_2 is the Young's modulus for the second blade that fits beneath the primary blade, which ends at point C. Due to this, an equivalent of Young's modulus was calculated to ease the calculations. Heights, h_1 and h_2 , are the minimum blade thickness and maximum of the two blades together respectively, and b is the thickness of the two blades.

Table A - 3: Input variables

F	800	[N]
b	45	[mm]
α	104	[°]
ζ	20	[°]
L_1	44	[mm]
L_2	175	[mm]
L_3	105	[mm]
L_4	70	[mm]
h_1	4.3	[mm]
h_2	7.4	[mm]
E_1	63.7	[MPa]
E_2	63.7	[MPa]

Initial Calculations

Some calculations were done before the superposition calculations could be completed.

The angle, φ , was used in the calculations to simplify them. This was determined as follows:

$$\varphi = \alpha - 90$$

$$\varphi = 104 - 90$$

$$\varphi = 14 [^\circ]$$

The force perpendicular to the segment BC is given by the following equation. The force was used to calculate the amount of displacement the blade would face as well as the moments found during the superposition calculations.

$$F_{x1} = -F * \sin(\zeta)$$

$$F_{x1} = -800 * \sin(20)$$

$$F_{x1} = -273.62 [N]$$

$$F_{y2} = F \times \cos(\zeta + \varphi)$$

$$F_{y2} = 800 \times \cos(20 + 14)$$

$$F_{y2} = 663.23 [N]$$

Using the rule of mixtures as explained in Chapter 4, an equivalent stiffness was found. In this case the equivalent stiffness was equal to E_2 , since E_1 and E_2 were equal. However, the calculation for E_{2eq} , is still illustrated below for thoroughness.

$$E = \sum_{i=1}^n V_i E_i$$

$$v_t = h_2 b$$

$$v_t = 7.4 \times 45$$

$$v_t = 334.35$$

$$V_1 = \frac{h_1 b}{v_t}$$

$$V_1 = \frac{4.3 \times 45}{334.35}$$

$$V_1 = 0.57$$

$$V_2 = \frac{(h_2 - h_1) \times b}{v_t}$$

$$V_2 = \frac{(7.4 - 4.3) \times 45}{334.35}$$

$$V_2 = 0.43$$

$$E_{2eq} = E_1 V_1 + E_2 V_2$$

$$E_{2eq} = (63.7)(0.57) + (63.7)(0.43)$$

$$E_{2eq} = 63.7 \text{ [GPa]}$$

The moment caused by the force up to point C is given by:

$$M_1 = F_{y2} L_4$$

$$M_1 = 663.23 \times 70$$

$$M_1 = 46\,426.10 \text{ [N} \cdot \text{mm]}$$

The moment responsible for bending AB is given by:

$$M_2 = M_1 + F_{y2} L_3$$

$$M_2 = 46\,426.10 + 663.23 \times 105$$

$$M_2 = 116\,065.26 \text{ [N} \cdot \text{mm]}$$

The area moment part 1 is given by:

$$I_1 = \frac{bh_1^3}{12}$$

$$I_1 = \frac{45 \times 4.3^3}{12}$$

$$I_1 = 298.15 \text{ [mm}^4\text{]}$$

The area moment part 2 is given by:

$$I_2 = \frac{bh_2^3}{12}$$

$$I_2 = \frac{45 \times 7.4^3}{12}$$

$$I_2 = 1\,519.59 \text{ [mm}^4\text{]}$$

Initial position

The calculations were done by using point locations on a Cartesian plane. First, the initial position of the points was calculated using the dimension lengths and angles. Thereafter, using superposition, the deflection of the segments and displacements of the point were calculated.

The position for each point was calculated, and the illustration of the points and foot is illustrated in Figure A - 7.

Point A:

$$A_0 = [0, 0]$$

Point B:

$$B_0 = [0, -L_1]$$

$$B_0 = [0, -44]$$

Point C:

$$C_{0x} = L_3 \cos(-\varphi)$$

$$C_{0x} = 105 \times \cos(-14)$$

$$C_{0x} = 101.88$$

$$C_{0y} = L_3 \sin(-\varphi) + B_{0y}$$

$$C_{0y} = 105 \times \sin(-14) - 44$$

$$C_{0y} = -69.40$$

$$C_0 = [101.88, -69.40]$$

Point D:

$$D_{0x} = L_2 \cos(-\varphi)$$

$$D_{0x} = 175 \times \cos(-14)$$

$$D_{0x} = 169.80$$

$$D_{0y} = L_2 \sin(-\varphi) + B_{0y}$$

$$D_{0y} = 175 \times \sin(-14) - 44$$

$$D_{0y} = -86.34$$

$$D_0 = [169.80, -86.34]$$

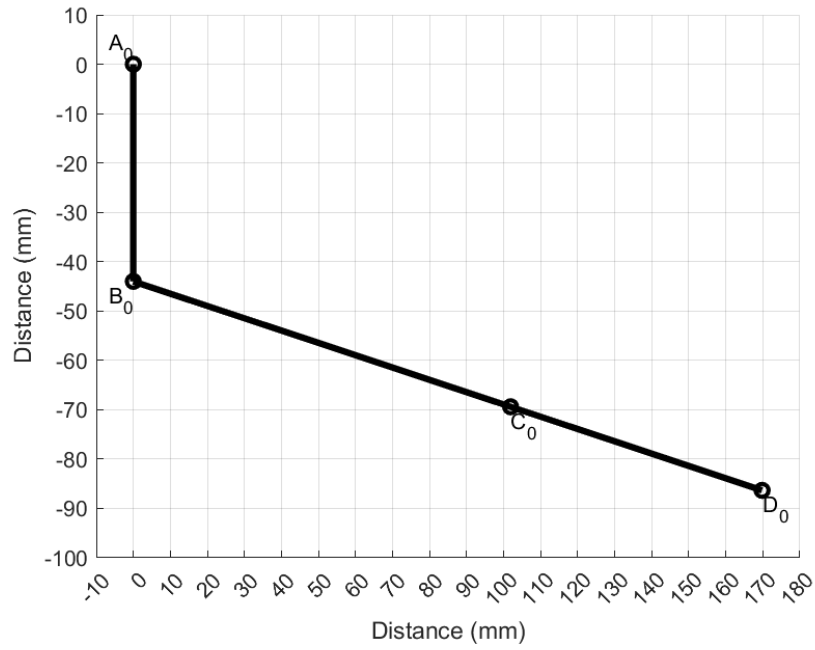


Figure A - 7: Initial position of the forefoot.

Deflection of Segment AB.

For the deflection of AB, the moment M_2 , was used. After the new location of B was calculated, the slope caused by the deflection was required to calculate the angle that the new BD segment has with the axis. With this angle, the new locations for C and D were calculated. The displacement due to the deflection of AB can be seen in Figure A - 8.

Point B:

$$B_{1x} = \frac{M_2 (-L_1)^2}{2E_{2eq}I_2} + \frac{F_{x1}(-L_1)^3}{3E_{2eq}I_2}$$

$$B_{1x} = \frac{116\,065.26 \times (-44)^2}{2 \times 63.7 \times 1000 \times 1519.59} + \frac{-273.62(-44)^3}{3 \times 63.7 \times 1000 \times 1519.59}$$

$$B_{1x} = 1.24$$

$$B_{1y} = -L_1$$

$$B_{1y} = -44$$

$$B_1 = [1.24, -44]$$

Slope and Angle:

$$\theta_1 = \frac{M_2(-L_1)}{E_{2eq}I_2} + \frac{F_{x1}(-L_1)}{2E_{2eq}I_2}$$
$$\theta_1 = \frac{116\,065.26 \times (-44)}{63.7 \times 1000 \times 1519.59} + \frac{-273.62(-44)}{2 \times 63.7 \times 1000 \times 1519.59}$$
$$\theta_1 = -0.055 \text{ [rad]}$$
$$\theta_1 = -3.15 \text{ [°]}.$$

$$\gamma_1 = \varphi + \theta_1$$
$$\gamma_1 = 14 - 3.15$$
$$\gamma_1 = 10.85 \text{ [°]}$$

Point C:

$$C_{1x} = L_3 \cos(\gamma_1) + B_{1x}$$
$$C_{1x} = 105 \times \cos(10.85) + 1.24$$
$$C_{1x} = 104.36$$

$$C_{1y} = -L_3 \sin(\gamma_1) + B_{1y}$$
$$C_{1y} = 105 \times \sin(10.85) - 44$$
$$C_{1y} = -63.77$$

$$C_1 = [104.36, -63.77]$$

Point D:

$$D_{1x} = L_2 \cos(\gamma_1) + B_{1x}$$
$$D_{1x} = 175 \times \cos(10.85) + 1.24$$
$$D_{1x} = 173.11$$

$$D_{1y} = -L_2 \sin(\gamma_1) + B_{1y}$$
$$D_{1y} = -175 \times \sin(10.85) - 44$$
$$D_{1y} = -76.94$$

$$D_1 = [173.11, -76.94]$$

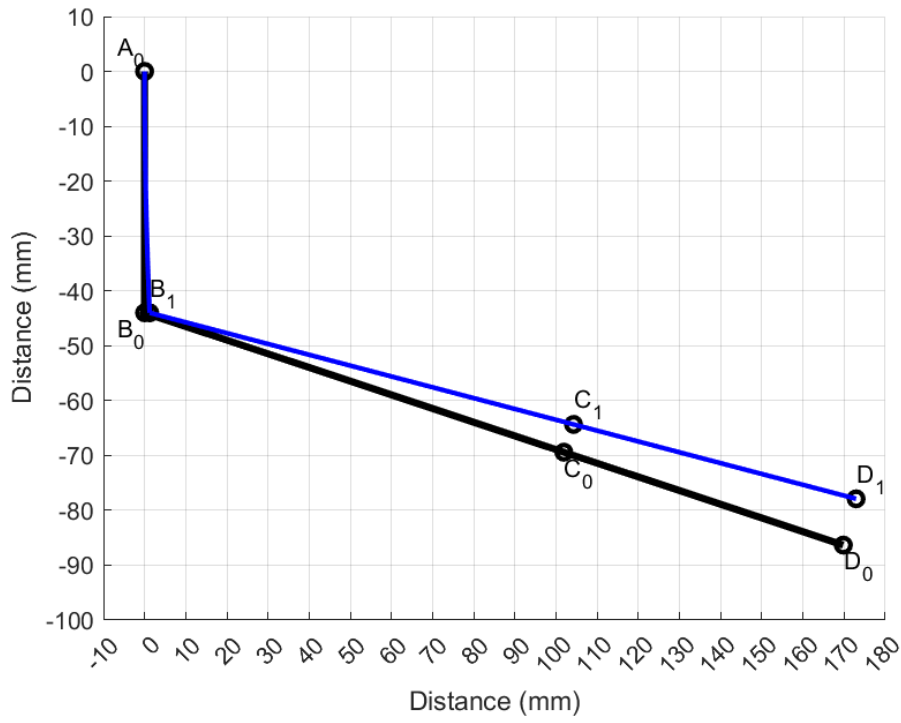


Figure A - 8: Displacement of the forefoot blade when AB deflects.

Deflection of segment CD

The deflection of segment CD was considered a simple cantilever beam with a force applied at the edge. Once the initial point, D_{2a} , was calculated, the point was rotated to fit the slope of BD and thereafter adjusted to ensure it fits at point C.

Point D:

Initial deflected point:

$$D_{2ax} = L_4$$

$$D_{2ax} = 70$$

$$D_{2ay} = \frac{F_{y2}(L_4)^3}{3E_1I_1}$$

$$D_{2ay} = \frac{663.23 \times (70)^3}{3 \times 63.7 \times 1000 \times 298.15}$$

$$D_{2ay} = 3.99$$

$$D_{2a} = [70, 3.99]$$

Rotation of initial point:

$$R = \begin{bmatrix} \cos(-\varphi) & -\sin(-\varphi) \\ \sin(-\varphi) & \cos(-\varphi) \end{bmatrix}$$

$$D_{2b} = R \times D_{2a}$$

$$D_{2b} = [68.89, -13.06]$$

Final point location:

$$D_{2x} = D_{2bx} + C_{0x}$$

$$D_{2x} = 68.89 + 101.88$$

$$D_{2x} = 170.77$$

$$D_{2y} = D_{2by} + C_{0y}$$

$$D_{2y} = -12.92 - 69.40$$

$$D_{2y} = -82.46$$

$$D_2 = [170.77, -82.46]$$

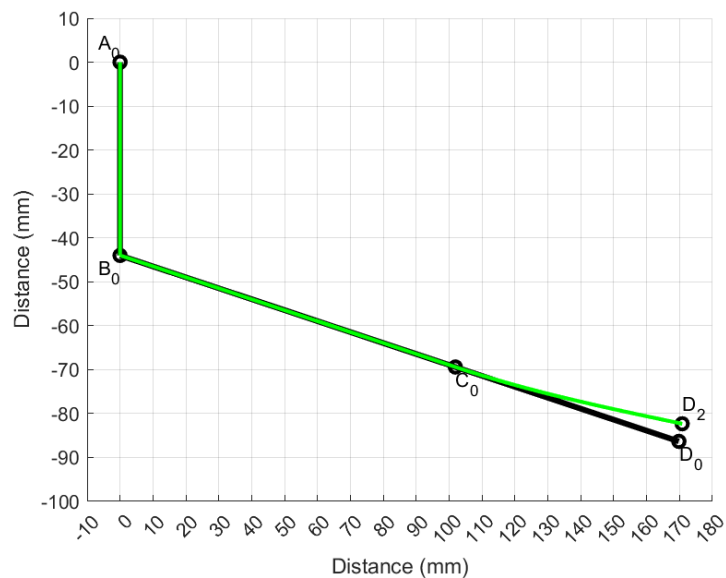


Figure A - 9: Deflection of segment CD

Deflection of Segment BC

This segment's method of calculation followed the methods from both the calculation of the deflection of Segment AB and the calculation of Segment CD. First, an initial point for point C was calculated, which was then rotated and adjusted. After this, the slope due to the deflection of BC was calculated, which was used to calculate the angle of CD with the x-axis. This angle was used to calculate where point D will be located. The displacement is illustrated in Figure A - 10.

Point C:

Initial deflected point:

$$C_{3ax} = L_3$$

$$C_{3ax} = 105$$

$$C_{3ay} = \frac{F_{y2}(L_3)^3}{3E_{2eq}I_2} + \frac{M_1(L_3)^2}{2E_{2eq}I_2}$$

$$C_{3ay} = \frac{663.23 \times (105)^3}{3 \times 63.7 \times 1000 \times 1519.59} + \frac{46\,426.10 \times (105)^2}{2 \times 63.7 \times 1000 \times 1519.59}$$

$$C_{3ay} = 5.29$$

$$C_{3a} = [105, 5.29]$$

Rotation of initial point:

$$R = \begin{bmatrix} \cos(-\varphi) & -\sin(-\varphi) \\ \sin(-\varphi) & \cos(-\varphi) \end{bmatrix}$$

$$C_{3b} = R \times C_{3a}$$

$$C_{3b} = [103.16, -20.27]$$

Final point location:

$$C_{3x} = C_{3bx} + B_{0x}$$

$$C_{3x} = 103.14 + 0$$

$$C_{3x} = 103.16$$

$$C_{3y} = C_{3by} + B_{0y}$$

$$C_{3y} = -20.27 - 44$$

$$C_{3y} = -64.27$$

$$C_3 = [103.16, -64.27]$$

Slope and Angle:

$$\theta_3 = -\frac{M_1(L_3)}{E_{2eq}I_2} + \frac{F_{y2}(L_3)^2}{2E_{2eq}I_2}$$

$$\theta_3 = -\frac{46\,426.10 \times (105)}{63.7 \times 1000 \times 1519.59} + \frac{663.23 \times (105)^2}{2 \times 63.7 \times 1000 \times 1519.59}$$

$$\theta_3 = 0.09 \text{ [rad]}$$

$$\theta_3 = 5.16 \text{ [}^\circ\text{]}$$

$$\gamma_3 = -\varphi + \theta_3$$

$$\gamma_3 = -14 + \text{atan}(5.16)$$

$$\gamma_3 = -8.86 \text{ [}^\circ\text{]}$$

Point D:

$$D_{3x} = L_4 \cos(\gamma_3) + C_{3x}$$

$$D_{3x} = 70 \times \cos(-8.86) + 103.16$$

$$D_{3x} = 172.32$$

$$D_{3y} = L_4 \sin(\gamma_3) + C_{3y}$$

$$D_{3y} = 70 \times \sin(-8.86) - 64.27$$

$$D_{3y} = -75.05$$

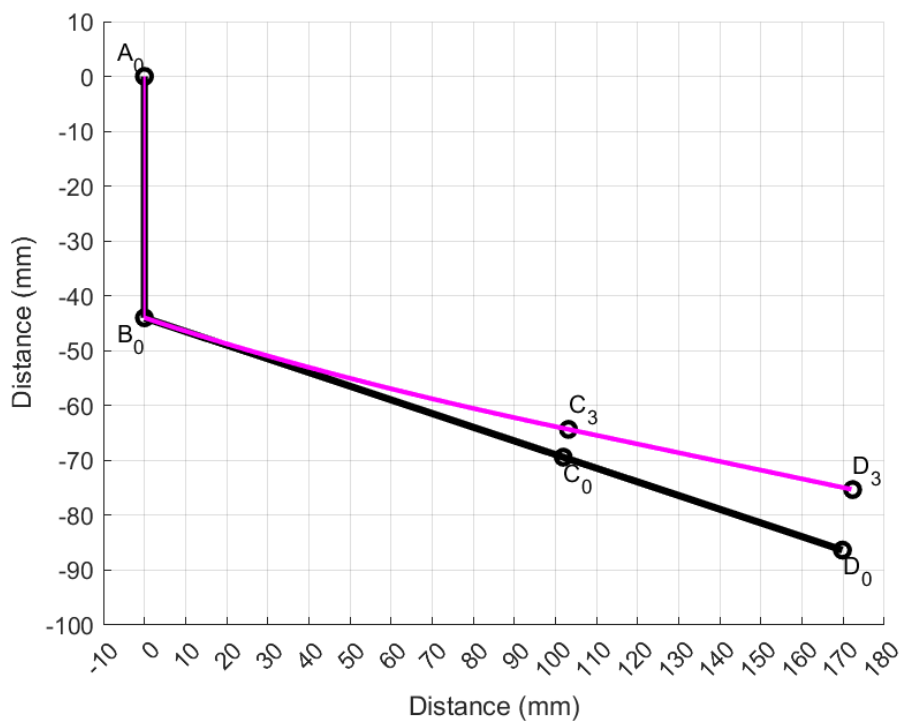


Figure A - 10: Deflection of segment BC

Final Displacement

For the final displacement calculation, the points calculated in the previous sections, C_{3a} and D_{2a} , were rotated to fit the displaced section due to AB's deflection. Point C was rotated to fit the displaced BC segment, whereas point D required the calculation of an angle to fit the slope. The final displacement of the prosthetic foot was illustrated in Figure A - 11.

