

Motion Tracking and Modelling of the Forefoot for Clinical Gait Analysis

Department of Engineering Science
University of Oxford



In partial fulfilment of the requirements for the degree of
Doctor of Philosophy

by

Po-Hsiang Chan

St Catherine's College

Trinity Term 2019

Supervised by

Dr Amy Zavatsky & Dr Julie Stebbins

Declaration

I declare that this thesis is entirely my own work, and except where otherwise stated, describes my own research.

Po-Hsiang Chan
St. Catherine's College
University of Oxford

Motion Tracking and Modelling of the Forefoot for Clinical Gait Analysis

Po-Hsiang Chan

Abstract

Multi-segment foot models (MSFMs) are often used in clinical gait analysis for the diagnosis and treatment planning for patients with foot deformities. Thus far, close to 40 MSFMs have been proposed, and each differs in terms of segmentation, marker placement protocol, tracking technology, and joint motion. By representing individual segments as a series of linked rigid body segments, the computation and interpretation of foot biomechanics can be greatly simplified. While this approach may be acceptable for lower-limb segments such as the thigh, shank, or hindfoot, the assumption that this is also valid for the forefoot has been met with criticism. Out of all the segments, there is least agreement on how the forefoot should be represented in an MSFM. In general, Whether or not the flexibility of the forefoot is clinically meaningful is controversial.

Therefore, the aims of this thesis were to investigate the most appropriate way of segmenting the forefoot based on a rigidity criteria and to develop a clinically relevant and practical forefoot model based on the recommended segmentation. It was hypothesised that minimising the violation of rigid body assumption in the proposed novel MSFMs will produce the most relevant forefoot representation that is both practical and useful for clinical gait analysis.

This study first validated the use of σ_{RBE} to quantify MSFM segment rigidity through a Monte-Carlo based global sensitivity analysis, which showed that for a typical forefoot segment the σ_{RBE} represented 86.7% of the deformation magnitude, with a linear relationship that can be predicted using a simple regression model. Experimental data were used to compute rigidity and track deformation of the Oxford Foot Model segments and the metatarsal markers (attached to the heads and bases of all metatarsals) over the full gait cycle. These data were used to determine appropriate segmentation and marker-placement strategies in the design of novel MSFMs.

Several novel forefoot models were proposed and tested through comparison of visual and descriptive statistics of kinematic output and repeatability analysis. A single novel forefoot model was selected based on practicality and kinematic advantages, and piloted in five individual clinical case studies. The research has contributed to the evidence that the forefoot should not be treated as a single segment, and a new forefoot model suitable for CGA has been proposed.

Acknowledgement

I would like to express my gratitude towards my supervisors, Professor Amy Zavatsky and Dr Julie Stebbins for their guidance, support, and supervision that enabled me to complete this project. Their immense knowledge on the subject and their constant encouragement and inspiration is what made this thesis possible.

I would also like to thank the Institute of Mechanical Engineering, St. Catherine's College (University of Oxford), and the Department of Engineering Science (University of Oxford) for their generous financial support.

To the team at the Oxford Gait Laboratory, in particular, Mr Tim Theologis, Mike Colloff, Amran Saddiq, Andrew Lewis, Jules Letherland, and Marian Harrington, for their technical support and guidance, I would like to express my sincere gratitude.

To all research volunteers who must not be named due to ethical reasons, you are what made this research possible, and for that I would like to express my deepest appreciation.

To all my friends, colleagues, and mentors, thank you for keeping me motivated and making my time in Oxford enjoyable, in particular, Tsai-Chang Wu, Yu Chou, Tsung-Wei Lai, Yung-Chih Liu, Vikas Bagla, Chao Jiang, Henry Lee, Laura Diment, Frank Penkava, Jerneja Uhan, and friends from the Oxford Robotics and Additive Manufacturing Society. Special thanks to Dr Tung-Wu Lu and Dr Spyros Masouros who inspired me to pursue this field in the first place.

Last but not least, I am extremely grateful to my best friend, Yingyi He, for her support and understanding that enabled me to complete this thesis. Her many personal sacrifices, constant encouragement, and unconditional support are what gave me strength throughout this journey.

Finally, I am deeply thankful to my parents, Li-Chen Hsu and Ren-Wu Chan, for their unending support and sacrifices throughout my life and the journey of completing this degree.

Table of Contents

Declaration	i
Abstract	ii
Acknowledgement	iii
Table of Contents	viii
List of Figures	xi
List of Tables	xii
Abbreviations & Nomenclatures	xiv
1 Introduction	1
1.1 Rationale	2
1.2 Aims and Objectives	3
1.3 Thesis Overview	3
2 Literature Review	5
2.1 Clinical Gait Analysis	6
2.1.1 Historical Development	6
2.1.2 Modern Instrumented Gait Analysis	8
2.1.3 Motion Tracking Technologies	9
2.1.4 Other Gait Measurements	10
2.1.5 Gait Cycle and Events	11
2.1.6 Temporal and Spatial Parameters	13
2.2 Rigid Body Kinematics	14
2.2.1 Euler Angles	16
2.2.2 Helical Angles	17
2.2.3 Joint Coordinate System	18
2.2.4 Summary	19
2.3 Foot Anatomy and Biomechanics	19
2.3.1 Basic Anatomical Terminologies	20
2.3.2 Bones of the Foot	22
2.3.3 Joints of the Foot	24

2.3.4	Arches of the Foot	29
2.4	Summary	31
3	Multi-segment Foot Models: An Overview of Design and Application	33
3.1	General Overview	34
3.2	Notable Foot Models	36
3.2.1	Milwaukee Foot Model	37
3.2.2	Oxford Foot Model	39
3.2.3	Rizzoli Foot Model	40
3.2.4	Model Comparison	42
3.3	Forefoot Representation in MSFM	44
3.3.1	Single-Segment Representations	44
3.3.2	Two-Segment Representations	48
3.3.3	Anatomical Forefoot Subdivision	51
3.4	MSFM Design Considerations	54
3.4.1	Practical Considerations	55
3.4.2	Methodological Considerations	57
3.5	Summary	59
4	Methodology	60
4.1	Ethical Considerations	61
4.2	Subjects and Participants	61
4.3	Equipment	63
4.4	Gait Analysis Data Collection	63
4.4.1	Systems Calibration	63
4.4.2	Subject Preparation	65
4.4.3	Experimental Protocol	68
4.5	Data Processing	68
4.5.1	Kinematic Data Processing	68
4.5.2	Biomechanical Model Processing	70
4.6	Model Computation	70
4.6.1	Plug-in-Gait Model	70
4.6.2	Oxford Foot Model	76
4.6.3	Oxford Forefoot Model	79
4.7	Gait Kinematics	80
4.7.1	Plug-in-Gait	80
4.7.2	Oxford Foot Model Kinematics	82
4.8	Quantifying Rigidity	82
4.9	Summary	84
5	Rigid Body Error: A Global Sensitivity Analysis	85
5.1	Introduction	86
5.2	Methodology	87
5.2.1	Sensitivity Analysis - General Methodology	88

TABLE OF CONTENTS

5.2.2	Sensitivity Analysis - Input Simulation	89
5.2.3	Regression Analysis	92
5.3	Results	94
5.3.1	Uncertainty Analysis	94
5.3.2	Sensitivity Analysis	95
5.3.3	Regression Analysis	99
5.4	Discussion	102
5.4.1	Sensitivity Analysis	102
5.4.2	Regression Analysis	104
5.5	Summary	105
6	Quantifying Oxford Foot Model	
	Segment Rigidity	106
6.1	Introduction	107
6.2	Methodology	108
6.2.1	Ethical Considerations	108
6.2.2	Subjects and Participants	108
6.2.3	Data Collection and Processing	108
6.2.4	Statistics	109
6.2.5	Repeatability Analysis	110
6.2.6	Clinical Case Studies	111
6.3	Results	111
6.3.1	Segment Rigidity	111
6.3.2	Repeatability Analysis	115
6.3.3	Clinical Case Studies	115
6.4	Discussion	117
6.5	Summary	119
7	Forefoot Segmentation and Rigidity of Metatarsal Pairs	121
7.1	Introduction	122
7.2	Methodology	122
7.2.1	Ethical Considerations	122
7.2.2	Subjects and Participants	123
7.2.3	Data Collection and Processing	123
7.2.4	Statistics	124
7.3	Results	125
7.4	Discussion	126
7.5	Summary	128
8	Novel Forefoot Modelling: An Exploratory Study	129
8.1	Introduction	130
8.2	Methodology	131
8.2.1	Ethical Considerations	131
8.2.2	Subjects and Participants	132
8.2.3	Data Collection and Processing	132

TABLE OF CONTENTS

8.2.4	Biomechanical Models	132
8.2.5	Inter-Segmental Kinematics	139
8.2.6	Statistics	139
8.3	Results	140
8.3.1	Inter-Segmental Kinematics	140
8.3.2	Planar-Angles	145
8.3.3	Descriptive Statistics	148
8.3.4	Repeatability Studies	151
8.4	Discussion	154
8.5	Summary	156
9	Novel Forefoot Modelling: Analysis of Hybrid Models	157
9.1	Introduction	158
9.2	Methodology	159
9.2.1	Ethical Considerations	159
9.2.2	Subjects and Participants	159
9.2.3	Data Collection and Processing	159
9.2.4	Biomechanical Models	160
9.2.5	Inter-Segmental Kinematics	164
9.2.6	Statistics	164
9.3	Results	165
9.3.1	Inter-Segmental Kinematics	165
9.3.2	Planar-Angles	170
9.3.3	Descriptive Statistics	172
9.3.4	Repeatability Analysis	174
9.4	Discussion	176
9.5	Summary	177
10	Clinical Case Studies	179
10.1	Introduction	180
10.2	Methodology	180
10.2.1	Ethical Considerations	180
10.2.2	Subjects and Participants	181
10.2.3	Data Collection and Processing	184
10.2.4	Abnormal Gait Identification	185
10.3	Results	186
10.3.1	Case A	186
10.3.2	Case B	189
10.3.3	Case C	192
10.3.4	Case D	195
10.3.5	Case E	198
10.4	Discussion	200
10.5	Summary	202

11 Conclusions	203
11.1 Key Contributions	204
11.1.1 Quantifying Segment Rigidity Using σ_{RBE}	204
11.1.2 Segment Rigidity and Forefoot Representation	204
11.1.3 Novel Forefoot Modelling	205
11.1.4 Clinical Case Studies	206
11.2 Limitations	207
11.2.1 Quantifying Segment Rigidity Using σ_{RBE}	207
11.2.2 Segment Rigidity and Forefoot Representation	208
11.2.3 Novel Forefoot Modelling	209
11.2.4 Clinical Case Studies	210
11.3 Future Directions	210
A Data Collection Documentations	212
A.1 Subject Measurement Sheet	212
B Additional Kinematic Graphs	213
B.1 Chapter 6 Pathological Gait - Patient 1	214
B.2 Chapter 6 Pathological Gait - Patient 2	215
B.3 Chapter 10 Clinical Case Study - Subject A	216
B.4 Chapter 10 Clinical Case Study - Subject B	217
B.5 Chapter 10 Clinical Case Study - Subject C	218
B.6 Chapter 10 Clinical Case Study - Subject D	219
B.7 Chapter 10 Clinical Case Study - Subject E	220
C Derivations	221
C.1 Orthogonal Procrustes Problem	221
C.2 Random Rotation Matrix	223
D Additional Analysis and Assessments	226
D.1 Convergence Assessment for Sensitivity Analysis	226
Bibliography	229

List of Figures

2.1	Muybridge’s horse in motion	7
2.2	Gait cycle and events during locomotion	12
2.3	Finite helical axis motion	18
2.4	Anatomical plane and foot motion	20
2.5	Bones of the foot	22
2.6	Midtarsal joint axes	27
2.7	Windlass mechanism	30
3.1	Subdivision of notable foot models based on tracking	37
3.2	Rizzoli Foot Model diagram	41
3.3	Nester’s walking simulator	52
3.4	EM sensors for metatarsal tracking	54
4.1	Vicon camera and AMTI force-plate	62
4.2	Oxford Gait Laboratory walkway	62
4.3	Data collection flow chart	64
4.4	Calibration wand and a 10 kg calibration weight	64
4.5	PiG and OFM marker-set diagram	66
4.6	Vicon Nexus camera reconstruction	69
4.7	PiG pelvis local coordinate system.	71
4.8	PiG femur local coordinate system	74
4.9	PiG tibia local coordinate system	75
4.10	OFM hindfoot local coordinate system	77
4.11	OFM forefoot local coordinate system	78
5.1	Sensitivity variable dependency flowchart	88
5.2	Randomly generated configuration examples	90
5.3	Sensitivity analysis scatter-plot 1	96
5.4	Sensitivity analysis scatter-plot 2	97
5.5	Diagram of σ_{RBE} vs. $ \delta D $ with outlier removed.	100
5.6	Log-log regression model plot	102
6.1	OFM marker set for rigidity quantification	109
6.2	OFM rigidity vs. gait cycle, static reference	112
6.3	OFM rigidity vs. gait cycle, mid-stance reference	113
6.4	Comparisons of pathological OFM segment rigidity	116

LIST OF FIGURES

7.1	Markers on the bases and heads of all five metatarsals	124
7.2	Bar chart with σ_{av} for each metatarsal (MT) pair	125
7.3	Forefoot segmentation recommendation based on σ_{RBE}	127
8.1	Type A OFFM forefoot segment definitions	135
8.2	Type B OFFM forefoot segment definitions	137
8.3	Type A OFFM medial forefoot inter-segmental kinematics	142
8.4	Type A OFFM lateral forefoot inter-segmental kinematics	143
8.5	Type B OFFM intermediate forefoot inter-segmental kinematics . . .	144
8.6	Type B OFFM projected planar angles	146
8.7	Type B OFFM angle between first and fifth metatarsals	147
9.1	Type C1 OFFM forefoot segment definitions	161
9.2	Type C2 OFFM forefoot segment definitions	162
9.3	Type C OFFM medial forefoot inter-segmental kinematics	167
9.4	Type C OFFM lateral forefoot inter-segmental kinematics	168
9.5	Type C OFFM intermediate forefoot inter-segmental kinematics . . .	169
9.6	Type C OFFM first metatarsal kinematics	171
9.7	Type C OFFM angle between first and fifth metatarsals	171
10.1	Static foot posture of case study subject A	181
10.2	Static foot posture of case study subject B	182
10.3	Static foot posture of case study subject C	182
10.4	Static foot posture of case study subject D	183
10.5	Static foot posture of case study subject E	183
10.6	Case study A inter-segmental forefoot kinematics	186
10.7	Case study A first and fifth metatarsals kinematics	187
10.8	Case study A MTI-V angle	187
10.9	Case study B inter-segmental kinematics	189
10.10	Case study B first and fifth metatarsals kinematics	190
10.11	Case study B MTI-V angle	190
10.12	Case study C inter-segmental kinematics	192
10.13	Case study C first and fifth metatarsals kinematics	193
10.14	Case study C MTI-V angle	193
10.15	Case study D inter-segmental kinematics	195
10.16	Case study D first and fifth metatarsals kinematics	196
10.17	Case study D MTI-V angle	196
10.18	Case study E inter-segmental kinematics	198
10.19	Case study E first and fifth metatarsals kinematics	199
10.20	Case study E MTI-V angle	199
B.1	OFM kinematic graphs for CP patient 1	214
B.2	OFM kinematic graphs for CP patient 2	215
B.3	OFM kinematics graph for Subject A	216
B.4	OFM kinematics graph for Subject B	217
B.5	OFM kinematics graph for Subject C	218

LIST OF FIGURES

B.6	OFM kinematics graph for Subject D	219
B.7	OFM kinematics graph for Subject E	220
C.1	Comparison of rotation methods	223
C.2	Z-Y-X Sequential rotation	224
D.1	Shifting convergence band example	228

List of Tables

3.1	List of single-segment foot models	46
3.2	List of two-segment foot models	50
4.1	List of PiG, OFM, and OFFM markers	67
4.2	PiG kinematic output	81
4.3	OFM kinematic output	82
5.1	Monte-Carlo computation steps	93
5.2	Descriptive statistics of input parameters	94
5.3	Descriptive statistics of output parameters	94
5.4	Sensitivity indices for σ_{RBE}	98
5.5	Sensitivity indices for $Rerr$	98
5.6	Log-log regression model parameters	102
6.1	Comparison of OFM segment rigidity, calculated with static reference	114
6.2	Comparison of OFM segment rigidity, calculated with mid-stance ref- erence	114
6.3	Comparisons of OFM repeatability	115
7.1	Pairwise MT pairs rigidity comparison	126
8.1	Type A OFFM segment definition	136
8.2	Type B OFFM segment definition	138
8.3	Type A OFFM descriptive statistics	149
8.4	Type B OFFM descriptive statistics	150
8.5	Repeatability of type A OFFM kinematics	152
8.6	Repeatability of type B OFFM kinematics	153
9.1	Type C OFFM segment definition	163
9.2	Type C OFFM descriptive statistics	173
9.3	Repeatability of type C OFFM kinematics	175
10.1	Clinical case subject information and clinical background.	181
D.1	Convergence assessment parameters & CB width	229
D.2	Converged sample size through ASCBR	229

Abbreviations & Nomenclatures

List of Abbreviations

AJC	Ankle joint centre
ASCBR	Acceptable shifting convergence band rule
ASIS	Anterior superior iliac spine
BTK	Biomechanical Toolkit
CP	Cerebral palsy
EMG	Electromyography
GA	Gait analysis
GCS	Global coordinate system
GC	Gait cycle
GSA	Global sensitivity analysis
HFMM	Heidelberg Foot Measurement Method
HJC	Hip joint centre
JCS	Joint Coordinate System
KJC	Knee joint centre
LCS	Local coordinate system
LFMI	Leardini Foot Model I
LLA	Lateral longitudinal arch
MFM	Milwaukee Foot Model
MLA	Medial longitudinal arch
MSFM	Multi-segment foot model
mSHCG	modified Shriners Hospitals for Children Greenville Foot Model

OAT	One-at-a-time (sensitivity analysis method)
OFFM	Oxford Forefoot Model
OFM	Oxford Foot Model
OGL	Oxford Gait Laboratory
PiG	Plug-in-Gait
RA	Regression analysis
RFM	Rizzoli Foot Model
ROM	Range of motion
SA	Sensitivity analysis
STA	Soft tissue artefact
SVD	Singular value decomposition
UA	Uncertainty analysis

List of Nomenclatures

σ_{av}	Average rigid body error	mm
σ_{ptp}	Peak-to-peak rigid body error	mm
σ_{RBE}	Rigid body error	mm
k_A	Condition number	-
R_{err}	Rotation error	-

Chapter 1

Introduction

In both clinical gait analysis (CGA) and foot biomechanical research, multi-segment foot models (MSFM) have been used extensively to collect and analyse motion data. However, much of the motivation behind the development of different MSFMs up until now have been descriptive in nature, with little justification on why the individual segments are designed the way they are, their purpose, or justification. This is particularly true for the forefoot region. This thesis intends to design and evaluate a number of new forefoot modelling protocols based on a set of objective criteria, with the final aim of proposing a forefoot model suitable for CGA.

1.1 Rationale

In stereophotogrammetry based CGA and biomechanical research, MSFMs have been utilised to capture and characterise movements of the foot [11, 154]. Diagnosis and treatment planning for locomotion disorders, especially central nervous system diseases associated with spasticity, have benefited enormously through the use of preoperative gait analysis [57, 68, 129, 275]. The methods used simplify the complex foot structure into a collection of rigid body segments, with each segment representing a group of bones and the motion between each segment represented by a specific joint angle. However, there is currently no consensus on how the foot should be represented. Many foot models were developed to be used ‘in-house’, posing difficulties for exchanging information, characterising pathological gait [208], consistent reporting [23], and validating clinical studies. In addition, most of the existing foot models lack external applications, which, according to Bishop et al. [23] is due to the absence of a clear purpose and design justification.

Kinematic models proposed in the 1990s have modelled the entire foot as a single segment rigid body [130, 178, 235, 239], with the most famous being the Plug-in-Gait (PiG) model [63, 126]. To this date, PiG is still widely used in many clinical gait laboratories and is available in most motion capture systems. While this model has managed to extensively describe the three major lower-limb joints, that is, the hip, knee, and ankle joints, they failed to capture the intricate movements within the foot. Since then, a number of MSFMs have been proposed with division of the foot into smaller rigid body segments, usually into hindfoot, forefoot, hallux, and sometimes mid-foot segments [42, 125, 143, 161, 240]. Amongst these segments, there is least consensus on how the forefoot should be modelled [210]. While some consider the forefoot as a single segment, such as the Oxford Foot Model [42, 250] and the Rizzoli Foot Model [143], others have divided it into a medial and a lateral segment, such as

the Kinfoot [161], Ghent Foot Model [65], Jenkyn & Nicol Foot Model [125], and the Salford Foot Model [187]. The lack of consensus in forefoot representation and the justification for subdivision motivated this research.

1.2 Aims and Objectives

The primary aim of this thesis was to first derive and validate a method for quantifying the rigidity of 3D motion capture segments, especially of the forefoot, and to show its use in defining and evaluating a model for the forefoot. Secondly, to develop a novel forefoot model that can be used for both clinical gait analysis and foot biomechanics research. To achieve this, this thesis addresses the following objectives:

- To assess the existing modelling techniques and identify essential design criteria.
- To validate and apply an appropriate method of forefoot segmentation.
- To justify subdivision of the forefoot based on rigidity criteria.
- To compare and evaluate different forefoot modelling strategies.
- To validate the proposed model through preliminary clinical case studies.

1.3 Thesis Overview

The overall structure of the thesis takes the form of eleven Chapters. **Chapter one** (the current chapter) outlines the overall aims and rationale for this study. This is then followed by a general review of the literature (**Chapter two**) of CGA, which includes the historical development and the established methods for collecting, processing, and presenting relevant parameters used to describe biomechanics, such as temporal and spatial parameters. In addition, Chapter two also introduces the fundamentals of foot biomechanics and various kinematic conventions to present inter-segmental motions. **Chapter three** provides an overview of the existing MSFMs, comparing the differences in marker-placement and segment definition strategies in

reference to clinical and validation studies. In **Chapter four**, the methods and materials used to achieve the research objectives are fully described, with procedures and statistical techniques limited to specific chapters documented within their respective chapters. **Chapter five** presents a Monte-Carlo based global sensitivity analysis to further investigate the efficacy of using σ_{RBE} to quantify rigidity of an MSFM. The result of this Chapter justifies the use of σ_{RBE} to quantify rigidity and track deformation of a MSFM marker-cluster. In **Chapters six and seven**, the rigidity of various segments of the lower-limb including the tibia, hindfoot, forefoot, and all combinations of the metatarsal pairs are quantified. The rigidity of each segment was quantified and tracked over a full gait cycle using rigid body error (σ_{RBE}). The results of these two chapters were used to guide the development of novel forefoot models reported in Chapter eight and nine. The next two chapters (**Chapters eight and nine**) detail the analysis of several novel forefoot models proposed. **Chapter ten** completes the main body of this thesis by presenting the result of five individual clinical case studies using a single forefoot model selected from the previous chapter. Finally, the conclusion, limitations, and future studies are presented in **Chapter eleven**.

Chapter 2

Literature Review

Multi-segment foot models (MSFM) have been developed for stereophotogrammetric-based motion analysis, which has a wide range of clinical and research applications. In this Chapter, the basic principles of clinical gait analysis (CGA), motion tracking technologies, and gait measurements are presented. This includes both established and novel methods of collecting, measuring, and interpreting gait data. In the interest of modelling the forefoot in the most anatomically and functionally appropriate way, an overview of foot anatomy and biomechanics will also be provided. In line with the aim and theme of this thesis, the content of this literature review will focus on the applications and biomechanics relevant to the forefoot. While greater attention will be paid to the kinematic analysis, it is acknowledged that CGA relies on other components, including electromyography, kinetics, and energy assessment, none of which are stand-alone in a typical gait analysis session.

2.1 Clinical Gait Analysis

2.1.1 Historical Development

The study of human locomotion dates back to the ancient Greeks. Notable philosophers including Hippocrates, Plato, and Socrates offered theoretical explanations of human movement. Throughout history, countless philosophers and scientists have contributed towards the advancement of human biomechanics, with the first known documentation of the analysis of human walking written by Aristotle (384-322 BC) in his book *De Motu Animalium* (Movement of Animals). Another significant contributor to the progression of biomechanics was by Galen (131-201 AD), who investigated the roles of muscles, nerves, and soft-tissues in relation to their contributions to motion [98].

Further advancement in biomechanics was not until the Renaissance period when scientists including Leonardo da Vinci (1452-1519), Galileo Galilei (1564-1642), Giovanni Borelli (1608-1679), and much later on, the Weber brothers (1795-1891), Étienne-Jules Marey (1830-1904), Eadweard Muybridge (1830-1904) undertook ground-breaking research that shaped biomechanics research as we know it today. Borelli's publication, *De Motu Animalium* (On the Motion of Animals), was the first to include forces and the estimation of centre of mass in the description of muscular movement and body dynamics. During the Age of Enlightenment, Wilhelm Eduard Weber and his brother Eduard Friedrich Weber studied the mechanism of walking in humans [267]. Since then, the study of human locomotion has progressed from a descriptive science to one based on quantitative measurements. Around the same period, Etienne-Jules Marey made an important breakthrough in analysing human movements using graphical methods and celluloid film chronophotography. In parallel, Eadweard Muybridge studied the movement of horses trotting and demonstrated that there are moments

when all four of its feet are simultaneously lifted from the ground (Fig. 2.1). The gait measurements were made using multiple cameras triggered successively as the horse ran into thin wires across the runway which paved the way for modern gait analysis methodology. His scientific achievements also include the the motion study of humans performing various tasks and the action of other animals from the Philadelphia zoo.

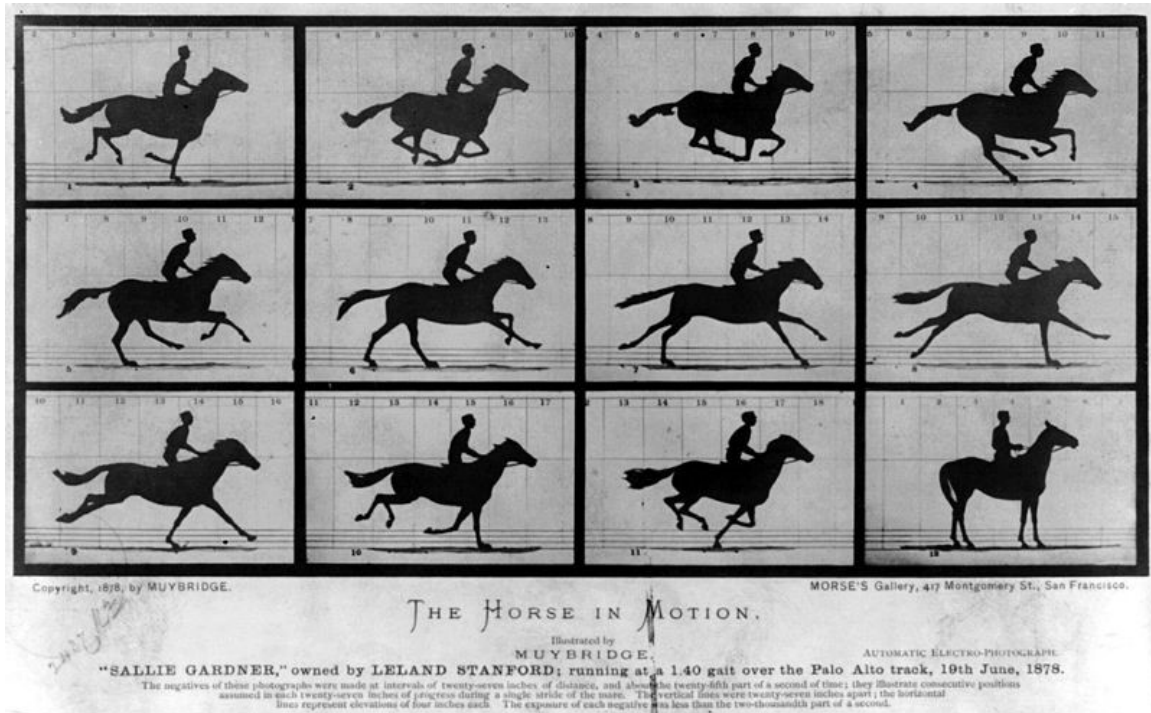


Figure 2.1: The horse in motion, illustrated by Muybridge [182].

There are other notable scientists who contributed to the study of human biomechanics. After returning from World War II, Verne Inman applied electromyography to study walking in disabled populations and developed the first clinical research programmes using movement analysis techniques [121]. This technology was adopted by the pioneers of clinical gait analysis, Jacquelin Perry, David Sutherland and Jurg Braumann [10], who modernised the clinical gait analysis procedure.

Human motion analysis has developed rapidly through advancement in technology and the understanding of anatomy and mechanics. While stereophotogrammetry has evolved to be the gold-standard of modern-day gait analysis methodology [37, 53],

novel techniques such as inertial sensor-based systems [180] and marker-less motion capture systems are also available [45, 60]. Together, the rapid advancement in computer technology and novel algorithms [102] further accelerated the development of motion analysis.