Point C:

Rotation of C_{3a} :

$$R = \begin{bmatrix} \cos(\gamma_1) & -\sin(\gamma_1) \\ \sin(\gamma_1) & \cos(\gamma_1) \end{bmatrix}$$

$$C_{5a} = R \times C_{3a}$$

$$C_{5a} = [104.12, -14.57]$$

Final point location:

$$C_{5x} = C_{5ax} + B_{1x}$$

$$C_{5x} = 104.12 + 1.24$$

$$C_{5x} = 105.36$$

$$C_{5y} = C_{5ay} + B_{1y}$$

$$C_{5y} = -14.57 - 44$$

$$C_{5y} = -58.57$$

$$C_5 = [105.36, -58.57]$$

Angle:

$$\gamma_5 = -\varphi + \theta_3 - \theta_1$$

$$\gamma_5 = -14 + 5.16 + 3.15$$

$$\gamma_5 = -5.71 [^\circ]$$

Point D:

Rotation of D_{2a} :

$$R = \begin{bmatrix} \cos(\gamma_5) & -\sin(\gamma_5) \\ \sin(\gamma_5) & \cos(\gamma_5) \end{bmatrix}$$

$$D_{5a} = R \times D_{2a}$$

$$D_{5a} = [70.05, -2.99]$$

Final point location:

$$D_{5x} = D_{5ax} + C_{5x}$$

$$D_{5x} = 70.05 + 105.18$$

$$D_{5x} = 175.41$$

$$D_{5y} = D_{5ay} + C_{5y}$$

$$D_{5y} = -2.99 - 58.57$$

$$D_{5y} = -61.56$$

$$D_5 = [175.41, -61.56]$$

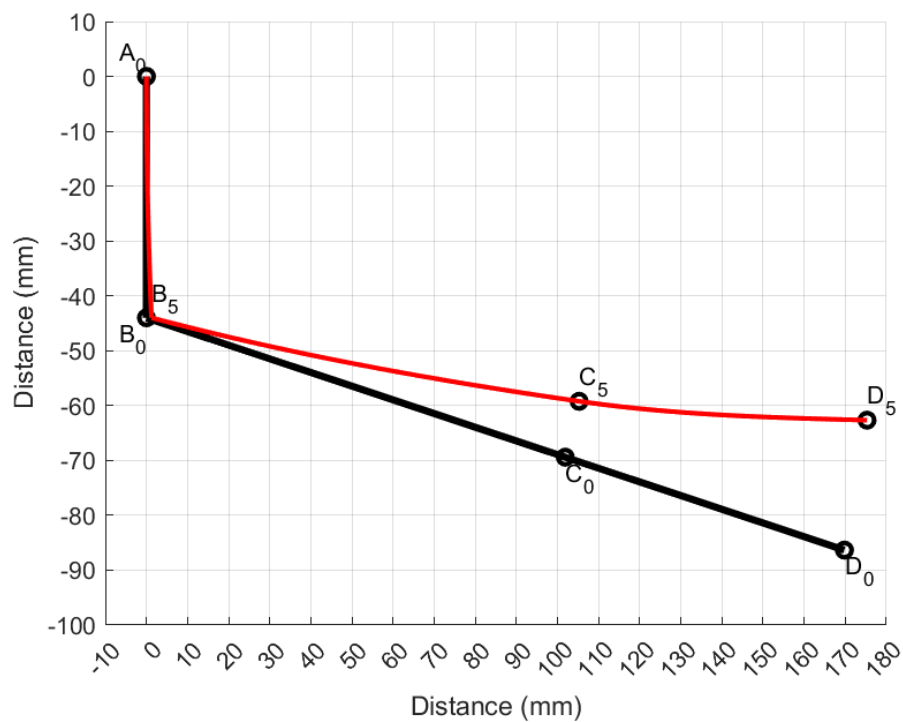


Figure A - 11: Final forefoot displacement

The total forefoot displacement was given by subtracting the end and initial D-y values.

$$\Delta_y = D_{5y} - D_{0y}$$

$$\Delta_y = -61.56 + 86.34$$

$$\Delta_y = 24.78$$

Heel

Figure A - 12 is a line illustration of the forefoot and heel blades. The heel blade's dimensions depend on the dimensions of the forefoot blade. Figure A - 13 illustrates the resulting forces and moments to consider for the calculations of the heel blade.

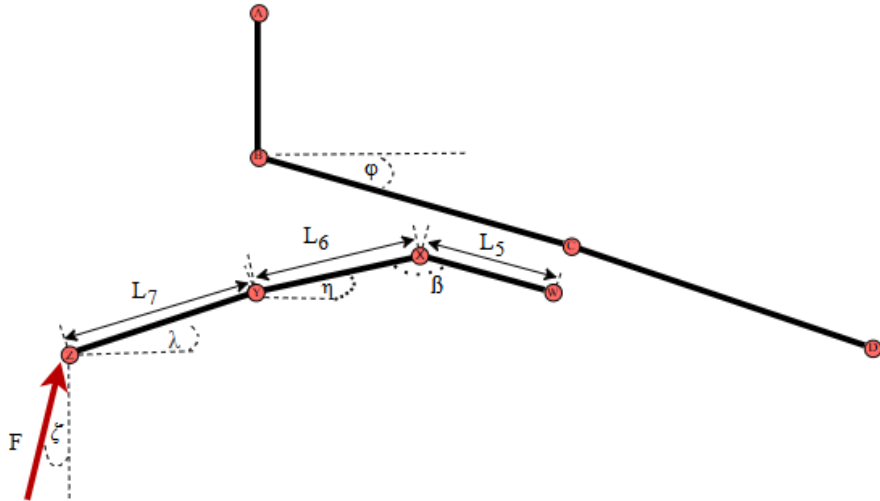


Figure A - 12 Segments of the heel blade with the forefoot blade.

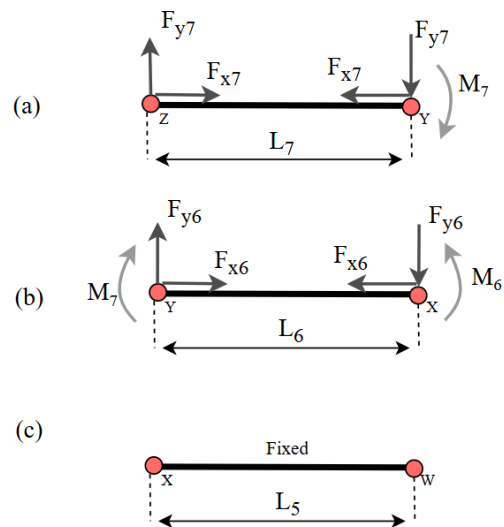


Figure A - 13: Resulting forces and moments on the heel blade.

Constants and Assumptions

The input variables required to complete the calculations of the heel's displacement were captured in Table A - 4. The variables were gathered from the results of the Designer.

Table A - 4: Input values for heel blade calculations

F	800	[N]
b	45	[mm]
α	104	[°]
ζ	15	[°]
λ	21	[°]
β	160	[°]
L ₅	30	[mm]
L ₆	50	[mm]
L ₇	49	[mm]
h ₂	7.4	[mm]
h ₃	5.6	[mm]
E ₃	24.24	[GPa]

Initial Calculations

The calculations for the moments and forces illustrated in Figure A - 13 are done below:

$$\eta = 90 - (360 - \alpha - \beta)$$

$$\eta = 90 - (360 - 104 - 160)$$

$$\eta = -6 [^\circ]$$

$$F_{y6} = F \times \sin (90 - \zeta - \eta)$$

$$F_{y6} = 800 \times \sin (90 - 15 + 6)$$

$$F_{y6} = 746.86 [N]$$

$$F_{y7} = F \times \sin (90 - \lambda - \zeta)$$

$$F_{y7} = 800 \times \sin (90 - 21 - 6)$$

$$F_{y7} = 647.21 [N]$$

$$M_7 = F_{y7} L_7$$

$$M_7 = 790.15 \times 49$$

$$M_7 = 31\,713.29 [N \cdot mm]$$

$$M_6 = F_{y6}L_6 + M_7$$

$$M_6 = 746.86 \times 50 + 31\,713.29$$

$$M_6 = 69\,056.29 [N \cdot mm]$$

$$I_3 = \frac{bh_3^3}{12}$$

$$I_3 = \frac{45 \times 5.6^3}{12}$$

$$I_3 = 658.56 [mm^4]$$

Initial Position.

As with the forefoot, the calculations were done by using point locations on a Cartesian plane. The input angles and length dimensions were used to calculate the initial position of the points. Using superposition, the deflection of the segments and displacements of the point were calculated.

The positions of the four heel blade points are calculated and then illustrated in Figure A - 14.

Point W:

$$W_{0x} = C_{0x} - h_2 \cos(\varphi)$$

$$W_{0x} = 101.88 - 7.4 \times \cos(14)$$

$$W_{0x} = 94.70$$

$$W_{0y} = C_{0y} - h_2 \sin(\varphi)$$

$$W_{0y} = -69.40 - 7.4 \times \sin(14)$$

$$W_{0y} = -71.19$$

$$W_0 = [94.70, -71.19]$$

Point X:

$$X_{0x} = -L_5 \cos(\varphi) + W_{0x}$$

$$X_{0x} = -30 \times \cos(14) + 94.70$$

$$X_{0x} = 65.59$$

$$X_{0y} = L_5 \sin(\varphi) + W_{0y}$$

$$X_{0y} = -30 \times \sin(14) - 71.19$$

$$X_{0y} = -63.93$$

$$X_0 = [65.59, -63.93]$$

Point Y:

$$Y_{0x} = -L_6 \cos(\eta) + X_{0x}$$

$$Y_{0x} = -50 \times \cos(-6) + 65.59$$

$$Y_{0x} = 15.86$$

$$Y_{0y} = L_6 \sin(\eta) + X_{0y}$$

$$Y_{0y} = -50 \times \sin(-6) - 63.93$$

$$Y_{0y} = -69.16$$

$$Y_0 = [15.86, -69.16]$$

Point Z:

$$Z_{0x} = -L_7 \cos(\lambda) + Y_{0x}$$

$$Z_{0x} = -49 \times \cos(21) + 15.86$$

$$Z_{0x} = -29.89$$

$$Z_{0y} = -L_7 \sin(\lambda) + Y_{0y}$$

$$Z_{0y} = -49 \times \sin(21) - 69.16$$

$$Z_{0y} = -86.72$$

$$Z_0 = [-29.89, -86.72]$$

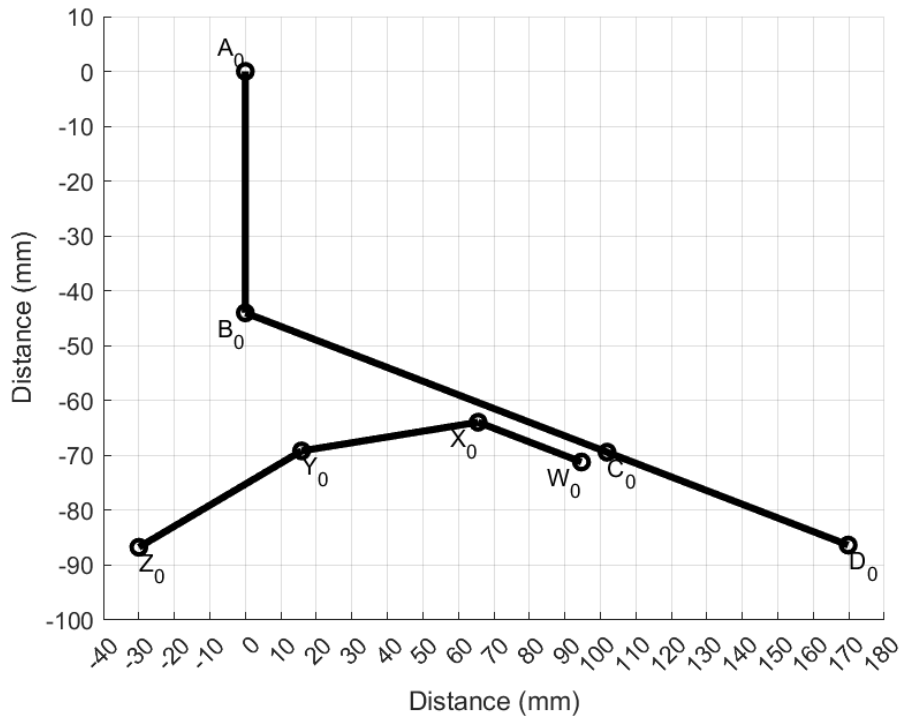


Figure A - 14: Initial points of the heel blade

Deflection of Segment ZY

To calculate the displacement of Z, first an initial point, Z_{6a} , was calculated, after which the point was rotated to fit the initial ZY slope and then adjusted to meet point Y. This deflection is illustrated in Figure A - 15.

Point Z:

Initial deflected point:

$$Z_{6ax} = -L_7$$

$$Z_{6ax} = -49.00$$

$$Z_{6ay} = \frac{F_{y7}L_7^3}{3E_3I_3}$$

$$Z_{6ay} = \frac{647.21 \times 49^3}{3 \times 24.24 \times 1000 \times 658.56}$$

$$Z_{6ay} = 1.59$$

$$Z_{6a} = [-49, 1.59]$$

Rotation of initial point:

$$R = \begin{bmatrix} \cos(\lambda) & \sin(\lambda) \\ -\sin(\lambda) & \cos(\lambda) \end{bmatrix}$$

$$Z_{6b} = R \times Z_{6a}$$

$$Z_{6b} = [-46.32, -16.08]$$

Final point location:

$$Z_{6x} = Z_{6bx} + Y_{0x}$$

$$Z_{6x} = -46.32 + 15.86$$

$$Z_{6x} = -30.46$$

$$Z_{6y} = Z_{6by} + Y_{0y}$$

$$Z_{6y} = -16.08 - 69.16$$

$$Z_{6y} = -85.24$$

$$Z_6 = [-30.46, -85.24]$$

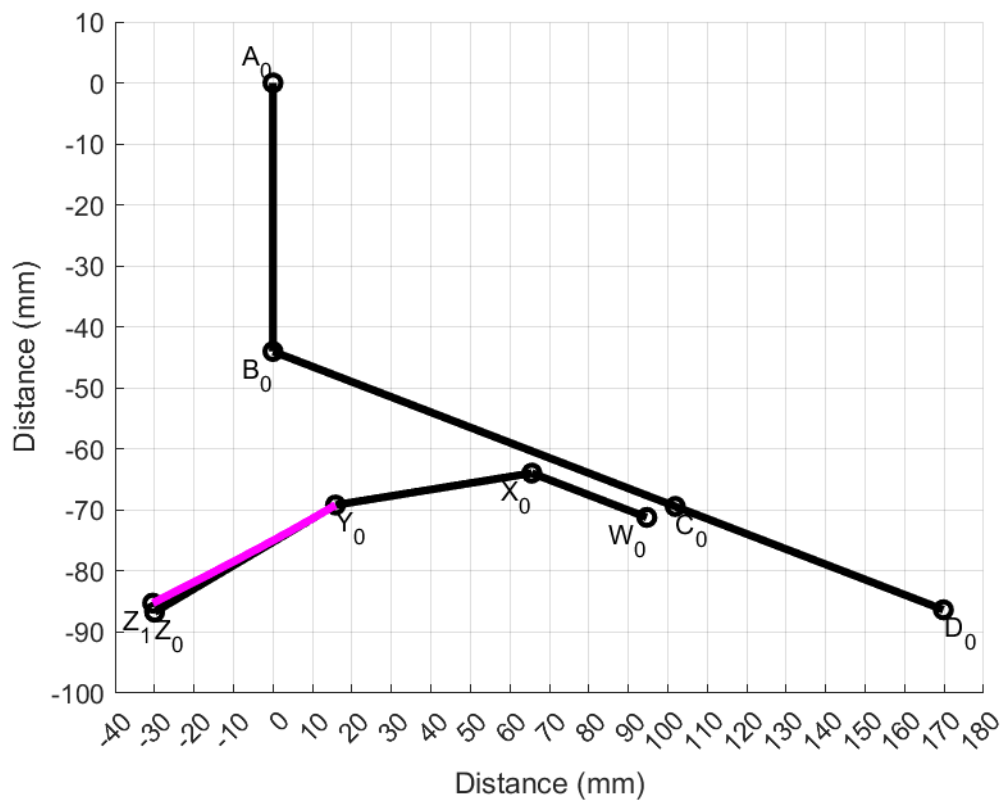


Figure A - 15: The deflection of segment YZ

Deflection of Segment YX

Next, the deflection of segment YX was determined. The point Z was calculated by the angle created due to the deflection of Y. The angle η is the angle between the X-axis and the segment YX. The input values provided an η angle of -6° .

Point Y:

Initial deflected point:

$$Y_{7ax} = -L_6$$

$$Y_{7ax} = -50$$

$$Y_{7ay} = \frac{M_7(L_6)^2}{2E_3I_3} + \frac{F_{y6}(L_6)^3}{3E_3I_3}$$

$$Y_{7ay} = \frac{31\,713.30 \times (70)^2}{2 \times 24.24 \times 1000 \times 658.56} + \frac{746.86 \times (70)^3}{3 \times 24.24 \times 1000 \times 658.56}$$

$$Y_{7ay} = 4.43$$

$$Y_{7a} = [-50, 4.43]$$

Rotation of initial point:

$$R = \begin{bmatrix} \cos(-\eta) & -\sin(-\eta) \\ \sin(-\eta) & \cos(-\eta) \end{bmatrix}$$

$$Y_{7b} = R \times Y_{7a}$$

$$Y_{7b} = [-50.19, 0.82]$$

Final point location:

$$Y_{7x} = Y_{7bx} + X_{0x}$$

$$Y_{7x} = -50.19 + 65.59$$

$$Y_{7x} = 15.40$$

$$Y_{7y} = Y_{7by} + X_{0y}$$

$$Y_{7y} = -0.82 - 63.93$$

$$Y_{7y} = -64.75$$

$$Y_7 = [15.40, -64.75]$$

Slope and Angle:

$$\theta_7 = \frac{M_7(L_6)}{E_3 I_3} + \frac{F_{y6}(L_6)^2}{2E_3 I_3}$$

$$\theta_7 = \frac{31\,713.30 \times (70)}{24.24 \times 1000 \times 658.56} + \frac{746.86 \times (70)^2}{2 \times 24.24 \times 1000 \times 658.56}$$

$$\theta_7 = 0.16 \text{ [rad]}$$

$$\theta_7 = 9.17 \text{ [}^\circ\text{]}$$

$$\gamma_7 = \lambda - \theta_7$$

$$\gamma_7 = 21 - 9.17$$

$$\gamma_7 = 11.91 \text{ [}^\circ\text{]}$$

Point Z:

Initial deflected point:

$$Z_{7x} = -L_7 \cos(\gamma_7) + Y_{7x}$$

$$Z_{7x} = -49 \times \cos(11.91) + 15.40$$

$$Z_{7x} = -32.55$$

$$Z_{7y} = -L_7 \sin(\gamma_7) + Y_{7y}$$

$$Z_{7y} = -49 \times \sin(11.91) - 64.75$$

$$Z_{7y} = -74.86$$

$$Z_7 = [-32.55, -74.86]$$

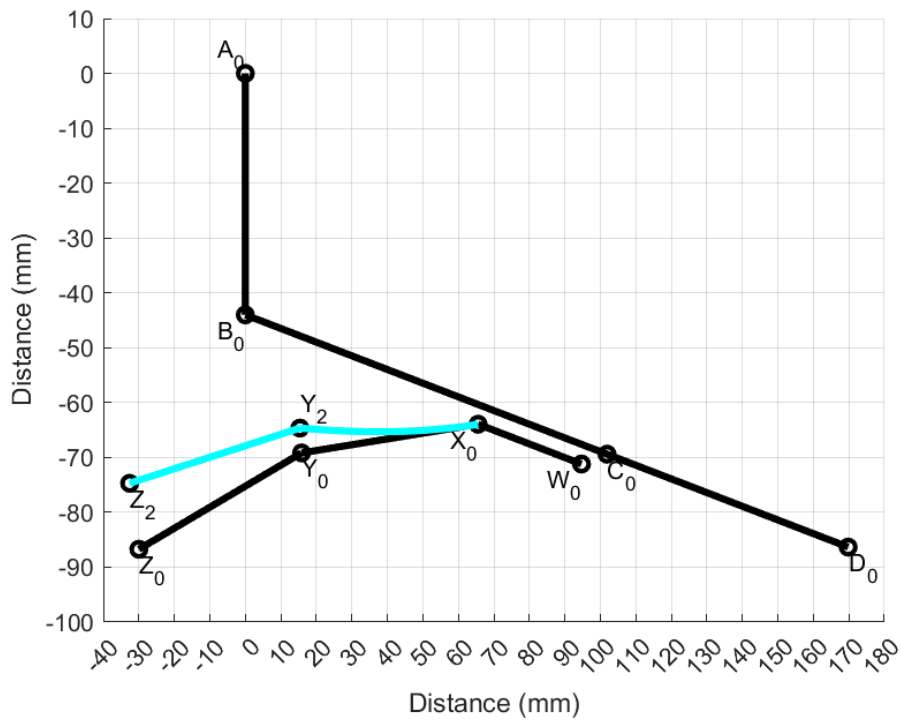


Figure A - 16: Deflection of segment YX

Final Displacement

The final displacement of the heel blade was calculated by rotating the Z value from ZY's deflection around the displacement segment of ZY from YX's deflection. The final displacement is illustrated in Figure A - 16.