2.1.2 Modern Instrumented Gait Analysis

Gait analysis utilises a range of technologies to reveal quantitative information about the biomechanics of the human musculo-skeletal system during locomotion. According to the result of a recent Italian consensus conference [21], CGA can be defined as follows:

“Gait analysis (GA) is a process of instrumented measurement and evaluation of walking ability in patients with impairments specific to locomotion. GA aims at answering questions subtending clinical decisions and/or monitoring, and is part of a broader technology of instrumented analysis of human movement. GA, as intended herein, does not refer to a generic method for recording gait parameters, but rather to the way to answer a specific clinical question. Measurements must be related to the individual clinical problem, with the aim of supporting medical decision-making.”

(Benedetti et al., 2017)

While there exists a number of technologies and biodynamic variables that can be measured, the current standard for a clinical gait laboratory requires a minimum of three pieces of equipment; that is, stereophotogrammetry, force-plates, and electromyography (EMG) [21]. To collect gait kinematics, the general procedure involves capturing the instantaneous positions of markers in three-dimensions, such as those specified for the Plug-in-Gait (PiG) model [63, 126] or the Oxford Foot Model (OFM) [42, 250], while the subject walks across a walkway instrumented with force-plates that measures ground-reaction force. Electrodes, typically on the skin, measure myoelectric signals that are used to interpret relevant muscle activities.

Rapid development in computer and motion capture technologies have prompted the widespread use of stereophotogrammetry-based CGA. Through accreditation programmes, such as the Clinical Movement Analysis Society and North American Commission for Motion Laboratory Accreditation, the use of instrumented gait analysis has been increasingly popularised. Today, there is a wealth of evidence supporting the use of CGA for diagnosis and treatment planning. According to a systematic review by Wren et al. [275], recommendations from CGA have often altered clinical decisions and resulted in a higher percentage of surgical procedures being followed up afterwards. This is especially true for the management of neuromuscular diseases such as cerebral palsy; many studies have supported the use of CGA as a diagnostic and treatment evaluation tool [47, 57, 86, 129]. However, it must be acknowledged that there are still many technical and methodological problems to be addressed [241], including the reduction of time and expense, reporting standards [23], and the variability in kinematic and kinetic measurements.

2.1.3 Motion Tracking Technologies

Modern CGA relies on the non-invasive technique of motion tracking through the use of retro-reflective markers attached to the skin via hypo-allergenic tapes. The instantaneous positions of these markers are tracked by either video cameras or optoelectronic sensors, available as commercial systems such as Vicon or Qualisys. In a gait analysis laboratory, several cameras are mounted on the tripods, walls, and ceilings surrounding the walk-way. Computer softwares is able to calculate the relative position of each camera from a calibration process that usually involves waving an object with several markers positioned in a pre-determined geometrical relationship. During motion capture, the calculation is reversed to locate the precise coordinates of markers in three-dimensions. A comprehensive technical review of the general gait analysis methodology based on this technique was given in a series of four papers by

Chiari [53, 52], Leardini [147], and Della Croce [67].

The main limitation of stereophotogrammetry is not the inherent errors in measuring the marker coordinates (i.e. system errors and camera resolution); it is instead the movement of the skin and muscles relative to the underlying bone. Although studies in the 1990s measured a mean system error of between 0.94 - 18.42 mm [87, 88], significant technological progress has been made since then. A more recent study [89] measured an average system error of 1.11 mm under normal gait laboratory conditions. Compared to this, the errors caused by the movement of soft-tissue relative to the skin, the soft-tissue artefact (STA), and incorrect positioning of the markers in relation to the bony anatomy are more problematic [32]. A recent review by Cereatti et al. [44] on STA showed that the markers attached to the thigh region are associated with a deviation magnitude of 27.7 mm along a single axis of the bone embedded anatomical frame (i.e. femur coordinate system).

Other systems are also available for gait analysis. Active marker systems using infrared light-emitting diodes also provide accurate readings [166], but suffer from occlusion and the need to carry batteries. The use of inertial measurement units that consist of 3D accelerometers and gyroscopes are also gaining popularity because of their increased measurement range and portability [31, 218]. More recently, there appears to be an increasing interest in the use of marker-less systems due to advantages in the reduction of preparation time and the elimination of the occlusion problem. Studies done recently have demonstrated modest results in obtaining reliable 3D gait kinematics [45, 225].

2.1.4 Other Gait Measurements

From the review on the evolution of CGA given in three companion papers by Sutherland [251–253], it is evident that a combination of different gait measurements complement each other, expanding the capacity of the assessment compared to each of

the components alone. With motion data obtained from marker trajectories, inter-segmental angles can be computed using rigid body kinematics [53]. With the addition of ground reaction force [90], inverse dynamics can be performed to measure joint moment, power, and energy. The inclusion of EMG is also essential in a well-equipped gait laboratory to measure the electrical activity of a contracting muscle [21]. The benefit of the use of EMG in recommending surgical operations [203, 95] and in understanding of biomechanics is well-documented [251]. Together, the combination of motion capture, EMG, and ground reaction forces forms the basis of modern CGA. For research-oriented gait laboratories, other equipment such as pedobarographs, oxygen consumption measuring systems, and heart-rate monitoring may also be available [154].

2.1.5 Gait Cycle and Events

Human locomotion is a cyclic process that is often referred to as the ‘gait cycle’ (GC), which is defined as the interval between two successive steps of the same foot during walking. It is a convenient way of associating the activation of specific muscle activity and changes of joint kinetics and kinematics at a specific point in time (as percentage gait cycle) during walking. In practice, a consensus is reached on the definition of most gait phases and events during locomotion, enabling research comparisons across the literature [11, 154, 202]. However, the exact timing of each phase differs from person to person due to the differences in physical characteristics and walking speed [136]. In this thesis, the description of the GC events and phases follows the convention described by Levine et al. [154] and Baker [11]. The timing of each phase and period reported here are based on a large sample size study of gait cycle timing by Blanc et al. [25], approximated from the average of male and female subjects. This section presents a general overview of the key gait events during the GC. A more detailed description of foot motion can be found in Section 2.3.

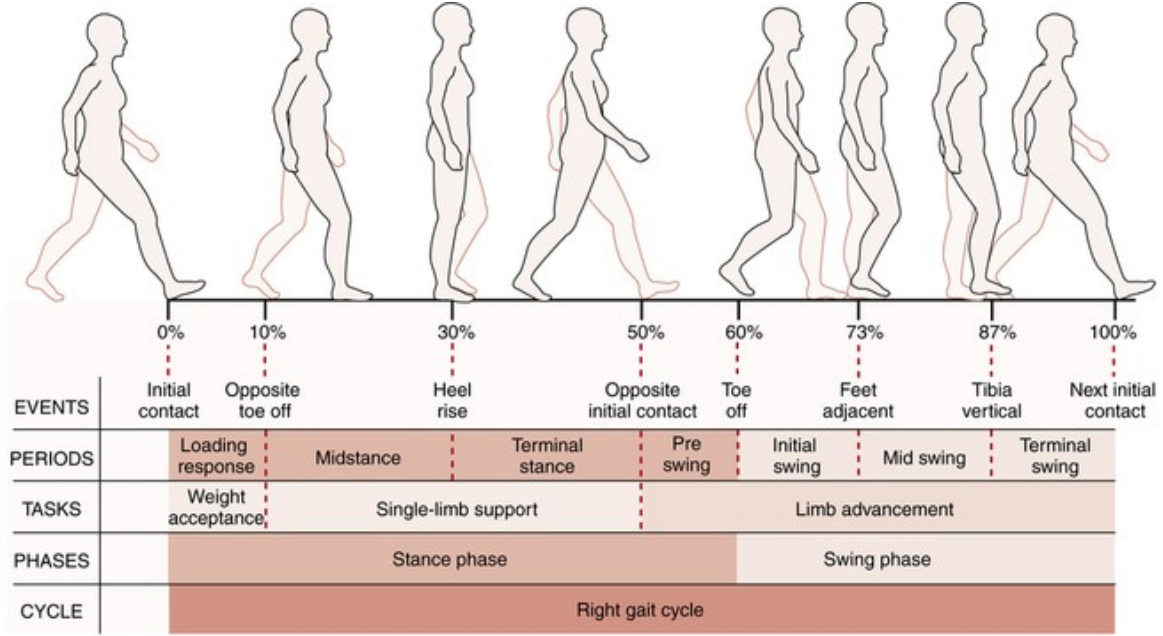


Figure 2.2: Gait cycle and events during locomotion (right foot) [131]

A full GC can be divided into two phases, stance phase and swing phase (Fig. 2.2). Stance phase refers to the period when the foot is in contact with the floor, usually occupying the first 60% of the gait cycle. The remaining 40% is the swing phase, representing the time of which the foot is lifted. While the GC describes the motion of a single leg, the GC of the contralateral limb is the same in healthy normal adults but time-shifted by 50%. The overlapping period of the stance phase in both limbs is called the double-support period, representing the phase when both feet are on the ground simultaneously.

Loading response corresponds to the double support period following initial contact (0 – 11.5% GC), which is also commonly called the ‘initial rocker’ [202]. Plantarflexion at the ankle is observed in this period as the forefoot is lowered to the ground, rapidly increasing the ground reaction force. The mid-stance period occupies approximately 20% of the GC, from opposite toe-off to heel rise. Note that sometimes mid-stance is used to describe the event when the swing phase foot passes the other foot, also referred to as ‘feet adjacent’ [46, 143]; or as defined here in this thesis,

50% stance phase. According to Perry [202], the ‘second rocker’ occurs between mid-stance and terminal-stance; Also called the ‘ankle-rocker’, it is characterised as the transitional ankle movement from plantarflexion to dorsiflexion as the shank rotates forward.

Shortly after heel-off, the ankle joint reaches maximum dorsiflexion at approximately 75% GC. Following this, the ankle rapidly plantarflexes along with tibia external rotation and forefoot supination, creating the ‘third rocker’ [202]. The pre-swing period begins at opposite initial contact, which is the second double support period lasting approximately 10% GC. This short duration is also known as the push-off phase or the propulsion. As the name appropriately suggests, the body accelerates forward as the foot act as a rigid lever [24], pushing the body forward through the power generated primarily at the ankle [154].

The swing phase starts around 60% GC as the foot begins to lift from the ground. Throughout the initial swing phase, the foot swings across the mid-line of the body acting as a ‘pendulum’ with the knee flexed to maintain ground clearance and advancement of the limb [185]. During the swing phase, the knee extends passively as the foot approaches the floor once again and the ankle progressively recovers from a plantarflexed position to a more neutral pose.

2.1.6 Temporal and Spatial Parameters

There are several temporal (time-related) and spatial (distance-related) parameters that are of particular interest in CGA. In the absence of expensive gait measurement equipment, there are four parameters that can be easily evaluated by using a stopwatch and a tape measure [215]:

- *Velocity*: The total distance between the first and last heel strikes (or foot contacts) divided by the elapsed time for the distance, measured in metres per second.
- *Stride Length*: The distance between two consecutive ipsilateral heel strikes.

- *Step Length*: The distance between two consecutive contralateral heel strikes. The side of the more forward heel strike is given credit for the step.
- *Cadence*: The number of steps taken during the sequence divided by the time elapsed for the sequence, with a unit of steps per minute.

Other parameters directly related to the gait cycle have already been mentioned in the previous Section (2.1.5), such as the time for double-support, stance phase, and single support periods, expressed in percentages GC. This information provides the simplest form of objective gait evaluation that can be used to assess the quality of a subject's gait with reported differences found between normal and pathological gait [28, 81, 168].

In gait analysis, spatial and temporal parameters are often normalised to account for variations in body size. According to the recommendations from Hof [115], quantities may be scaled using mass, gravitational acceleration, and leg length to reduce variability in gait data. This is especially important for the analysis of paediatric gait [114, 196, 205]. However, normalisation must be applied with caution when regression or correlation analysis is involved. A recent investigation by Chai & Sangeux [51] reported that dimensionless normalisation using leg length does not fully remove the correlation in spatial and temporal parameters. In the same publication, the author also highlighted the conflicting result between Dixon et al. [78] and Pinzone et al. [205] on the dependency of walking speed and leg length after data normalisation. Nevertheless, normalisation remains a valid approach to remove the dependency of physical characteristics on spatial and temporal parameters, as well as other gait measurements such as kinetic variables [174] and rate of oxygen consumption [233].

2.2 Rigid Body Kinematics

In classical mechanics, rigid body kinematics refers to the description of idealised motion between bodies whose size and shape are permanently fixed. It is assumed

that the distance between any two points of the body remains unchanged [134]. In the context of CGA, joint motions are often described using segmental kinematics by assuming the body segments act as non-deformable structures. This process greatly simplifies the calculation while being able to provide clinically relevant angles [73]. The most widely measured and reported inter-segmental angles are of the hip, knee, and ankle joints [214], with the pelvis, thigh, shank, and the entire foot (in many models, but not all) assumed as rigid bodies. Biomechanical models, such as the PiG model [63, 126], Rizzoli Foot Model (RFM) [143], and OFM [42, 250] are amongst the most widely used and validated biomechanical models in tracking and modelling these assumed ‘rigid body segments’ in CGA[14, 146].

The main objective of segmental kinematics is the collection of data that allows the reconstruction of its motion relative to another body segment (e.g. femur, tibia, hindfoot, etc.). In stereophotogrammetry, each segment is tracked using a marker cluster, also called the *marker cluster technical frame* or *technical markers set* [53]. Three non-collinear technical markers are required to fully describe a segment’s position and orientation in three dimensions. In this thesis, ”*technical markers*” are referred to and defined as the markers directly tracked by the motion capture system, that is used to provide further information for the construction of the segment coordinate system. The positioning of these markers is designed to be highly visible to the motion capture system, move as little as possible relative to the underlying bone, and can be identified by palpation in a repeatable manner. Della Croce [66] and Cappozzo [38] have identified a number of anatomical landmarks that are suitable for this purpose.

Since the technical markers are selected based on practicality, this implies that the technical frame does not normally represent the orientation of a body segment. Instead, another coordinate system, called the anatomical frame, is constructed for this purpose. A coordinate transformation is required to determine the anatomical

frame from a technical frame, most often, through a fixed (non-time-varying) calibration procedure [38]. For the purpose of interpretation and relevance to the functional anatomy (See Sec. 2.3.1), inter-segmental kinematics are usually described between two anatomical frames. Note that technical markers directly placed on the skin are often used to define an anatomical frame to economise the number of markers used and simplify the calculations, as seen in the OFM [42] and the RFM [143].

2.2.1 Euler Angles

There are various representations of three-dimensional joint movements based on rigid body kinematics [49, 101, 274]. Euler angle representation of (relative) orientation is probably the most widely used as seen in many biomechanical models [65, 135, 161, 211, 259]. Euler angles can be expressed in terms of a rotation matrix R_{ijk} (Eq. 2.1) that is parameterised by three independent rotational components that map one segment coordinate system to another, based on the Euler rotation theorem.

$$R_{ijk} = R_i(\varphi_x) R_j(\varphi_y) R_k(\varphi_z) \quad (2.1)$$

where i , j , and k denotes the rotation axes. For an xyz rotation sequence with the component matrices are defined as:

$$R_i(\varphi_x) = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos \varphi_x & -\sin \varphi_x \\ 0 & \sin \varphi_x & \cos \varphi_x \end{bmatrix}; \quad R_j(\varphi_y) = \begin{bmatrix} \cos \varphi_y & 0 & \sin \varphi_y \\ 0 & 1 & 0 \\ -\sin \varphi_y & 0 & \cos \varphi_y \end{bmatrix};$$

$$R_k(\varphi_z) = \begin{bmatrix} \cos \varphi_z & -\sin \varphi_z & 0 \\ \sin \varphi_z & \cos \varphi_z & 0 \\ 0 & 0 & 1 \end{bmatrix}; \quad (2.2)$$

The non-commutative nature of matrix multiplication means that the joint kinematics represented by Euler (if $i \neq k$) or Cardan angles (if $i = k$) is sequence-dependent. This method also suffers from Gimbal lock (when $\varphi_y = \pm \pi/2$) [273], as one of the axes is driven into a parallel configuration with another axis, locking the rotational system into a two-dimensional system. Nevertheless, Euler angle representation remains a popular choice due to its simplicity. Furthermore, with a suitable choice of axes and rotation sequence, anatomical movements are readily interpretable.

The use of Cartesian coordinate systems and the Euler angle conventions is influenced by the promotion from the standardisation committee of the International Society of Biomechanics led by Dr Ge Wu and Prof Peter Cavanagh, who recognised that the Cartesian coordinate system is the most appropriate system to describe kinematic data [278]. Even earlier, Lewis and Lew [155] identified the Cartesian based Euler angles as being clinically relevant, referred to them as the orthopaedic angles. However, they also acknowledged the possible confusion due to the many sequences (e.g. X-Y-Z, Z-X-Y) that can be used. To address this issue, Wu et al. [280] later proposed that the joint coordinate system (as discussed below in Sec. 2.2.3) be used as the standard method of computing joint kinematics.

2.2.2 Helical Angles

The use of helical angles to represent biomechanical angles was proposed by Woltring as an alternative method to overcome the disadvantages of Euler angle representation [273, 274]. Helical angle representation of joint motion describes the rotation (θ) of a body about and translation (d) along an instantaneous axis $\vec{n}$ (Fig. 2.3). Due to its ease of computation, much biomechanical research today utilises the helical axis system to calculate segmental kinematics [6, 26, 219, 280]. However, most biomechanical models and foot models do not choose this representation as it is not translatable to anatomical terms of motion, making physical interpretation difficult in the clinical

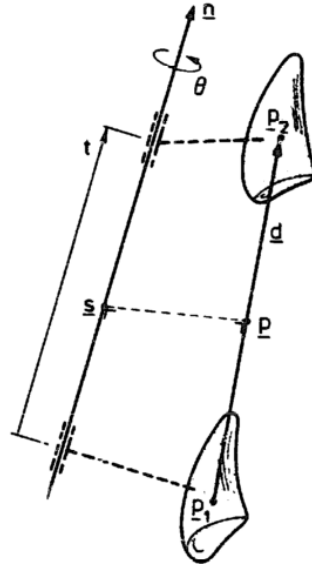


Figure 2.3: Description of motion using a finite helical axis that represents a displacement ($\vec{d}$) from p_1 to p_2 . The motion can be characterised by a translation along the direction of a unit vector ($\vec{n}$), followed by a rotation (θ) about $\vec{n}$, adopted from Woltring et al. [274].

setting [73]. Moreover, the helical system is also known to be sensitive to noise and measurement error [64, 274].

2.2.3 Joint Coordinate System

Grood and Suntay in 1983 [101] proposed a clinically relevant joint coordinate system (JCS) for the knee joint that is currently adopted in many MSFMs [42, 125, 130, 143, 156, 250, 276]. This system utilises two segment embedded axes and a “floating axis” that is mutually perpendicular to other two. While the original paper presented the JCS system specifically for the knee joint, it can be adapted to describe the kinematics of other joints [55].

The advantage of using such a system is that the joint orientation is identical to those described for Euler angles and overcomes the issue of sequential dependency. In addition, JCS could provide a more clinically relevant physical interpretation. As Nigg & Herzog note [191]: “For a given joint orientation, the sequential nature

of the Cardan angle approach implies that a movement from the neutral position is occurring to obtain the joint orientation of interest, which is not necessarily the case. The JCS on the other hand, is conceptually an actual representation of a specific joint orientation". The ease of computation, equivalence to the Euler system, and clinical relevancy made the JCS one of the most adopted methods to calculate joint kinematics in MSFMs [73].

2.2.4 Summary

In summary, each method of describing joint motion has its own virtues. The selection of each approach will depend on the research (or clinical) question and application. Despite receiving much less attention in representing three-dimensional joint angles, other methods such as projected angles [143, 240] and quaternion rotation [254] are also available and can be advantageous in the right circumstances.

2.3 Foot Anatomy and Biomechanics

The human foot is a highly complex structure that has evolved through the necessity to perform different tasks, such as walking, running, and turning. The ability to perform any functional tasks is a result of joint movements between different rigid bones that is controlled by muscles and supported/or restricted by other soft-tissues. With more than 28 irregularly shaped bones, 30 synovial joints, 30 muscles, and over 100 tendons and ligaments involved, the foot and its biomechanics are not easy to describe, with many areas lacking consensus and harbouring controversies [104].

The primary aim of this thesis is to seek the most appropriate manner to represent the forefoot in gait analysis. To achieve this, the prerequisite of this study is to understand the dynamic behaviour of the foot during human locomotion. Therefore, an overview of foot anatomy and biomechanics is presented here. This section will

begin by setting out the common terminologies and reference planes used to describe human movement, followed by an overview of the functional anatomy of the foot bones, joints, soft-tissues, and mechanical models used to describe foot biomechanics.

2.3.1 Basic Anatomical Terminologies

The study of human biomechanics uses anatomical terms when describing a particular motion. All anatomical descriptions are expressed in relation to a standard anatomical position, in which a person is standing upright, with their upper limbs by their sides and their feet parallel.

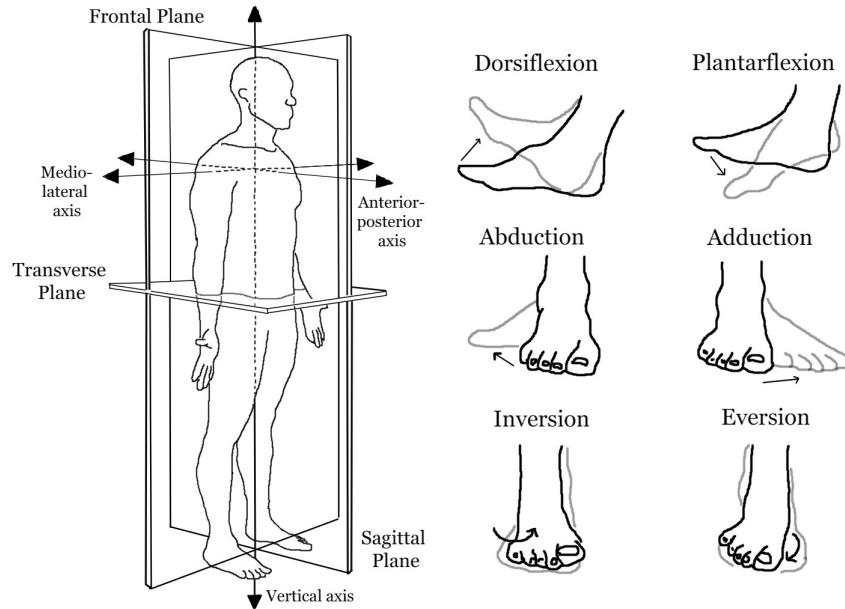


Figure 2.4: Anatomical planes (left) and foot motion terminologies (right), adapted from Nordin & Frankel [192].

Directional Terms

Directional terms are grouped in pairs of opposites based on the standard anatomical position (Fig. 2.4):

- *Superior/inferior*: Superior means above and inferior means below.
- *Anterior/posterior*: Anterior means towards the front (chest side) and posterior towards the back.

- *Medial/lateral*: Medial means towards the midline of the body and lateral means away from the midline. *Ipsilateral* means on the same side and *contralateral* on the opposite side.
- *Proximal/distal*: Proximal means closest to the point of origin or the trunk of the body, distal means furthest away.

Anatomical Reference Planes

There are three anatomical planes (Fig. 2.4):

- *Sagittal plane*: The plane that divides the body into a right and a left part.
- *Coronal (or frontal) plane*: The plane that divides the body into a front and a back part.
- *Transverse plane*: The plane that divides the body into an upper and a lower part.

Foot Motion Terminologies

- *Flexion and extension*: These motions take place in the sagittal plane. Flexion describes a decrease in angle relative to the proximal segment, and extension is in the opposite. In the ankle, these movements are referred to as *dorsiflexion* and *plantarflexion*, where the toes are brought close to or away from the shin respectively.
- *Abduction and adduction*: These motions take place in the frontal plane. Abduction occurs when the distal segment moves away from the midline, relatively to the proximal segment, and adduction is the opposite.
- *Inversion and eversion (or varus/valgus)*: Inversion (varus) describe an angulation of a joint towards the midline, and eversion (valgus) the opposite.
- *Pronation and supination*: Pronation is a tri-planar motion of the subtalar joint (eversion + dorsiflexion + abduction), moving the foot up and away from the centre of the body. In the opposite, supination (inversion + plantarflexion + adduction) describes the movement of the foot down and towards the centre of the body. Note that different terminologies may be encountered in the literature.

2.3.2 Bones of the Foot

There are in total 28 bones in the foot which can be thought of as being divided into four anatomical parts [154, 176], namely:

- The **hindfoot** consists of the talus and calcaneus.
- The **midfoot** consists of the three cuneiforms, navicular, and cuboid.
- The **forefoot** consists of the five metatarsals.
- The **toes** consists of five sets of phalangeals composed of sixteen individual bones, including the two sesamoids.

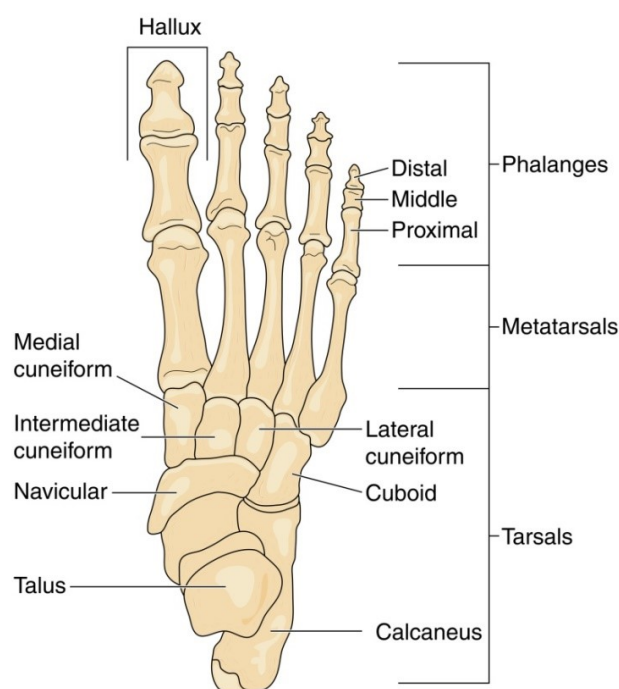


Figure 2.5: Bones of the right foot, with sesamoids hidden from sight under the hallux [1]

Such groupings are useful to help us understand how the function of the foot relates to its structure. However, it is important to realise that there is no consensus on the terminology for the groupings. Some anatomist groups consider the forefoot as only the metatarsus, whereas others define forefoot as a combination of the metatarsals

and the phalanges [104]. In this thesis, the forefoot is defined as only consisting of the five metatarsals, as commonly referred to in the field of CGA [210].

The talus is located just above the calcaneus, which together forms the hindfoot. Its superior surface (ankle joint) articulates above and medially with the tibia and laterally with the fibula. The calcaneus forms the heel of the foot and lies below the talus. It is the largest bone of the foot and articulates with the talus through the subtalar joint.

The navicular bone is situated on the medial side of the foot and forms a proximal articulation with the talus and distal articulation with the three cuneiform bones. In addition, it also articulates laterally with the cuboid bone. It is an important part of the medial longitudinal arch which plays a significant role in the even distribution of body weight and propulsion [192]. The cuboid is positioned laterally to the cuneiforms and the navicular. Anteriorly, the cuboid articulates with the fourth and fifth metatarsal and posteriorly with the calcaneus.

In the forefoot, five metatarsals lie approximately parallel to each other. Proximally, the three medial metatarsals articulate with the three cuneiform bones, whereas the two lateral metatarsals articulate with the cuboid. Metatarsals are long, convex in shape, and they provide the foot with its characteristic arch. The first metatarsal is the shortest and thickest. The second, third, and fourth metatarsals are thinner and their bases are wider dorsally than on their plantar sides. The fifth metatarsal bone differs from the others in that it has a tuberosity on the lateral side of its base. The head also appears to be more spherical than those of the second, third, and fourth metatarsals [124]. The function of the metatarsals includes creating the supporting structure of the foot as well as stabilisation in mid-stance [2].

Lastly, the phalanges are the bones of the toes, with two in the first digit and three in the rest. All five digits articulate with the distal head of the metatarsals at the metatarsophalangeal joint and within them, interphalangeal joints with a mostly

dorsi/ plantarflexion motion. The toes are regarded as a forward extension of the foot with the function of maintaining balance as the body's centre of gravity changes. They also act as an important force transmission mechanism during the propulsion phase [120] and assist adaptation of the foot to different terrains.

2.3.3 Joints of the Foot

The human foot is a highly complex structure with more than 30 individual joints, with each contributing to the overall motion of the foot at different stages during gait [217] through a combination of active motion and passive mechanisms that support stability [149, 248]. This section presents the biomechanics of some of the key joints with a focus on the kinematic and functional aspects.

Ankle Joint (Talocrural Joint)

The talocrural joint can be classified as a synovial hinge joint that moves primarily in the sagittal plane at the distal tibiofibular joint and the trochlea of the talus. An earlier anthropometric study by Isman & Inman [123] recognised that the ankle joint axis can be approximated as a single axis through the two malleoli and the transverse talus. Because the medial malleolus lies slightly anterior and superior to the lateral malleolus, the axis also has an oblique component so that some abduction and eversion accompanies dorsiflexion while adduction and inversion occur in plantarflexion. Whilst the single-axis model was adopted in many investigations as reported in a comprehensive literature review on ankle motion by Leardini et al. [149], others have considered the ankle joint with two axes of rotation (one plantarflexion axis and one dorsiflexion axis) that changes approximately at the neutral position [16, 111]. Subsequent studies have also represented motion of the ankle joint as an instantaneous helical axis that continuously changes throughout flexion [148].