Point C:

Rotation of Z_{6a} :

$$R = \begin{bmatrix} \cos(\gamma_7) & \sin(\gamma_7) \\ -\sin(\gamma_7) & \cos(\gamma_7) \end{bmatrix}$$

$$Z_{8a} = R \times Z_{6a}$$

$$Z_{8a} = [-48.27, -8.56]$$

Final point location:

$$Z_{8x} = Z_{8ax} + Y_{7x}$$

$$Z_{8x} = -48.27 + 15.40$$

$$Z_{8x} = -32.87$$

$$Z_{8y} = Z_{8ay} + Y_{7y}$$

$$Z_{8y} = -8.56 - 64.75$$

$$Z_{8y} = -73.31$$

$$Z_8 = [-32.87, -73.31]$$

The total vertical heel displacement is given by subtracting the end and initial Z-y values.

$$\Delta_y = Z_{8y} - Z_{0y}$$

$$\Delta_y = -73.31 + 86.72$$

$$\Delta_y = 13.41$$

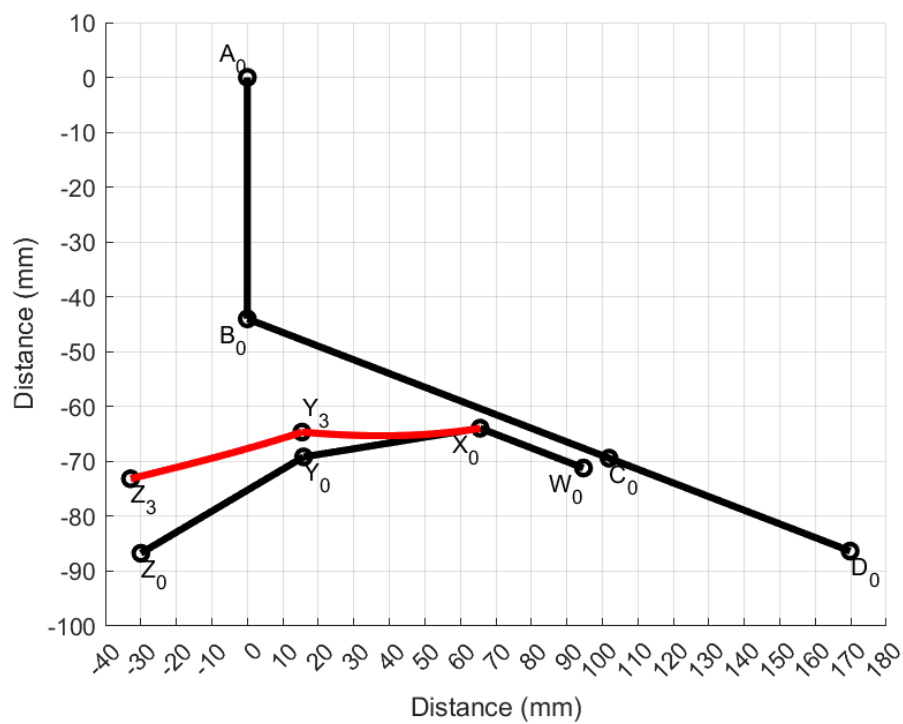


Figure A - 17: Final heel displacement

Results comparison

As can be seen in Table A - 5 the results from the Designer and the hand calculations are highly comparable. The largest difference from the Designer, was between the heel results, with a percentage deviation less than 1%. The most probable cause for the differences was the usage of significant figures. The hand calculations only considered 2 significant figures, whereas the Designer used Matlab's default number of figures, which was 16.

The deviations for the forefoot and for the heel comparison are below 1% and the dimensions gathered from the Designer were considered acceptable dimensions for the prosthetic foot.

Table A - 5: Comparison between the Designer and hand calculation results

		Values
Forefoot	Designer (mm)	24.73
	Hand Calculation (mm)	24.78
	Difference (mm)	0.05
	Difference (%)	0.22
Heel	Designer (mm)	13.31
	Hand Calculation (mm)	13.41
	Difference (mm)	0.10
	Difference (%)	0.76

To further verify the results from the Designer, the results were compared to that from a finite element analysis. The FEA was set up exactly as explained in Appendix A.1, only using the value set as presented by the designer in Table A - 3 and Table A - 4. The results can be seen in Figure A - 18. Table A - 6 presents the results from the comparison of the FEA and designer. The percentage deviations from both comparisons were revealed to be around 3%. NX, being a CAD software, uses different methods of calculating.

Overall, the error deviations were acceptable and verified the output variables of the Designer as a possible solution for a prosthetic foot.

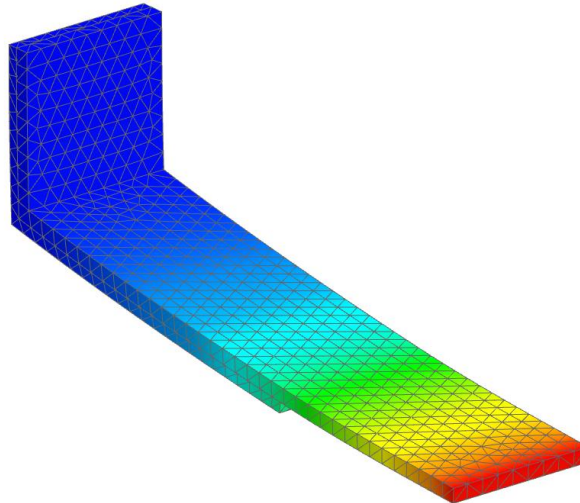
Table A - 6: Comparison between the Designer and FEA results for the forefoot blade

		Values
Forefoot	Designer (mm)	24.73
	FEA (mm)	24.13
	Difference (mm)	0.60
	Difference (%)	2.47
Heel	Designer (mm)	13.31
	FEA (mm)	13.78
	Difference (mm)	0.47
	Difference (%)	3.42

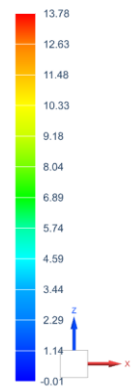
Blade Assembly_sim1 : Solution 1 Result
 Subcase - Static Loads 1, Static Step 1
 Displacement - Nodal, Z
 Min : -0.19, Max : 24.13, Units = mm
 Deformation : Displacement - Nodal Magnitude



[mm]



Blade Assembly_sim1 : Solution 2 Result
 Subcase - Static Loads 1, Static Step 1
 Displacement - Nodal, Z
 Min : -0.01, Max : 13.78, Units = mm
 Deformation : Displacement - Nodal Magnitude



[mm]

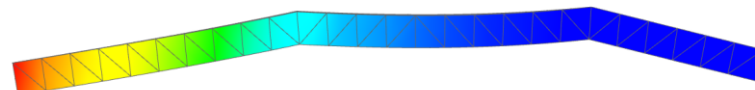


Figure A - 18: FEA results for the verification of designer results

The figure provides the FEA results when using the designer dimension values, with the forefoot results above and the heel results below.

A.3. Verification of Gas Spring Calculations

FEA was used to verify the calculations around the addition of gas springs and gas spring forces to the prosthetic forefoot. The FEA was set up the same as for the hand calculation verification in Appendix A.1 and the designer result verification in Appendix A.2. The difference came in with the addition of the gas spring forces. As was done with the applied loads, the gas spring forces were split into their respective components, G_x and G_y as explained in Chapter 4 with equations 86 and 87.

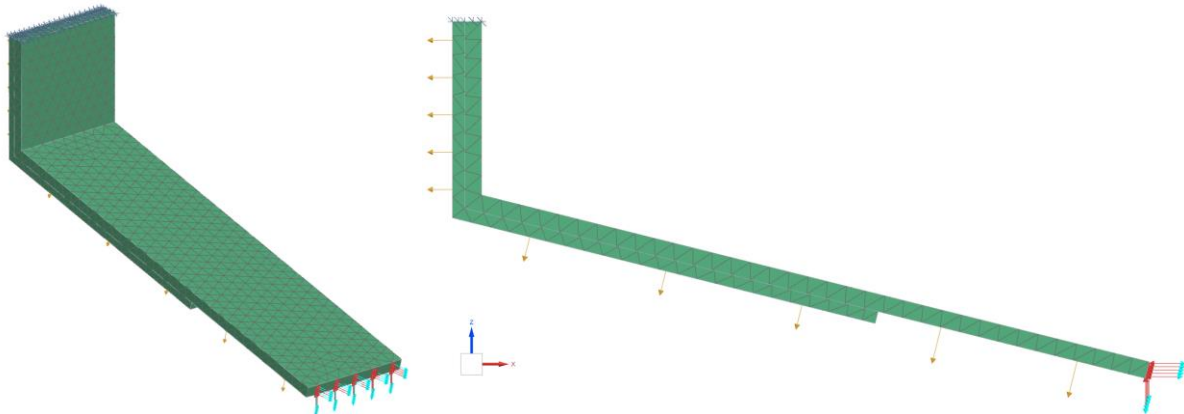


Figure A - 19: FEA setup for gas spring verification

The calculations done to verify the applied gas spring loads followed the method as described in Chapter 4. In the following tables, the results of the calculations and of the FEA are recorded. The comparison of the results for the forefoot blade without gas springs applied (Table A - 7), the 300 N gas spring force applied (Table A - 8), the 700 N (Table A - 9), and the 1000 N gas spring forces applied (Table A - 10). The results of the FEA with an applied force of 800 N and a gas spring force of 700 N can be seen in Figure A - 20. The FEA results were considered comparable to the calculations and verified the calculation method.

Table A - 7: Calculation vs FEA for gas spring verification: no gas spring

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
200	6.13	6.03	0.09	1.54
500	15.40	15.08	0.32	2.10
800	24.73	24.13	0.60	2.47
1000	30.96	30.17	0.79	2.63
1200	37.20	36.20	1.00	2.76

Table A - 8: Calculation vs FEA for gas spring verification: 300 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
200	1.93	2.00	0.07	3.27
500	10.98	11.29	0.31	2.72
800	20.17	20.34	0.17	0.86
1000	26.39	26.38	0.01	0.05
1200	32.63	32.41	0.22	0.68

Table A - 9: Calculation vs FEA for gas spring verification: 700 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
500	5.52	5.69	0.17	3.05
800	14.61	15.29	0.68	4.47
1000	20.70	21.32	0.62	2.91
1200	26.81	27.35	0.54	1.97

Table A - 10: Calculation vs FEA for gas spring verification: 1000 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
800	10.49	10.79	0.30	2.77
1000	16.57	17.53	0.96	5.49
1200	22.67	23.56	0.89	3.79

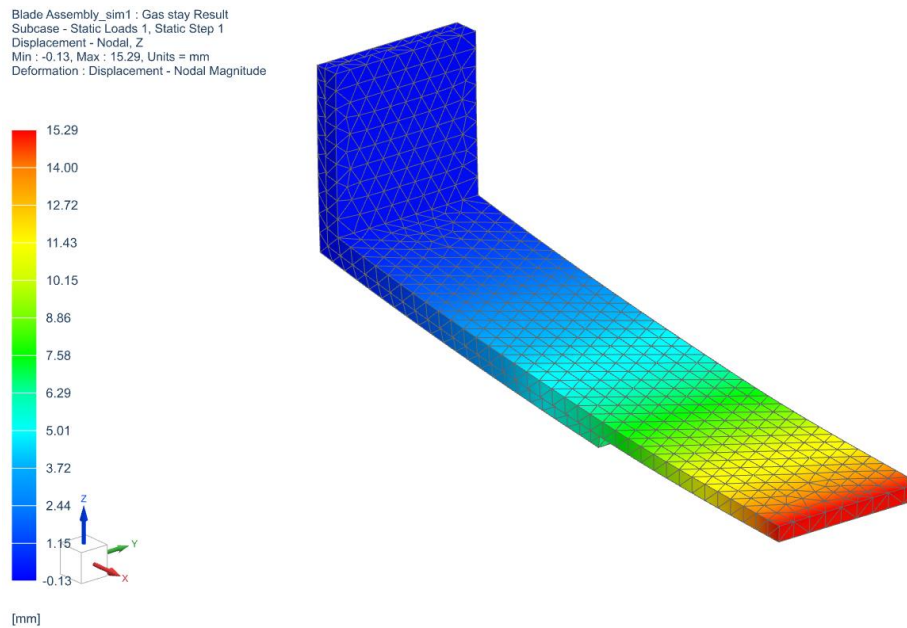


Figure A - 20: FEA results for a forefoot with 800 N applied load and 700 N GSF

A.4. Verification of Layup FEA results

Finite element analysis (FEA) was performed on thin-faced models matching the dimensions gathered from the Designer from Table 5.

A CQUAD4 mesh was used for the composite components, and the element size was refined to 1 mm to improve accuracy in stress and displacement predictions. The material properties not mentioned within Chapter 4 are considered confidential and not made available within this study. The laminate tool was used to create ply layups for each blade. The plies were set up in a staggered manner, as can be seen in Figure A - 21. The staggered lengths as well as the ply orientation were achieved by multiple iterations and inspection of FEA. The final ply layup schedule can be seen in the layup drawings for each blade in Appendix B.

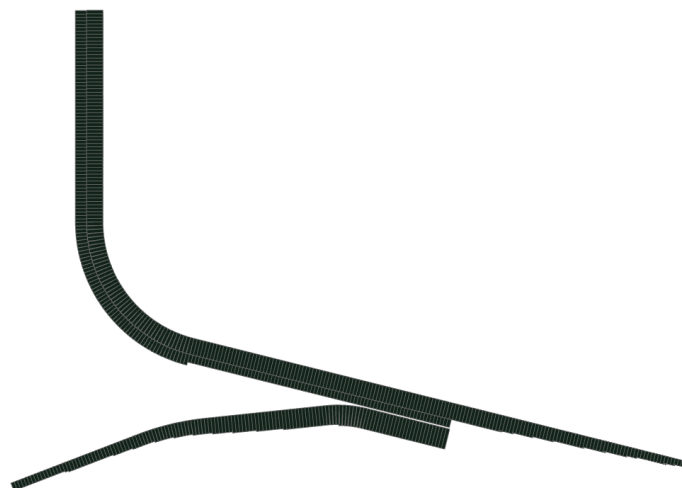


Figure A - 21: Display of ply layup with stagger

Although all the plies are two-dimensional and are illustrated as thin faces in NX, there is a display option that allows for three-dimensional visualisation. The ply layup is visualised in this figure, and the stagger effect can be seen on the three blades, as they are thinner on the edges where the loads are applied and thicker in the places where the most bending occurs.

Fixed supports were applied as described in the calculation method:

- The top edges of the forefoot blades, as seen in Figure A - 22.
- The two forefoot blades were bonded together by a glue constraint.
- The face of the heel blade was bonded to the bottom forefoot blade by a glue constraint.

As explained in the previous FEA setups, the loads were applied in their respective components to the free edges of the forefoot and heel blades.

The results of the forefoot and heel with 800 N applied can be seen in Figure A - 23 and Figure A - 24 respectively. The forefoot had a displacement of 23.6 mm and a maximum ply index of 0.419, as seen in Figure A - 23. On closer inspection, however, the average around the maximum ply index was only 0.391, which was allowable given the 0.45 cut-off for the PI. The heel blade has a displacement of 13.58 and a max ply index of 0.38, which was also acceptable.

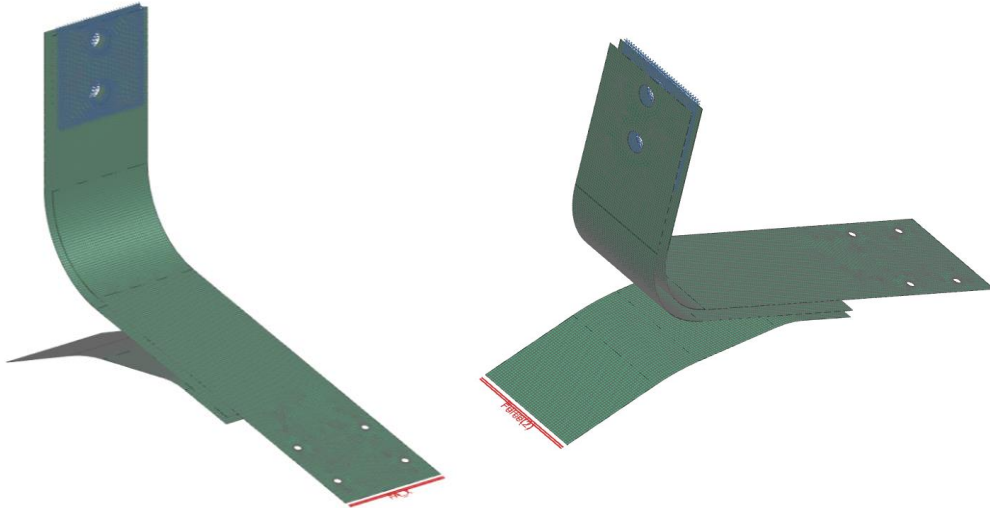


Figure A - 22: FEA setup of applied forces to the forefoot (left) and heel blade (right)

The fixed face can be seen at the top front of the foot. This is where the pyramid adapter is mounted. The forces are applied to the face edge of the forefoot and heel. The forces are split into horizontal and vertical components as done with the other FEA setups.