The primary function of the ankle is to act as a pivoting point to allow plan-

tar/dorsiflexion at various stages during gait to achieve clearance and propulsion. The largest motion occurs in the sagittal plane, with plantarflexion range of motion (ROM) from 23° to 56° , and a dorsiflexion ROM from 13° to 33° [149]. However, more recent in-vivo cadaveric studies utilising bone-pins have also shown evidence of movements in other planes, with a mean ROM of 8.1° in the frontal plane and 7.8° in the transverse plane observed by Lundgren et al. [160] and a corresponding value of 15.3° and 10.0° by Nester et al. [189].

Another important function of the ankle is to provide stability through its extensive ligamentous support on both medial and lateral sides. The ligaments that surround the ankle limit plantar/dorsiflexion, anterior/posterior movement of the foot, tilting of the talus, and inversion and eversion [104]. Each of the ligaments has a specific role in stabilising the ankle depending on its orientation, type of loading condition, and the position of the ankle at the time of loading [113, 149]. For the ankle joint, the most important stabiliser is the anterior talofibular ligament [149].

Subtalar Joint (Talocalcaneal Joint)

The subtalar, or talocalcaneal joint is defined as the articulation between the calcaneus and talus superiorly and the navicular anteriorly [216]. Motion about the subtalar joint is more difficult to conceptualise compared to the ankle joint due to its lack of external landmarks and irregular shape. Many authors have described the subtalar joint axis as a single hinge joint or screw-like rotation with an orientation that runs obliquely from the posterior lateral plantar surface to the anterior dorsal medial aspect of the talus [92, 104, 123, 167, 192]. Manter [167] considered the joint axis as having one degree of freedom with an axis oriented upward by 42° and anteriorly and medially by 16° , with an average helical angle ROM of 12° .

However, some authors have rejected the view of a single axis representation. In two publications, Lundberg and Svensson [158, 159] provided evidence of a parallel translation of the joint axis and criticised the single-axis treatment. A subsequent

study by Leardini et al. [150] confirmed this viewpoint by demonstrating that the subtalar joint axis moves along a repeatable path.

Due to the oblique orientation of the joint axis, subtalar joint motion has often been described as pronation and supination, as described in Section 2.3.1. Functionally, it has been recognised that the movement of the subtalar joint is essential for load transmission from the foot to the leg in early stance through a coupling mechanism with the ankle joint and tibia rotation [111, 277]. It is worth noting that even in modern CGA, it is almost impossible to distinguish between the ankle and subtalar joint motion because the talus cannot be tracked using skin markers, and therefore it is often referred to as a combined movement of the ‘ankle/subtalar complex’ [154].

Midtarsal Joint (Transverse Tarsal Joint)

Themidtarsal joint, also known as the transverse tarsal joint or Chopart’s joint, consists of two joints, the calcaneocuboid joint on the lateral side and the talonavicular joint on the medial side of the foot [139]. In combination, the two joints form an S-shaped joint line whose motion is often presented as a two-axis system, one oblique and one longitudinal (Fig. 2.6). According to Manter [167], the longitudinal axis is oriented 15° upward from the horizontal and 9° medially from the long axis of the foot (from calcaneus to the second metatarsal head) and gives a ROM of about 10° . The oblique axis is oriented 52° upward from the horizontal and 57° anteromedially. The inversion/eversion movement occurs primarily about the longitudinal axis and flexion/extension about the oblique axis. However, the upward inclination the longitudinal axis also implied that the motion of abduction and adduction is coupled with eversion and inversion respectively [167].

The movements of the subtalar andmidtarsal joints are known to be a coupled dynamic mechanism that is of a great functional significance [192]. Pronation and supination about the subtalar axis cause the midfoot to transit from a rigid formation to a flexible configuration. During loading response, the subtalar joint moves into a

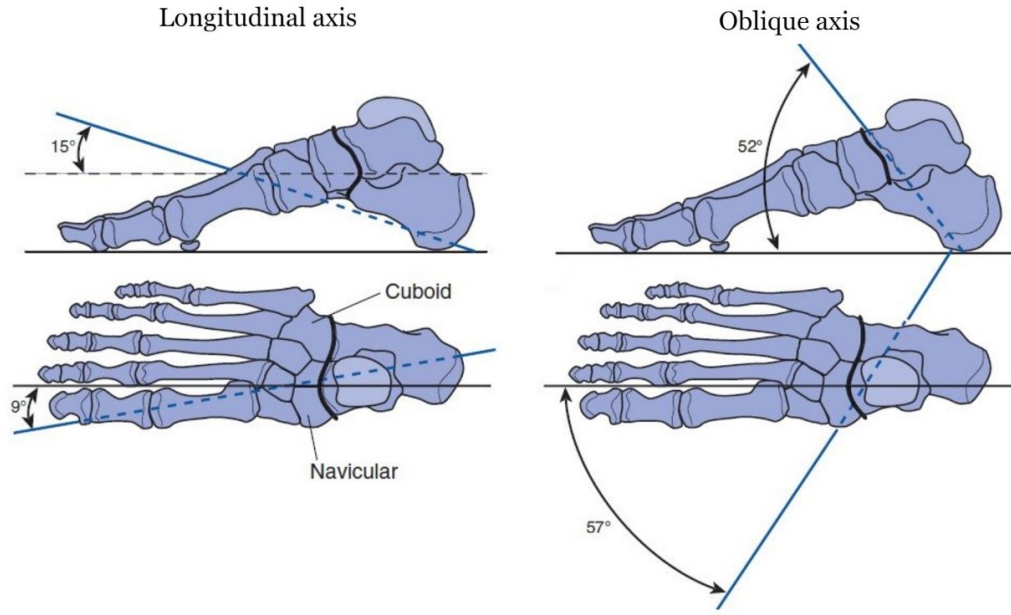


Figure 2.6: Two-axis representation of the midtarsal joint motion, the longitudinal axis (left) and the oblique axis. Adapted from Norton & Frankel [192].

pronated position, driving the calcaneocuboid and the talonavicular joint axes into a more parallel configuration [91]. As a consequence, more joint motion is allowed due to the loosely packed formation of the mid-foot bones, transforming the foot into a mobile adaptor to accommodate landing, particularly in case of uneven terrains. As the subtalar joint inverts, the axes of the two midtarsal joints becomes skewed, locking the midtarsal joint and thus providing rigidity to the midfoot, transforming the foot into a lever which allows effective propulsion and force transmission in terminal stance. This phenomenon is known as midfoot lock. Blackwood [24] reported an increased sagittal ROM of the metatarsals when the calcaneus is everted compared to inversion, consistent with the earlier observations by Elftman [91] and Manter [167] on the coupling between the midtarsal and the subtalar joints. However, a recent study by Okita et al. [195] questioned the claim of a “rigid joint” producing a rigid lever during push-off as a result of an inverted subtalar joint. In their in-vitro investigation, the midtarsal joints were observed to undergo gradual dorsiflexion and inversion during

propulsion, indicating a compliant mid-foot structure during terminal stance.

Intertarsal and Tarsometatarsal Joints

The intertarsal joints consist of the three cuneinavicular joints, the intercuneiform joints, the cuboideonavicular joints, and the cuneocuboid joint [192]. The motion of these joints is not typically described [111, 172] due to their highly packed configuration and limited movement between each other from the inter-connected ligamentous network. The tarsometatarsal joint, also known as the Lisfranc joint (although it is technically not a joint [140]) marks the border between the forefoot and the mid-foot bones. It is a highly stable structure and again with very a limited motion. An in-vitro cadaveric study by Ouzounion & Shereff [198] showed that motion of the first and second metatarsals is significantly smaller than that of the fourth and fifth, with a mean dorsi/plantarflexion ROM of 10.2° at the cuboid-fifth metatarsal, 9.6° in cuboid-fourth metatarsal, 3.5° for cuneiform-first metatarsal, and 0.6° for cuneiform-second metatarsal. In CGA, the intertarsal and tarsometatarsal joints are often treated as a transmission mechanism [42, 250] and it is almost impossible to measure their motion.

Metatarsophalangeal Joints

Each of the five metatarsophalangeal joints is composed of a convex metatarsal head and a concave proximal phalanx. The convex-shaped distal metatarsal head guides the motion of the concave proximal phalanx ends, allowing a gliding motion primarily in the sagittal plane. Although there are in total five individual joints, most MSFMs only measure the motion of the hallux [42, 65, 135, 144, 187, 212] due to interest in the windlass mechanism [110] and clinical applications for disorders such as hallux valgus [69, 197] and hallux rigidus [33, 184].

In a cadaveric study involving fifteen specimens, Shereff et al. [238] measured the sagittal plane motion of the first metatarsophalangeal joint to be an average of 111° ,

accompanied by a small motion in the transverse plane as shown by the translation of the joint axis. Measurements made later by Hopson et al. [116] in an in-vivo study with twenty-six adult participants also obtained similar values using a variety of techniques and loading conditions. This sagittal plane motion is crucial to the force transmission during the propulsion phase and adaption to uneven terrains. Furthermore, the windlass mechanism, which will be discussed shortly in the subsequent subsection (See Sec. 2.3.4), is fundamental to our current understanding of foot biomechanics and is only possible through motion of this joint.

Contributions of the Soft Tissues

The motion of the joints mentioned here above is controlled through a combination of twelve extrinsic and nineteen intrinsic muscles, supported (and limited) by a network of ligaments and connective tissues to provide stability. The activation of muscle groups during gait and their function in controlling specific movements is well-documented in many biomechanics textbooks, including Nordin & Frankel [192], Levine et al. [154], and Baker [11]. The passive role of ligaments in the foot and ankle has also been studied extensively [149, 248], and is also described in the aforementioned textbooks. It is acknowledged that the understanding of the muscle and soft-tissues mechanics is crucial to gait analysis [253]. However, it is not within the scope of this thesis to provide a comprehensive review of the roles of each individual anatomical structure. Nevertheless, the contribution of soft-tissues will be discussed in the thesis where relevant.

2.3.4 Arches of the Foot

There are three important arches in the foot, one in the transverse and two in the longitudinal directions [128]. The medial longitudinal arch (MLA) is comprised of the calcaneus, talus, navicular, medial cuneiform, and the first three metatarsals [192],

and has been credited with the function of load-bearing, elastic energy storage, and stabilisation of the foot throughout gait. The lateral longitudinal arch (LLA) lies in contact with the ground via soft-tissue, and comprises the calcaneus, cuboid, and the lateral two metatarsals. The transverse arch span across the heads of the five metatarsals and is relatively flat in the neutral standing position. The definition of these three arches is based on the physical observation of a concave arrangement of the bones across the plantar aspect of the foot. While some anatomists and textbooks have defined the three arches as mentioned above [192, 104], in practice, measuring and tracking of the MLA and LLA often uses only the location of the first and fifth metatarsals respectively [40, 270]. Despite lack of a precise anatomical definition, the characteristics of these arches have substantial biomechanical and clinical significance, particularly the MLA [40, 224, 270].

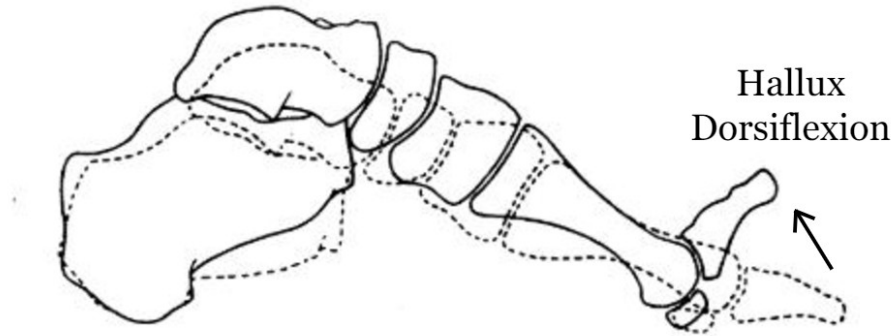


Figure 2.7: Tracings from radiographs of a foot in neutral standing position. Dotted line: hallux flexed with MLA lowered. Adapted from Hicks et al. [112]

A classical research study by Hicks et al. [110] described the arch as a ‘windlass mechanism’, where the plantar fascia contributes to the stability of the MLA through a tightening process (Fig. 2.7). According to Hicks the phenomenon of the arch rising and tightening is a result of the passive extension of the big toe at the metatarsophalangeal joint, along with supination of the posterior part of the foot, lateral rotation of the leg, dorsiflexion of the hindfoot relative to the tibia, plantarflexion of the fore-

foot relative to the hindfoot, and the tightening of the plantar aponeurosis [112]. The importance of the plantar fascia was later confirmed in an investigation by Huang et al. [119] on the stability of the MLA through sequential sectioning of the different ligaments. Murphy et al. [181] and Thordarson et al. [258] have also observed the collapse of the MLA after plantar fascia release.

Several other functional models have been used to describe the MLA mechanism. Kapandji [128] described the three arches as a triangular plantar vault, where the joint, ligament, and muscles are blended into a unified system that allows for adaptation to uneven terrain due to its structural curvature and elasticity. Sarrafian [229] have described the MLA using a truss model, where the arch is modelled as a triangular structure with the tie rod (i.e. plantar fascia) connecting (in tension) the two ends of the base. Despite the differences in analogy, all models accept the role of MLA in shock absorption, energy storage, and load transfer, all of which are essential for the functionality of gait [41, 249]. Therefore, any pathological condition that exaggerates or flattens its curvature will inevitably affect locomotion [151, 259].

2.4 Summary

This Chapter has presented a review of the current status of CGA, including the historical development, general methodology, equipment used, and the parameters used to characterised human locomotion. Different mathematical techniques of calculating joint kinematics, including Euler angles, helical angles, and JCS are also presented in this Chapter. Lastly, a comprehensive overview of our current understanding of foot biomechanics is also included. The human foot is a highly complex structure controlled by an intricate system of bones, muscles, and soft-tissues, with each anatomical component playing an essential role in locomotion. Dysfunction of any one of them could damage the entire biomechanical system due to the inter-dependency of one

anatomical part to another. The development of CGA and biomechanical models have allowed clinicians and engineers to gain insight into patient-specific function that enables more accurate diagnosis and personalised treatment. However, it remains unclear how the foot should be represented in an MSFM and how the foot motion should be described in the most anatomically, functionally appropriate way that is also clinically feasible and practical. The following Chapter explores some of the MSFMs used in CGA, their unique design features, and their associated validation and clinical studies.

Chapter 3

Multi-segment Foot Models: An Overview of Design and Application

The fundamentals of clinical gait analysis (CGA) and foot biomechanics were described in the previous Chapter. Traditionally, multi-segment foot models (MSFM) have been used to track and analyse major foot segments kinematics during gait. However, there is currently no agreement on how the forefoot structure should be best represented and how to describe the articulations of the relevant foot joints. The purpose of this Chapter is to present an overview of the existing MSFMs, including an in-depth view of those that are widely used for both foot biomechanics research and clinical applications. In line with the aim of this thesis, the second section will be dedicated to the discussion of forefoot representations in MSFMs, more specially, the current strategies in tracking the motion of metatarsal bones. Finally, this Chapter seeks to identify crucial design factors that link to the utility and efficacy of a MSFM. A number of practical and methodological limitations of the existing MSFM will also be explored.

3.1 General Overview

In CGA, human body motion is often analysed based on segmental rigid body kinematics (see Sec. 2.2). Traditionally, the foot is modelled as a single segment rigid body [62, 126, 130]. This representation is useful in identifying motion and forces of the ankle joint complex and to identify relevant pathological conditions. [85, 268, 281]. However, this approach has been met with criticism, highlighting its inadequacy in measuring the intricate motion occurring within the foot structure. Therefore, a number of MSFMs have been introduced over time.

In the context of CGA, MSFM is a marker placement protocol that can be used to track and analyse foot joint kinematics (and kinetics) of the foot by simplifying the human foot structure as a system of rigid body segments. The development of MSFMs has made it possible to capture the motion of the joints within the foot through dividing the foot into multiple rigid body segments, and this approach is becoming increasingly popular in part due to the progression in measurement technologies. Such division of the foot has been found to be very useful for the in-vivo analysis of dynamic kinematics and clinical examination of foot deformities that were previously not possible using single-segment representation [73, 208].

Several review articles have already been published on the performance and clinical utility of MSFMs. Rankine et al. [210] were the first to publish a systematic review describing twenty-four MSFMs proposed for both biomechanics research and clinical applications. Each model was classified and evaluated based on the number of segments and the joint kinematics convention (e.g. Euler sequence, joint coordinate system, or projection angles). In addition, Rankine et al. included a thorough comparison of the kinematic output of each MSFM at each phase of the gait cycle. From the article, it is not surprising to learn that there was little agreement regarding the segment definition and kinematics convention. After all, many models were

proposed not long before the publication, with only a few associated validation and clinical papers available at the time. Deschamps et al. [73] later published another systematic review that focused on the clinical application of MSFMs that have at least three segments. In total, fifteen models were reviewed and discussed based on the model’s physical characteristics, kinematic convention, and robustness from their validation and clinical studies. Both Deschamps et al. [73] and Rankine et al. [210] expressed views on the lack of consensus on both segment and kinematic descriptions, particularly with the forefoot segment.

In response to the consensus issue as described above, Bishop et al. [23] proposed five reporting standards to promote transparency and usability of MSFMs in the scientific community, as well as to allow validation studies by third parties. The five reporting standards are stated as follows:

1. Report the location, accuracy, and reliability of marker placement.
2. Report the definition of segment(s).
3. Report the definition of segment coordinate systems.
4. Report the definition of joint parameters.
5. Report the reliability of joint kinematics.

In this paper, the authors reviewed the validation experiments conducted for all the MSFMs proposed up to 2012, which revealed that most of the proposed models did not satisfy the five standards or had a high degree of ambiguity that rendered the studies irreproducible. Many MSFMs were built “in-house” and the algorithms used to process the kinematics data were convoluted and not available to the public. Because none of the MSFMs passed all five of these proposed standards at the time, Bishop’s paper recommended more robust validation and clinical studies since then [145].

It is not within the scope of this review Chapter to conduct another systematic review on MSFMs, especially considering that a thorough review was published very

recently by Leardini & Caravaggi [145] in 2018. This publication was followed closely by another review by Leardini et al. [146], explicitly aimed at clinical applications of MSFMs. Dr Stebbins, who supervises the completion of this thesis, also contributed to the aforementioned systematic review. From these surveys, it is evident that a large number of MSFMs have been proposed up to date, increased from twenty-four in 2008 [210] to thirty-nine in 2019 [146]. However, most of the models still lack evidence for clinical use and validation studies. Out of all thirty-nine, there are only a few that consistently demonstrate clinical utility, that is, the Oxford Foot Model (OFM), Rizzoli Foot Model (RFM), and Milwaukee Foot Model (MFM) [135]. Therefore, a closer inspection of these models will be provided in Section 3.2.

Another MSFM that is worth noting due to its unique modelling strategy is the Heidelberg Foot Measurement Method (HFMM) proposed by Simon et al. [240]. Instead of the usual tri-planar inter-segmental kinematics, the HFMM uses a variety of angles that are theorised to be more clinically relevant and applicable to severely deformed feet. Since the HFMM was developed based on a different modelling strategy, limited comparison could be made against other models in terms of marker placement strategy, foot segmentation, and kinematic output. Hence, an in-depth review of this model is omitted from the following Section.

3.2 Notable Foot Models

In order to identify crucial design factors for the development of a novel forefoot model, an overview of some of the most successful and extensively studied clinical MSFMs are presented in this Section. These are, as identified by Leardini et al. [146], the MFM [135], OFM [42, 250], and RFM [143], together representing 2/3 of all the clinical papers. Figure 3.1 shows the subdivision of the foot bones based on the definitions of these models. Please note that the segment definition of each model

differs to each other and our earlier bone groupings (Sec. 2.3.2). The inconsistencies between the segment definition and the actual bone tracked raises many questions, but most fundamentally, what anatomical unit does the model segment represent? Is this tracked with appropriately placed markers? Is the rigid body assumption justified? From a clinical and practical point of view, is the anatomical accuracy of a modelled segment important if pathological gait can be repeatably identified? This thesis seeks answer to these questions.

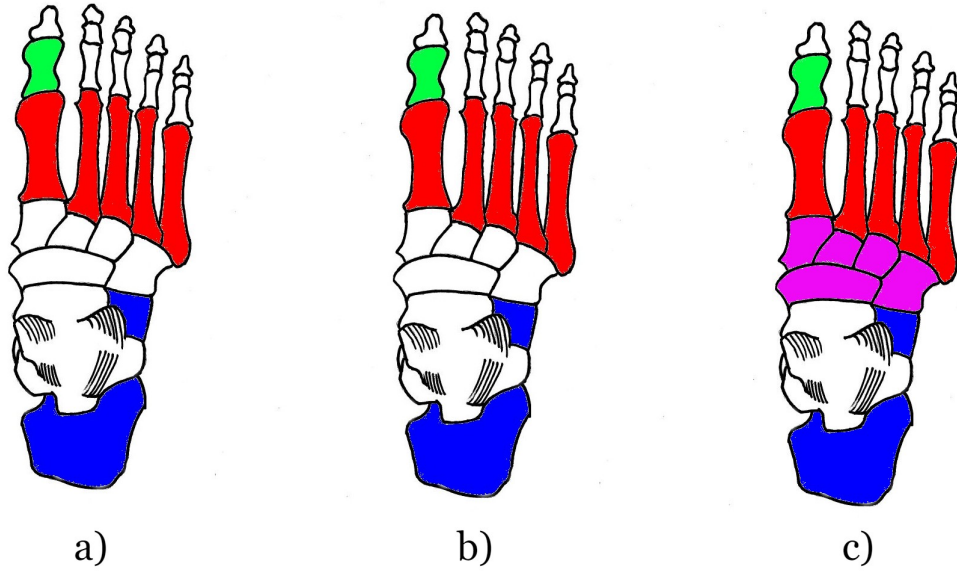


Figure 3.1: Subdivision of the foot models in a) Milwaukee Foot Model. b) Oxford Foot Model. and c) Rizzoli Foot Model in terms of foot segments tracked. The white regions are the locations without any tracking markers. Adapted from Leardini & Caravaggi [145].

3.2.1 Milwaukee Foot Model

Of the three aforementioned models, the MFM proposed by Kidder et al. [135] was the earliest. The MFM is composed of tibia, hindfoot, forefoot, and hallux (Fig. 3.1 a) and is tracked using twelve markers attached to the anatomical landmarks of interest on each foot. In the original paper, the authors explicitly defined the combination of

bones represented by each segment, with the shank segment composed of tibia and fibula; the hindfoot segment composed of calcaneus, talus, and navicular; the forefoot composed of cuneiforms, cuboid, and metatarsals; and the hallux just the proximal phalanx of hallux. However the segmentation defined by the author is different to the actual segment tracked (Fig. 3.1). In addition, it should be noted that no markers were attached directly to the midfoot bones. Moreover, later in-vivo and in-vitro foot kinematic studies using intercortical pins have provided evidence that suggests this representation may not be ideal, as a significant motion was found between calcaneus and mid-foot, and between midfoot and forefoot [186, 188].

The MFM employs the Euler sequential rotation convention to compute the kinematics of the foot as tri-planar joint motions, a standard approach utilised in many existing foot models [210]. To date, several clinical studies have utilised the MFM to analyse pathological gait ranging from foot deformity to the arthritic foot. Canseco et al. assessed post-surgical outcome of patients with hallux rigidus [33] and characterised the pathological gait of hallux valgus [34] using this model. Khazzam et al. applied this model in the study of gait performance in patients with a degenerative joint disease [132], and later, analysed the effect of rheumatoid arthritis on the overall foot motion [133]. More examples of clinical studies using the MFM can be found in a recent review by Leardini et al. [146].

As one of the earliest developed MSFMs, there are several methodological limitations associated with the MFM. The model has been criticised for its reliance on radiographic calibration to establish the underlying foot anatomy relative to the positions of the technical frames of the segments [193]. Although this approach can be seen as advantageous in correcting the model's alignment, additional equipment is required which may not be available in a standard gait laboratory [21] and it also exposes subjects to ionising radiation. Furthermore, a more robust repeatability study is required since the original paper only reported the gait for one adult male subject.

A subsequent validation study by Myers [183] recruited only three children to study paediatric gait. However, it is worth noting that many original papers proposing new models often have a small sample size, and this is acceptable as a proof of concept.

3.2.2 Oxford Foot Model

The original OFM is a three-segment MSFM proposed by Carson et al. [42] in 2001 composed of hindfoot, forefoot, and hallux (and shank). Stebbins et al. [250] subsequently made some modifications to make the model more applicable to children with deformed feet, as well as to improve the tracking and reliability of the model by better position the markers for visibility and practicality purposes. The OFM hindfoot is defined as the calcaneus and talus, the forefoot composed of all five metatarsals, and the hallux is just the proximal phalanx (Fig. 3.1 b). The OFM requires a total of eighteen markers on each foot, five more than the MFM. Although only three markers are required to define and track each rigid segment in stereophotogrammetry, having one extra marker allows for reconstruction if any one of them is missing due to technical reasons, such as occlusion, system error, or otherwise.

Additional markers are used in the static trial to define the anatomical frames. Once static calibration is completed, these extra markers are removed and reconstructed as virtual markers in the dynamic trials. This eliminates the need for additional calibration equipment and reduces the number of tracking markers. Both the hindfoot and forefoot static anatomical frames are aligned to the floor during static calibration. Novak et al. [193] argued that this approach might hinder the model's ability to distinguish pathological gait in the event of foot deformity. A similar argument can be made if "neutral reference angle offset" is used to "zero" the kinematic angles. This, however, is not implemented in the OFM protocol. Although zeroing joint kinematics has the advantage of improved repeatability and reduction in marker placement error [118, 276], a static standing posture is not always possible for patients

with a foot deformity.

Originally, the OFM was developed as a general model for both normal and deformed feet. Most of the clinical applications have been in children with foot deformities, such as cerebral palsy or clubfoot [255, 256]. However, over time, the range of disorders examined using the OFM has been extended to rheumatoid arthritis [262, 263], flatfoot [117, 138] and many other pathologies [146]. To this date, the OFM remains one of the most widely-used MSFMs available, and this can be attributed to the extensive validation studies and to its implementation in the Vicon motion capture software [266]. The repeatability of the OFM kinematic output have been reported for different age groups [61], pathological conditions [171], motion tasks [15, 157], and in different laboratories [190].

One of the main criticisms of the OFM is that it did not include a mid-foot segment [65, 73, 146, 193, 143], which is necessary if conditions such as mid-foot break were to be examined [170]. The representation of the forefoot as a single segment is also questionable; more on this topic will be discussed in Section 3.3. In summary, a general overview of the OFM is presented in this sub-section. Since this model was used as a platform for the development of a novel forefoot model in this thesis, detailed calculations required for this method have been included in the Methodology Chapter (See Sec. 4.6.2).

3.2.3 Rizzoli Foot Model

The RFM is a three-segment MSFM (plus hallux in 2D) proposed by Leardini et al. [143] that measures tri-planar motion between the three-dimensional segments as well as functional angles commonly measured in the clinics. In this model, the hindfoot comprises the calcaneus; the forefoot is composed of all five metatarsals; the midfoot represents the navicular, cuboid, and all three cuneiforms; and the hallux is just the proximal phalanx of the hallux (Fig. 3.1 c). In addition to the tri-planar

angles commonly measured between two independent segments, RFM also measures the projection of the first, second, and fifth metatarsal vectors to the ground, as well as relative to each other in the forefoot transverse plane (i.e. the planar and functional angles) (Fig. 3.2 b). The dynamic change in medial longitudinal arch (MLA) and navicular drop can also be determined using this protocol.

The RFM adopted the joint coordinate system as recommended by the International Society of Biomechanics [279]. In contrast with the OFM, the RFM applies a kinematic offset (if applicable) using the neutral static pose to all of its kinematic angles. However, the planar and functional angles are not adjusted, which is useful for comparing to clinical radiographic measurements [207].

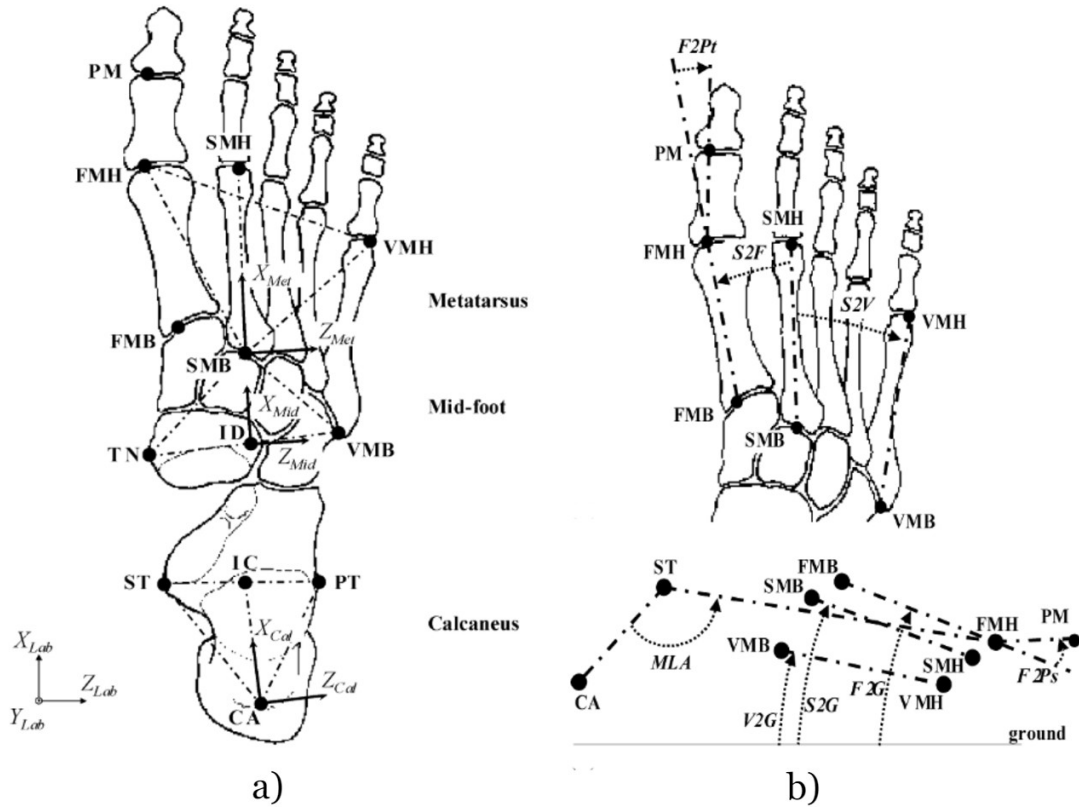


Figure 3.2: Schematic of a) Rizzoli Foot Model segment definition and b) functional and planar angle in the forefoot transverse plane (top) and sagittal plane (bottom). Adapted from Leardini et al. [143]

There have been extensive methodological studies on the RFM. Repeatability of the kinematic parameters have been assessed in both regular [39, 71] and pathological gait [70, 207]. In addition, the model was also validated for applications in paediatric gait analysis by Mahaffey et al. [162], highlighting its superior performance over other foot models. A standard error of measurement higher than 5° was found in 15.2% of the kinematic angles in RFM in contrast to the 26.2% in OFM. The RFM was also validated for analysing gait in populations over 50 years old [7].