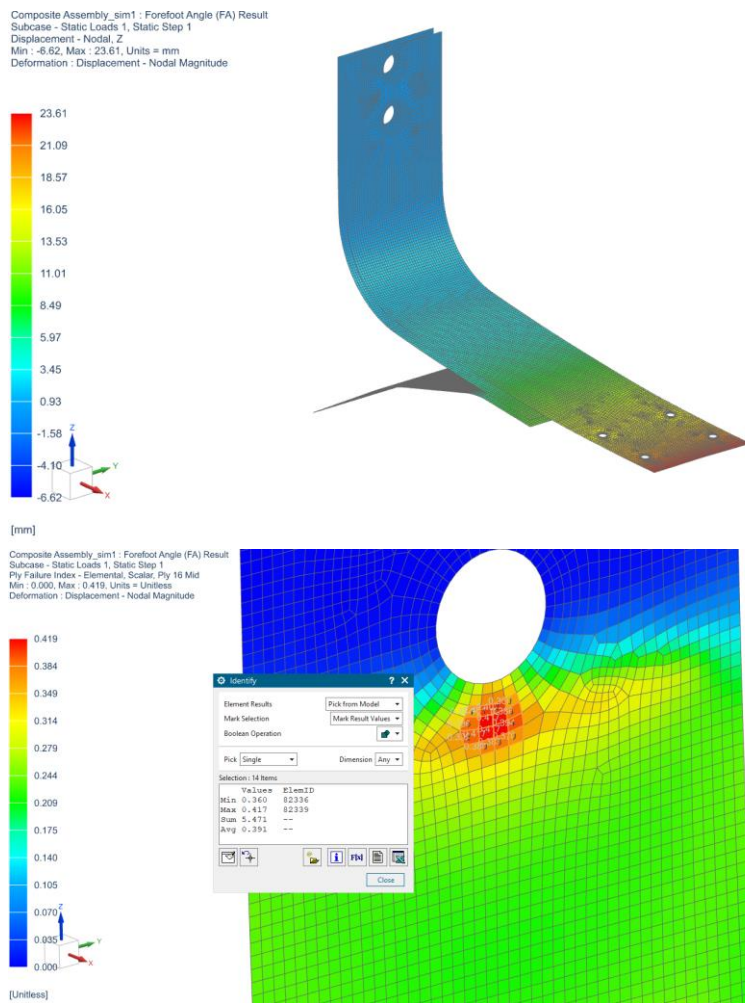
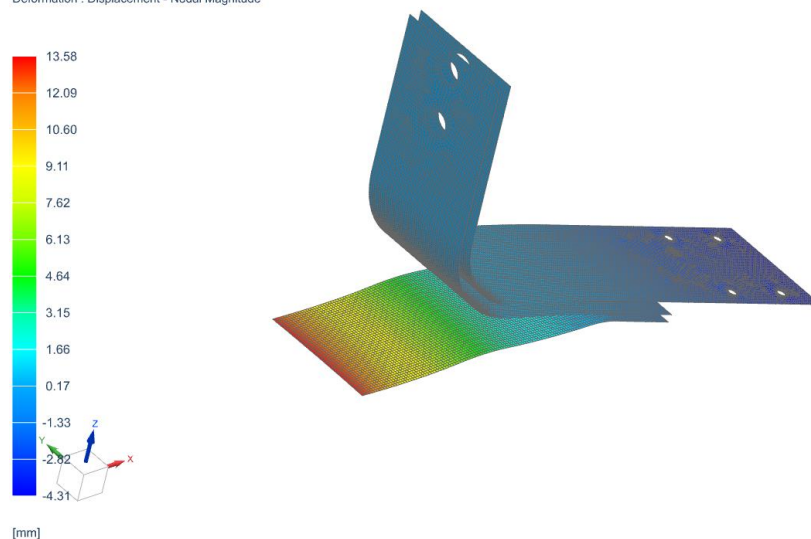


Figure A - 23: Results of FEA of the ply layup forefoot

The displacement results (top) and ply index results (bottom) for FEA on the forefoot with 800 N applied.

Composite Assembly_sim1 : Copy of Heel Angle (HA) Result
 Subcase - Static Loads 1, Static Step 1
 Displacement - Nodal, Z
 Min : -4.31, Max : 13.58, Units = mm
 Deformation : Displacement - Nodal Magnitude



Composite Assembly_sim1 : Copy of Heel Angle (HA) Result
 Subcase - Static Loads 1, Static Step 1
 Ply Failure Index - Elemental, Scalar, Ply 319 Mid
 Min : 0.000, Max : 0.380, Units = Unitless
 Deformation : Displacement - Nodal Magnitude

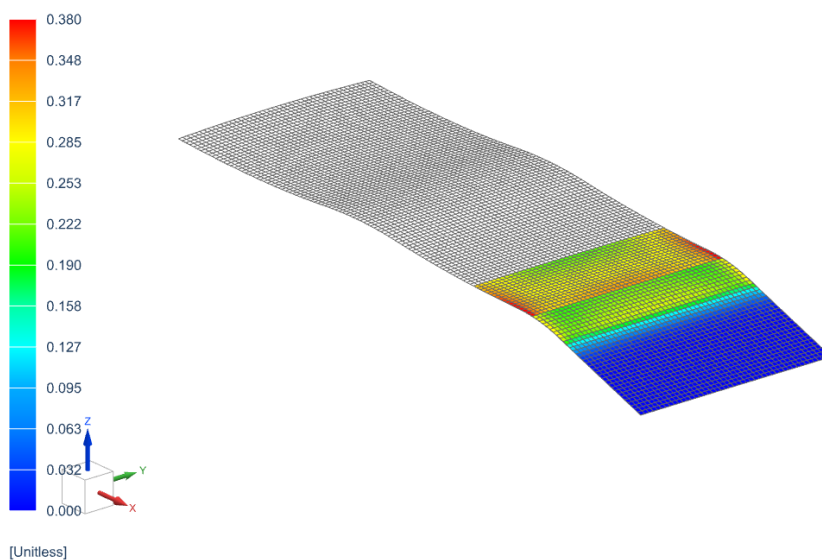


Figure A - 24: Results of FEA of the ply layup heel

The displacement results (top) and ply index results (bottom) for FEA on the heel with 800 N applied.

In the tables below, the displacement calculations and the displacements from the FEA are recorded for the heel blade (Table A - 11) and the forefoot blade (Table A - 12) as well as the three gas spring conditions (Table A - 13, Table A - 14, Table A - 15). The results were found to be highly comparable and thus verified the calculations.

Table A - 11: Calculation vs FEA for layup verification: heel

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
0	0.00	0.00	0.00	0.00
100	1.64	1.70	0.05	3.03
200	3.30	3.39	0.09	2.83
300	4.95	5.08	0.13	2.55
400	6.62	6.78	0.16	2.34
500	8.29	8.47	0.18	2.18
600	9.96	10.17	0.21	2.06
700	11.63	11.86	0.23	1.90
800	13.31	13.58	0.27	1.99
900	14.98	15.25	0.27	1.75
1000	16.65	16.95	0.30	1.75
1100	18.32	18.64	0.32	1.70
1200	19.99	20.33	0.34	1.69
1300	21.64	22.03	0.39	1.75

Table A - 12: Calculation vs FEA for layup verification: forefoot with no gas spring

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
0	0.00	0.00	0.00	0.00
100	3.06	2.95	0.11	3.63
200	6.13	5.90	0.23	3.69
300	9.21	8.85	0.36	4.02
400	12.30	11.80	0.50	4.23
500	15.40	14.75	0.65	4.39
600	18.50	17.71	0.79	4.47
700	21.61	20.66	0.95	4.61
800	24.73	23.61	1.12	4.73
900	27.84	26.56	1.28	4.83
1000	30.96	29.51	1.45	4.92
1100	34.08	32.46	1.62	4.99
1200	37.20	35.41	1.79	5.05
1300	40.31	38.36	1.95	5.09

Table A - 13: Calculation vs FEA for layup verification: forefoot with 300 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
100	0.00	0.00	0.00	0.00
200	1.93	1.96	0.02	1.24
300	4.94	4.99	0.05	1.02
400	7.96	8.11	0.15	1.85
500	10.98	11.06	0.08	0.68
600	14.02	14.01	0.01	0.07
700	17.06	16.96	0.10	0.61
800	20.17	19.91	0.26	1.28
900	23.28	22.86	0.42	1.83
1000	26.39	25.81	0.58	2.26
1100	29.51	28.76	0.75	2.61
1200	32.63	31.71	0.92	2.90
1300	35.75	34.67	1.08	3.11

Table A - 14: Calculation vs FEA for layup verification: forefoot with 700 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
400	2.51	2.54	0.03	0.00
500	5.52	5.58	0.06	1.00
600	8.54	8.61	0.07	0.81
700	11.57	12.03	0.46	3.83
800	14.61	14.98	0.37	2.50
900	17.65	17.93	0.28	1.56
1000	20.70	20.88	0.18	0.87
1100	23.75	23.83	0.08	0.32
1200	26.81	26.78	0.03	0.11
1300	29.87	29.74	0.13	0.43

Table A - 15: Calculation vs FEA for layup verification: forefoot with 1000 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
500	1.44	1.46	0.02	1.25
600	4.45	4.50	0.05	1.06
700	7.46	7.53	0.07	0.87
800	10.49	10.79	0.30	2.77
900	13.53	14.23	0.70	4.95
1000	16.57	17.19	0.62	3.63
1100	19.61	20.14	0.53	2.61
1200	22.67	23.09	0.42	1.83

A.5. Adjustment calculation verification

Below both the calculated displacements and the FEA displacements were recorded for the adjusted calculation and FEA model to match the manufactured blades. The results for the heel blade (Table A - 16), forefoot (Table A - 17), and the three gas spring cases (Table A - 18, Table A - 19, and Table A - 20) are found in the tables below. The calculations were considered verified since the deviations were minimal.

Table A - 16: Calculation vs FEA for adjustment verification: heel

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
0	0.00	0.00	0.00	0.00
100	0.96	0.95	0.01	1.44
200	1.92	1.89	0.03	1.52
300	2.88	2.84	0.05	1.72
400	3.85	3.78	0.07	1.87
500	4.82	4.73	0.09	1.98
600	5.79	5.67	0.12	2.12
700	6.76	6.62	0.15	2.24
800	7.74	7.56	0.18	2.33
900	8.71	8.51	0.21	2.43
1000	9.69	9.45	0.24	2.55
1100	10.67	10.40	0.27	2.58
1200	11.65	11.34	0.31	2.70
1300	12.62	12.29	0.33	2.72

Table A - 17: Calculation vs FEA for adjustment verification: forefoot with no gas spring

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
0	0.00	0.00	0.00	0.00
100	1.90	1.89	0.01	0.55
200	3.81	3.79	0.03	0.68
300	5.73	5.68	0.05	0.82
400	7.65	7.58	0.07	0.93
500	9.57	9.47	0.10	1.03
600	11.49	11.36	0.13	1.16
700	13.42	13.26	0.16	1.21
800	15.35	15.15	0.20	1.33
900	17.29	17.04	0.25	1.44
1000	19.22	18.94	0.28	1.48
1100	21.16	20.83	0.33	1.58
1200	23.10	22.73	0.37	1.62
1300	25.04	24.62	0.42	1.70

Table A - 18: Calculation vs FEA for adjustment verification: forefoot with 300 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
100	0.00	0.00	0.00	0.00
200	0.19	0.21	0.02	8.10
300	2.10	2.09	0.01	0.25
400	4.00	3.98	0.02	0.53
500	5.92	5.88	0.04	0.70
600	7.84	7.77	0.07	0.85
700	9.76	9.67	0.09	0.92
800	11.68	11.56	0.12	1.08
900	13.61	13.45	0.16	1.21
1000	15.54	15.35	0.19	1.27
1100	17.48	17.24	0.24	1.38
1200	19.41	19.13	0.28	1.49
1300	21.35	21.03	0.32	1.53

Table A - 19: Calculation vs FEA for adjustment verification: forefoot with 700 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
400	0.00	0.00	0.00	0.00
500	1.08	1.10	0.02	1.50
600	2.99	2.98	0.00	0.09
700	4.90	4.88	0.02	0.42
800	6.81	6.77	0.04	0.63
900	8.73	8.67	0.07	0.79
1000	10.66	10.56	0.10	0.92
1100	12.58	12.45	0.13	1.08
1200	14.51	14.35	0.16	1.14
1300	16.45	16.24	0.21	1.27

Table A - 20: Calculation vs FEA for adjustment verification: forefoot with 1000 N GSF

Force (N)	Displacement (mm)		Deviation	
	Calculation	FEA	[mm]	[%]
600	0.00	0.00	0.00	0.00
700	1.27	1.30	0.03	2.53
800	3.18	3.18	0.00	0.10
900	5.09	5.07	0.02	0.31
1000	7.01	6.97	0.04	0.55
1100	8.93	8.86	0.06	0.73
1200	10.85	10.75	0.10	0.93
1300	12.78	12.65	0.13	1.00