The RFM has demonstrated applications in both clinical and foot biomechanics research. Portinaro et al. [207] have characterised pes planus using a modified version of this model. The foot kinematics of patients with chronic plantar fasciitis was investigated using the RFM by Chang et al. [48], who reported an increase in hindfoot eversion and peak first metatarsal phalangeal joint angle. Other applications include the developmental study in typically developing boys [72] and studies of foot motion in dancing [43]. More applications can be found in the previously mentioned review articles [73, 145, 146].

3.2.4 Model Comparison

There is currently an overabundance of MSFMs, and comparison between them is often difficult due to the differences in segment definition, kinematic convention, and the use of reference postures and offsets. However, there have been several comparative studies aimed at evaluating the relative performance of the existing MSFMs in different scenarios, usually in the form of a repeatability analysis. Mahaffey et al. [163] have compared the repeatability of the OFM, RFM, and Kinfoot [161] in 17 children using intraclass correlation coefficients. The percentage of kinematic parameters with a standard error of greater than 5 deg was used as an indication of the overall performance, of which a value of 15%, 26%, and 44% was reported for the RFM, OFM, and Kinfoot respectively. While the RFM appeared to be more repeatable than

the other two, all MSFMs demonstrated a good level of repeatability in the analysis of the paediatric foot. In another comparative study by Di Marco [74], this time with thirteen adult subjects with an average age of 27, the RFM also demonstrated better repeatability over OFM and the modified Shriners Hospitals for Children Greenville Foot Model (mSHCG) [228], on par with the Sawacha model [230]. Two types of variability index were used in this assessment, the medium absolute deviation and maximum difference. A more recent study by Nicholson et al. [190] performed a comparative study with a number of MSFMs, comparing the duPont [109], HFMM, OFM, RFM, and mSHCG in ten adults averaging 19.8 years of age. All MSFMs demonstrated a similar level of repeatability based on intra-rater, inter-session, and inter-trial standard deviations, with a few exceptions. The RFM appeared to be the least repeatable in the hindfoot and forefoot sagittal plane motion, but this was only observed in one rater and not the other, which could be attributed to the inexperience in marker placement. As noted earlier, repeatability could be improved by introducing a neutral offset, with a possible trade off of reduced clinical utility. Overall, there is insufficient evidence to rank MSFM's performance based on repeatability.

While it is useful to compare repeatability in order to rank or select the best MSFM, it is important to realise that their relative performance depends on the intended use and the research or clinical questions that the user is trying to answer. The selection of an appropriate model also depends on the usability and technical restrictions. In a gait laboratory with a lower budget, MSFMs with a fewer number of markers may be better than some of the more complex protocols that require a higher camera resolutions. In the case of foot deformity, the choice of neutral reference posture alignment and the application of joint offset could also affect clinical interpretation and thus must be carefully considered.

3.3 Forefoot Representation in MSFM

3.3.1 Single-Segment Representations

The majority of MSFMs represent the forefoot as a single unified rigid body segment [145]. MFM [135] was the first model to represent an independent forefoot segment tracked using three markers attached to the distal heads of the first metatarsal and fifth metatarsal and the proximal base of the first metatarsal. Based on the same marker placement protocol, Rattanaresert [212] proposed another model with a forefoot segment tracked using markers placed on the same locations as the MFM. However, instead of using radiography, the anatomical frame was defined using the static position aligned with a spirit-levelled sliding calliper. Both Kidder and Rattanaresert's studies utilised an anatomical alignment procedure to define the anatomical frame of the forefoot. However, neither papers provided the information on how their MSFM's segments are tracked and defined.

In the interest of investigating general foot dysfunction, Cornwall & McPoil [58] proposed a foot model that utilised electromagnetic sensors instead of passive markers. The forefoot segment was tracked by attaching one sensor to the medial aspect of the first metatarsal shaft, leaving the second through fifth metatarsals untracked. Another model that only tracked the motion of the first metatarsal is the Leardini Foot Model I (LFMI) [148]. In this model, a cluster-based marker plate was attached directly to the shaft of the first metatarsal. The movement profile of the first metatarsal in the Cornwall & McPoil Foot Model and LFMI appeared to have a good agreement with the models that track the forefoot as a whole [135, 212].

To study the effect of ankle arthrodesis on gait, Wu et al. [280] proposed a MSFM composed of a four-marker forefoot segment, with markers attached to the heads and bases of the first and fifth metatarsals. Unfortunately, the definition for the segment

coordinate system was not reported in the original paper. In the following year, the OFM was proposed by Carson et al. [42]. The forefoot marker placement is the same as the Wu model, with an additional marker placed in between the second and third metatarsal heads. This marker is used to define the origin of the forefoot segment and the forward-pointing axis. Other authors, including, Kitaoka et al. [137], Pohl et al. [206], and Bruening et al. [29] all utilised a marker placed near the second and third metatarsal heads to define and align the forward axis of the forefoot segment.

In 2007, the RFM model was proposed by Leardini et al. [143]. It utilises even more passive markers to track and define the forefoot segment. The additional markers were used to measure the deformation of the forefoot through a range of “functional angles” as mentioned in Section 3.2.3. The Sawacha model [231] proposed in 2009 and aimed at characterising diabetic foot motion. This model also adopted a very similar strategy to OFM and RFM in their definition of the forefoot segment. The comparison of the forefoot segment definition of all the single segment models can be found in Table 3.1.

From the existing models, it is clear that the measurement of the first metatarsal motion is of primary interest in many MSFMs, with some models ignoring the second through fifth metatarsals all together [58, 148]. In models that consider the motion of the entire forefoot moving as a whole, the markers tend to be placed on the first and fifth metatarsals. This is to be expected as this is the most logical way to model and track the forefoot for several reasons. Firstly, the ease of identifying anatomical landmarks on the first and fifth metatarsal’s heads and bases translates to better marker placement repeatability. Secondly, those markers can also be used as anatomical landmarks or to establish the forefoot transverse plane that is approximately parallel to the floor in a neutral standing position [42]. In addition, from a modelling perspective, being sandwiched in between, the motion of the second through fourth metatarsals could be reasonably estimated using the first and fifth.

Model	No. forefoot markers	Forefoot articulation	Tracking Markers	Kinematic convention	Anatomical frame
Milwaukee Foot Model [135]	3 passive	Hindfoot (P); Hallux (D)	Proximal base of 5th MT; Distal head of 1st MT and 5th MT.	Euler; Ref. to radiographic; No angle offset	Determined through radiographic data.
Rattanapresets [212]	3 passive	Hindfoot (P); Hallux (D)	Proximal base of 5th MT; Distal head of 1st MT and 5th MT.	JCS; Ref. aligned position; Zero angle offset	Determined through anatomical alignment.
Cornwall & McPoil [58]	1 EM Sensor	Tibia (P)	Medial side of the 1st MT shaft	JCS; Ref. static trial; Zero angle offset	Not reported
Leardini Foot Model I [148]	1 cluster plate	Midfoot (P); Hallux (D)	Cluster on 1st MT; CAST anatomical landmarks: Medial projection of 1st MT base (TB); Medial (TM) and lateral (TL) projection of 1st MT head	JCS; Ref. static trial; Zero angle offset	Origin: Base of 1st MT; x-axis: Vector from TB to TM; y-axis: Perp. to the other two axes; z-axis: Parallel to the transverse plane defined by x-axis and TL, positive from left to right.
Wu [280]	4 passive	Hindfoot (P)	Proximal base of 1st MT and 5th MT; Distal head of 1st MT and 5th MT	Euler; Ref. static trial; Zero angle offset	Not reported
Oxford Foot Model [42]	5 passive	Hindfoot (P); Tibia (P); Hallux (D)	Proximal base of 1st MT (P1) and 5th MT (P5); Distal head of 1st MT (D1) and 5th MT (D5); Mid-point of 2nd and 3rd MT head (TOE).	JCS; Ref. static trial; No angle offset	Origin: TOE projected to transverse plane; x-axis: Perp. to z-axis and transverse plane; y-axis: Perp. to the other two axes; z-axis: From PROXFF to TOE, where PROXFF is a fraction of distance between P1 and P5 projected to the transverse plane.
Kitaoka [137]	4 passive	Hindfoot (P)	Proximal base of 1st MT and 5th MT; Distal head of 1st MT and 5th MT;	Not reported	Not reported

Table 3.1: Comparisons of forefoot definition in models with a single forefoot segment. The kinematics convention column also includes information on the reference measurement position and whether a zero angle offset is applied (or alternatively, do not require/or not reported with an angle offset). Abbreviations: P, Proximal segment; D, Distal segment; JCS, Joint coordinate system; Ref, Reference; Perp, Perpendicular; MT, Metatarsal.

Model	No. forefoot markers	Forefoot articulation	Tracking Markers	Kinematic convention	Anatomical frame
Pohl [206]	5 passive	Hindfoot (P)	Proximal base of 1st MT (P1) and 5th MT (P5); Distal head of 1st MT (D1) and 5th MT (D5); Mid-point of 2nd and 3rd MT head (D3).	Cardan; Ref. static trial; Zero angle offset.	Origin: D3 projected to transverse plane; x-axis: Perp. to z-axis and transverse plane; y-axis: Perp. to the other two axes; z-axis: From PROXFF to TOE, where PROXFF is a fraction of distance between P1 and P5 projected to the transverse plane.
Rizzoli Foot Model [143]	6 passive	Hindfoot (P)	Distal head of 1st MT (FMH), 2nd MT (SMH), and 5th MT (VMH); Proximal base of 1st MT (FMB); 2nd MT (SMB), and 5th MT (VMB).	JCS; Ref. static trial; Zero angle offset.	Origin: SMB; x-axis: Project of line joining SMB and SMH on the transverse plane passing through origin, FMH, and VMH. y-axis: Perp. to x-axis and lies in the transverse plane. z-axis: Perp. to the other two axes.
Sawacha [230]	4 passive	Midfoot (P)	Distal head of 1st MT (IMH) and 5th MT (VMH); Proximal base of 5th MT (VMB); Proximal epiphysis of the 2nd TOE (IIT).	Cardan; Ref. static trial.	Origin: Mid-point of IMH and VMH; x-axis: Perp. to the two other axis; y-axis: Perp. to the plane formed by line connecting VMH and IIT and the z-axis; z-axis: Parallel to the line connecting IMH and VMH, with positive from left to right.
Bruening [29]	6 passive	Hindfoot (P); Midfoot (P)	Proximal base of 1st MT (B1) and base of 5th MT (B5); Distal head of 1st MT (H1); Mid-point of 2nd and 3rd MT head (H2); Medial prominence of navicular (NV); Lateral centroid of cuboid (CU).	Cardan; Ref. static trial; No offset.	Origin: Mid-point of CU and NV; x-axis: Origin to the projection of H2 1/2 distance to the floor; y-axis: Perp. to the other two axes; z-axis: Perp. to x-axis and the transverse plane formed by the origin, the projection of H2 1/2 distance to the floor, and H2

Table 3.1 continued

3.3.2 Two-Segment Representations

One of the earliest MSFMs that divided the forefoot into two segments mediolaterally is the Kinfoot, proposed by MacWilliams et al. [161]. The MSFM aimed to address the oversimplification of the foot anatomy in previous models that assumed a single segment forefoot. Although the motion of both the medial and lateral forefoot relative to hindfoot was reported, the variability was high, especially in the coronal and transverse planes. Unfortunately, the segment definition is not available in the original paper, and the cited references that possibly contain this information cannot be found, which prohibits further investigation and validation.

Like the Kinfoot, Jenkyn and Nicol’s [125] five-segment MSFM also separated the forefoot into a medial and a lateral segment. In this MSFM, the medial forefoot represents the motion of the combined first and second metatarsals, and the lateral forefoot represents the motion of the third through the fifth metatarsals. As noted by the authors, the splitting of the forefoot into a medial and a lateral segment allowed the supination/pronation twisting to be described in the coronal plane. However, there is a fundamental problem with the forefoot modelling assumption. The interpretation of the medial and lateral forefoot motion needs to be made with caution due to a shared landmark in their segment definition. In both medial and lateral forefoot segments, the mediolateral axis (in the original paper – the second axis) is defined as the vector from the fifth metatarsal base to the first metatarsal base.

The Ghent Foot Model (GFM) [65] is a recently proposed MSFM that uses a combination of individual and cluster markers to track and define the forefoot segments. In this model, the medial forefoot is defined as the first metatarsal and the lateral segment as the second through fifth metatarsals. This model was designed with a particular focus on accurately tracking the first metatarsal, using three markers attached to the base, head, and shaft of the first. The lateral forefoot is defined by markers

attached to the head and base of the fifth metatarsal, and the lateral side of the first metatarsal. In addition, a cluster of three markers is attached to aid tracking of the lateral side. One interesting finding from the original paper is that there appears to be an opposite motion of the medial and lateral forefoot segment in late stance. This phenomenon was not observed in any of the earlier models.

The Salford Foot Model [187] is another MSFM that considered the forefoot divided into two segments. Like the Jenkyn and Nicol Model [125], two clusters were attached to the medial and lateral sides of the forefoot to track their movement. The medial forefoot was theorised to represent the first metatarsal, and the lateral forefoot the fourth and fifth, leaving the second and third untracked. However, from the original paper, the placement of the medial forefoot marker cluster appeared to have extended to the position of the second metatarsal. This means the medial forefoot that represents only the first metatarsal also captures the motion of the second metatarsal. The motion of the medial and lateral forefoot segments was measured with reference to the mid-foot. In contrast to the GFM, the sagittal motion of the medial and lateral forefoot is very similar to each other, with most of the differences observed in the frontal and transverse planes.

In general, there is currently no agreement on how the forefoot should be divided, as summarised in Table 3.2, with most models lacking a clear segment definition and having a grouping of the metatarsals that appears to be arbitrary and misleading [65, 161, 187]. In the Jenkyn & Nicol Model, the motion of the first and second metatarsals are tracked with markers attached only to the first metatarsal; motion of the third through fourth metatarsals are only tracked with markers attached to the fifth metatarsal. In the GFM, the marker attached to the first metatarsal is used to define the anatomical frame of the lateral forefoot segment. In the Salford Foot Model, the first metatarsal segment is tracked by a marker cluster that also extends to the second metatarsal.

Model	No. forefoot markers	Forefoot articulation	Tracking Markers	Kinematic convention	Anatomical frame
Kinfoot [161]	6 passive	Cuboid (P - lat. forefoot); Talus/navicular/cuneiform (P - med. forefoot)	Proximal base of 1st MT, 3rd MT and 5th MT; Distal head of 1st MT, 3rd MT, and 5th MT.	Euler; Ref. static trial.	Not reported.
Jenkyn and Nicol Model [125]	2 clusters	Midfoot (P)	Medial forefoot: First MT cluster. Lateral forefoot: Fifth MT cluster	JCS; Ref. static trial.	Medial forefoot Origin: First MT base; x-axis: First MT head to first MT base y-axis: Fifth MT base to first MT base z-axis: Perp. to the other two axes. Lateral forefoot: Origin: Fifth MT base; x-axis: Fifth MT head to fifth MT base y-axis: Fifth MT base to first MT base z-axis: Perp. to the other two axes.
Ghent Foot Model [65]	6 passive + 1 cluster	Midfoot (P); Hallux (D - medial forefoot)	Medial forefoot: Base of 1st MT; Medial and lateral side of 1st MT; wand marker on the shaft of 1st MT. Lateral forefoot: Lateral head of 1st MT; lateral side of the head of the base of 5th MT;	Euler; Ref. static trial.	Not reported
Salford Foot Model [187]	6 passive	Midfoot (P); Hallux (D - medial forefoot)	Medial forefoot cluster; Lateral forefoot cluster.	Cardan; Ref. static trial; Zero angle offset.	Not reported

Table 3.2: Comparisons of forefoot definition in models with a divided forefoot segment. The kinematics convention column also includes information on the reference measurement position and whether a zero angle offset is applied (or alternatively, do not require/or not reported with an angle offset). Abbreviations: P, Proximal segment; D, Distal segments; JCS; Joint coordinate system; Ref, Reference; Perp, Perpendicular; MT, Metatarsal.

3.3.3 Anatomical Forefoot Subdivision

From the previous Sections, it is clear that there is currently no consensus on how the forefoot should be divided. It may be intuitive to group the first three metatarsals together as the medial forefoot and the rest as the lateral forefoot based on the bone structure of the forefoot since the first three metatarsals are directly neighbouring the three cuneiforms, whereas the fourth and fifth metatarsals are connected to the cuboid. However, in reality the complex system of connective tissues that maintains alignment and imposes restrictions in motion, the muscular system controlling the motions of the foot, along with the interaction between the floor and the plantar surface of the foot makes it difficult to draw a definitive line of functional division by merely looking at the anatomical structure.

In addition to repeatability studies, studies of in-vivo and in-vitro foot motion have shed light on how to appropriately group the midfoot and forefoot bones in MSFM. Nester et al. [189] presented the first detailed description of foot kinematics during walking using a cadaver specimen actuated by a walking simulator composed of a system of motor-controlled cables and a pneumatic piston (Fig. 3.3). In this experiment, marker-clusters attached to wires were drilled into the tibia, talus, navicular, cuboid, all three cuneiforms, and all metatarsals. The motion of all metatarsals was measured relative to the immediate proximal bone. According to Nester et al., the motion between the fourth and fifth metatarsal and the cuboid is larger than the motion between the other metatarsals and their respective cuneiforms. Fourth and fifth metatarsals have a range of motion (ROM) of 11.4° and 11.6° in the sagittal and frontal planes with respect to the cuboid, whereas the first through third metatarsals moved on average 6.1° and 6.6° relative to their respective cuneiforms. This observations indicate that a unified forefoot segment oversimplifies the complexity of the forefoot motion.

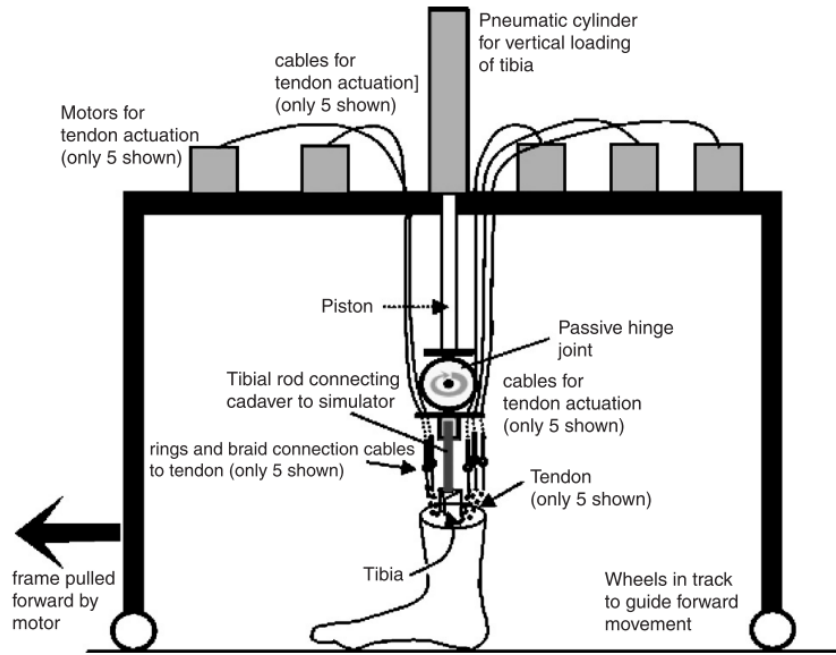


Figure 3.3: Diagram of the walking simulator used by Nester et al. [189] to study the foot kinematics. A hinged joint is mounted at the base of the piston to allow sagittal plane motion of the tibia. A cadaver foot specimen is attached to the base of the hinge joint via an aluminium rod, controlled by applying force through nine tendons. Adapted from Nester et al. [189].

Wolf et al. [271] investigated the presence of “functional units” within the foot using a previously reported protocol in which intra-cortical pins were inserted into a selection of foot bones [6]. Functional units represented the coupled motion of neighbouring joints within the foot, which can be visually identified as a recognisable and repeatable pattern in an angle versus angle plot. For example, a linear relationship was identified between the joint motion between the fifth metatarsal relative to the cuboid, and cuboid relative to the calcaneus in the transverse plane. In their investigation involving six male subjects (averaging 38 years old), a distinct and random motion was revealed between the first and fifth metatarsals, suggesting that there exists a unique movement profile in each of them and therefore they should not be combined as a single segment. This finding of Wolf et al. was supported by a cadaver study by Okita et al. [194] where a significant difference was found in kinematic

profile between the first metatarsal, fifth metatarsal, and a unified forefoot in the transverse and frontal plane.

In 2010, Nester et al. [188] published a paper looking specifically at the violation of rigid body error in the groupings of the metatarsals using a previously published dataset [189]. The errors were quantified as the mean and maximum difference between the kinematics of an assumed rigid body segment and the individual kinematics of the bones comprising each segment. Unsurprisingly, the largest error was found when all five metatarsals were grouped together. Based on their investigation, they suggested that the first, second, and third metatarsals should be grouped along with all the cuneiforms, and the fourth and fifth as another segment. However, Nester et al. did not compare the difference in errors when different combinations of metatarsals were grouped together, such as the grouping of only the first and second, or grouping of the third and fourth.

A recent study by DiLiberto et al. [75] suggested a different subdivision to Nester et al. [188] based on the individual movement profile of the first, third, and fifth metatarsal. The author reported a strong coupling motion between the third and the fifth metatarsal bone using electromagnetic sensors (Fig. 3.4). However, it must be noted that the sensors were attached in different positions along the course of the metatarsal rather than directly drilled into the bone. Therefore, these results are influenced by soft tissue artefact.

It is evident that it may not be optimal to treat the forefoot as a unified segment in a MSFM, in particular, the grouping of the first and fifth metatarsals due to their unique movement profiles. However, none of the existing studies provided a definitive answer on how exactly to subdivide the forefoot. It is important to also realise the best way to subdivide of the forefoot anatomically might not be the best way of representing the forefoot bones in a MSFM due to technical and methodological limitations of stereophotogrammetric-based clinical gait analysis. The next Section

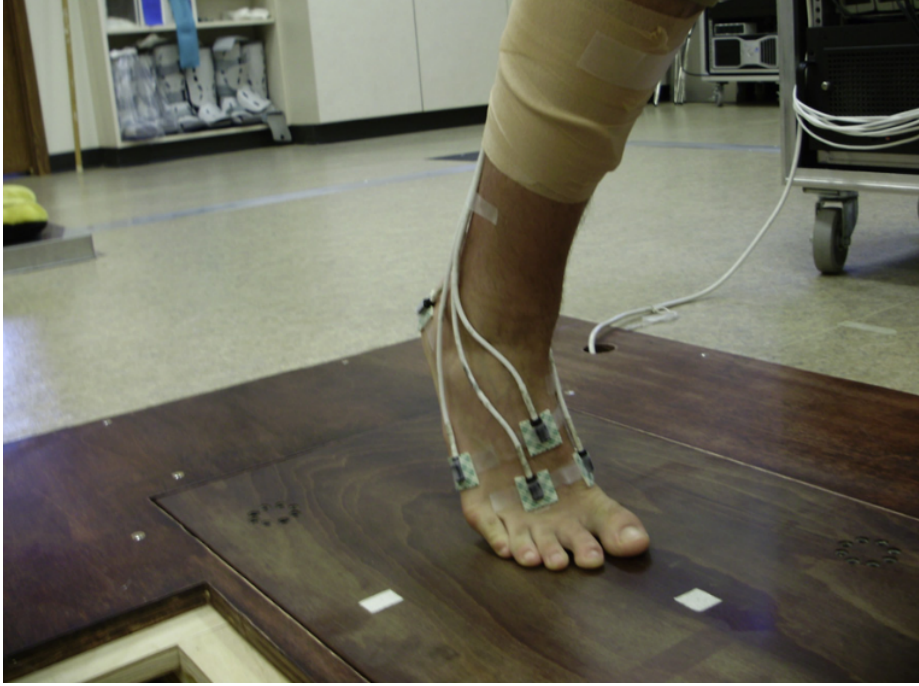


Figure 3.4: EM Sensors attached directly onto the skin to track the first, third, and fifth metatarsal bones. Adapted from DiLiberto et al. [75]

will seek to identify some crucial design factors that affect the utility of a MSFM.

3.4 MSFM Design Considerations

Due to the lack of consensus on forefoot segmentation and definition, this thesis aims to develop a novel MSFM with a forefoot in the most appropriate way in terms of practicality and rigidity. To aid the design and development process, the goal of this Section is to discuss existing limitations and summarise the factors that would influence the efficacy of a MSFM. It is acknowledged that some of the limitations in one MSFM might not be applicable to another, depending on the tracking technologies used and the clinical or research question that the user intends to answer. Therefore, the design factors discussed here will be limited to the scope of stereophotogrammetric-based gait analysis using passive reflective markers under a standard laboratory setting [21]. From review of the literature, the design factors of MSFM can be broadly

categorised into two aspects, practical and methodological. Practical considerations include the tracking of the passive markers, identifying reliable anatomical landmarks, and defining anatomical and technical frames in a repeatable manner. Methodological considerations include consistency with general knowledge in foot biomechanics, rigid body assumption, interpreting and reporting of kinematic data, validation, and additional features such as kinetic analysis or backward compatibility to the existing models (which could also be a practical factor). In the design of a MSFM, both practical and methodological considerations are equally important and valid with regards to the proposal and utilisation of a MSFM, as long as it is clinically relevant.

3.4.1 Practical Considerations

Practical challenges imposed on a MSFM are a consequence of limitations in the motion capture technology. As technology progresses over time, these restrictions are often relaxed. In the early development of MSFM, Cappozzo [38] recognised several considerations when designing local frames (applicable to both technical and anatomical frames), which can be summarised as follows:

1. Reflective markers should be visible to the minimum number of cameras required to track the marker at any given time (usually two).
2. The distance between three markers associated with each segment and offset of any marker from the line joining the other two should be large enough to minimise error from segment coordinate reconstruction.
3. Soft tissue artefact should be minimised.
4. Marker placement procedure should be as fast and effortless as possible.
5. Markers placement should be adaptable to the presence of orthoses, prostheses, or external fixators (listed for completeness, but not applicable to this study).

The first point is relatively straightforward. Stereophotogrammetry is a technique based on the reconstruction of several images in order to obtain instantaneous positions of the markers in three-dimensions, and therefore it is essential that the markers

are placed in locations most visible to the cameras. The second point concurs with an investigation by Södervist & Wedin [244], which showed that the determination of the segment orientation can be inaccurate if the markers are positioned close to a straight line and should be avoided in the design of a marker cluster. The spacing between the markers is also an important consideration. Caravaggi et al. [39] noted that the error caused by small deviations in the marker position was magnified as a result of the close proximity of markers placed on the calcaneus.

The third point concerning the movement of the skin between the bone and the marker is a well-recognised phenomenon, called the soft-tissue artefact (STA). STA is known to cause problems in the estimation of the bone pose, joint centre, and joint rotations [59]. The STA is considered as one of the major issues in gait analysis and has received considerable attention over the years [32]. Many studies have tried to quantify the magnitude of soft tissue artefact [44, 204], with the highest value recorded for the thigh region of 31 mm in marker deviation by Stagni et al. [247]. Most of the studies have reported values for the thigh and shank region where the STA is more problematic due to thick layers of soft tissue [44, 204]. Thus far, only a few have reported values for the foot. In a radiographic study, Maslen [169] observed a translation error in the foot and ankle ranging from 0.6 to 6.8 mm. In the forefoot, a range of 0.8 to 3 mm was observed for a marker placed on the fifth metatarsal, consistent with an earlier study by Tranberg & Karlsson [260] who reported a maximum STA value of 1.8 mm in the same location.