Appendix B : Drawings

B.1. Manufacturing Layup Drawings

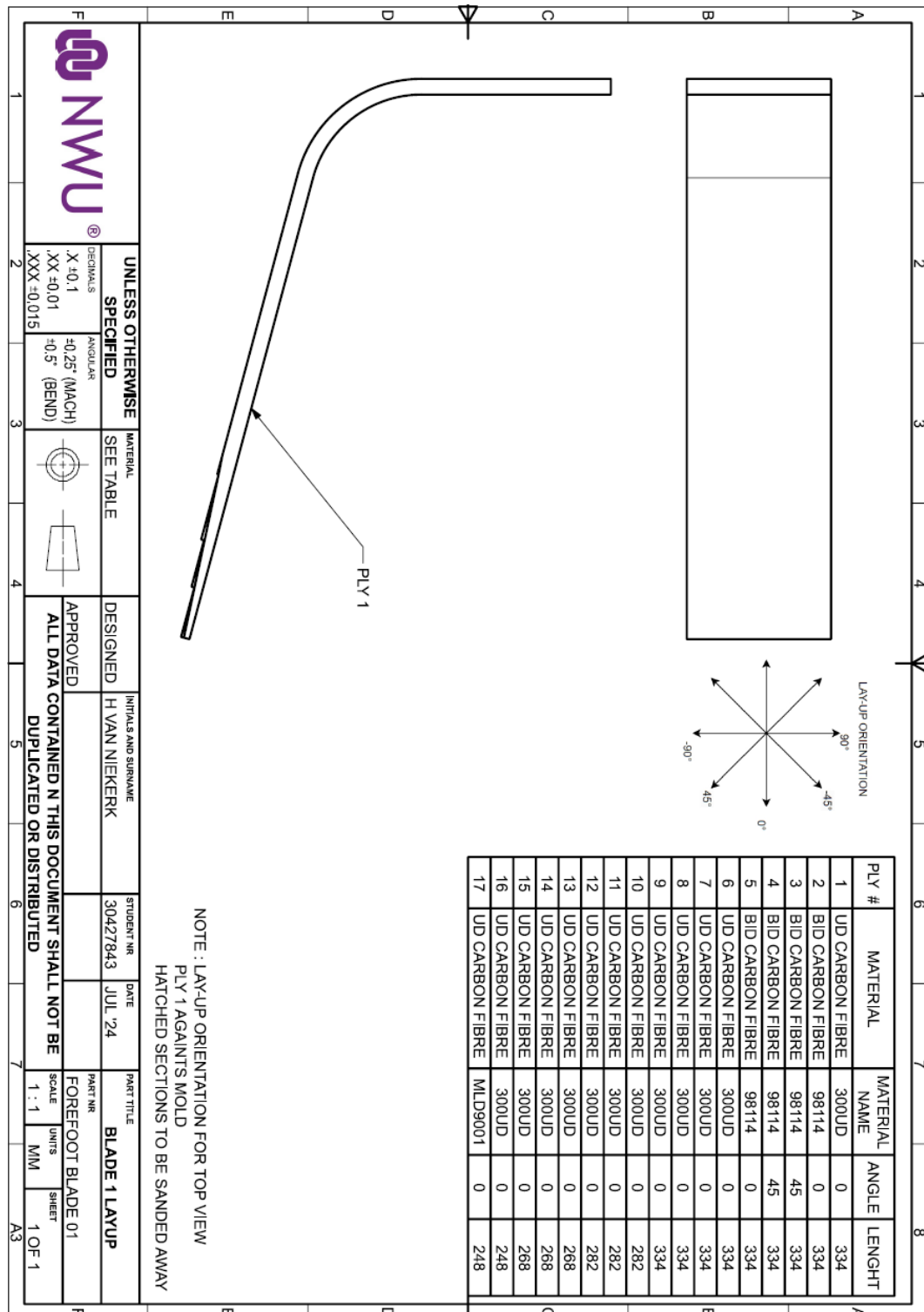


Figure B - 1: Layup drawing for primary forefoot blade

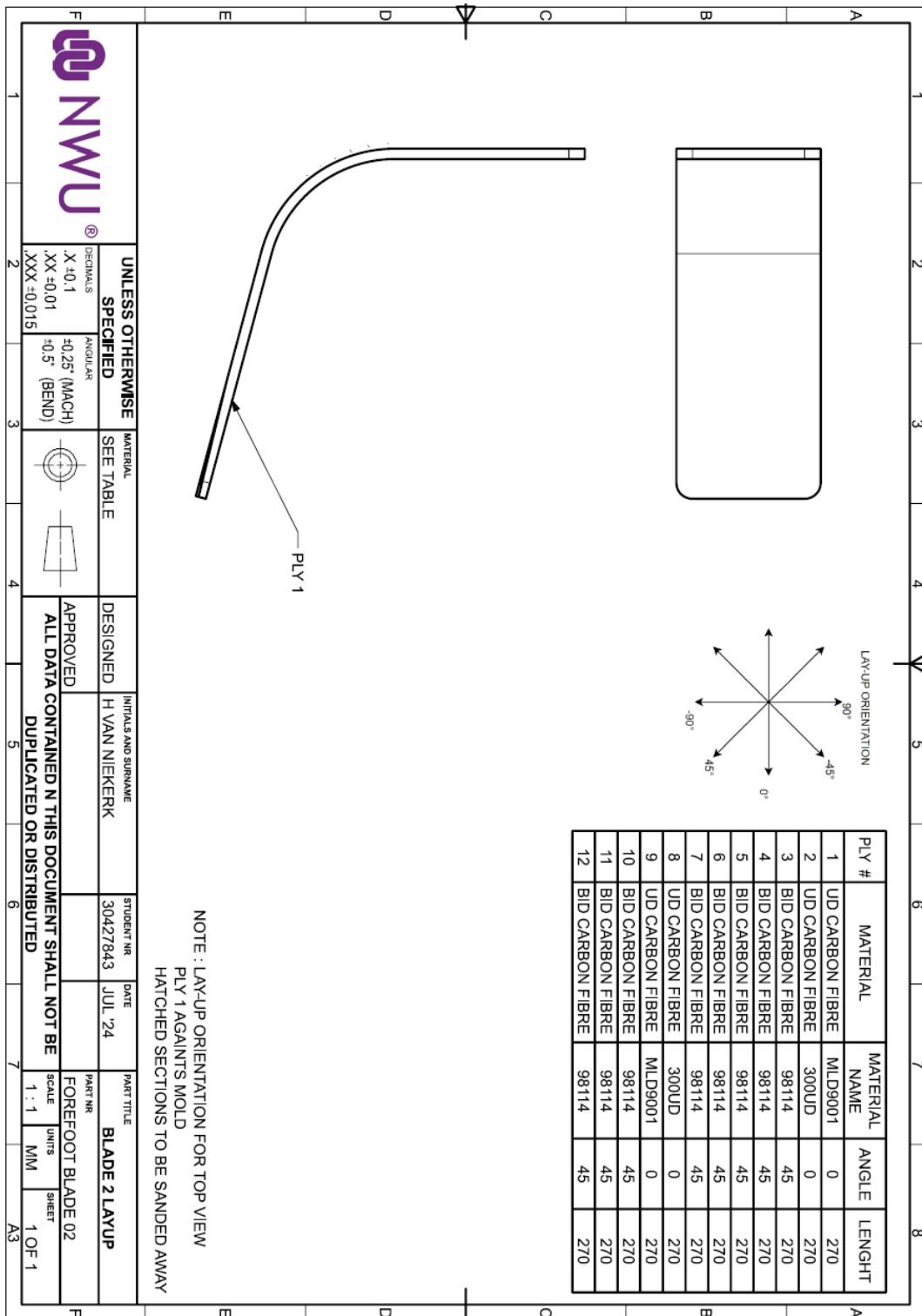


Figure B - 2: Layup drawing for secondary forefoot blade

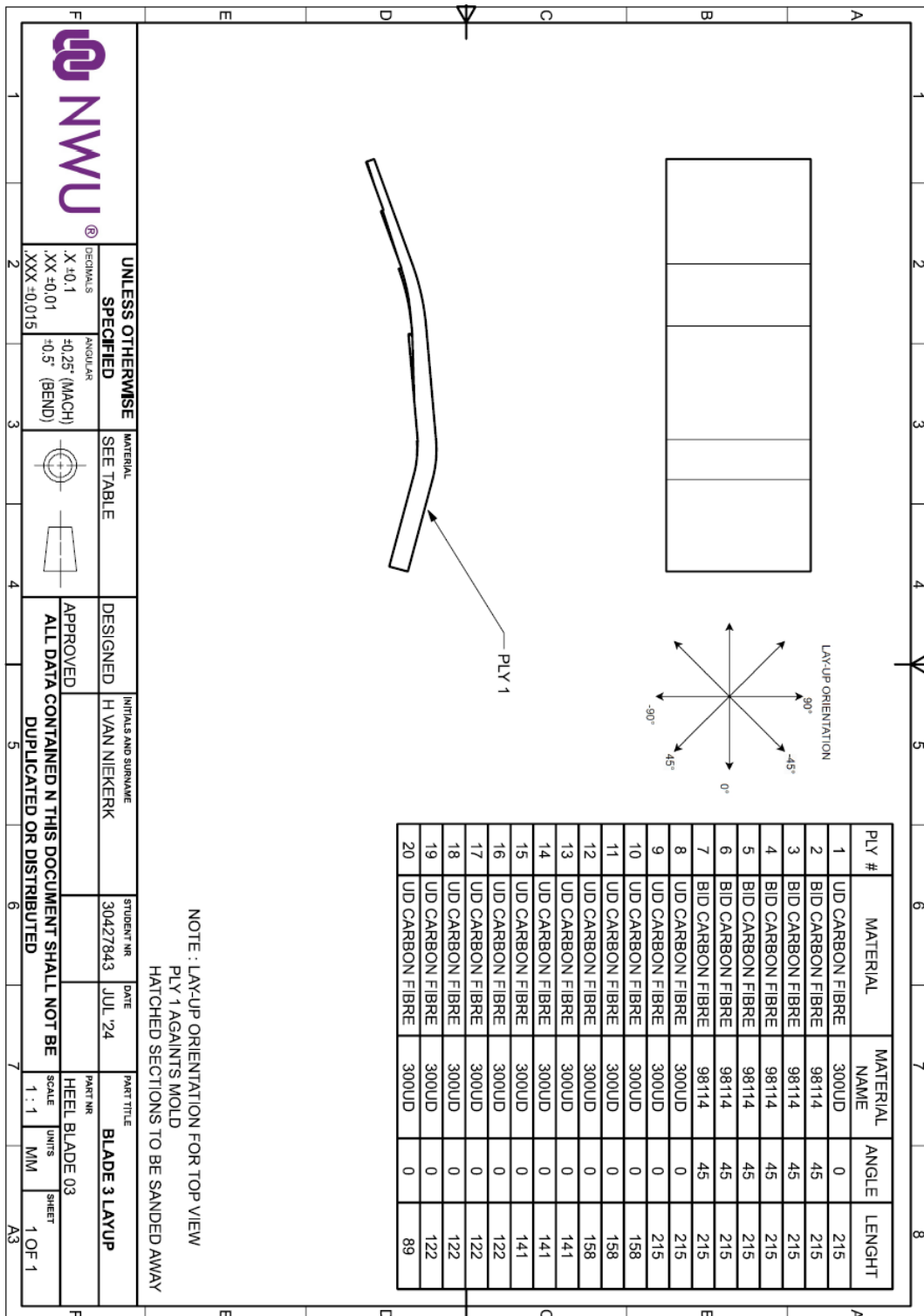


Figure B - 3: Layup drawing for heel blade

B.2. Mould Drawings

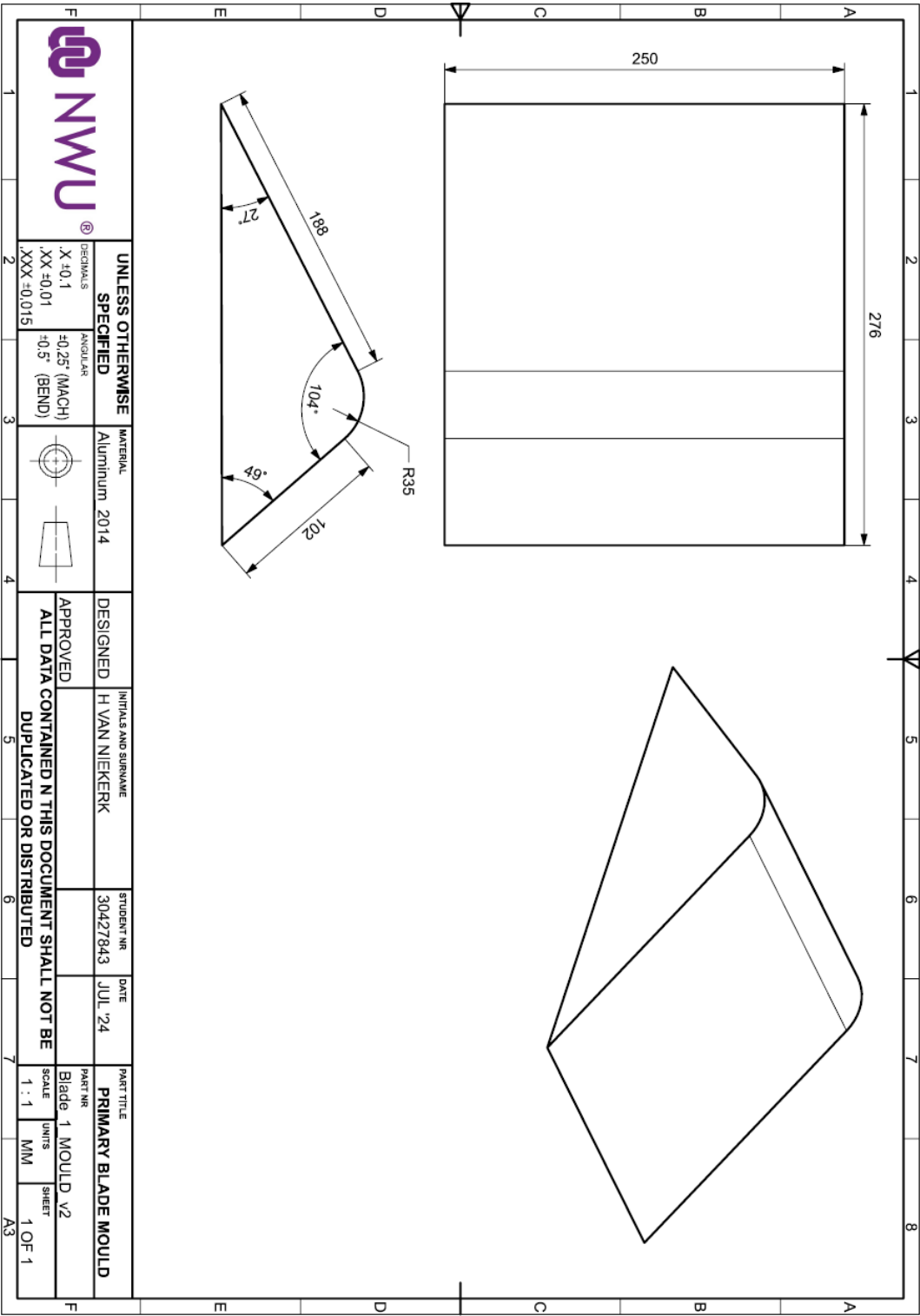


Figure B - 4: Mould drawing for primary forefoot blade

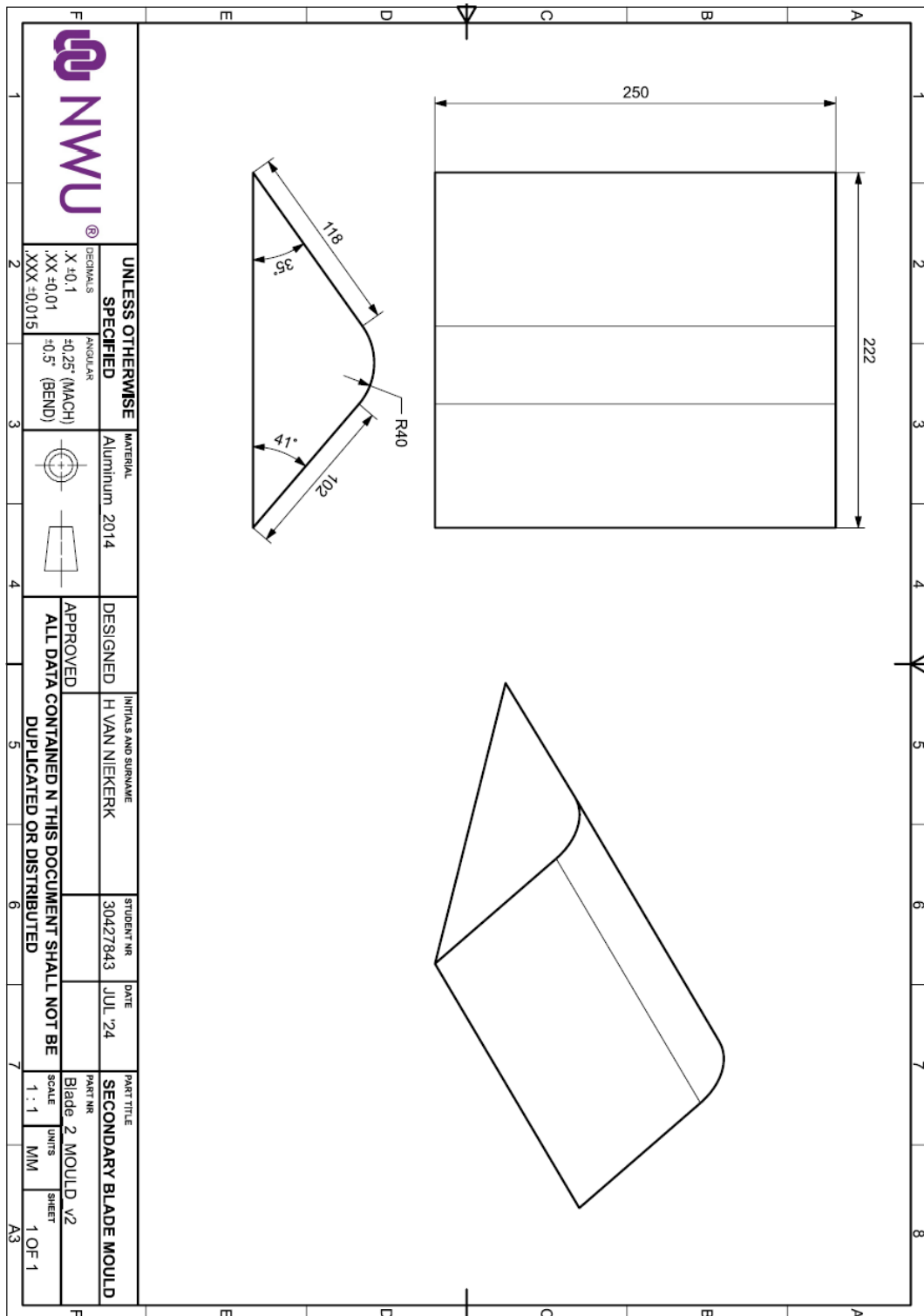


Figure B - 5: Mould drawing for secondary forefoot blade

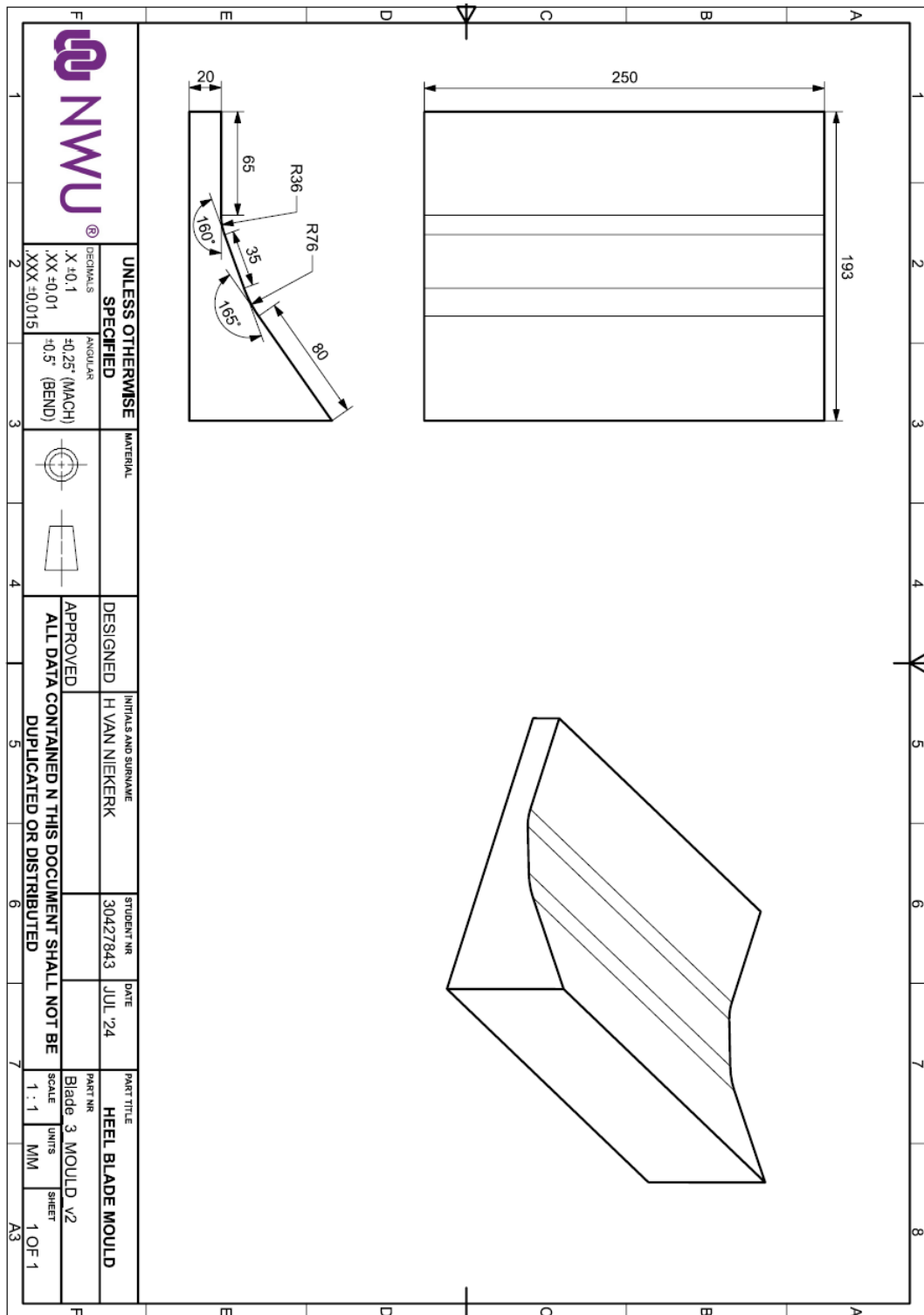


Figure B - 6: Mould drawing for heel blade

Appendix C : Graphs and Tables of Results

C.1.Forefoot: No Gas Springs

C.1.1. Experiment Results

Below are the experiment results for the forefoot blade without gas springs. Table C - 1 has the recorded data from the experiment runs at every millimetre along with the standard deviation over the three runs. Figure C - 1 illustrates the data from Table C - 1. Table C - 2 and Table C - 3 presents the maximum deviation from the average data from each millimetre increment for all three runs.

Table C - 1: Force at displacement of experiment runs: forefoot with no GSF

Displacement (mm)	Force (N)			Standard Deviation
	Forefoot 0N R01	Forefoot 0N R02	Forefoot 0N R03	
0	0	0	0	0
1	54.12	54.32	52.72	0.87
2	104.49	103.94	102.81	0.86
3	156.88	158.41	155.17	1.62
4	210.33	209.82	205.63	2.58
5	256.15	253.06	252.68	1.9
6	304.64	301.12	303.11	1.76
7	355.82	353.15	351.63	2.12
8	422.51	422.01	421.09	0.72
9	482.86	465.91	473.07	8.51
10	531.27	524.42	526.54	3.51
11	586.09	573.79	585.43	6.92
12	625.71	617.52	622.51	4.13
13	662.7	663.97	659.92	2.07
14	719.46	716.5	709.25	5.25
15	787.69	795.2	791.87	3.77

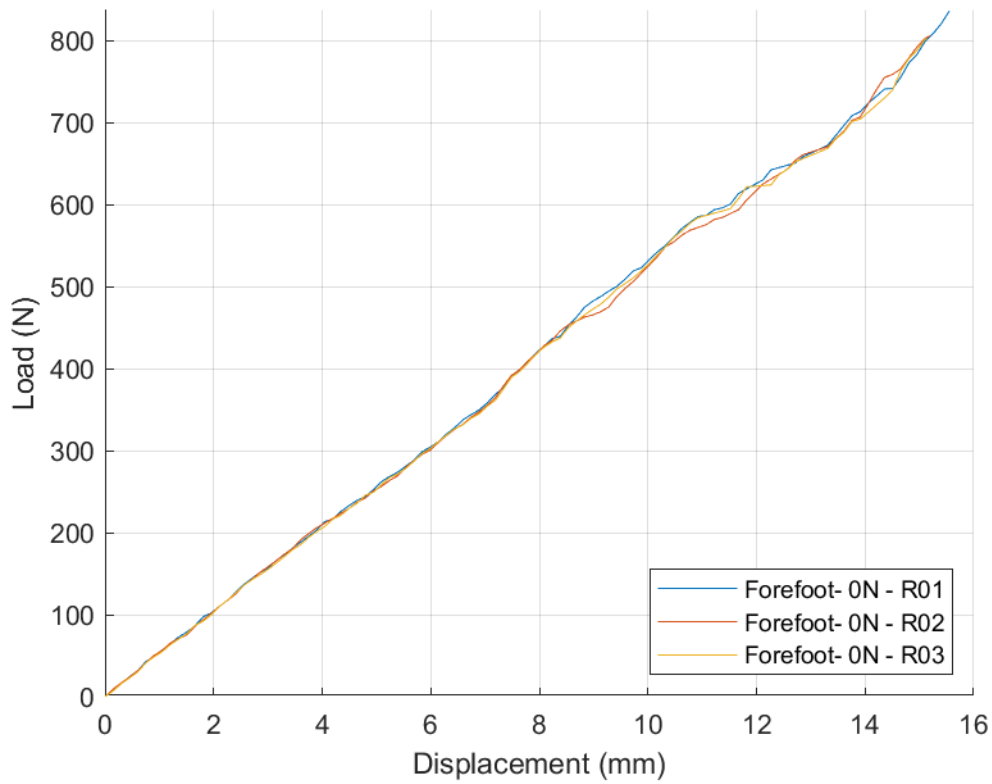


Figure C - 1: Results of the experiment tests: forefoot with no GSF

Table C - 2: Deviation from average: forefoot with no GSF

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
1	0.75	42.83	40.44	2.39
	1.80	97.34	93.94	3.40
	2.55	136.55	133.08	3.47
	3.15	166.08	164.65	1.43
	4.05	213.79	210.80	2.99
	5.10	262.38	258.62	3.76
	6.59	337.44	332.83	4.62
	7.49	390.02	386.72	3.30
	8.83	474.69	465.11	9.58
	9.13	487.94	480.92	7.02
	10.62	571.07	561.36	9.71
	11.67	613.72	608.68	5.05
	12.27	642.34	632.68	9.67
	13.31	672.47	678.74	6.27
	14.51	741.51	754.06	12.55

Table C - 3: Deviation from average: forefoot with no GSF - Continued

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
2	0.90	50.46	48.49	1.97
	1.50	75.05	78.88	3.83
	2.55	135.54	133.04	2.49
	3.60	191.08	187.81	3.26
	4.80	242.32	244.71	2.39
	5.39	269.49	273.22	3.74
	6.89	347.04	347.88	0.84
	7.48	391.71	386.65	5.06
	8.98	465.37	472.90	7.53
	9.28	475.12	488.87	13.75
	10.92	572.07	577.53	5.45
	11.67	593.60	608.63	15.03
	12.86	661.08	656.71	4.36
	13.31	670.59	678.70	8.11
	14.36	754.92	742.37	12.54
3	0.60	30.18	32.36	2.18
	1.50	76.67	78.90	2.23
	2.55	135.53	133.04	2.48
	3.60	185.34	187.97	2.63
	4.05	207.57	210.74	3.17
	5.54	278.56	280.60	2.05
	6.89	344.71	347.87	3.16
	7.19	361.20	366.28	5.08
	8.38	437.23	441.73	4.50
	9.13	478.46	480.86	2.41
	10.77	577.05	569.43	7.62
	11.52	595.15	602.65	7.51
	12.27	623.96	632.68	8.72
	13.31	668.52	678.73	10.21
	14.51	740.23	754.10	13.87

C.1.2. Comparison Plots and Tables

The calculation values were compared to the experiment average values, as illustrated in Figure C - 2. Table C - 4 captures the plotted data and presents the deviation values in both millimetres and percentages.

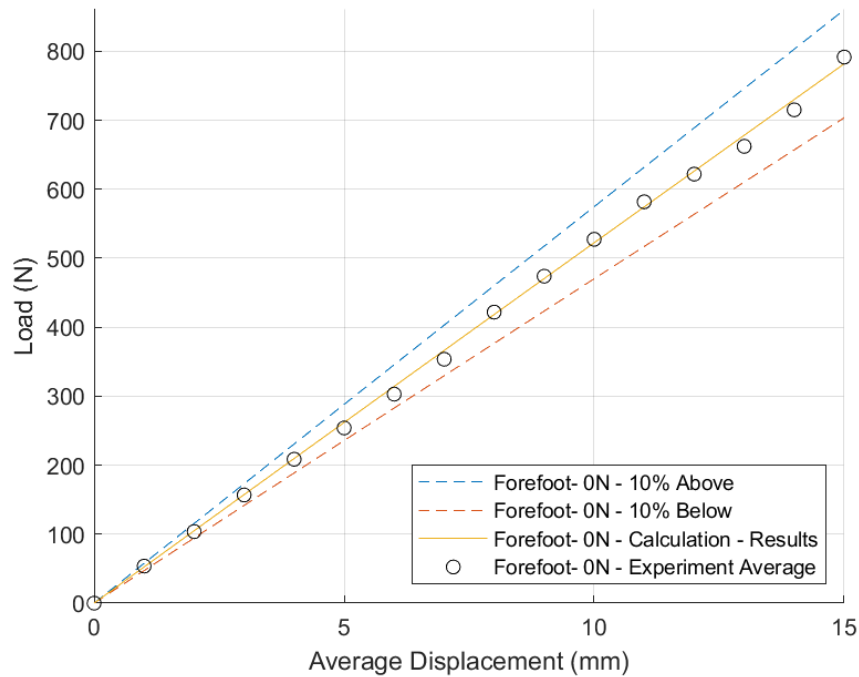


Figure C - 2: Calculations and experiment results: forefoot with no GSF

Table C - 4: Calculations and experiment deviation values: forefoot with no GSF

Displacement (mm)	Calculations (N)	Experiment (N)	Deviation (N)	Deviation (%)
1	52.51	53.72	1.21	2.31
2	105.00	103.75	1.25	1.19
3	157.38	156.82	0.56	0.35
4	209.73	208.59	1.14	0.54
5	261.98	253.96	8.02	3.06
6	314.21	302.96	11.25	3.58
7	366.35	353.53	12.82	3.50
8	418.45	421.87	3.42	0.82
9	470.49	473.94	3.45	0.73
10	522.49	527.41	4.92	0.94
11	574.43	581.77	7.34	1.28
12	626.33	621.91	4.42	0.71
13	678.22	662.2	16.02	2.36
14	730.01	715.07	14.94	2.05
15	781.81	791.58	9.77	1.25

C.2. Forefoot: 300 N GSF

C.2.1. Experiment Results

The experiment results for the forefoot blade with two 150 N gas springs (300 N GSF) attached are presented in this section. Table C - 5 presents the experiment run data at every millimetre along with the standard deviation over the three runs. This data is illustrated in Figure C - 3. Table C - 6 presents the maximum deviation from the average data from each millimetre increment for all three runs.

Table C - 5: Force at displacement of experiment runs: forefoot with 300 N GSF

Displacement (mm)	Force (N)			Standard Deviation
	Forefoot 0N R01	Forefoot 0N R02	Forefoot 0N R03	
0	210.20	201.34	192.47	8.87
1	253.55	247.91	254.12	3.44
2	302.34	301.33	309.96	4.72
3	360.21	357.37	364.07	3.36
4	416.16	416.23	421.93	3.31
5	458.21	456.51	460.97	2.25
6	502.74	498.57	508.42	4.94
7	537.96	537.73	545.20	4.25
8	569.92	573.51	587.58	9.34
9	611.12	618.04	631.73	10.49
10	655.08	659.47	674.03	9.92
11	699.89	700.64	721.61	12.33
12	743.98	752.29	766.64	11.46
13	777.94	784.46	803.98	13.55
14	823.37	831.87	NaN	NaN

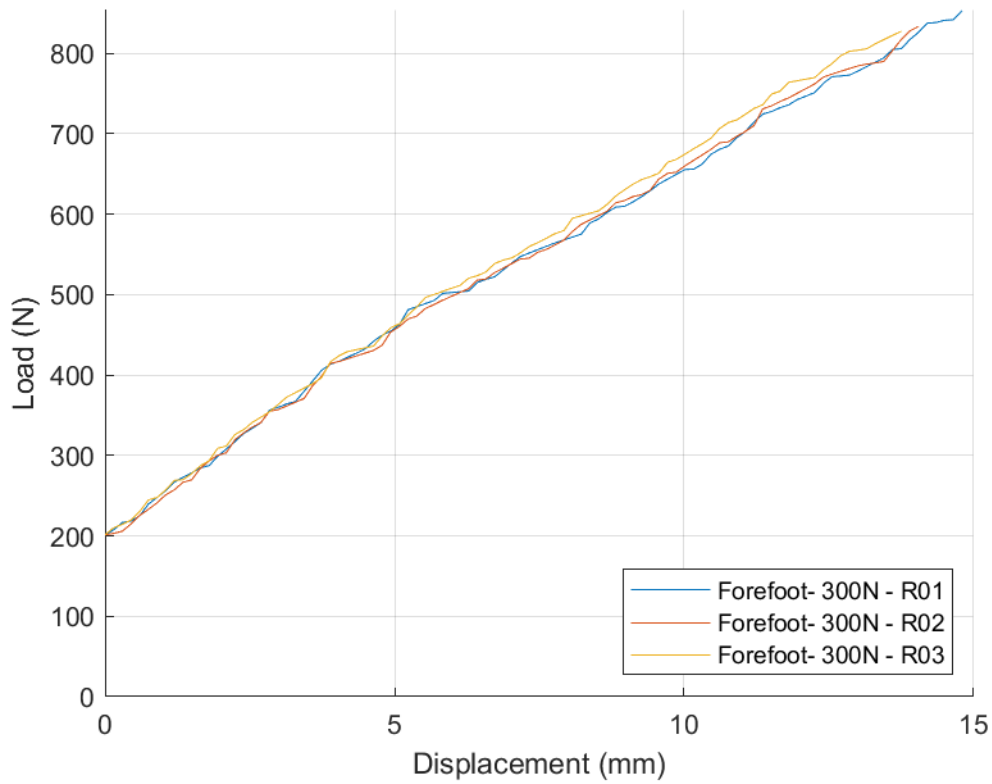


Figure C - 3: Results of the experiment tests: forefoot with 300 N GSF

Table C - 6: Deviation from average: forefoot with 300 N GSF

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
1	0.60	225.19	231.66	6.47
	1.80	287.48	293.86	6.39
	2.84	356.71	351.81	4.90
	3.29	367.20	377.24	10.04
	4.49	432.17	437.81	5.64
	5.23	481.25	469.03	12.22
	6.28	504.50	513.55	9.05
	7.77	564.45	568.65	4.20
	8.22	575.31	586.71	11.40
	9.12	615.47	625.32	9.86
	10.31	662.53	676.83	14.29
	11.96	742.75	752.30	9.55
	12.26	751.42	763.12	11.70
	13.75	805.83	817.90	12.07

Table C - 7: Deviation from average: forefoot with 300 N GSF - Continued

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
2	0.30	205.78	216.38	10.59
	1.49	269.61	277.91	8.30
	2.09	303.29	309.75	6.45
	3.44	370.83	385.74	14.91
	4.63	430.61	443.79	13.19
	5.83	493.06	495.60	2.53
	6.28	507.44	513.45	6.02
	7.62	556.27	563.13	6.86
	8.67	603.56	605.88	2.32
	9.41	629.87	637.94	8.07
	10.76	690.01	696.69	6.69
	11.06	703.03	710.10	7.07
	12.40	770.43	768.25	2.18
	13.45	790.15	806.29	16.15
3	0.75	244.60	239.17	5.42
	1.95	308.94	301.68	7.27
	2.24	326.17	318.22	7.95
	3.74	396.68	403.07	6.39
	4.64	436.00	443.83	7.83
	5.53	496.45	482.36	14.09
	6.73	538.69	530.20	8.49
	7.33	559.82	552.25	7.58
	8.07	594.98	580.11	14.87
	9.72	664.74	650.79	13.95
	10.76	713.91	696.75	17.16
	11.81	764.07	745.27	18.80
	12.85	802.72	783.77	18.96
	13.00	804.00	788.89	15.11

C.2.2. Comparison Plots and Tables

Figure C - 4 illustrates the experiment average values and the calculation values for comparison purposes. This data is captured in Table C - 8 along with the deviation values in both millimetres and percentage.

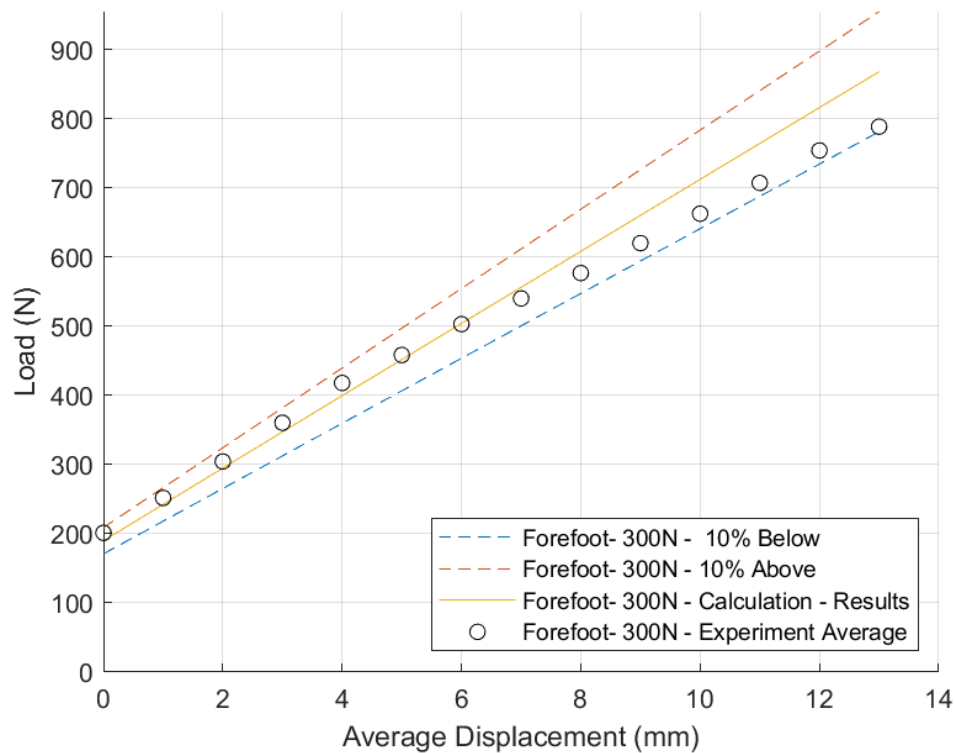


Figure C - 4: Calculations and experiment results: forefoot with 300 N GSF

Table C - 8: Calculations and experiment deviation values: forefoot with 300 N GSF

Displacement (mm)	Calculations (N)	Experiment (N)	Deviation (N)	Deviation (%)
0	190.01	201.34	11.33	5.96
1	242.51	251.86	9.35	3.86
2	295.00	304.55	9.55	3.24
3	347.38	360.55	13.17	3.79
4	399.74	418.11	18.37	4.59
5	451.99	458.56	6.57	1.45
6	504.22	503.24	0.98	0.19
7	556.35	540.3	16.05	2.88
8	608.46	577	31.46	5.17
9	660.49	620.3	40.19	6.09
10	712.50	662.86	49.64	6.97
11	764.44	707.38	57.06	7.46
12	816.34	754.3	62.04	7.60
13	868.24	788.79	79.45	9.15

C.3.Forefoot: 700 N Gas Springs

C.3.1. Experiment Results

The experimental data for when two 350 N gas springs (700 N GSF) were attached to the prosthetic foot is given in this section. The displacement data at every millimetre is recorded in Table C - 9 along with the standard deviation over the three runs. Figure C - 5 illustrates the displacement and force curves for the three runs. The maximum deviation for every millimetre increment for each run is recorded in Table C - 10.

Table C - 9: Force at displacement of experiment runs: forefoot with 700 N GSF

Displacement (mm)	Force (N)			Standard Deviation
	Forefoot 0N R01	Forefoot 0N R02	Forefoot 0N R03	
0	449.36	459.86	454.61	5.25
1	530.42	515.01	526.08	7.95
2	594.85	596.14	588.13	4.30
3	625.76	632.69	641.99	8.14
4	673.66	666.52	682.78	8.15
5	754.80	743.85	746.24	5.76
6	822.10	827.46	816.68	5.39
7	848.71	NaN	NaN	NaN

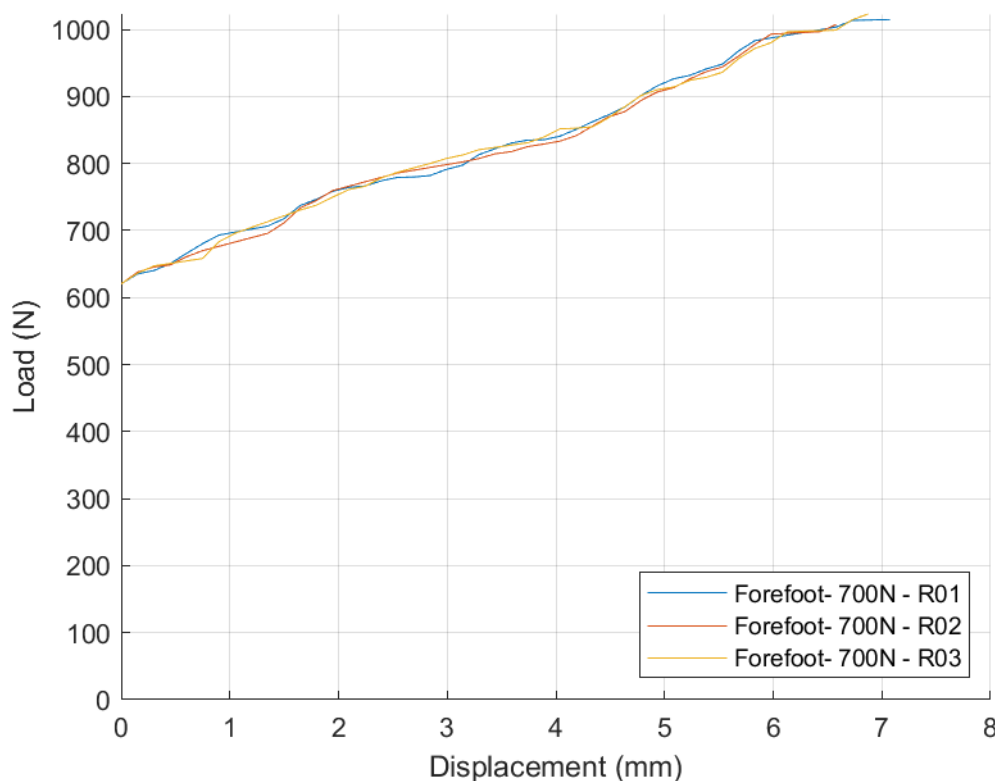


Figure C - 5: Results of the experiment tests: forefoot with 700 N GSF

Table C - 10: Deviation from average: forefoot with 700 N GSF

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
1	0.90	527.38	516.81	10.56
	1.35	540.80	547.80	7.00
	2.84	616.26	627.07	10.80
	3.14	631.77	639.24	7.47
	4.94	750.32	743.55	6.77
	5.83	817.49	809.57	7.91
	6.73	847.72	841.45	6.27
2	0.15	472.09	465.00	7.09
	1.35	530.08	547.99	17.91
	2.54	620.10	615.02	5.08
	3.89	663.25	669.78	6.53
	4.19	675.78	688.22	12.44
	5.53	778.43	787.68	9.25
	6.58	841.57	837.49	4.08
3	0.75	492.50	506.45	13.95
	1.79	572.02	578.85	6.83
	2.99	641.70	633.14	8.56
	3.29	654.83	645.40	9.43
	4.34	689.06	699.41	10.35
	5.53	770.38	787.60	17.21
	6.87	857.32	845.38	11.95

C.3.2. Comparison Plots and Tables

In the following section the calculation values and experiment average values are illustrated (Figure C - 6) and captured (Table C - 11). The deviations between the two data sets are presented in Table C - 11.