Various strategies have been developed to cope with STA, including the use of multiple anatomical landmark calibration [35], which is based on the calibrated anatomical system technique (CAST) [38]; the solidification technique, based on making rigid body assumption to each individual segment using the Procrustes method [50, 264]; the use of point-clusters to determine the level of soft-tissue artefacts of each landmarks [5], and many more [53]. Advanced algorithms have also been de-

veloped to characterise and compensate for the movement of the soft tissue artefact [4, 22, 83, 100]. With regards to the MSFM design, the most direct way is to avoid placing markers on locations with an excessive layer of soft-tissue in the first place. These are usually the places with bony landmarks that can be easily and reliably palpated [38].

In addition to the STA, the ability to demonstrate good marker placement repeatability is essential to obtain reliable dynamic data. This mostly comes in the form of a repeatability analysis [42, 74, 97]. Repeatability analysed under different conditions is used to reveal different sources of errors. Intrinsic errors measured through inter-trial and inter-subject protocols are often used as a baseline for comparison [234]. Extrinsic errors such as inter-rater or intra-rater can be used to indicate the variability due to marker placement errors. In CGA, it is now customary to conduct a repeatability analysis for any new models proposed, and in some cases, the application of MSFM in new scenarios, such as a new disorder, a new laboratory condition, or a different age group.

3.4.2 Methodological Considerations

Methodological factors concern how the MSFM is utilised to solve research or clinical problems. The utilisation of the model is governed by multiple factors, most fundamentally, the assumptions made about the model in the first place, that is, the rigid body assumption, and the consistency between the MSFM kinematic output and our general knowledge of foot biomechanics. Additional features such as kinetic analysis [77], backward compatibility to the earlier model version [250], utilisation of additional landmarks to define anatomical frames [230, 250], or for calibration purposes [226], may be advantageous.

Compared to repeatability studies, the validity of an MSFM based on its consistency with the general knowledge of foot biomechanics is much harder to demonstrate

due to differences in research methodology and the convention used to describe joint motion (See Sec. 2.3.3). The current gold standard is to compare the differences in joint angle outputs based on passive markers to the outputs based on intra-cortical pins instrumented with markers [160, 189, 188, 194, 271]. Several studies using this approach questioned the appropriateness of a unified forefoot segment have already been discussed in Section 3.3.3. Nevertheless, comparative studies between the motion of skin mounted markers and intra-cortical pins have demonstrated reasonable consistency between them [186, 269]. Despite some concerns over the alteration of gait as a result of local anaesthesia and the invasive nature of intra-cortical pins, Maiwald et al. [164] have demonstrated the validity of comparative studies by showing a good consistency in gait pattern before and after the insertion of bone pins.

Overall, the design of a MSFM is a compromise of practical and methodological considerations. Additional markers may be used to gain insight into more detailed anatomical motion, however, the camera system must be able to capture the level of detail. The development of an optimal MSFM that can be applied for both clinical and research purposes and for both normal and pathological feet is an attractive proposition. However, as noted by Bishop et al. [23], “The complexity of the marker set and model used should always be indicative of the complexity of the analytical question being asked as well as the number of individual segments needed to be analysed.” Today, Bishop’s statement is reflected by the wealth of MSFMs available and many modifications made to the models in order to make them suitable for the intended purpose. However, whether an existing MSFM’s complexity is matched with the clinical or analytical question that the model is trying to answer remains debatable.

3.5 Summary

The aim of this research is to determine how the forefoot segment can be defined in the most appropriate way in terms of rigidity, practicality, and foot biomechanics so that it can be applied in a clinical setting and wider applications such as sport science, injury prevention, etc. Therefore, an understanding of foot anatomy, biomechanics, and the historical development of MSFMs is required to determine important design features that constitute a robust foot model. This Chapter presented an overview of the notable foot models that are currently in use for both foot biomechanics research and clinical applications. An in-depth review of how the forefoot is defined and the evidence for the inadequacy of treating the forefoot as a unified single segment is also presented. Finally, important design considerations and limitations of the models are also outlined. In the next Chapter, the general methodology and equipment used in the present study will be described.

Chapter 4

Methodology

The current state of multi-segment foot models (MSFMs) and their clinical utility was presented in the previous Chapter. Evidence from the literature has provided the rationale for proposing a more appropriate method of representing the forefoot segment in a MSFM. Designing a novel MSFM involves justification, preliminary studies, validation, and finally, clinical case studies to demonstrate its potential value. This Chapter outlines the general methods and materials used to collect and process the biomechanical motion data needed to conduct this research. Study-specific techniques and statistical analysis unique to each study will be described within the relevant chapters. The method of rigidity quantification using the rigid body error (σ_{RBE}) forms a significant part of the initial model design process, which is the justification for appropriate segmentation. Therefore, the algorithm for computing the σ_{RBE} will also be described in this chapter. Supplementary analysis and mathematical proofs are available in Appendices C.1 and C.2.

4.1 Ethical Considerations

Existing motion capture data on healthy adults and patients was obtained from the Oxford Gait Laboratory (OGL), Nuffield Orthopaedic Centre, OUH Hospitals, Oxford, UK. The data had been collected with the approval of the National Research Ethics Service (RED reference: 14/LO/1975, London - South East NRES Committee). For the data collection that used a range of unique marker placement protocols (Chapters 7 and 8), an additional ethical approval was obtained from the Medical Sciences Interdivisional Research Ethics Committee, University of Oxford (Reference: R57227/RE001) prior to the study.

4.2 Subjects and Participants

The first technical chapter (Chapter 5) of this thesis did not involve analysis using motion analysis data. For Chapter 6, motion data from OGL's normal database was used to quantify segment rigidity. Subjects included 64 healthy normal adults (32 male and 32 female) with an average age of 27.0 ± 7.0 years performing straight line walking at a self-selected speed. For Chapter 7, motion data of 23 adult volunteers (12 male and 11 female) with an average age of 27.6 ± 7.3 were collected using the OFFM marker-set listed in Table 4.1 of Section 4.4.2. In combination with the Chapter 7 data set, an additional 22 volunteers were recruited for the study of a novel forefoot modelling approach. In total, the data of 45 subjects (25 male and 20 female) with an average age of 28.2 ± 6.8 were analysed in Chapter 8. All participants included in this study were healthy normal adults with no known lower-limb and foot injuries, deformities, or surgical history according to a self-reporting survey that also exclude both flexible and rigid flatfoot. Five further subjects with pathology were recruited for the clinical case studies. Details will be provided in Chapter 10.



Figure 4.1: Vicon MX-T40 Camera (left) and AMTI OR6-6 force-plate (right).



Figure 4.2: Oxford Gait Laboratory motion capture walkway.

4.3 Equipment

In this study, motion data was captured using 10 mm retro-reflective passive markers attached directly onto the anatomical locations of interest via hypoallergenic double-sided tape. Sixteen infra-red Vicon MX-T40 cameras (Vicon Motion Systems, Oxford, UK) were used to track the instantaneous positions of the markers, sampling at a rate of 100 Hz with a resolution of four megapixels (Fig. 4.1). To identify gait events (on foot contact and foot off time points) and measure the foot contact force with the ground during heel-strike, two AMTI OR6-6-1000 and one OR6S6 (Advanced Mechanical Technology Inc., MA, USA) force-plates were used in this study. Data from them was collected at a rate of 1000 Hz and synchronised to the motion tracking of the Vicon hardware.

4.4 Gait Analysis Data Collection

To capture three-dimensional gait data, an experimental protocol was developed based on the OGL protocol of capturing walking data from normal healthy adult participants and recommendations from the literature on gait analysis methodology [11, 154]. See Figure 4.3 for an overview of the data collection procedure.

4.4.1 Systems Calibration

Prior to subject preparation and data collection, systems calibration was conducted using a calibration wand and a 10 kg weight (Fig. 4.4). The frame of the active wand aligns with the edges of the force-plate which establishes the global coordinate system of the capture volume. Calibration of the cameras was achieved by waving the wand from ground to shoulder-height up and down the walkway, as instructed by the manufacturer's user guide. Distances between the LEDs on the wand (as measured

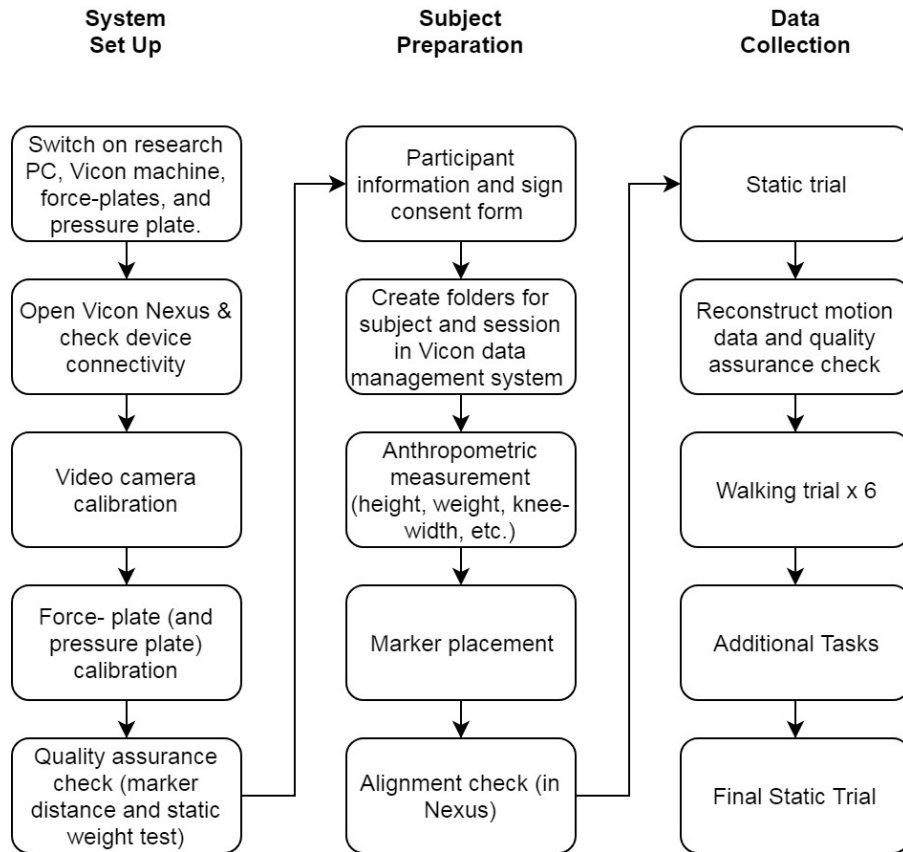


Figure 4.3: Data collection flow chart

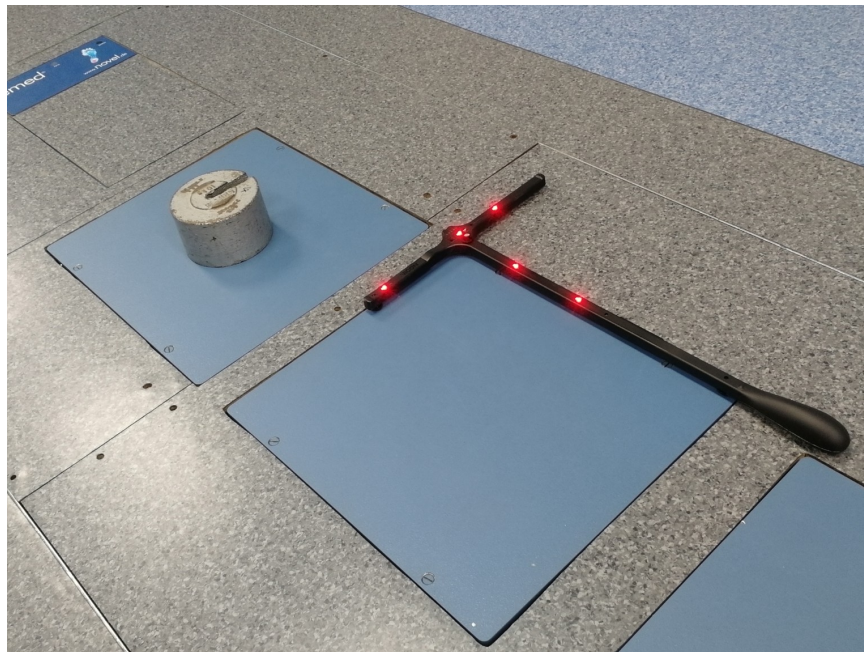


Figure 4.4: T-shaped calibration wand for camera calibration and a 10 kg weight for force-plate checking.

by the cameras) were also checked in the data capture software for quality assurance. The force-plates, used for the determination of gait events, were zeroed and checked using a pre-determined weight (10 kg). The camera was calibrated to an accuracy of ± 1.5 mm for the motion capture system and ± 1.5 N for the force-plate according to the manufacturer.

4.4.2 Subject Preparation

Volunteers were recruited based on the procedure outlined in the ethical application (Sec. 4.1), through departmental email and posters around the campus of the University of Oxford. A questionnaire with exclusion criteria was sent to potential participants. The exclusion criteria for screening the subjects were as follows:

- Over the age of 18.
- Does not wear (or have a history of wearing) orthotics or insoles.
- Has not had surgeries or significant injuries to legs or feet.
- No foot deformity.
- Does not have pain in the joints and muscles of the lower limb.

Before motion capture, consent forms were signed and anthropometric measurements (height, weight, hip-width, knee-width, ankle-width, and foot-length) were taken as required by the gait analysis models adopted for this study, the Plug-in-Gait (PiG) model [126, 63] and the Oxford Foot Model (OFM) [42, 250]. See Appendix A.1 for the subject measurement sheet. Following this, 53 retro-reflective markers (10 mm in diameter) were attached to specific anatomical landmarks of interest on the lower limbs using double-sided hypoallergenic tape. Additional markers were also attached in the interest of measuring the motion of individual metatarsals. Since the primary interest of this thesis is to develop a forefoot model, these additional metatarsal markers shall be named the "*Oxford Forefoot Model*" (OFFM). See Figure 4.5 and Table 4.1 for the locations of the marker placement.

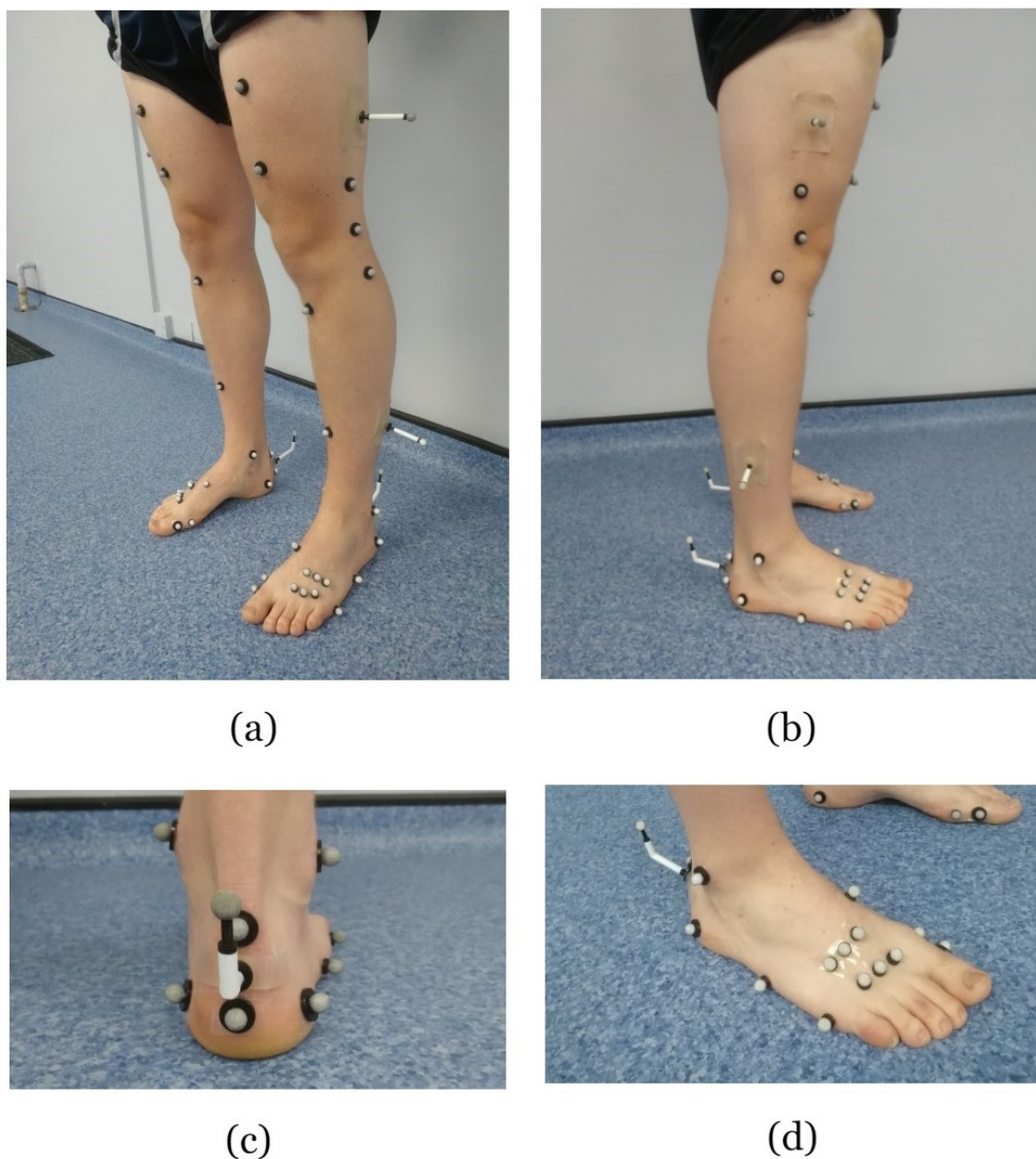


Figure 4.5: Plug-in-Gait and Oxford Foot Model marker-set attached to the lower leg and from a) isometric view. b) side view. c) zoomed-in back view of the right foot. d) zoomed-in isometric view of the foot. Oxford Forefoot Model markers were added and the TOE markers were removed.

	Marker	Anatomical Location
<i>Pelvis & Thigh</i>	SACR	Midway between the posterior superior iliac spine
	L/ RASI	Left/ Right anterior superior iliac spine
	L/ RTHI	Left/ Right lateral aspect of the thigh (wand marker)
	L/ RKNE	Lateral epicondyle of the knee
<i>Shank</i>	L/ RHFB	Lateral aspect of the head of fibula
	L/ RTUB	Anterior aspect of the tibial tuberosity
	L/ RSHN	Anterior crest of the tibia
	L/ RTIB	Lateral surface of the shank (wand marker)
<i>Foot & Ankle</i>	L/ RMMA	Most prominent point of the medial malleolus
	L/ RANK	Most prominent point of the lateral malleolus
<i>Hindfoot</i>	L/ RCPG	Placed between PCA and HEE (wand marker)
	L/ RPCA	Posterior aspect of calcaneus
	L/ LCA	Lateral aspect of calcaneus
	L/ RSTL	Sustentaculum tali
	L/ RHEE	Middle of the calcaneus, same height as the TOE marker
<i>Forefoot</i>	L/ RTOE	Between distal ends of the 2 nd and 3 rd metatarsals. Defined as virtual marker when OFFM markers are used.
	L/ RD1M	Medial side of the head of the 1st metatarsal
	L/ RHLX	Medial side of the proximal phalanx
	L/ RD5M	Lateral side of the head of the 5 th metatarsal
	L/ RP5M	Lateral side of the base of the 5 th metatarsal
	L/ RP1M	Proximal base of the 1 st metatarsal
<i>OFFM</i>	L/ RD2M	Directly on the head of the 2 nd metatarsal
	L/ RD3M	Directly on the distal head of the 3 rd metatarsal
	L/ RD4M	Directly on the distal head of the 4 th metatarsal
	L/ RP2M	Directly on the proximal base of the 2 nd metatarsal
	L/ RP3M	Directly on the proximal base of the 3 rd metatarsal
	L/ RP4M	Directly on the proximal base of the 4 th metatarsal

Table 4.1: Full Plug-in-Gait, Oxford Foot Model, and Oxford Forefoot Model (OFFM) test marker set. Note that the TOE marker is removed when the full set of OFFM is applied

4.4.3 Experimental Protocol

Once consent was given and anthropometric measurements were taken, participants were asked to perform one standing static trial followed by six walking trials in a straight line. Some additional motion tasks, which have not been used in this research were also performed for the purpose of exploring different ways in which the forefoot can deform. These included heel-rise, turning, and sit-to-stand motion.

In a static trial, the subject was asked to stand on double leg on one of the force-plates in a natural position with their hands falling naturally at their sides. For dynamic walking trials, subjects were instructed to walk bare-footed across a 10.0 m walkway (Fig. 4.2) within the capture volume at a self-selected speed. Both ends of the walkway were marked with guiding lines as starting positions in order to ensure that the subject walked over the force-plate without purposefully aiming for it. Finally, another static trial was also taken at the end of the session as quality control.

4.5 Data Processing

4.5.1 Kinematic Data Processing

Capturing and part of post-processing the kinematic data was performed within Vicon Nexus software 2.6.0 (Vicon Motion Systems, Oxford, UK). Positions of the passive retro-reflective markers were captured and reconstructed by the system using two-dimensional images obtained from a minimum of two cameras (Fig. 4.6). All markers were manually labelled for the static trials, and all dynamic trials were labelled through an automated process (auto-labelling function) within Nexus that recognises the marker locations from the static trial. Gait events (heel-strike and toe-off) were detected using the force-plate, which marks the start and finish of a full

stance phase.

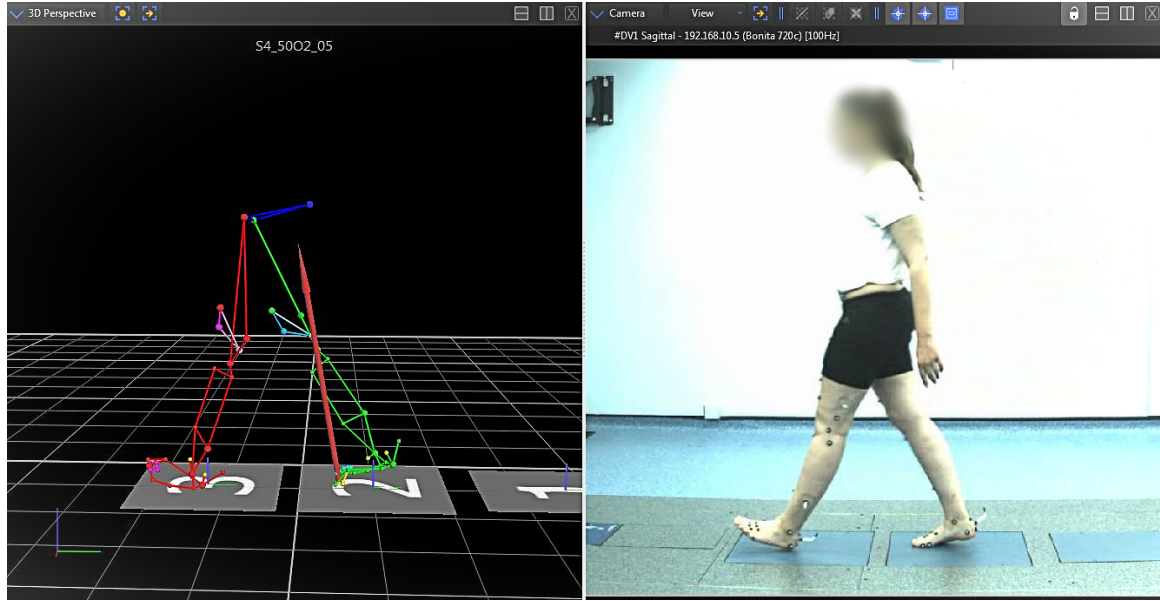


Figure 4.6: Reconstructed markers at one time point in a dynamic trial and ground-reaction force visualisation in Vicron Nexus, along with an accompanying video frame.

Numerous studies have been conducted with various methods proposed for signal processing of movement analysis data [36, 175, 242]. In this study, the Nexus-integrated method of spline smoothing (mean square error of 20 mm) described by Woltring et al. [272] was adopted to filter the raw data, which is equivalent to a double Butterworth filter with the additional advantage of the ability to process data with unequal sampling intervals. Gaps in marker trajectories below 5 frames (0.05 s) were filled using the ‘spline-fill’ interpolation function embedded within Nexus. For gaps larger than 5 frames, depending on the location and timing during the gait cycle, other gap-filling algorithms were chosen according to the OGL Gap Filling Protocol v1.2. Trials with gaps larger than 15 frames were discarded. The raw marker coordinates were exported in a c3d file format [179] along with information on anthropometric parameters, analogue force-plate data, and gait events.

4.5.2 Biomechanical Model Processing

Processing of the biomechanical models and computation of kinematic angles were carried out using MATLAB (R2018a, MathWorks Inc., Natick, USA). Two open-source packages were used to process and analyse kinematic data. Biomechanical Toolkit [17] was used to extract and modify data from the raw c3d. files. Modified BiomechZoo [79] was used to process parts of the PiG and OFM kinematic angles. Implementation of the OFFM was also within the MATLAB environment, written using the scripts adapted from the BiomechZoo.

4.6 Model Computation

This section outlines the methods of computing biomechanical models in this study. The Plug-in-Gait (PiG) model [63, 126], also known as the Conventional Gait Model or the Helen Hayes Model, is a popular validated model used widely in both clinical gait analysis and biomechanical research [14]. In this study, PiG was used to compute the hip and knee kinematics. In conjunction, the Oxford Foot Model [42, 250] was used to estimate foot kinematics, and to provide a framework and reference to which the novel forefoot model was compared.

4.6.1 Plug-in-Gait Model

Over the past 30 years since the publication of the original PiG model [63, 127], the method of implementation has changed considerably with many proposed modifications [14]. Here, the method of computing the PiG kinematics will follow the procedure outlined in the OGL clinical protocol.

The PiG models the lower-limb as seven segments and six joints using fifteen markers and several anthropometric measurements. Firstly, the laboratory axes, or the

global coordinate system (GCS) is defined with $\hat{x}_{lab}$ parallel to the walkway and the direction of walking, $\hat{z}_{lab}$ pointing vertically, and $\hat{y}_{lab}$ medio-laterally. The pelvis local coordinate system (LCS) is defined using the LASI, RASI, and the SACR marker, as shown in Figure 4.7.

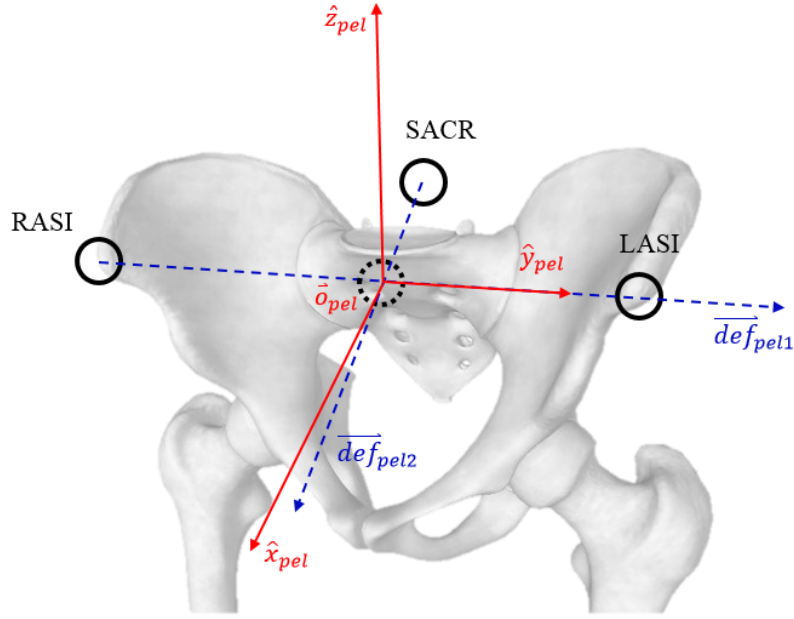


Figure 4.7: PiG pelvis local coordinate system.