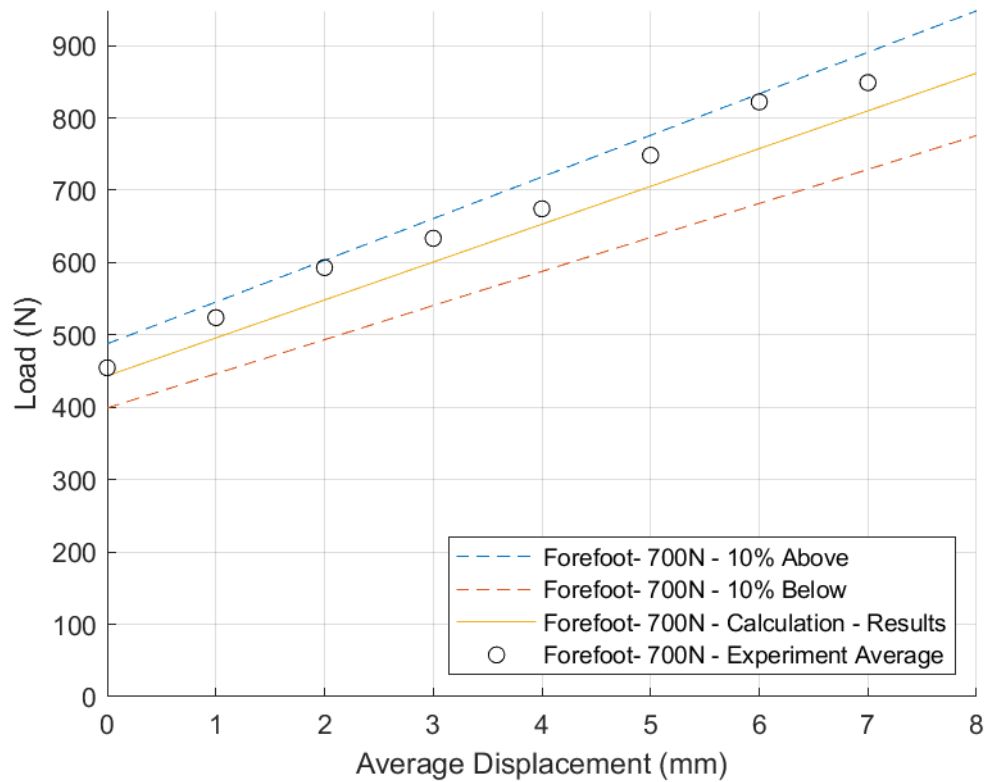


Figure C - 6: Calculations and experiment results: forefoot with 700 N GSF

Table C - 11: Calculations and experiment deviation values: forefoot with 700 N GSF

Displacement (mm)	Calculations (N)	Experiment (N)	Deviation (N)	Deviation (%)
0	443.45	454.61	11.16	2.52
1	495.88	523.83	27.95	5.64
2	548.31	593.04	44.73	8.16
3	600.74	633.48	32.74	5.45
4	653.05	674.32	21.27	3.26
5	705.34	748.3	42.96	6.09
6	757.54	822.08	64.54	8.52
7	809.71	848.71	39.00	4.82
8	861.79	NaN	NaN	NaN

C.4.Forefoot: 1000 N Gas Springs

C.4.1. Experiment Results

The recorded experiment data for the experiment with two 500 N gas springs (1000 N GSF) attached is captured in Table C - 12 and illustrated in Figure C - 7. The deviation from the average of the three runs is captured in Table C - 13.

Table C - 12: Force at displacement of experiment runs: forefoot with 1000 N GSF

Displacement (mm)	Force (N)			Standard Deviation
	Forefoot 0N R01	Forefoot 0N R02	Forefoot 0N R03	
0	642.64	644.15	641.13	1.51
1	705.62	697.14	715.91	9.40
2	782.16	750.12	779.23	17.72
3	826.10	828.41	841.47	8.29
4	879.51	873.55	894.63	10.86
5	929.08	929.23	950.94	12.58
6	NaN	1010.07	1043.75	NaN

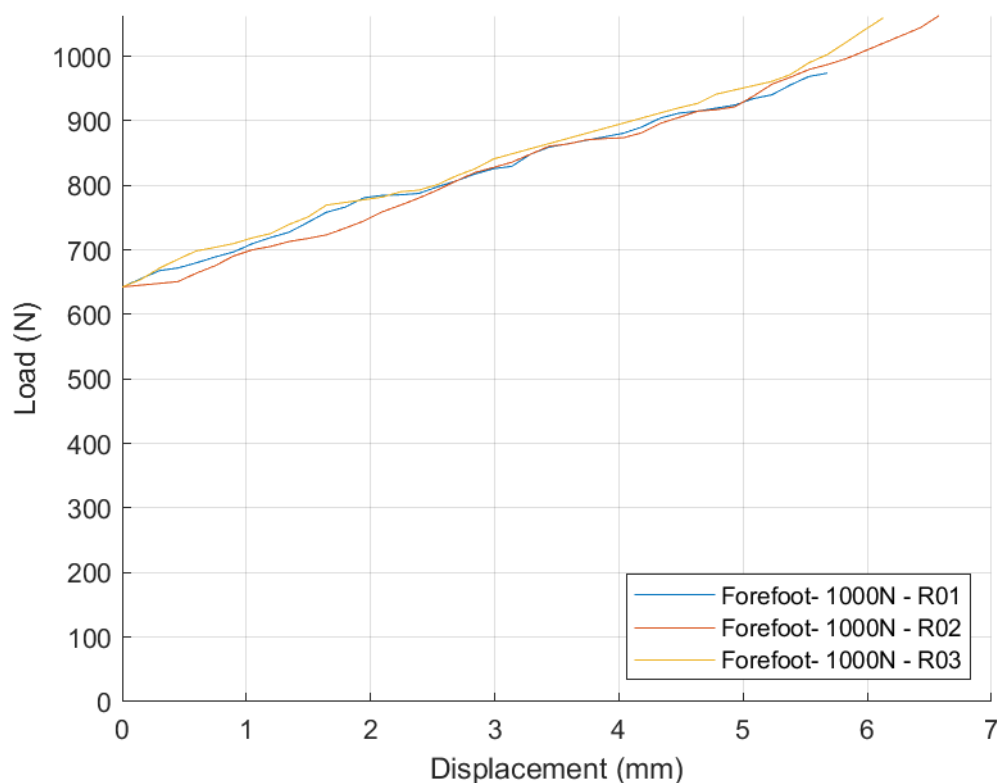


Figure C - 7: Results of the experiment tests: forefoot with 1000 N GSF

Table C - 13: Deviation from average: forefoot with 1000 N GSF

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
1	0.30	667.97	661.68	6.29
	1.94	780.69	766.84	13.85
	2.09	784.49	776.05	8.44
	3.14	829.62	838.96	9.34
	4.93	924.50	932.74	8.24
	5.68	974.01	997.82	23.81
2	0.45	650.90	671.12	20.21
	1.65	723.40	747.74	24.35
	2.09	759.02	776.22	17.20
	3.44	860.85	854.16	6.68
	4.93	921.77	932.81	11.04
	5.83	996.23	1011.31	15.09
3	0.60	698.50	680.56	17.94
	1.64	769.46	747.59	21.88
	2.99	840.89	831.34	9.56
	3.89	888.63	876.84	11.79
	4.78	941.30	924.64	16.65
	5.98	1040.96	1024.87	16.08

C.4.2. Comparison Plots and Tables

There was some deviation between the calculation values and experiment average values, as can be seen in Figure C - 8. These deviation values are captured in Table C - 14.

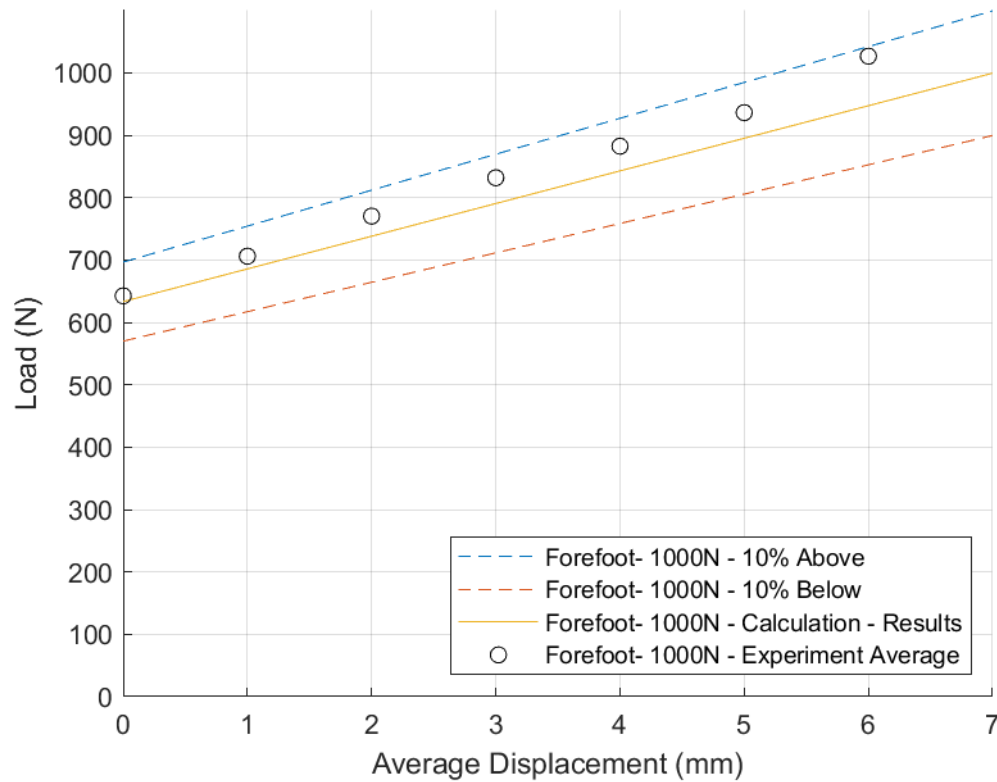


Figure C - 8: Calculations and experiment results: forefoot with 1000 N GSF

Table C - 14: Calculations and experiment deviation values: forefoot with 1000 N GSF

Displacement (mm)	Calculations (N)	Experiment (N)	Deviation (N)	Deviation (%)
0	633.47	642.64	9.17	1.45
1	685.89	706.22	20.33	2.96
2	738.32	770.5	32.18	4.36
3	790.74	832	41.26	5.22
4	843.05	882.56	39.51	4.69
5	895.35	936.41	41.06	4.59
6	947.54	1026.91	79.37	8.38
7	999.72	NaN	NaN	NaN

C.5. Heel

C.5.1. Experiment Results

For the heel blade experiment the experiment, results of the three runs are captured in Table C - 15. This data is illustrated in Figure C - 9.

Table C - 16 presents the largest deviation in each millimetre interval for all three runs.

Table C - 15: Force at displacement of experiment runs: heel

Displacement (mm)	Force (N)			Standard Deviation
	Forefoot 0N R01	Forefoot 0N R02	Forefoot 0N R03	
0	0.00	0.00	0.00	0.00
1	111.86	110.37	106.99	2.49
2	228.82	224.06	223.63	2.88
3	361.91	355.28	355.38	3.80
4	462.55	451.44	478.04	13.36
5	565.32	565.12	576.73	6.64
6	678.93	676.00	679.74	1.97
7	NaN	781.54	774.96	NaN

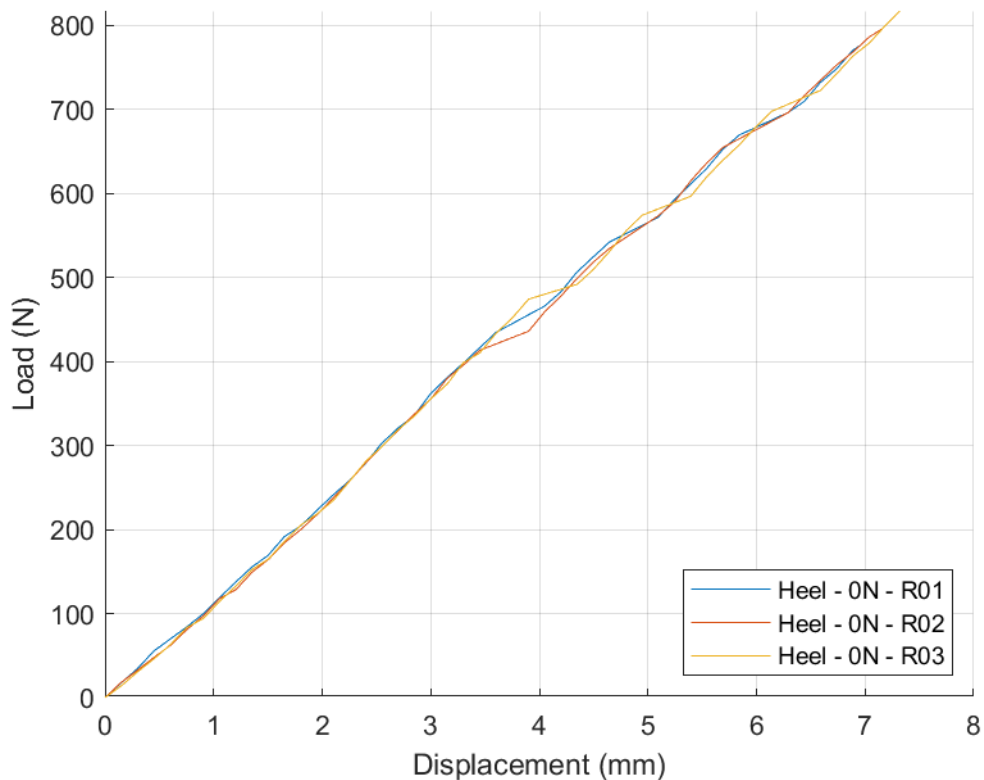


Figure C - 9: Results of the experiment tests: heel

Table C - 16: Deviation from average: heel

Run Number	Displacement at Max Deviation (mm)	Force at Max Deviation (N)	Force of the average Data (N)	Deviation (N)
1	0.45	55.38	49.26	6.12
	1.65	191.50	185.16	6.33
	2.55	302.94	298.21	4.73
	3.60	434.45	421.32	13.14
	4.65	542.04	531.77	10.27
	5.84	669.38	660.77	8.61
	6.44	709.39	722.14	12.74
2	0.60	62.44	66.37	3.93
	1.20	128.47	133.46	4.99
	2.10	237.50	239.01	1.51
	3.90	436.00	453.42	17.42
	4.05	458.99	469.12	10.13
	5.69	654.69	644.58	10.11
	6.29	696.15	707.38	11.22
3	0.91	94.13	99.36	5.24
	1.50	164.84	168.15	3.31
	2.10	234.92	239.35	4.43
	3.90	474.08	453.40	20.68
	4.95	573.99	563.35	10.64
	5.39	596.57	612.08	15.51
	6.59	722.63	737.13	14.50

C.5.2. Comparison Results

The calculation values for the heel blade experiment as well as the experiment average values are recorded and illustrated in Table C - 17 and Figure C - 10 respectively.

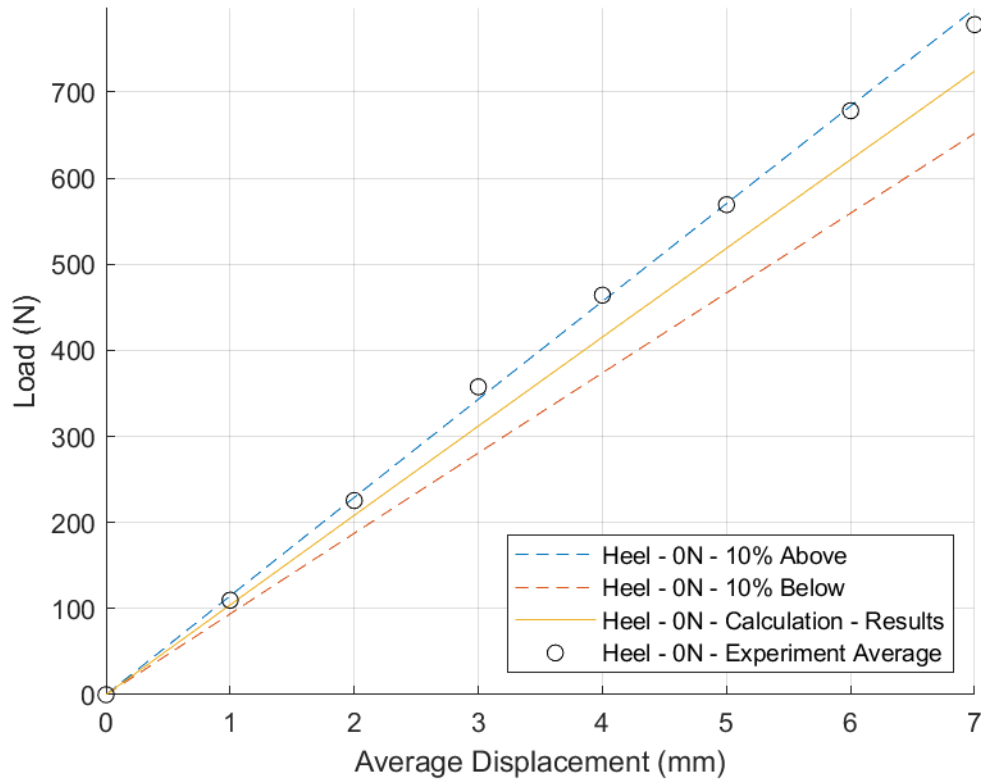


Figure C - 10: Calculations and experiment results: heel

Table C - 17: Calculations and experiment deviation values: heel

Displacement (mm)	Calculations (N)	Experiment (N)	Deviation (N)	Deviation (%)
1	104.32	109.74	5.42	5.20
2	208.29	225.5	17.21	8.26
3	311.96	357.52	45.56	14.61
4	415.34	464.01	48.67	11.72
5	518.49	569.06	50.57	9.75
6	621.42	678.22	56.80	9.14
7	724.16	778.25	54.09	7.47

Appendix D : Cost Evaluation

The following appendix breaks down the cost estimation done in Chapter 4. Unless the costs were publicly available, like on websites, the company that issued the quote was kept anonymous.

Firstly, the material costs of the bidirectional and unidirectional carbon fibres were calculated. The breakdowns are found in Table D - 1 and Table D - 2, respectively. The costs per metre used in the breakdown were from the ATM Composites website. Screenshots of the products can be seen in Figure D - 1 and Figure D - 2.

The cost per square metre was then calculated, and the cost for the required material was calculated. The required material was calculated using the lengths from the layup schedule in Appendix B and a width of 85 mm for a single foot blade. The additional 20 mm tabs on the sides of the 45 mm blade were added to ensure the blade can be fished properly. The calculated weight was used to determine how much resin will be required.

The total cost for the two fibres sums up to R 1226.50 for a single foot.

Table D - 1: Bi-directional carbon fibre cost breakdown

Bi- Directional Cost	526.70	R/m
Bi- Directional Weight	200.00	g/m ²
Bi- Directional Calculated	526.70	R/m ²
Bi- Directional Required	0.41	m ²
Weight Required	81.36	g
Bi- Directional Cost	214.27	R



Carbon Fabric 200g/M² (Plain Weave) 100cm

R526.70 incl VAT

All prices shown include VAT

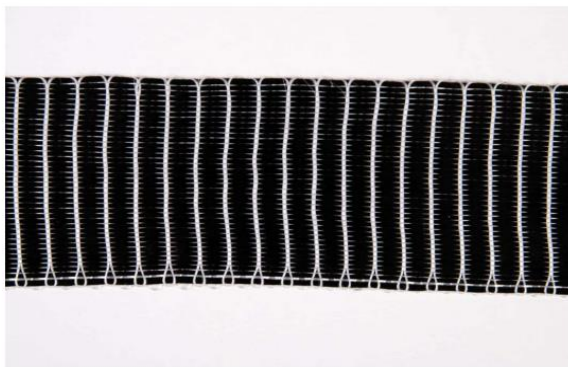
The 200 grams per square meter plain weave fabric is suitable for use in wet-lay, vacuum bagging and resin infusion manufacture. Each ply of this carbon fabric, wet out with an equal mass of resin will be approximately 0.22MM in thickness.

Availability: 659 in stock

Figure D - 1: Bi-directional carbon fibre cost

Table D - 2: Unidirectional carbon fibre cost breakdown

Unidirectional Cost	86.02	R/m
Unidirectional Weight	343.00	g/m ²
Unidirectional Calculated	1720.40	R/m ²
Unidirectional Required	0.59	m ²
Weight Required	201.81	g
Unidirectional Cost	1012.23	R



Carbon Fabric Tape 343g/M² (Unidirectional) 50mm

R86.02 incl VAT

All prices shown include VAT

AMT Composites stock a wide range of woven and unidirectional carbon tapes, and are sold per linear meter.

Availability: 1907 in stock

Figure D - 2: Unidirectional carbon fibre cost

The weights of the fibres were used in the resin cost breakdown in Table D - 3. The amount of resin used, in kilograms, is equal to the weight of the composite materials. The resin and hardener are mixed in the ratio 100:27, and therefore less hardener is required. The unit prices for the resin and hardener were estimated from a quote received. An extract from the quote is seen in Figure D - 3. From the breakdown, the total cost of the resin mixture would be about R90.

Table D - 3: Resin cost breakdown

Composite Weight	0.28	kg
Resin Price	238.44	R/kg
Hardener Price	289.94	R/kg
Resin Required	0.28	kg
Hardner Required	0.08	kg
Resin Cost	67.52	R
Hardener Cost	22.17	R
Total Resin Matrix Cost	89.69	R

Description	Qty	Unit of Measure	Unit Price
Sicomin GreenPoxy 33 Resin - Laminating	1.00	KG	238.44
Sicomin SD 4775 Laminating Medium Hardener	0.27	KG	289.94

Figure D - 3: Extract from quote for resin and hardner.

The moulds on which the composite blades are built are cut from supawood. The material cost for a 16 mm thick sheet of supawood can be seen in the screenshot from Builder's Warehouse's website in Figure D - 4. The cost breakdown for the moulds, Table D - 4, was for a single blade. Although no official hourly rate for CNC cutting or workshop rates could be found, multiple forums suggested that R375-R450 could be expected. In order to be conservative with this estimation, an hourly rate of R500 was used in the breakdown. Since the moulds are quite small, the estimated labour time was one hour, and the total for the moulds was estimated to be R559.68.

Table D - 4: Mould costs breakdown

Supawood price	899.00	R
Supawood price	178.64	R/m ²
Required for moulds	0.33	m ²
Mould wood cost	59.68	R
Wood Cut rate	500.00	R/h
Hours estimated	1.00	h
Mould manufacturing cost	500.00	R
Total Mould cost	559.68	R

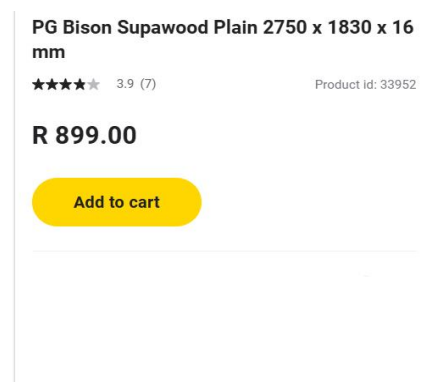


Figure D - 4: Cost of Supawood material

The manufacturing of the blades was estimated the same hourly rate as that of the mould manufacturing to remain conservative with the cost calculation, as seen in the breakdown in Table D - 5. Four hours for the blade manufacturing and the finishing were estimated for a single foot. This includes the layout procedure, glueing of the blades, appropriate drilling and sanding, and final assembly. The total estimation for the manufacturing was R4 000.

Table D - 5: Manufacturing cost breakdown

Composite build rate	500.00	R/h
Hours estimated	4.00	h
Blade manufacturing cost	2000.00	R
Composite Finish rate	500.00	R/h
Hours estimated	4.00	h
Blade finishing cost	2000.00	R
Total blade manufacture costs	4000.00	R

For the pyramid adapter, a quote was received, to manufacture a similar adapter (Figure D - 5), was used. The pyramid adapter for the foot would have a bit more material and cutting work. Therefore, an additional R200 is added to the cost estimation of the pyramid adapter, making the estimated total for the pyramid adapter R1000.

Stock code	Quantity	Unit price	V.A.T.	Gross amount
PYRAMID ADAPTER	1	800.000	120.00	800.00

Figure D - 5: Extract of pyramid adapter quote.

Assuming that on average a user would want three stiffness settings on their foot, no gas springs, and two different gas spring setups, a total of 4 gas springs would be required. The unit costs of the gas springs used in Table D - 6 are from a quote received for gas springs and their end fittings; see Figure D - 6. The total costs for gas springs and end fittings were estimated to be R694.32.

Table D - 6: Gas springs cost breakdown

Gas spring price	122.76	R/unit
Amount of springs required	4.00	unit
Gas spring cost	491.04	R
End fitting price	25.41	R/unit
Amount of fittings required	8.00	unit
End fitting cost	203.28	R
Total Gas spring assembly cost	694.32	R

3 SA4135-103	Gas Spring 150mmx22x10-150N Without End Fittings	2 EA	122.76	0.00	36.83	245.52
4 FTS-LX144R-22*10 (258252)	Steel End Fitting with Removable ball stud for 22x10 gas spring	4 EA	25.41	0.00	15.25	101.64

Figure D - 6: Gas springs quote extract

The totals calculated above were then summed together to get a total cost of R 7570.19 for the manufacturing of the prosthetic foot. An additional 50% was added to the total for any unforeseen costs, like additional equipment or the costs for testing required before a prosthetic may be medically used. These costs would typically be split over multiple products. A profit margin of 50% is added to the total cost, bringing the estimated cost of the prosthetic foot to about R17 000. Whether or not the profit margin is acceptable would have to be considered within a detailed cost estimation study.

Table D - 7: Total cost breakdown

Total Cost before additions	7 570.19	R
Unforeseen Costs margin	50.00	%
Prosthetic foot estimated cost	11 355.28	R
Profit margin	50.00	%
Estimated Prosthetic foot price	17 032.93	R

Appendix E : Experiment Methodology

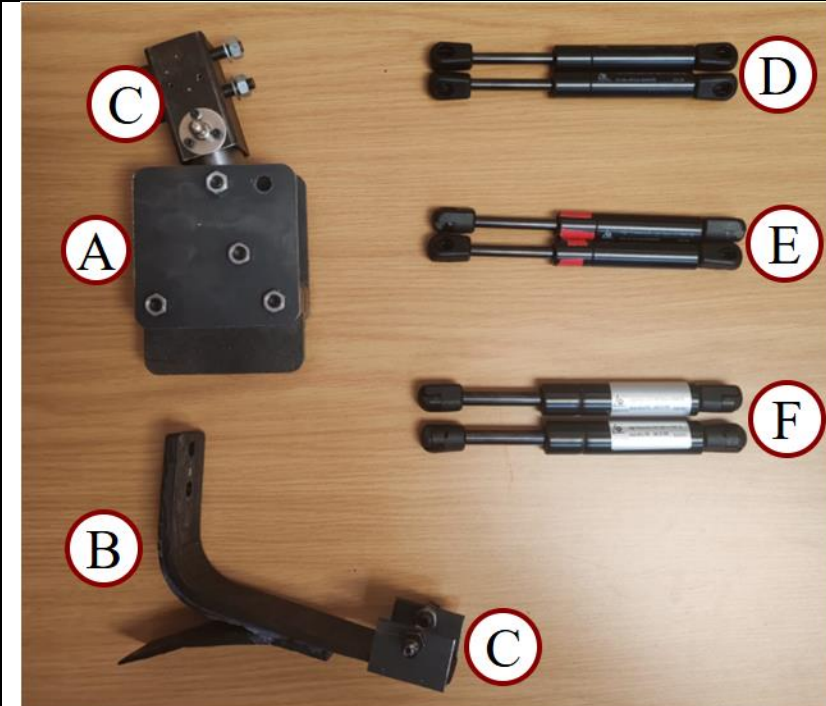
E.1. Equipment and Tools Used

To test the prosthetic feet, some tools, machines, and parts are needed. The list below encapsulates the required components, and the next section explains where they are used to build the experimental setup.

The components used, excluding the tooling, can be seen in Table E - 1.

- MTS machine
- Testing Jig
- The prosthetic foot that will be tested
- Two 150 N gas springs
- Two 350 N gas springs
- Two 500 N gas springs
- Two circular gas spring connector joints
- Two bent gas spring connector joints
- Allen Keys
- Wrench
- Bolts
- Nuts and Washers

Table E - 1: Components used during experiment

	A	Testing Jig
	B	Prosthetic foot
	C	Gas Spring Connectors
	D	150 N Gas Springs
	E	350 N Gas Springs
	F	500 N Gas Springs

E.2. Experimental Procedure

Multiple experimental runs were done, such as forefoot testing and heel testing on multiple prosthetic feet, gas spring testing at different height attachment points and a forefoot and heel fail test. The steps below explain how these tests were conducted.

Testing procedure for foot with no springs

1. The MTS machine was switched on and the necessary programs were started.
2. The top hydraulic grip was closed. This made a surface which would press down on the foot.
3. The jig was placed in the MTS machine and clamped in the bottom hydraulic grip.
4. The jig was adjusted to the 20° position. This position was for testing the forefoot.
5. The prosthetic foot blade was attached to the testing jig.
6. The forefoot was preloaded.
7. The MTS machines program was started, and the machine pressed down on the prosthetic foot.
8. The program was stopped manually at a displacement of approximately 20 mm for the forefoot and 15 mm for the heel.
9. Steps 5 to 7 were repeated for 3 successful runs.
10. The prosthetic foot was removed from the testing jig, and the jig was set to the 15° position. This position was for testing the heel.
11. The prosthetic foot blade was attached to the testing jig.
12. Steps 4 to 9 were repeated for the test of the prosthetic foot's heel.
13. The prosthetic foot was removed from the testing jig.

The MTS hydraulic press was stopped just shy of the designed displacement to prevent failure and allowed the specimens to be intact for multiple testing to prove the repeatability of the foot's testing. The displacement data up until the stopping point was sufficient for making data comparisons.

Slightly preloading the foot prevented the program from reading negative load values which would have lead to unsuccessful testing runs.

Testing procedure for foot with gas springs

The testing continued:

14. The testing jig was moved to the 20° position for forefoot testing.
15. This prosthetic foot was attached to the testing jig.
16. A 150 N gas spring was attached to both sides of the prosthetic foot and jig.
17. With the two springs attached, the prosthetic foot was loaded slightly.
18. The MTS machines program was started, and the machine pressed down on the prosthetic foot.
19. The program was stopped around 800 N manually or if the gas springs were pressed to its limit.
20. The 150 N gas springs were removed, and steps 16 -19 were repeated for 350 N gas springs and for 500 N gas springs.
21. The gas spring tests were concluded when there were three tests done for each setup.
22. The foot was removed from the jig.
23. The results of all the runs were saved on an external USB.
24. The MTS machine was switched off.

This concluded the experimental procedure.

Appendix F : Standards, Theory and Additional Data

F.1. International Standards (ISO10328)

According to ISO 10328 for a prosthetic foot to meet international standards, there are two tests that the foot must pass two tests, namely a test of torsion and a test of the foot unit itself. The foot unit includes static tests and cyclic tests. There are three loading conditions for the testing procedures: P3, P4, and P5. These conditions are based on data of amputees regarding their body weight. The P5 loading condition is based on data from amputees with a body mass greater than 100 kg, whereas the P4 and P3 loading conditions are based on data from amputees with body weights less than 80 kg and 60 kg, respectively.

The P5 loading condition has a test force of 2240 N, the P4 condition has a test force of 2065 N, and the P3 condition has a force of 1610 N. This load acts on both the forefoot and the heel. A minimum of two tests are required: the static proof test, the static ultimate strength test, and the cyclic test.

The recommended heel height, captured as h_r in Figure F - 1, for the foot under test, is 20 mm. This is the difference between the heel and the toe part.

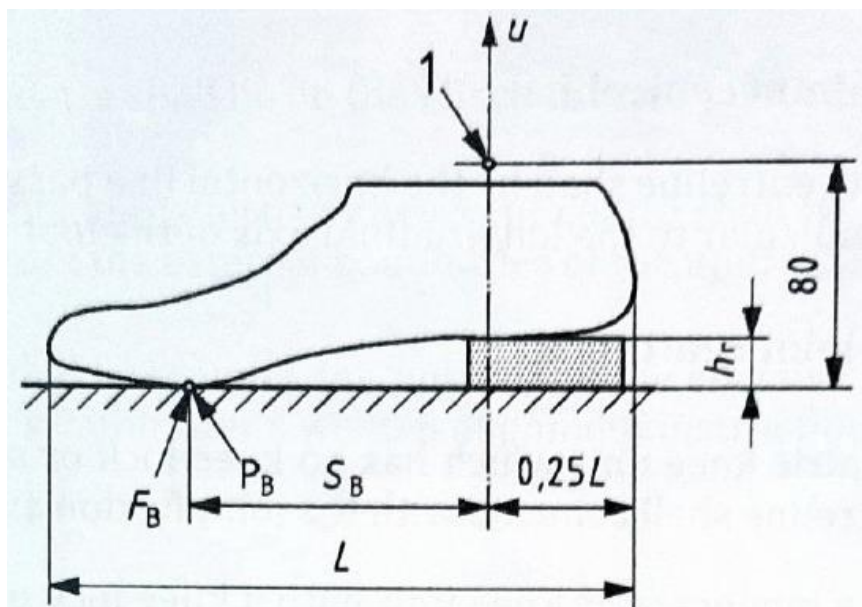


Figure F - 1: Position of setup of the prosthetic foot during testing.

Figure F - 2 shows the set-up of the prosthetic foot during testing. For loading conditions P3, P4, and P5, α , β , and γ are 15° , 20° , and 7° , respectively.

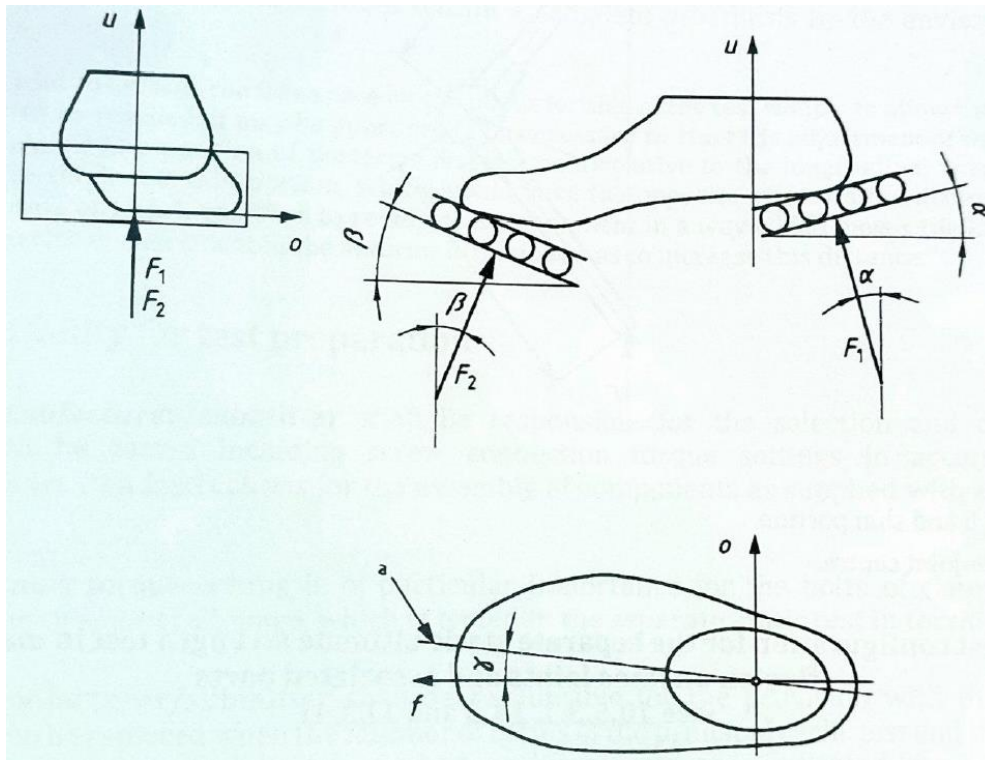


Figure F - 2: Angles of the loading platforms during testing.

The testing equipment must be able to provide the load at a rate of 100 N/s and 250 N/s. The force should be transferred to the heel and forefoot using a loading platform, as seen in Figure F - 2. The platforms are long enough to allow the toe and heel to be loaded at the same time. If it is necessary to use an attachment to test a prosthetic component, in this case the foot, the ISO standards make it clear that the stiffness of this attachment is not less than that of the component being tested. All cosmetic attachments are removed from the prosthetic components prior to testing, unless they contribute to the strength of the foot.

F.2. Ottobock Foot Size Chart

ottobock.

Foot length cm	Shoe size <i>Europe</i>	Shoe size <i>England</i>	Foot length cm	Shoe size <i>USA</i>	
				MALE	FEMALE
14	21	5	14	7 - 7 1/2	8 - 8 1/2
15	22	6	15	8 - 8 1/2	9 - 9 1/2
16	23	7	16	9 - 9 1/2	10 - 10 1/2
17	24	8	17	10 - 11 1/2	11 - 12 1/2
18	25	9	18	12 - 12 1/2	13 - 13 1/2
19	26	10	19	13 - 13 1/2	1 - 1 1/2
20	27	11	20	1 - 1 1/2	2 - 2 1/2
21	28	12	21	2 - 3 1/2	3 - 4 1/2
22	29	13			
23	30				
24	31				
25	32				
26	33	1	22	MALE	FEMALE
27	34	2	23	4 - 5	5 1/2 - 6 1/2
28	35	3	24	5 1/2 - 6	7 - 7 1/2
29	36	4	25	6 1/2 - 7	8 - 8 1/2
30	37	5	26	7 1/2 - 8	9 - 9 1/2
31	38	6	27	8 1/2 - 9	10 - 10 1/2
32	39	7	28	9 1/2 - 10	11 - 11 1/2
33	40	8	29	10 1/2 - 11	12 - 12 1/2
34	41	9	30	11 1/2 - 12	13 - 13 1/2
35	42	10	31	12 1/2 - 13	14 - 14 1/2
36	43	11		13 1/2 - 14	15 - 15 1/2
37	44	12			
38	45				
39	46				
40					
41					
42					
43					
44					
45					
46					

Shoe sizes vary. For most accurate sizing, measure insole length of patients shoe in cm.