Pelvis Segment Coordinate System

The origin of the pelvis ($\vec{o}_{pel}$) is taken as the mid-point of the two anterior superior iliac spine (ASIS) markers.

$$\vec{o}_{pel} = (\vec{r}_{LASI} + \vec{r}_{RASI})/2 \quad (4.1)$$

The y-axis (medio-lateral) is the direction from the RASI to LASI marker:

$$\hat{y}_{pel} = \frac{\vec{r}_{LASI} - \vec{r}_{RASI}}{|\vec{r}_{LASI} - \vec{r}_{RASI}|} \quad (4.2)$$

The z-axis (proximal-distal) is in the direction perpendicular to two defining lines:

$$\begin{cases} \vec{def}_{pel1} = \vec{r}_{LASI} - \vec{r}_{RASI} \\ \vec{def}_{pel2} = \vec{r}_{\vec{o}_{pel}} - \vec{r}_{SACR} \end{cases} \quad (4.3)$$

where $\vec{def}_{pel1}$ is the vector from RASI to LASI, and $\vec{def}_{pel2}$ is the vector from SACR to $\vec{o}_{pel}$. Finally, the x-axis (anterior-posterior) is mutually perpendicular, with the forward direction pointing towards the direction of progression.

$$\hat{x}_{pel} = \hat{y}_{pel} \times \hat{z}_{pel} \quad (4.4)$$

Hip Joint Centres

The hip joint centres (HJC) are defined using the Davis Hip Model [63] which begins by computing several parameters. Firstly, the distance between the ASIS and the greater trochanter is estimated as (left: d_L ; right: d_R):

$$\begin{cases} d_L = 0.1288 LL_L - 48.56 \\ d_R = 0.1288 LL_R - 48.56 \end{cases} \quad (4.5)$$

where LL_L and LL_R are the left and right leg length. Next, a constant c_{pel} is calculated from the average leg length (LL_{av}) as:

$$c_{pel} = 0.115 LL_{av} - 15.3 \quad (4.6)$$

Based on the Davis Hip Model, the offset vector for the left hip joint centre ($\vec{o}_{LHJC}$) relative to the pelvic LCS is defined as:

$$\begin{cases} HJC_x = c_{pel} \cos(\theta) \cos(\beta) - (d_L + d_m/2) \cos(\beta) \\ HJC_y = -c_{pel} \sin(\theta) + d_{IA}/2 \\ HJC_z = -c_{pel} \cos(\theta) \sin(\beta) - (d_L + d_m/2) \sin(\beta) \end{cases} \quad (4.7)$$

where $\theta = 0.5$ rad and $\beta = 0.314$ rad. For the right hip joint centre ($\vec{o}_{RHJC}$), the Y-offset HJC_y is multiplied by -1 since Y is in the medio-lateral direction. To find the location of the left and right hip joint centres $\vec{o}_{LHJC}$ and $\vec{o}_{RHJC}$ relative to the global coordinate system (${}_g\vec{o}_{LHJC}$ and ${}_g\vec{o}_{RHJC}$), the vectors (Eq. 4.7) need to be transformed from the pelvic segment coordinate system ($\hat{x}_{pel}, \hat{y}_{pel}, \hat{z}_{pel}$) to the global coordinate system ($\hat{i}_g, \hat{j}_g, \hat{k}_g$).

$$\begin{bmatrix} HJC'_x \\ HJC'_y \\ HJC'_z \end{bmatrix} = \begin{bmatrix} \hat{x}_{pel} \cdot \hat{i}_g & \hat{y}_{pel} \cdot \hat{i}_g & \hat{z}_{pel} \cdot \hat{i}_g \\ \hat{x}_{pel} \cdot \hat{j}_g & \hat{y}_{pel} \cdot \hat{j}_g & \hat{z}_{pel} \cdot \hat{j}_g \\ \hat{x}_{pel} \cdot \hat{k}_g & \hat{y}_{pel} \cdot \hat{k}_g & \hat{z}_{pel} \cdot \hat{k}_g \end{bmatrix} \begin{bmatrix} HJC_x \\ HJC_y \\ HJC_z \end{bmatrix} \quad (4.8)$$

$$\begin{cases} {}_g\vec{o}_{LHJC} = \vec{o}_{pel} + LHJC' \\ {}_g\vec{o}_{RHJC} = \vec{o}_{pel} + RHJC' \end{cases} \quad (4.9)$$

Femur Segment Coordinate System and Knee Joint Centre

The femur LCS (Fig. 4.8) is defined using the positions of the HJC, KNE, THI, and the knee joint centre (KJC). The KJC is estimated to be at a distance (δ_{KNE}) medially from KNE, aligned with the plane formed by HJC, THI, and KNE, such that the vector from HJC to KJC is perpendicular to the vector from KNE to KJC. This is also known as the CHORD function, outlined in the PIG manual [266]. The δ_{KNE} is calculated for individual legs as:

$$\delta_{KNE} = (KW + d_m)/2 \quad (4.10)$$

where KW is the knee width and d_m the marker diameter. The position of the THI marker is critical in the estimation of the KJC as it defines the $y - z$ plane of the femur segment. Therefore, the DynaKAD function [13] was implemented to define a thigh rotation offset through minimising the varus-valgus knee angle range during the walking trial.

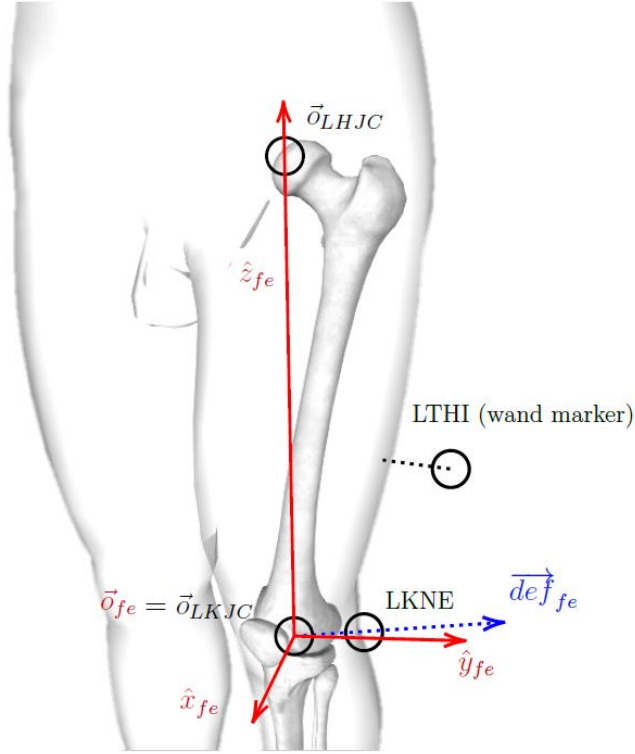


Figure 4.8: PiG femur local coordinate system

The femur origin ($\vec{o}_{fe}$) is taken as the KJC, and the z-axis taken as the direction from the femur origin to the HJC:

$$\hat{z}_{fe} = \frac{\vec{o}_{LHJC} - \vec{o}_{fe}}{|\vec{o}_{LHJC} - \vec{o}_{fe}|} \quad (4.11)$$

The x-axis (medio-lateral) is mutually perpendicular to the z-axis and the $\vec{def}_{fe}$:

$$\hat{x}_{fe} = \frac{\vec{def}_{fe} \times \hat{z}_{fe}}{|\vec{def}_{fe} \times \hat{z}_{fe}|} \quad (4.12)$$

where the (left) segment defining line ($\vec{def}_{fe}$) is the vector from the femur origin ($\vec{o}_{fe}$) to the KNE marker. For the right femur LCS, the defining line is negated. Finally, the y-axis is mutually perpendicular:

$$\hat{y}_{fe} = \hat{z}_{fe} \times \hat{x}_{fe} \quad (4.13)$$

Tibia Segment Coordinate Systems and Ankle Joint Centres

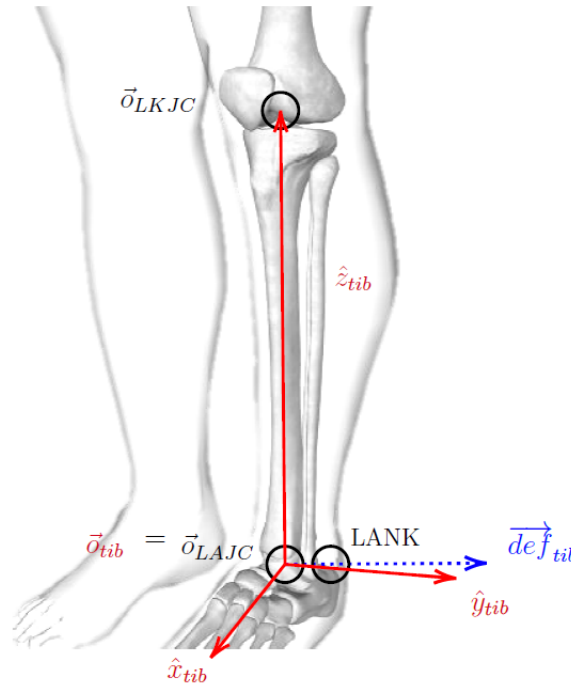


Figure 4.9: PiG tibia local coordinate system

The tibia LCS (Fig. 4.9) and the ankle joint centres (AJCs) were determined in a similar manner to the femur LCS and KJC, using the positions of the KJC, ANK markers, and the ankle offsets (δ_{ANK}):

$$\delta_{ANE} = (AW + d_m)/2 \quad (4.14)$$

where AW is the ankle width and d_m the marker diameter. The AJC is determined

through the CHORD function in a similar function to how the KJC is computed. The equations defining the set of orthogonal axes are modified from Eqs. 4.11-4.13, with the only change being the use of KJC, TIB, and ANK instead of HJC, THI, and KNE. A second LCS for the tibia segment is also defined, termed the “untortioned tibia segment coordinate system”. This represents the proximal end of the tibia that is used to calculate the knee joint kinematics. The untortioned tibia LCS is defined as the reorientation of the tibia LCS (Fig. 4.9) by the angle made between the projection of knee flexion axis ($\hat{y}_{fe}$) and ankle dorsiflexion axis ($\hat{y}_{tib}$) in the transverse plane. This angle, called the “tibial torsion (τ_{tib})”, is measured in static pose (neutral standing position) and applied in the torsioned tibia LCS in the dynamic trials.

Foot Segment Coordinate System

Although the foot motion is measured with the OFM in this thesis, the PiG foot is also defined here for completeness. The PiG foot segment is defined with HEE, ANK, TOE, and two offset angles calculated from the static trial. The two offset angles, the static plantar-flexion offset (ψ_{pf}) and foot rotation offset (ψ_{fr}) are used to reorient the foot LCS to its ‘aligned position’ in the dynamic trial to remove the need to track with the HEE marker, described in the PiG manual [266]. The foot segment has its origin at the AJC, with the z-axis defined as the vector from AJC to TOE, the y-axis mutually perpendicular to the z-axis and the vector from AJC to KJC, and the x-axis mutually perpendicular to the y- and z-axes.

4.6.2 Oxford Foot Model

The OFM kinematics are calculated based on the model described by Stebbins et al. [250], which modified the original method published by Carson et al. [42]. Defined using eighteen markers on each foot, the OFM consists of four segments, tibia, hind-foot, forefoot, and hallux. The OFM was designed to be compatible with the PiG

such that it can be used simultaneously to collect motion of the entire lower-limb. In this study, the hindfoot segment is measured with reference to the untorsioned PiG tibia LCS (Fig. 4.9).

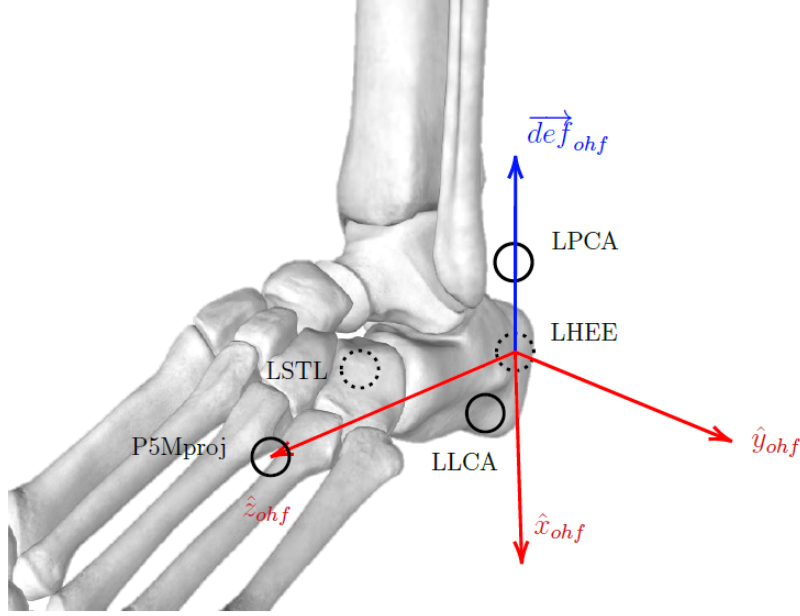


Figure 4.10: OFM hindfoot local coordinate system

Hindfoot Segment Coordinate Systems

The OFM hindfoot LCS is designed using five tracking markers, P5M, PCA, HEE, LCA, STL, and a virtual marker P5Mproj. Here, P5Mproj is defined as the projection of P5M onto the quasi-sagittal plane formed by PCA, HEE, and the mid-point between LCA and STL ($\vec{r}_{MidHeel}$):

$$\vec{r}_{P5Mproj} = \vec{r}_{P5M} - [(\vec{r}_{P5M} - \vec{r}_{PCA}) \cdot \vec{n}_{sag}] \vec{n}_{sag} \quad (4.15)$$

where $\vec{n}_{sag}$ is the normal vector of the projecting plane defined as:

$$\vec{n}_{sag} = (\vec{r}_{HEE} - \vec{r}_{MidHeel}) \times (\vec{r}_{PCA} - \vec{r}_{HEE}) \quad (4.16)$$

Following the same set of calculations as the PiG Segments (Eqs. 4.11-4.13), the

origin of the OFM hindfoot LCS is first defined as the position of HEE (Fig. 4.10). Then, the z-axis is defined as the direction from the origin ($\vec{o}_{ohf}$) to P5Mproj, and y-axis defined as being mutually perpendicular to z-axis and the defining line ($\vec{def}_{ohf}$), that is, the vector from HEE to PCA. Finally, the x-axis is mutually perpendicular to the other y- and z-axes.

Forefoot Segment Coordinate System

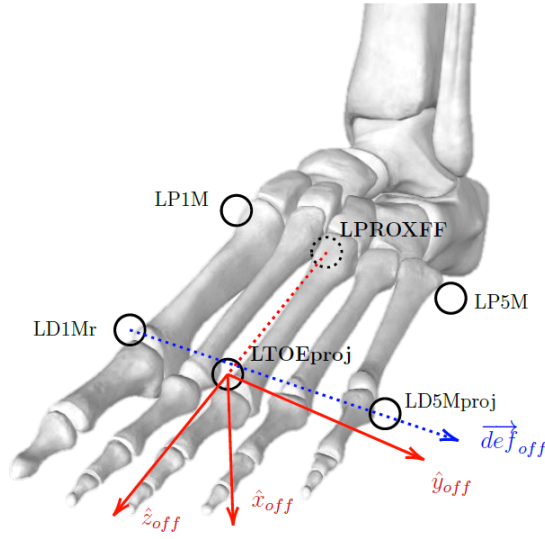


Figure 4.11: OFM forefoot local coordinate system

The OFM forefoot segment is defined using four tracking markers (P5M, D5M, TOE, and P1M) and one virtual marker (D1Mr). To align the forefoot coordinate system to the plantar aspect of the foot, projected markers and virtual markers were used to define the transverse plane of the LCS. The computation begins by defining a vector parallel to the ground in the standing static trial (reconstructed in dynamics).

$$\vec{r}_{PROXFF} = \vec{r}_{P1M} + (\vec{r}_{P5M} - \vec{r}_{P1M}) \times ProxFFscale. \quad (4.17)$$

where *ProxFFscale* is a parameter defined as a fraction of the distance between P1M

and P5M:

$$ProxFFscale = \frac{\|\vec{r}_{P5M} - \vec{r}_{P1M}\| - d_m/2}{2 \times \|\vec{r}_{P5M} - \vec{r}_{P1M}\|} \quad (4.18)$$

The origin of the forefoot LCS (Fig. 4.11) is TOEproj, with the forward z-axis defined as the vector from PROXFF to TOEproj. The x-axis is defined as being mutually perpendicular to the z-axis and the defining vector, the direction from D1Mr to D5Mproj. Lastly, the y-axis is mutually perpendicular. Here, the TOEproj and D5Mproj is the projected marker (TOE and D5M) onto the transverse plane of the forefoot, and D1Mr is the virtual marker reconstructed from the static position using the positions of the P1M, D5M, and P5M.

Hallux Segment Coordinate System

The origin of the hallux LCS is taken as the D1Mr marker, the z-axis direction from D1Mr to HLX, the x-axis perpendicular to the z-axis and the vector from D1Mr and D5Mproj, and the y-axis mutually perpendicular to the other two.

4.6.3 Oxford Forefoot Model

With a secondary aim of being compatible and adaptable to the current foot model, the forefoot model(s) were defined based on the framework of the OFM and PiG, defined as a set of three orthogonal vectors in order to measure the tri-planar motion. Detailed descriptions of the Oxford Forefoot Model, of which the naming is inspired by the OFM, will be provided in Chapter 8.

4.7 Gait Kinematics

4.7.1 Plug-in-Gait

PiG uses a YXZ Euler rotation sequence between two segments to compute the rotation of a three-dimensional segment with reference to another set of axes, which can be either the global coordinate system (GCS) or those of a different segment. The method of computing the rotational sequence is outlined in this section, based on rigid body dynamics [134]. The transformation from coordinate $[B]$ to $[A]$ can be expressed as:

$$[A] = [T][B] \quad (4.19)$$

where the transformation matrix $[T]$ corresponding to the Y-X-Z Euler angles can be expressed as:

$$[T] = R_z(\gamma)R_x(\beta)R_y(\alpha) = [R_z][R_x][R_y] \quad (4.20)$$

$$[T] = \begin{bmatrix} C_\gamma & S_\gamma & 0 \\ -S_\gamma & C_\gamma & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} 1 & 0 & 0 \\ 0 & C_\beta & S_\beta \\ 0 & -S_\beta & C_\beta \end{bmatrix} \begin{bmatrix} C_\alpha & 0 & -S_\alpha \\ 0 & 1 & 0 \\ S_\alpha & 0 & C_\alpha \end{bmatrix} \quad (4.21)$$

$$[T] = \begin{bmatrix} C_\gamma & S_\gamma & 0 \\ -S_\gamma & C_\gamma & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} C_\alpha & 0 & -S_\alpha \\ S_\alpha S_\beta & C_\beta & C_\alpha S_\beta \\ S_\alpha C_\beta & -S_\beta & C_\alpha C_\beta \end{bmatrix} \quad (4.22)$$

$$[T] = \begin{bmatrix} C_\alpha C_\gamma + S_\alpha S_\beta S_\gamma & C_\beta S_\gamma & C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma \\ S_\alpha S_\beta C_\gamma - C_\alpha S_\gamma & C_\beta C_\gamma & S_\alpha S_\gamma + C_\alpha S_\beta C_\gamma \\ S_\alpha C_\beta & -S_\beta & C_\alpha C_\beta \end{bmatrix} \quad (4.23)$$

where α , β , and γ corresponds to the first, second, and third sequential rotations, and operators C and S corresponds to sine and cosine trigonometric functions. Knowing

the elements of $[T]$ from Eq. 4.23, the angles can be computed as:

$$\beta = \arctan \frac{-T_{3,2}}{\sqrt{T_{1,2}^2 + T_{2,2}^2}}; \quad \alpha = \arctan \frac{T_{3,1}}{T_{3,3}}; \quad \gamma = \arctan \frac{T_{1,2}}{T_{2,2}} \quad (4.24)$$

where the subscripts i and j in $T_{i,j}$ indicate the matrix element of the i^{th} row and j^{th} column.

	Segments in motion [A]	Proximal/ Reference segment [B]	Kinematic output (deg)
Pelvis kinematics	Pelvis LCS	GCS	• Tilt (α) • Obliquity (β) • Rotation ($-\gamma$)*
Hip joint kinematics	Femur LCS	Pelvis LCS	• Flexion (α) • Adduction (β)* • Rotation ($-\gamma$)
Knee joint kinematics	Tibia LCS	Femur LCS	• Flexion (α) • Adduction ($-\beta$) • Rotation ($-\gamma$)*$-\tau_{tib}$
Ankle joint kinematics	Foot LCS	Tibia LCS	• Dorsiflexion ($-\alpha - \pi/2$) • Inversion ($-\beta$)* • Rotation ($-\gamma$)*
Foot/ lab kinematics	Foot LCS	GCS	• Progression ($-\beta$)

Table 4.2: PiG kinematics output for the left side, negate * marked angles for right side.

In PiG, the motion of the pelvis is measured with reference to the GCS, and the joint motions are described based on clinically relevant angles, also known as the orthopaedic angles described by Lewis and Law [155]. All joint kinematic outputs are made between the proximal and distal segments of the respective joints. Full PiG kinematic outputs are listed in Table 4.2.

4.7.2 Oxford Foot Model Kinematics

The OFM angles are computed similarly to the PiG model, where inter-segmental joint kinematics are described as tri-planar motions. However, the OFM angle employs the joint coordinate system (JCS) described by Grood and Suntay [101]. The JCS is equivalent to the YXZ Euler sequence as implemented in the PiG model in this study [49, 55]. Full OFM kinematic outputs are listed in Table 4.3.

	Segments in motion [A]	Proximal/ Reference segment [B]	Kinematic output (deg)
Knee joint kinematics	Tibia LCS	Femur LCS	• Flexion (α) • Adduction (β)* • Rotation ($-\gamma$)*
Ankle joint kinematics	Hindfoot LCS	Tibia LCS	• Dorsiflexion ($\alpha + \pi/2$) • Rotation (β)* • Inversion ($-\gamma$)*
Forefoot/ Hindfoot	Forefoot LCS	Hindfoot LCS	• Dorsiflexion ($-\alpha$) • Supination (β)* • Adduction ($-\gamma$)*
Forefoot/ Hindfoot	Forefoot LCS	Hindfoot LCS	• Dorsiflexion ($-\alpha + \pi/2$) • Supination (β) • Adduction ($-\gamma$)*
Hallux/ Forefoot	Hallux LCS	Forefoot LCS	• Dorsiflexion ($-\alpha$) • Varus (β)*

Table 4.3: OFM kinematics output for the left side, negate * marked angles for right side.

4.8 Quantifying Rigidity

In this study, rigid body error (σ_{RBE}) was used to quantify the rigidity of the foot segments (Chapters 6 and 7). This section outlines the rationale for using this method, while the complete algorithm is available in Appendix C.1.

There are many approaches to quantifying rigidity (or flexibility) of a point cloud,

which is the current problem. In classical mechanics, a rigid body is characterised by the requirement that the distance between any two points of the body remains fixed [134]. Therefore, the simplest form of quantifying rigidity is by taking the change in Euclidean distance between the landmarks, or by calculating strain to obtain deformation per unit length.

The deformation quantified by changes in the distance is easy to measure and requires minimal computation. However, there are disadvantages to using this method. Since the deformation is measured with reference to a particular point, the result will be different from point to point, and if this location is moved. Averaging over multiple measurements may over- or underestimate the real amount of deformation. Furthermore, the direction in which the marker deviates also affects the measurement. The problem lies in the determination of the reference position from which deviation is measured.

Rigid body error (σ_{RBE}) is derived from a method that computes the residual of the best-fitting transformation matrix between two set of points (of equal numbers) [264]. The residual of the segment tracking algorithm has been utilised as a method to quantify soft-tissue artefact [264], minimise rotation error [152], and quantify MSFM segment rigidity [29, 246]. Based on its past application, it appears that σ_{RBE} is an effective method of quantifying rigidity of an MSFM and tracking how it deforms over a full gait cycle. Compared to more intuitive methods such as Euclidean distance or strain, the advantage of σ_{RBE} is that it takes into account the deformation of all points simultaneously.

σ_{RBE} originated from the orthogonal Procrustes problem, that is, a least-squares problem to minimise the expression:

$$\|[T][A] - [B]\| \tag{4.25}$$

where $[A]$ is a $N \times 3$ matrix for the *initial marker configuration* (N - number of landmarks), $[B]$ the *deformed configurations* of the same size as $[A]$, and $[T]$ is a 4×4 *orthogonal transformation matrix* for which the Eq. 4.25 is minimised. Through the solving of this least-squares problem, the rigid body movement (rotation and translation) of the segment can be estimated using only the three-dimensional coordinates of the markers [237, 246, 265]. The Procrustes problem was first treated by Green [99], who used the method of Lagrange multipliers to impose constraints to the minimisation equation (Eq. 4.25), then consequently solved using eigendecomposition. A more generalised solution was later derived by Schönemann [232], followed by improvements in the algorithm utilising the method of singular value decomposition [105, 244, 245]. The Procrustes algorithm allowed a reference configuration to be superimposed onto another configuration as precisely as possible. Therefore, the residual of Eq. 4.25 represents the mismatch between the two configurations and thus is an indication of deformation. Proposed by van den Bogert et al. [264], the σ_{RBE} can be calculated as:

$$\sigma_{RBE} = \sqrt{\frac{1}{3N-6} \|[T][A] - [B]\|} \quad (4.26)$$

where N is the number of landmarks, $\sqrt{\frac{1}{3N-6}}$ is the adjustment for the degrees of freedom, and $\|x\|$ is the Frobenius norm (or matrix norm) of x .

4.9 Summary

This Chapter presented the general methodology utilised in this thesis to collect, process, and analyse the biomechanical data required to conduct this research. The next Chapter presents the analysis of σ_{RBE} , followed by the evaluation of segment rigidity in the OFM and the metatarsals, and the proposal and case studies of novel MSFMs.

Chapter 5

Rigid Body Error: A Global Sensitivity Analysis

As discussed in previous Chapter, residuals of a segment tracking algorithm (presented here on the form of the σ_{RBE}) has been utilised as a method to quantify rigidity. Past applications included quantifying soft-tissue artefact (STA) [264], minimise rotation error [152], and quantify MSFM segment rigidity [29, 246]. Previous studies have shown that the σ_{RBE} is a direct indication of the level of rigidity [80, 244, 245]. However, there is currently no study looking specifically at the sensitivity of σ_{RBE} to changes in the variables used in its calculations. More specifically, to what extent does segment size, geometric configurations, and other factors affect the correlation between the σ_{RBE} and the magnitude of deformation? This Chapter attempts to answer these questions, and validate the use of σ_{RBE} in the subsequent Chapters (6 and 7)

5.1 Introduction

In general, σ_{RBE} quantifies the misfit between two sets of points after an optimised transformation from one set to another. Here, the ‘sets of points’ refer to the landmark configuration of a MSFM segment and the ‘misfit’ is a result of measurement error and marker-cluster deformation. Despite having been successfully applied to quantify segment rigidity, to our knowledge, there is currently no study looking at the efficacy of using σ_{RBE} to quantify rigidity.

There are various methods to assess the relationship between the dependent and independent variables in a mathematical equation. The most common methods are uncertainty analysis (UA), sensitivity analysis (SA), and regression analysis (RA), which have been applied diversely from the study of economic models [199] to modelling chemical reactions [223] and structures [107]. While SA is used to determine the contribution of individual parameters to the output (here, the σ_{RBE}), UA is conducted to validate the input parameters and to estimate the uncertainty of the outputs. The most basic and intuitive approach to conduct a SA is ‘one-at-a-time’ (OAT) based local sensitivity analysis (LSA) due to its simplicity and ease to perform [93]. As the name implies, the OAT method assesses the sensitivity of an input parameter by taking derivatives at a single point. However, in order for this method to be effective, the input to output ratio has to be linear and the data points should also come from a normal distribution [122]. While this method is easy to implement, OAT suffers from a lack of coverage of the ranges of each input variable [221, 222].

Global sensitivity analysis (GSA) using regression or variance-based methods can be employed as a more robust way of testing the properties of a mathematical model [221]. Many methods have been developed for GSA, including the widely used Monte-Carlo analysis, response surface methodology, and Fourier amplitude sensitivity test (FAST) [103, 108]. The FAST method, based on Fourier transformation to estimate

sensitivity [173], has the advantage of covering the full range of input variables, efficiency, and low computational cost [108]. However, this method does not provide the distribution function of the input and output variables, and it is not possible to specify a correlation between variables. For the analysis of σ_{RBE} in this thesis, the Monte Carlo method was chosen due to some of its desirable properties that are appropriate for this study. The Monte Carlo technique samples the full coverage of the ranges of each input variable, therefore exposing the worst-case scenarios of the algorithm. Moreover, this method also reveals any nonlinearities, thresholds, and discontinuities of the algorithm [108].

While studies have shown that transformation residuals (hence σ_{RBE}) can be used as a direct indication of the non-rigid deformation of a geometry [80, 244, 245], there is currently no study that specifically investigates the effectiveness of quantifying segment rigidity of a MSFM. Therefore, the goal of this study is to determine how sensitive is σ_{RBE} to its input variables. This was achieved by conducting a Monte-Carlo based GSA, followed by the proposal of a regression model for the σ_{RBE} .

5.2 Methodology

The methodology of this chapter is composed of three parts. The first two Sections (5.2.1 and 5.2.2) outline the procedure for conducting the SA and the justification for selecting the range and distribution of each input variables for a Monte-Carlo simulation. The third part (Section 5.2.3) outlines the method of fitting a regression model based on the information obtained from the SA. The full GSA was conducted in MATLAB (vR2017a, MathWorks Inc., Natick, USA).

5.2.1 Sensitivity Analysis - General Methodology

To evaluate the whether σ_{RBE} is a suitable and effective variable to quantify rigidity, it can be thought of as a mathematical model with a range of inputs that describes the geometrical characteristics of the marker-cluster configuration and the deformation that it undergoes. The purpose of SA is to determine the contribution of the individual input variables to the total σ_{RBE} . In the present study, the sensitivity of the σ_{RBE} was assessed by correlation coefficients and scatter plots as they provide an immediate visual depiction of the relationship between it and the input variables.

The variables examined for sensitivity were not limited to the raw inputs or outputs, but also the secondary effect as a result of the random generation of data points, including the segment size (S), the condition number (kA), and rotation error (R_{err}), all of which have physical characteristics associated with σ_{RBE} and may reveal important information that contributes to model accuracy (Fig. 5.1). Note that introducing additional parameters will not affect the regression model output negatively, as insensitive parameters will not be included in the final model.

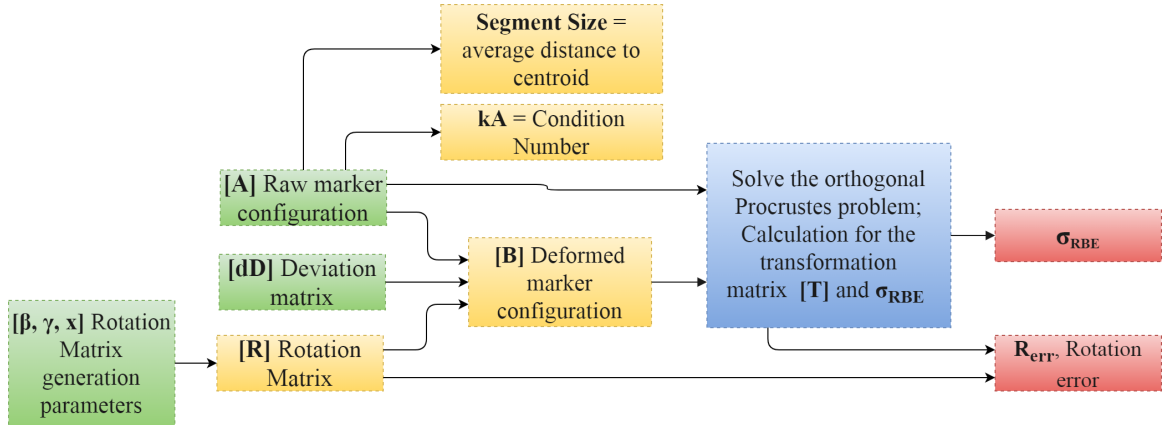


Figure 5.1: Sensitivity variable dependency flowchart. The direction of the arrows indicates the dependency of the input parameters

Raw marker configuration $[A]$, deviation matrix $[\delta D]$, and rotation matrix $[R]$ were treated as random variables in this study. The raw marker configuration $[A]$ simulates

the landmark coordinates of a MSFM segment, and hence determines the segment size S and the condition number kA . Here, S is calculated as the average distance from all the markers to their geometrical centroid, where the centroid is the average of all the marker coordinates. The kA is an indication of how well the landmarks are configured in three-dimensions [244]. To simulate a movement in three-dimensions, $[A]$ was rotated by a randomly generated $[R]$. Finally, $[\delta D]$ was added to the rotated configuration $[R][A]$ to simulate a deformation.

5.2.2 Sensitivity Analysis - Input Simulation

For each Monte-Carlo iteration, all three inputs ($[A_N]$, $[R_N]$ and $[\delta D]$) were randomised based on a pre-defined range and distribution, where N is the total number of Monte-Carlo samples ($N = 5000$). First, the raw marker configuration matrix $[A_N] = a_{ij}$ was determined, defining the three-dimensional positions of the skin markers to simulate all possible geometrical configurations. In this study, $[A]$ imitates a four-marker based MSFM segment as a 4×3 matrix. The primary objective was to validate the use of σ_{RBE} to quantify rigidity for a four-marker MSFM segment, to be compatible with the investigations in Chapters 6 and 7. Each matrix element, a_{ij} , where the lower case letter with subscript ij refers to the entry of the matrix that represents the i^{th} marker and j^{th} the axes of the Cartesian coordinates of the marker $j \in (x, y, z)$, is randomly sampled over $[-250, 250]$ mm through a uniform distribution to simulate all possible geometrical positions that a landmark can take. A sampling region of a 500 mm^3 cube was defined, as a typical MSFM segment size is most likely to fall within this space. Therefore, the segment size will range from 0, where all simulated point coincide at a single point, to a theoretical maximum value of 433 mm, where all four points are generated on the vertices of the cube and calculated as: $4 \sqrt{250^2 + 250^2 + 250^2}/4 = 433$. Figure 5.2 shows examples of two randomly generated configurations.

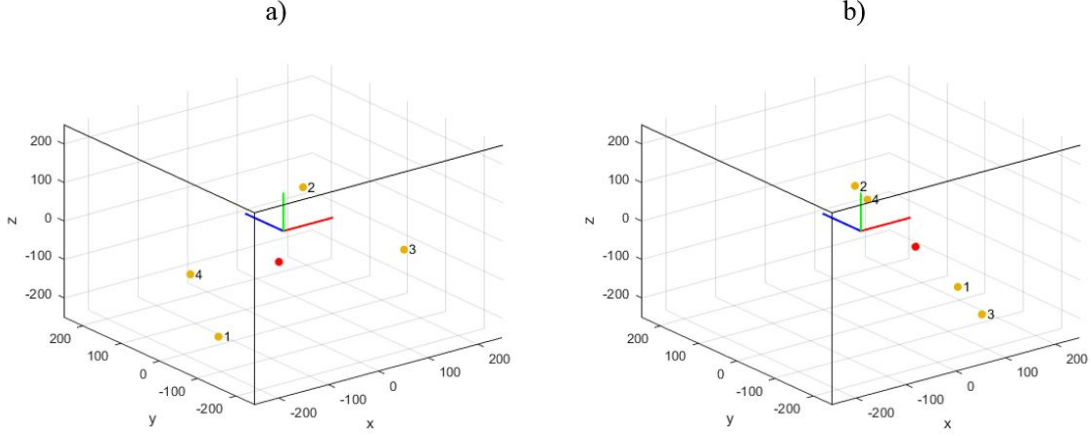


Figure 5.2: Example of randomly generated marker distributions. The red point is the centroid. a) shows a configuration that is well distributed, b) shows a marker configuration that is almost a straight line.

Once the raw marker configuration is generated, all four points are rotated about the origin by a random amount using the algorithm described by Arvo [8] to simulate movement that changes orientation. The simplest and most intuitive method of generating a random rotation matrix that covers all possible angles is by extracting three random Euler angles, each from a uniform distribution, followed by the computation of the rotation matrix through multiplying the three elemental rotation matrices. However, this method will not result in a rotation matrix that is uniformly distributed. Arvo [8] utilises a random reflection after the first rotation to produce a truly random rotation (Fig. C.1 b). In this study, the random rotation matrix is composed using three separate uniformly generated random inputs, two angles (β_N, γ_N) and one constant (X_N). The first angle, β_N , describes the random rotation about its z-axis and the second angle together with a constant (γ_N, x_N) describes the orientation of the hyperplane that reflects the point to any orientation (See Appendix C.2 for the full algorithm).

Finally, a deviation matrix $[\delta D]$ was added to the rotated marker configuration $[R][A]$ to simulate segment deformation. $[\delta D]$ is a $N \times 3$ matrix composed of one single uniformly distributed randomly generated element $\delta d_{1,1} \in [-50, 50]$, with the

rest of the elements equal to 0. Geometrically, this means that the first marker will deviate up to 50 mm away from its original position. This design allows for uniform distribution of deformation magnitude to be obtained across the full range of samples. Here the deformation magnitude is calculated as the matrix Frobenius norm of the deviation matrix, $\|\delta D\|$. If all the elements $\delta D_{i,j}$ of the deviation matrix $[\delta D]$ are allowed to vary, the deformation magnitude will not be uniformly distributed. In a typical MSFM segment, the major source of deformation would be the soft-tissue artefact (STA), with the largest being reported at the thigh region of up to 28.4 mm during walking [3, 44]. Therefore, the choice of up to 50 mm will certainly cover the maximum deviation magnitude resulting from any single marker in a realistic scenario. As $[A]$ is already a randomly generated configuration, the direction of the deviation does not have to be randomised. Table 5.1 outlines the computation steps for generating the Monte-Carlo samples and the algorithm for calculating the outputs.

Monte-Carlo based techniques provide direct estimation for the mean and standard deviation required to perform UA, as well as the sensitivity parameters based on various regression coefficients. The major drawback for the Monte-Carlo method is the computation cost and the requirement of a larger sample size [122]. Therefore, a convergence assessment was also conducted in this research to ensure parameter stabilisation (see Appendix D.1). The minimum number of samples required to reach our convergence criteria in this study was 3197, and therefore a sample size of $N = 5000$ was chosen for this study. This reduces the computation cost by avoiding an unnecessarily large sample size while ensuring the credibility of the sensitivity parameters.

Six parameters were examined in this SA: the direct input parameters, $\|\delta D\|_N$, β_N , γ_N , x_N , and those that characterise the geometrical distribution of $[A]_N$ with a likely influence on the σ_{RBE} ; that is, S and kA . The association of R_{err} with each input-variable was also investigated. The R_{err} is a direct measure of the difference between

$[R]$ and $[T]$ (See Table 5.1 - Step 6), which indicates the quality of the shape alignment. The sensitivity of the input variables will be determined using the two correlation coefficients, namely the Pearson's correlation coefficient (R) and the Spearman's ρ , due to the ease of computation and effectiveness in ranking sensitivity [96]. Coefficients of determination (R^2) and scatter-plots were used to identify the relationships between all variables and to confirm the validity of the sensitivity of input variables. A two-tailed t-test was used to establish the significance of the correlation coefficients (both R and ρ). All statistical analysis for the SA was completed using SPSS (v24, IBM Corp., Armonk, USA).

5.2.3 Regression Analysis

Linear regression is commonly used to describe the relationship between the dependent variable (here, the σ_{RBE}) and the independent variables [30, 213]. In this study, the independent variables are $\|\delta D\|_N, \beta_N, \gamma_N, x_N$, and $[A]_N$, which is characterised by S and kA . Initially, SA was conducted to identify non-sensitive variables (i.e. variables not correlated with the σ_{RBE}) and thus eliminate them from the subsequent regression analysis. Based on the relationship between sensitive variables and the σ_{RBE} , an appropriate regression model was developed along with a least square approximation to estimate its parameters. If the relationship between the dependent and the independent variable is not linear, curvilinear regression or data transformation may be applied. All statistical analysis for SA and regression analysis was completed using SPSS (v24, IBM Corp., Armonk, USA).

Steps	Mathematical representation
Generating Monte-Carlo Samples	
1. Generate random marker configuration $[A]$	$[A]$, where $a_{ij} \in [-250, 250]$
2. Apply random rotation $[R]$	$[R][A]$, where $\beta, \gamma \in [0, 2\pi]; x \in [0, 1]$
3. Add a random deviation $[\delta D]$	$[B] = [R][A] + [\delta D]$, where $\delta d_{1,1} \in [-50, 50]$
Compute best-fitting transformation matrix and all outputs	
4. Compute transformation matrix $[T]$	$[T]$, where $\min \ [T][A] - [B]\ $ describes the transformation from $[A]$ to $[B]$
5. Compute σ_{RBE}	$\sigma_{RBE} = \sqrt{\frac{1}{3N-6}} \ [T][A] - [B]\ $
6. Compute rotation error R_{err} based on the difference between $[R]$ and $[T]$	$\left\ R_{err} = \begin{bmatrix} R_{11} & R_{12} & R_{13} & t_x \\ R_{21} & R_{22} & R_{23} & t_y \\ R_{31} & R_{32} & R_{33} & t_z \\ 0 & 0 & 0 & 1 \end{bmatrix} - [T] \right\ $ where the translation vector $t = (t_x, t_y, t_z) = (0, 0, 0)$ since no translation was applied
7. Compute condition number kA	$kA = (\sigma_2^2 + \sigma_3^2)^{-1/2}$, where σ_2^2 and σ_3^2 are the two smallest singular values of $[A]$
8. Compute segment Size S	$S = \frac{1}{n} \sum_{i=1}^n [a]_{ij} - [\bar{a}] $, where $\bar{a} = \sum_{i=1}^n a_{ij}$, the geometric centroid of all the markers considered. (Here, $n = 4$ markers)

Table 5.1: Summary of Monte-Carlo computation steps and the computation of all outputs. Note that the computed $[T]$ in step 6 may contain non-zero t to give best-fit transformation.

5.3 Results

5.3.1 Uncertainty Analysis

The uncertainty analysis (UA) was conducted to validate the generated input parameters and to estimate the uncertainty of the outputs based on the samples generated.

Notation	Description	Unit	Range	Mean (S.D.)	Distribution
$ \delta D $	Deformation magnitude	mm	[0.008, 49.999]	25.21(14.52)	Uniform
a_{11}	Raw configuration elemental coordinate	mm	[-249.7, 249.9]	1.2 (146.1)	Uniform
β	Input for rotation matrix	rad	[0.00, 6.28]	3.15 (1.81)	Uniform
γ	Input for rotation matrix	rad	[0.00, 6.28]	3.12 (1.81)	Uniform
x_n	Input for rotation matrix	-	[0.00, 1.00]	0.50 (0.29)	Uniform

Table 5.2: Descriptive statistics of input parameters chosen for Monte-Carlo sampling. Note that only the first element of the marker configuration matrix is listed on the table for reference.

Notation	Description	Unit	Range	Mean (S.D.)
σ_{RBE}	Rigid body error	mm	[0.001, 17.47]	7.05 (4.31)
R_{err}	Rotation error	-	[0, 1.92]	0.100 (0.101)
S	Segment size	mm	[58.76, 332.63]	203.15 (39.55)
kA	Condition number	1/mm	[0.002, 0.026]	0.005 (0.002)

Table 5.3: Descriptive statistics of the output parameters estimated by Monte-Carlo Sampling.

The spread of the input parameters appears to be within the expected range and distribution, both statistically (Tables 5.2 & 5.3) and visually from the probability density functions (PDF) (Figs. 5.3 & 5.4). Figure 5.4 also revealed that the S matched a normal distribution, and R_{err} and kA appeared to match a lognormal distribution. Finally, the PDF of the σ_{RBE} appeared to be a uniform distribution skewed towards

higher values, differing from the distribution of $\|\delta D\|$. Based on a mean $\|\delta D\|$ of 25.21 ± 14.52 mm, the corresponding σ_{RBE} was estimated to be 7.05 ± 4.32 mm.

5.3.2 Sensitivity Analysis

Spearman's ρ was used as the primary measure of sensitivity, as the linearity between dependent and independent variables is unknown. The result indicates that the deformation $\|\delta D\|$ is the most dominant factor influencing the variability of the σ_{RBE} (Table 5.4). The correlation coefficients suggest that the relationship between $\|\delta D\|$ is highly linear ($R = 0.9311$) and highly monotonic ($\rho = 0.9393$). The sum of R^2 of all the sensitivity variables listed in Table 5.4 indicates that 86.77% of the σ_{RBE} variance is accounted for in this analysis, of which 86.70% is due to $\|\delta D\|$ alone. Apart from $\|\delta D\|$, none of the other variables ($\beta_N, \gamma_N, x_N, kA, S$) had any significant influence on or association with σ_{RBE} .

The relationships between the inputs and the σ_{RBE} were also visualised using scatter plots as shown in Figures 5.3 and 5.4. From both figures, only $\|\delta D\|$ appeared to have a strong relationship with the σ_{RBE} , as indicated by the high concentration of data points along the upward sloped line of best-fit. R_{err} was also plotted against the input variables and kA , as it indicates the quality of the transformation matrix fitting, hence affecting the accuracy of quantifying rigidity using σ_{RBE} . As expected, there seems to be no effect of the rotation matrix inputs (Fig. 5.3). However, a higher R_{err} appeared to be associated with samples with a higher $\|\delta D\|$, a smaller S , and a higher kA , with a p values of 0.7494, -0.2247, and 0.2256 respectively (Table 5.5).

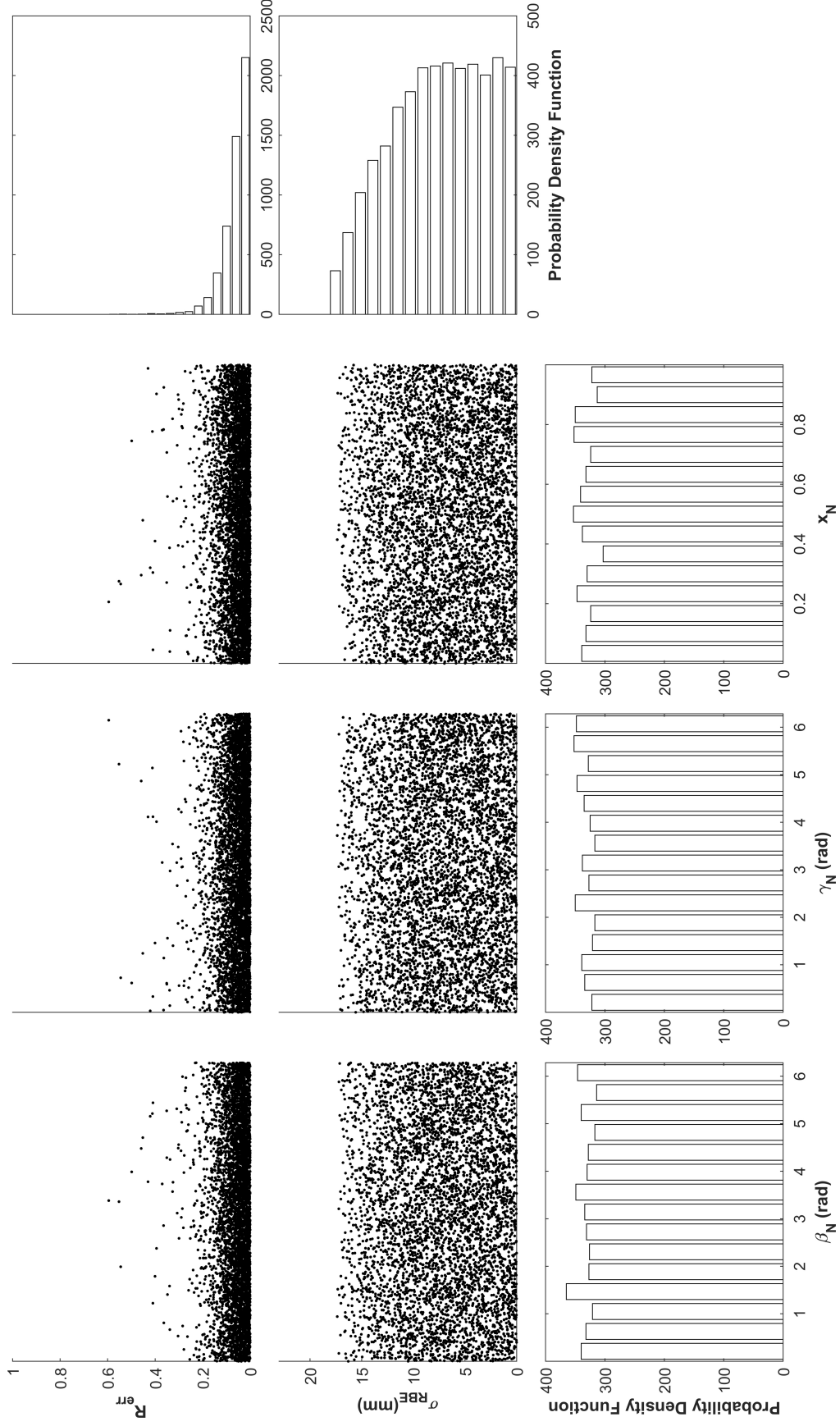


Figure 5.3: Scatter-plot of rotation error (R_{err}) and rigid body error (σ_{RBE}) against the three rotation matrix inputs β_N , γ_N , and x_N . The bottom and right most columns show the probability density function of the corresponding samples generated from Monte-Carlo sampling.

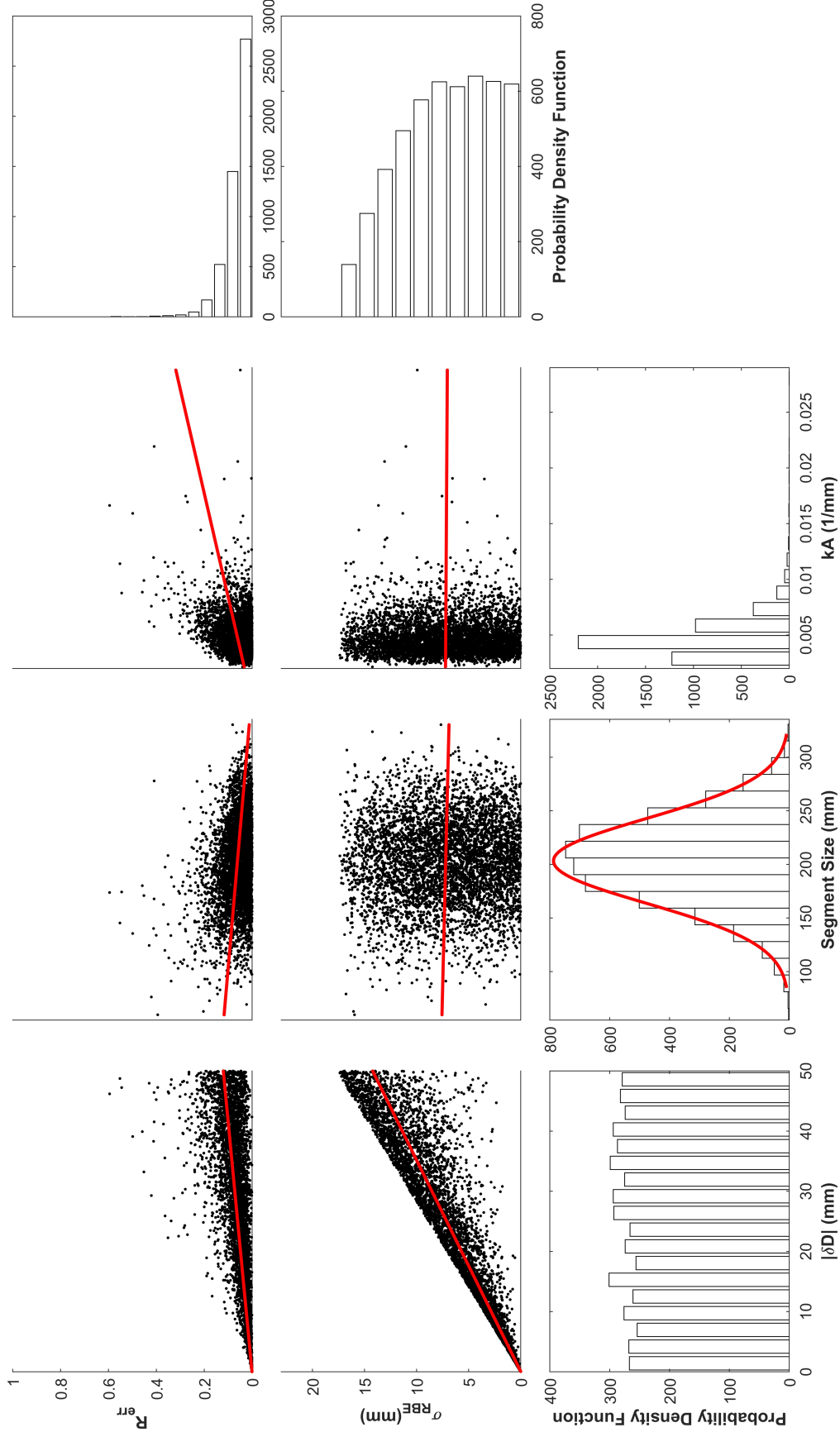


Figure 5.4: Scatter-plot of rotation error (R_{err}) and rigid body error (σ_{RBE}) against deviation magnitude ($||\delta D||$), segment size (S), and condition number (kA). The bottom and right most columns show the probability density function of the corresponding samples generated by Monte-Carlo sampling.

Sensitivity Parameter	Unit	Pearson's R	Spearman's ρ	R^2
$ \delta D $	mm	0.9311*	0.9393*	0.8670
S	mm	-0.0091	-0.0008	0.0001
kA	1/mm	0.0078	0.0098	0.0001
β_N	rad	0.0155	0.0159	0.0002
γ_N	rad	0.0127	0.0074	0.0002
x_N	-	0.0086	0.0067	0.0001
Sum	-	-	-	0.8677

*p-value < 0.01

Table 5.4: Sensitivity indices for the σ_{RBE} estimated by Monte Carlo sampling ($N = 5000$).

Sensitivity Parameter	Unit	Pearson's R	Spearman's ρ	R^2
$ \delta D $	mm	0.5968*	0.7494*	0.3561
S	mm	-0.2532*	-0.2247*	0.0642
kA	1/mm	0.3337	0.2256*	0.1113
β_N	rad	0.0294*	0.0159	0.0002
γ_N	rad	0.0127	0.0074	0.0002
x_N	-	0.0086	0.0067	0.0001
Sum	-	-	-	0.8677

*p-value < 0.01

Table 5.5: Sensitivity indices for the R_{err} estimated by Monte Carlo sampling ($N = 5000$).

5.3.3 Regression Analysis

In the current study, the technique of Monte-Carlo simulation was used to examine the relationship between σ_{RBE} and $\|\delta D\|$. It was hypothesised that a correlation exists between the two variables such that σ_{RBE} can be used to reasonably estimate $\|\delta D\|$. The result of the GSA (Section 5.3.2) showed that only $\|\delta D\|$ significantly contributed to the variation of σ_{RBE} . Therefore, the use of a single variable linear regression model is justified. The modelling process started with the assessment of assumptions for the fitting of a linear regression equation [30, 213], which includes:

1. The relationship between the independent and dependent variable is approximately linear.
2. Observations are independent.
3. There are no significant outliers.
4. The errors are normally distributed (homoscedasticity).

Assumption 1 can be considered as met based on observation from the scatter-plot (Fig. 5.4). The nature of using a large sample ($N = 5000$) Monte-Carlo simulation means that **Assumption 2** is also met due to random generation. To further justify this claim, a Durbin-Watson statistic of 1.956 was calculated. The Durbin-Watson statistic is a test statistic that detects the presence of autocorrelation, with a range between 0 and 4. A value of 2 suggests no autocorrelation, a value between 0 and 2 indicates positive autocorrelation, and a value between 2 and 4 indicates negative autocorrelation [84]. As predicted, its proximity to a value of 2 suggests there is no autocorrelation detected in this sample. From the σ_{RBE} vs. $\|\delta D\|$ scatter-plot (Fig. 5.4), it is clear that the assumption of homoscedasticity was violated. The variance of the error increases with increasing σ_{RBE} . Therefore, **Assumptions 3 & 4** were addressed prior to model-fitting.

Addressing Outliers

Depending on the location of the outliers in the xy space, outliers can have moderate to severe effects on the linear regression model [30, 56]. As a result of the random nature of a Monte-Carlo simulation, extreme geometries such as: when all the points form a straight line resulting in a poorly estimated transformation matrix (Fig. 5.2 a); or when all the points coincide to a single point, can be problematic in the estimation of a regression model. Therefore, it is essential to remove these points without removing natural data despite its unusual characteristics. A high standardised residual threshold of -3 (i.e. 3 standard deviations) was used to identify and mask any existing outliers, a validated strategy in identifying outliers [30, 56, 213]. In this study, 85 data points were individually identified, investigated, and excluded as outliers using the standardised residual threshold of -3, as shown in Figure 5.5 below:

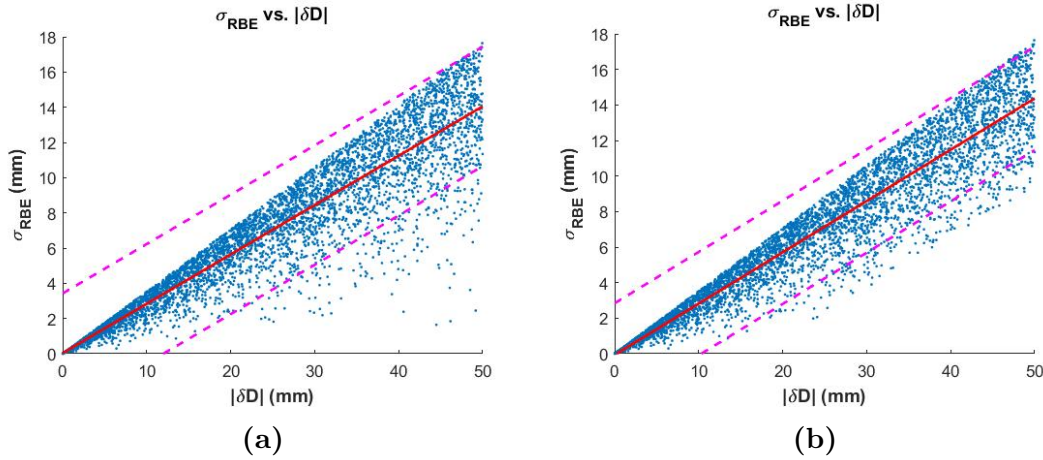


Figure 5.5: Dataset before (a) and after (b) the removal of outliers shown in the σ_{RBE} vs. $\|\delta D\|$ scatterplot.

Addressing Heteroscedasticity

Although the technique used in this study (ordinary least-square) to estimate the regression parameters does not require variance normality, heteroscedasticity affects variance prediction. Traditionally, data transformation can be applied to stabilise variance [19, 261] to address the linear regression assumption. However, data trans-

formation must be applied with caution and with interpretation as a goal in mind to prevent changing the nature of the study solely in the interest of satisfying the statistical assumptions [201].

Regression Equation

In this analysis, both the $\|\delta D\|$ and σ_{RBE} were log (natural logarithm) transformed, and their functional relationship can be expressed as an equation of a straight line:

$$\log(\sigma_{RBE}) = \beta_0 + \beta_1 \log(\|\delta D\|) + \epsilon \quad (5.1)$$

where β_0 is the intercept, β_1 the slope of the line, and ϵ the random error. Analysis of variance of the regression model showed that there is significant evidence ($p < 0.01$) to suggest that there is a linear association between the natural logarithm of the $\|\delta D\|$ and the natural logarithm of the σ_{RBE} (Table 5.6). This can also be observed in Figure 5.6, which includes a line of best fit and 95% confidence band. In addition, the obtained adjusted $R^2 = 0.959$ indicates that the model can explain 94% of the variation in the σ_{RBE} . The transformation is a clear improvement over the untransformed model ($R^2 = 0.867$), which is already a strong effect size according to Cohen [54]. Furthermore, both β_0 and β_1 are significant, showing that the model $\log(\sigma_{RBE}) = \beta_0 + \beta_1 \log(\|\delta D\|)$ is appropriate to describe the relationship between $\|\delta D\|$ and σ_{RBE} . In terms of the effects of changes in $\|\delta D\|$ on σ_{RBE} , with a value of β_1 equal to 1.004, this means that a 10% increase in the $\|\delta D\|$ would lead to a 10.04% increase in σ_{RBE} .

Term	Estimate	Std. Error	95% CI		t-Stat	p-value
			Lower	Upper		
β_0	-1.288	0.009	-1.308	-1.270	-141.1	0.000
β_1	1.004	0.003	0.998	1.010	336.0	0.000

Table 5.6: Coefficients of regression model expressed in Eq. (5.1)

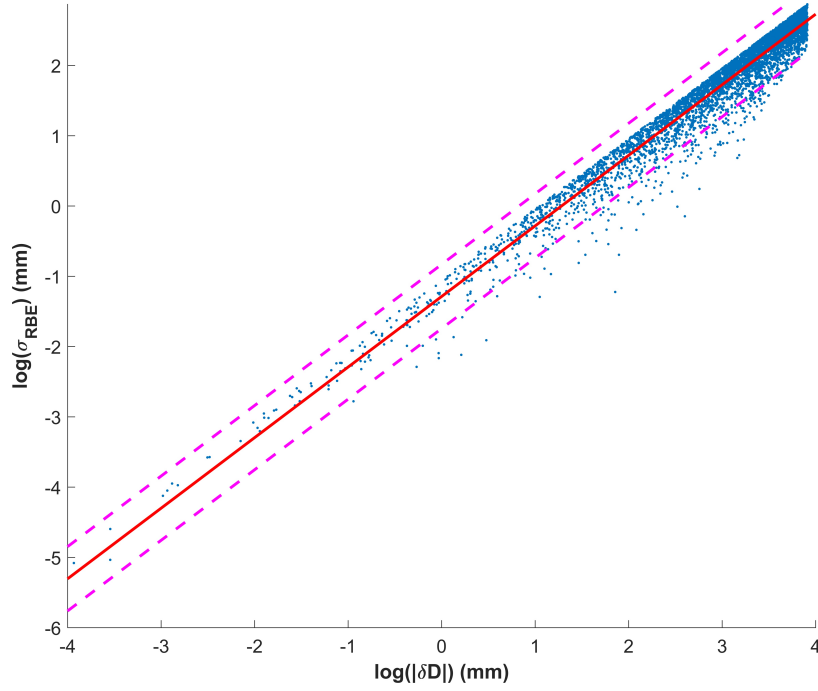


Figure 5.6: Fitted plot of $\log(\sigma_{RBE})$ vs. $\log(\|\delta D\|)$, with a line of best-fit (Solid line) and a 95 % confidence interval (dotted line).

5.4 Discussion

5.4.1 Sensitivity Analysis

In this study, Monte-Carlo based techniques have been utilised to conduct GSA and UA, which showed that σ_{RBE} is mostly sensitive to deformation magnitude $\|\delta D\|$. Furthermore, this study confirmed that σ_{RBE} is associated linearly with $\|\delta D\|$, based on a linear RA. The main objective of the SA was to identify the most critical variables that influences the output through obtaining a ranking of the input variables using

a specific ‘importance’ or ‘sensitivity’ index [199]. Hamby [103] outlined 14 possible sensitivity indices for cases where only a single output variable is to be evaluated, and Saltelli et al. [220] reviewed a total of 15. Although the baseline strategy for conducting a global sensitivity analysis has already been established in the literature, the result may be different depending on the techniques used (OAT, Monte-Carlo, FAST, etc.) and the indices that rank the variables. In this analysis, R and Spearman’s ρ were used to rank and compare the sensitivity of the variables to determine the parameter sensitivity as the coefficients can be easily computed and effective in ranking sensitivity [96, 222]. Since linearity cannot be assumed between the input variables and the σ_{RBE} , Spearman’s ρ was chosen to be the main factor in ranking the sensitivity.

All the input variables were defined to represent the variability of the different geometries and motions. By randomly generating four points within the space of 500 mm³, we allowed various geometries to be generated, with an average segment size of 203 mm. In gait analysis, segments are defined differently and can be in different shapes and sizes [73, 210]. Therefore, the robustness of this assessment depends on the ability of the algorithm to generate as many geometries as possible. By utilising random marker configuration generation within a confined space to increase the range of possible shapes, it also allows for bad geometries (i.e. abnormal shapes that do not realistically represent a segment) to be included in the assessment. This resulted in an underestimation of the variable sensitivity and model accuracy. Söderkvist & Wedin [244] have shown that the quality of the transformation matrix fitting depends on the initial landmark configuration. The problem of determining the rotation matrix can be very ill-conditioned when the singular values of $[A]$ are very small, i.e. the landmarks are almost on a straight line. Therefore, when quantifying rigidity of an MSFM segment, it is also important to make sure that the general marker configuration is well dispersed and the value of kA is low for quality assurance. A previous

study by Dorst [80] confirmed that the rotation error is oriented with the major axis of the deformed configuration $[B]$.

It is acknowledged that there were several limitations to the SA. In this study, the deformation has only been applied to one single marker. The focus of this study has been to investigate the correlation between the applied deformation magnitude $\|\delta D\|$ with the σ_{RBE} . With this in mind, the experiment was designed to generate a uniformly distributed $\|\delta D\|$, which is not possible when introducing perturbations in multiple markers at the same time. Finally, the UA and SA were only performed on a MSFM segment with four-markers. Although adding more markers will only increase the accuracy of determining $[T]$ [244], further analysis is required to understand the precise characteristic of how different configuration shape, deformation, and the number of markers affect the efficacy of using σ_{RBE} to quantify rigidity. The current scope of this analysis only considers MSFM with a set of four tracking markers.

5.4.2 Regression Analysis

To show that σ_{RBE} can be used to quantify rigidity, a regression model was proposed to approximate the relationship between σ_{RBE} and the simulated deformation magnitude $\|\delta D\|$. The regression model (Eq. 5.1) together with the approximated parameters (Table 5.6) indicated a very good fitting of the simulated data with an adjusted coefficient of determination $R^2 = 0.959$. However, as with all mathematical models, there are some imperfections.

It is acknowledged that there are several limitations to this model. Despite the log-log transformation, the scatter-plot (Fig. 5.6) still indicates that the residuals are negatively skewed, with a gap between the upper bound and the majority of the samples points. Although homoscedasticity does not cause bias in the estimation of the regression model parameters [213], it will likely underestimate the variance of the σ_{RBE} at a large $\|\delta D\|$ and overestimate the variance at a small $\|\delta D\|$. Finally, some

outliers were also observed at the bottom right of the line of best fit, which could also potentially impact the estimation of the model parameters. Overall, it is noted that a better regression is needed if the primary goal is to predict the specific values and variances of σ_{RBE} given the $\|\delta D\|$. Nevertheless, the result still clearly indicates a strong and predictable relationship between σ_{RBE} and $\|\delta D\|$, which justifies the use of σ_{RBE} to quantify and compare rigidity amongst MSFM segments as a more robust and validated method to the existing approach [219]. Furthermore, this study also validates the conclusions of Chapters 6 and 7.

5.5 Summary

In this Chapter, a Monte-Carlo based GSA have been successfully applied to show that the deformation magnitude $\|\delta D\|$ is the most important variable of the σ_{RBE} , and that it accounts for 86.7% of the total variation as indicated by R^2 . The sensitivity analysis further showed that the σ_{RBE} does not depend on the segment size or rotational motion. However, the geometric distribution of the markers appeared to have an effect on the computation of the σ_{RBE} . Nevertheless, the highly linear and monotonic relationship between the σ_{RBE} and $\|\delta D\|$, along with the proposed linear regression model, suggests that σ_{RBE} is an effective method to quantify rigidity and track deformation caused by either bone motion or soft tissue artefact in a MSFM segment. Therefore, the result of this study have justified the use of σ_{RBE} to quantify segment rigidity in a MSFM. In the following two Chapters (6 and 7), MSFM segment rigidity will be quantified and tracked using σ_{RBE} , followed by the proposal of novel MSFM based on the guidelines recommended from the result of these Chapters.

Chapter 6

Quantifying Oxford Foot Model Segment Rigidity

A multi-segment foot model (MSFM) is a skeletal model that represents a simplified biofidelic human foot structure as a system of rigid body segments and joint articulations. In order to provide an accurate and clinically useful representation of the underlying anatomy and its function, the rigid body assumption should be sufficiently satisfied. This chapter investigates the validity of such a claim on marker clusters used to define the individual segments (tibia, hindfoot, forefoot) of the Oxford Foot Model (OFM). In addition, this chapter characterises, for the first time, the deformation of OFM segments throughout a full gait cycle using rigid body error [264]. Furthermore, to demonstrate potential clinical applications, cases studies are also included to compare normal to pathological gait patterns. This Chapter presents a more comprehensive version of a manuscript published by the Journal of Biomechanics in 2019 [46], with the addition of clinical case studies.

6.1 Introduction

Although MSFMs are designed to reduce the complex foot anatomy to a small number of rigid bodies in order to simplify the measurements and calculations, an oversimplified model may not be representative of the underlying anatomy, and hence models may not be versatile enough to adequately represent foot deformity. There is a debate as to whether it is appropriate to consider the whole forefoot as a single segment, particularly given the anatomical division of the metatarsals into medial and lateral functional units based on their connections to the bones of the mid- and hindfoot. Whether the forefoot should be modelled as a rigid body is questionable, given the known flexibility of the medial longitudinal arch and the mediolateral spread of the forefoot during stance phase [82].

This study aimed to use the rigid body error (σ_{RBE}) to quantify the rigidity of forefoot, hindfoot, and tibia segments of a MSFM widely used in clinical gait analysis (OFM)[42, 250]. For normal healthy adults, it was hypothesised that on average the forefoot marker-cluster would be less rigid than the hindfoot and the tibia. In addition, the patterns of deformation of each marker-cluster would differ over the gait cycle. To explore the possible clinical utility of quantifying tracking of segment rigidity, comparisons were made between normal subjects and two patients with cerebral palsy (CP). It was expected that less deformation and an abnormal deformation pattern would be observed in forefoot segment of the CP patients, where bone deformity, restricted range of motion (ROM), and stiffness due to contractures and muscle weakness are often present [177, 236].

6.2 Methodology

6.2.1 Ethical Considerations

As outlined in Section 4.1, here, the motion data were obtained from the Oxford Gait Laboratory’s (OGL) normal database, approved by a local ethical committee. For the pathological gait case studies, data was obtained under the same ethical approval that allowed research use of clinical data.

6.2.2 Subjects and Participants

Sixty-four healthy adult subjects (32 males and 32 females) with an average age of 27.3 ± 7.0 years participated in this study, as described in Section 4.2.

6.2.3 Data Collection and Processing

The data collection and processing procedure can be found in Sections 4.4 and 4.5 respectively using equipment detailed in Section 4.3. One of six walking trials (right side only) was selected for further analysis based on how close its joint rotation patterns were to the average for that subject. The quantity used for the comparison was the root mean square error. For quality assurance, the temporal-spatial parameters and joint angles were inspected to ensure that there were no abnormal features [250, 255].

The σ_{RBE} for three lower-limb body segments (tibia, hindfoot, and forefoot) was calculated over one full gait cycle using MATLAB (v2015a, MathWorks Inc., Natick, USA). See Section 4.8 for details on the computation of σ_{RBE} . For each segment, four markers were selected to calculate the σ_{RBE} , as shown in Figure 6.1.

Here, the reference pose for the calculation of σ_{RBE} was taken to be the marker configuration at mid-stance (50% stance phase). This foot position was preferred to

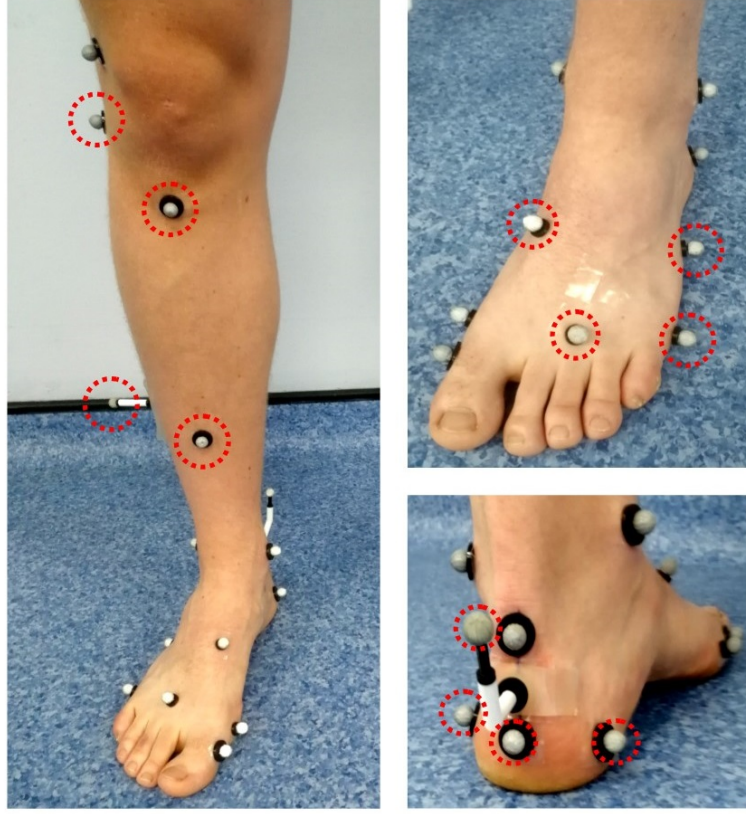


Figure 6.1: The OFM marker set. A cluster of four markers on each segment (shown circle in red) was used in the calculation of σ_{RBE} . On the tibia, markers TUB, HFB, TIB, and SHN were used to calculate rigidity. For the hindfoot, it was CPG, STL, LCA, and HEE. For the forefoot, it was P1M, TOE, P5M, and D5M. See Table 4.1 for detailed marker description

the static standing position since the use of the latter resulted in a more variable σ_{RBE} , as shown the increased standard deviation (Figs. 6.2 and 6.3). The average rigid body error (σ_{av}) was used to indicate the average rigidity of the segment, and the peak-to-peak rigidity σ_{ptp} was chosen to characterise the maximum deformation of the marker-cluster.

6.2.4 Statistics

For the comparison of the σ_{RBE} between foot segments in the normal population, the data were analysed using a two-way repeated measures ANOVA with gender (male

or female) as the between-subject factor and body segment (forefoot, hindfoot, and tibia) as the within-subject factor. The motivation behind this investigation was that there are known anthropometric differences between the male and female foot [94], which may influence the dynamic shape deformation throughout gait.

On the basis of the anticipated mean (here, used 2.0), a sample size of 14 is required to obtain statistical power at the recommended 0.8 level [54]. This study involved 32 subjects, and thus sufficient. Mauchley’s test was conducted to inspect the homogeneity of variance across within-subject factors. If significant, the Greenhouse-Geisser corrected p-value was reported, otherwise, the uncorrected p-value was used. If significant interactions ($p < 0.05$) were found, post-hoc comparisons were performed using an unpaired t-test with Bonferroni corrections. All statistical calculations were completed using SPSS (v24, IBM Corp., Armonk, USA).

6.2.5 Repeatability Analysis

To assess the repeatability of the measurement protocol for different levels of expertise, as well as the variability of the σ_{RBE} , one normal male participant underwent eight gait analysis sessions with data collected as described in Section 6.2.3. Sessions were conducted by two staff members: one (senior) who had 16 years of experience in clinical gait analysis, and another (junior) who had one year of gait analysis and marker-placement experience and no clinical experience. Each staff member performed two sessions on the same day, two hours apart, with the process repeated two weeks later. All markers were removed and reapplied for each new session. The repeatability of σ_{RBE} , defined as the variation in repeated measurements made on the same subject under identical conditions [18], were analysed and used to compute the inter-trial error (subject variability), intra-rater, and inter-rater error [234]. The ratio of the inter-rater error to the inter-trial error was also calculated since it reveals the effect of methodological errors.

6.2.6 Clinical Case Studies

To explore the possible clinical utility of segment rigidity as quantified by σ_{RBE} , two adult male patients with cerebral palsy (CP) were chosen from a clinical research database (Oxford Gait Laboratory, Oxford, UK). Both patients had spastic diplegia. Their walking data was analysed and compared to that of the normal healthy subjects. Patient 1 (age 27 years) had a negative windlass mechanism (arch not re-gained upon toe-rise) on his left foot. Without a functioning windlass mechanism, a decrease in forefoot deformation was expected. Patient 2 (age 27 years) had bilateral fixed ankle equinus (10° left, 15° right) and a midfoot break bilaterally. It was hypothesised that the forefoot deformation pattern in both patients would be different from healthy adult males.

6.3 Results

6.3.1 Segment Rigidity

As hypothesised, the patterns of deformation of the marker clusters differed over the gait cycle (Figs. 6.2 and 6.3). Use of the static standing position for the reference marker cluster configuration resulted in a more variable σ_{RBE} , as indicated by the standard deviations in Tables 6.1 and 6.2. Regardless of the reference position, the hindfoot cluster stayed relatively undeformed in comparison to the forefoot and tibia clusters.

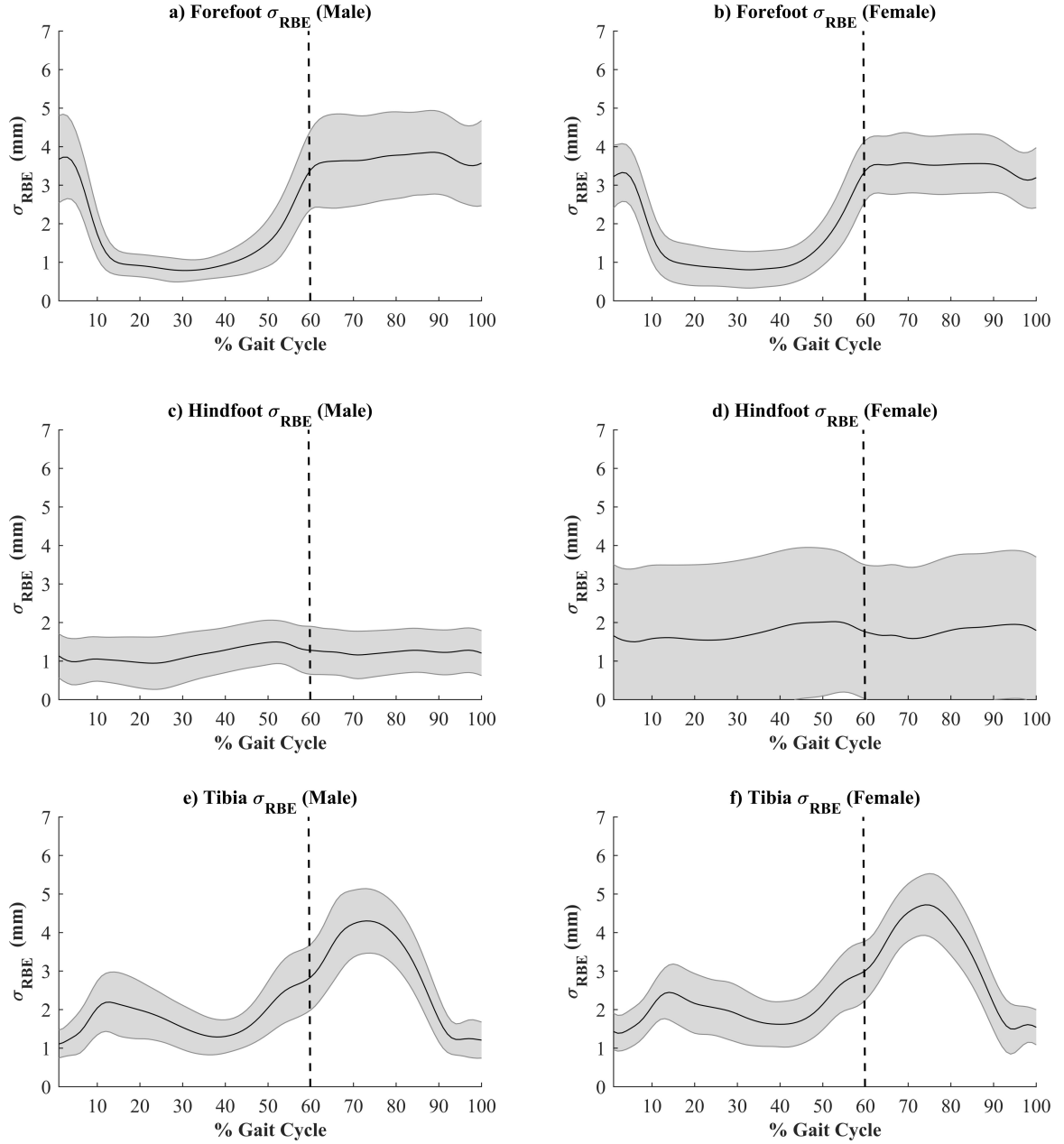


Figure 6.2: Forefoot, hindfoot, and tibia marker cluster rigid body error (σ_{RBE}) over one full gait cycle with the cluster configurations in the static trial as the reference. Means ± 1 standard deviation (mm). The left column shows kinematic graphs for male subjects and the right column for female subjects. The dotted line indicates the separation between stance and swing phase, with an average of 59.8% for the population tested in this study. Note that, on average, σ_{RBE} is never equal to zero, since the marker cluster configurations during gait never match their configurations in the reference position.

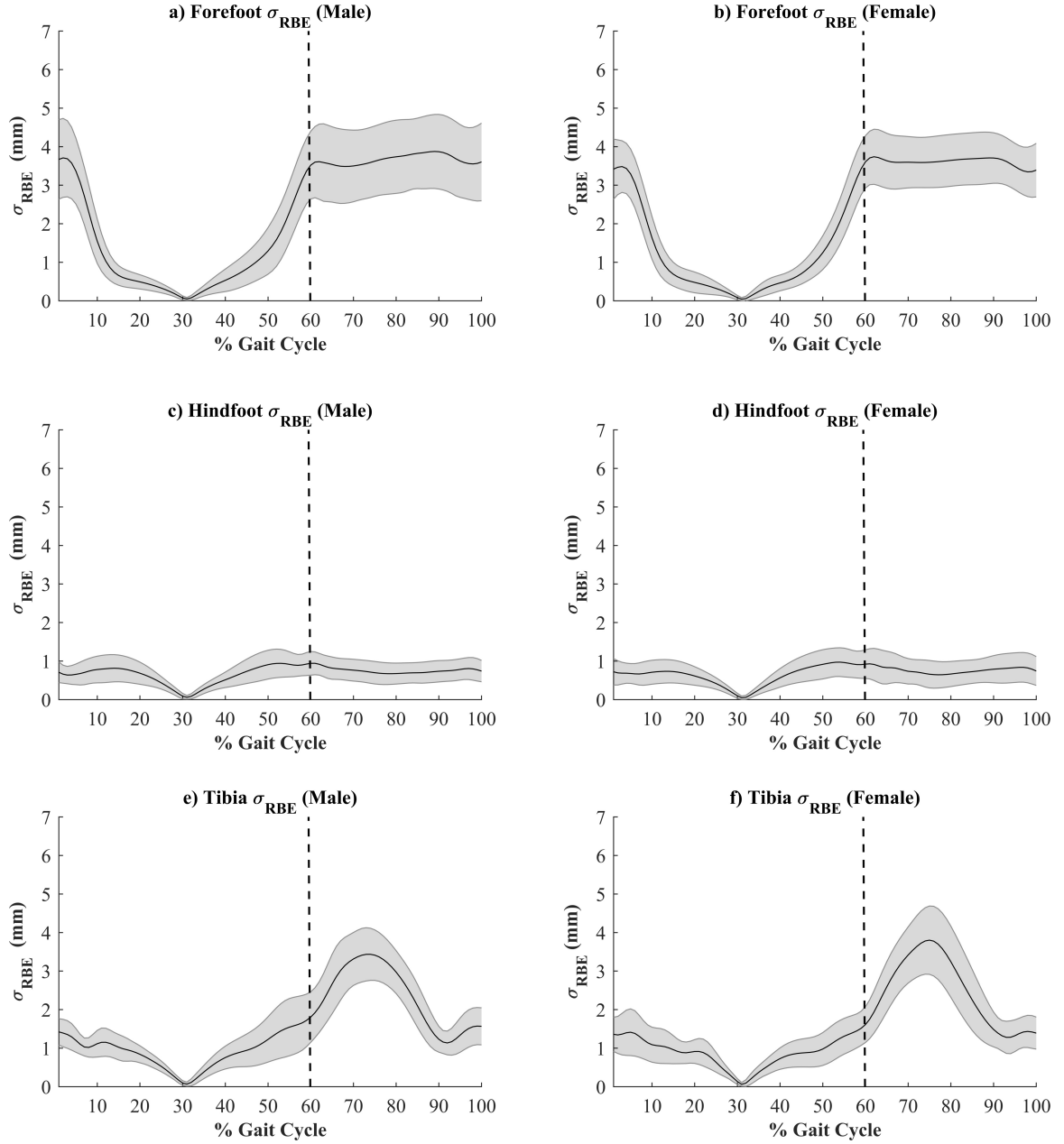


Figure 6.3: Forefoot, hindfoot, and tibia marker cluster rigid body error (σ_{RBE}) over one full gait cycle with the cluster configurations in mid-stance(50 % stance) as the reference. Means ± 1 standard deviation (mm). The left column shows kinematic graphs for male subjects and the right column for female subjects. The dotted line indicates the separation between stance and swing phase, with an average of 59.8% for the population tested in this study. Note that, by definition, $\sigma_{RBE} = 0$ at the reference position.

	n	Forefoot		Hindfoot		Tibia	
		σ_{av}	σ_{ptp}	σ_{av}	σ_{ptp}	σ_{av}	σ_{ptp}
Male	32	2.43 (0.61)	3.44 (1.12)	1.20 (0.55)	0.90 (0.29)	2.29 (0.50)	3.64 (0.84)
Female	32	2.33 (0.52)	3.24 (0.70)	1.76 (1.89)	1.01 (0.33)	2.53 (0.51)	3.75 (0.73)
Total	64	2.73 (0.57)	3.34 (0.94)	1.47 (1.40)	0.96 (0.32)	2.40 (0.52)	3.69 (0.78)

Table 6.1: Average rigid body error (σ_{av}) and peak-to-peak rigid body error (σ_{ptp}) [shown as mean (standard deviation)] for the forefoot, hindfoot, and tibia marker-cluster over one full gait cycle, with the static trial used as the reference configuration. Units of σ are mm.

	n	Forefoot		Hindfoot		Tibia	
		σ_{av}	σ_{ptp}	σ_{av}	σ_{ptp}	σ_{av}	σ_{ptp}
Male	32	2.24 (0.50)	4.11 (0.95)	0.68 (0.15)	1.21 (0.29)	1.50 (0.24)	3.52 (0.69)
Female	32	2.21 (0.34)	4.14 (0.67)	0.68 (0.21)	1.26 (0.37)	1.53 (0.24)	3.87 (0.85)
Total	64	2.22 (0.43)	4.12 (0.82)	0.68 (0.18)	1.24 (0.33)	1.51 (0.24)	3.67 (0.79)

Table 6.2: Average rigid body error (σ_{av}) and peak-to-peak rigid body error (σ_{ptp}) [shown as mean (standard deviation)] for the forefoot, hindfoot, and tibia marker-cluster over one full gait cycle, with the cluster configurations at mid-stance (50 % stance) as reference. Units of σ are mm.

For the forefoot cluster, the largest deformation occurred near the beginning and end of the stance phase, as indicated by the sharp fall and rise in σ_{RBE} (Figs. 6.2 and 6.3). As opposed to what was expected, the tibia cluster also deformed throughout the entire gait cycle. There was no statistically significant interaction between the effects of gender and σ_{av} ($F(2, 62) = 0.229, p = 0.80$); however, there were statistically significant differences between the average rigidities of the three segments $p < 0.01$ (Table 6.2). As hypothesised, the forefoot was the least rigid segment. It had the largest σ_{av} , followed by the tibia and then the hindfoot. The p -value was less than 0.01 in all pairwise comparisons between the three segments. The hindfoot segment exhibited the lowest σ_{ptp} ($p < 0.01$). The tibia σ_{ptp} was significantly lower than that of the forefoot ($p < 0.01$).

6.3.2 Repeatability Analysis

	Inter-trial error (mm)			Intra-rater error (mm)		Inter-rater error (mm)	$\frac{\text{Inter-rater}}{\text{Inter-trial}}$
	Junior	Senior	Mean	Junior	Senior	Mean	
Forefoot	0.157	0.130	0.143	0.230	0.154	0.257	1.79
Hindfoot	0.080	0.075	0.077	0.105	0.094	0.110	1.42
Tibia	0.178	0.154	0.166	0.299	0.245	0.238	1.43

Table 6.3: Repeatability of σ_{RBE} , as quantified by the inter-trial, intra-rater, and inter-rater errors, and the inter-rater/ trial ratio.

Of the three segments considered by the two assessors, the hindfoot had the most repeatable σ_{RBE} , with the lowest intra- and inter-rater errors (Table 6.3). The least reliable cluster for σ_{RBE} was at the forefoot segment, with the highest inter-rater error and inter-rater/ trial ratio. The rater with more experience in gait analysis achieved a more consistent results, as indicated by their lower intra-rater error for all three segments.

6.3.3 Clinical Case Studies

Both patients showed abnormal segment deformations, particularly at the forefoot and tibia (Fig. 6.4). The length of the stance phase for patient 1 was similar to normal, as in general were the deformation patterns for the right leg. His left forefoot and tibia deformations were much reduced compared to the healthy population. Patient 2 had a much longer-than-normal stance phase, and the rigid-body errors were reduced in both legs, most noticeably in the second half of stance phase.

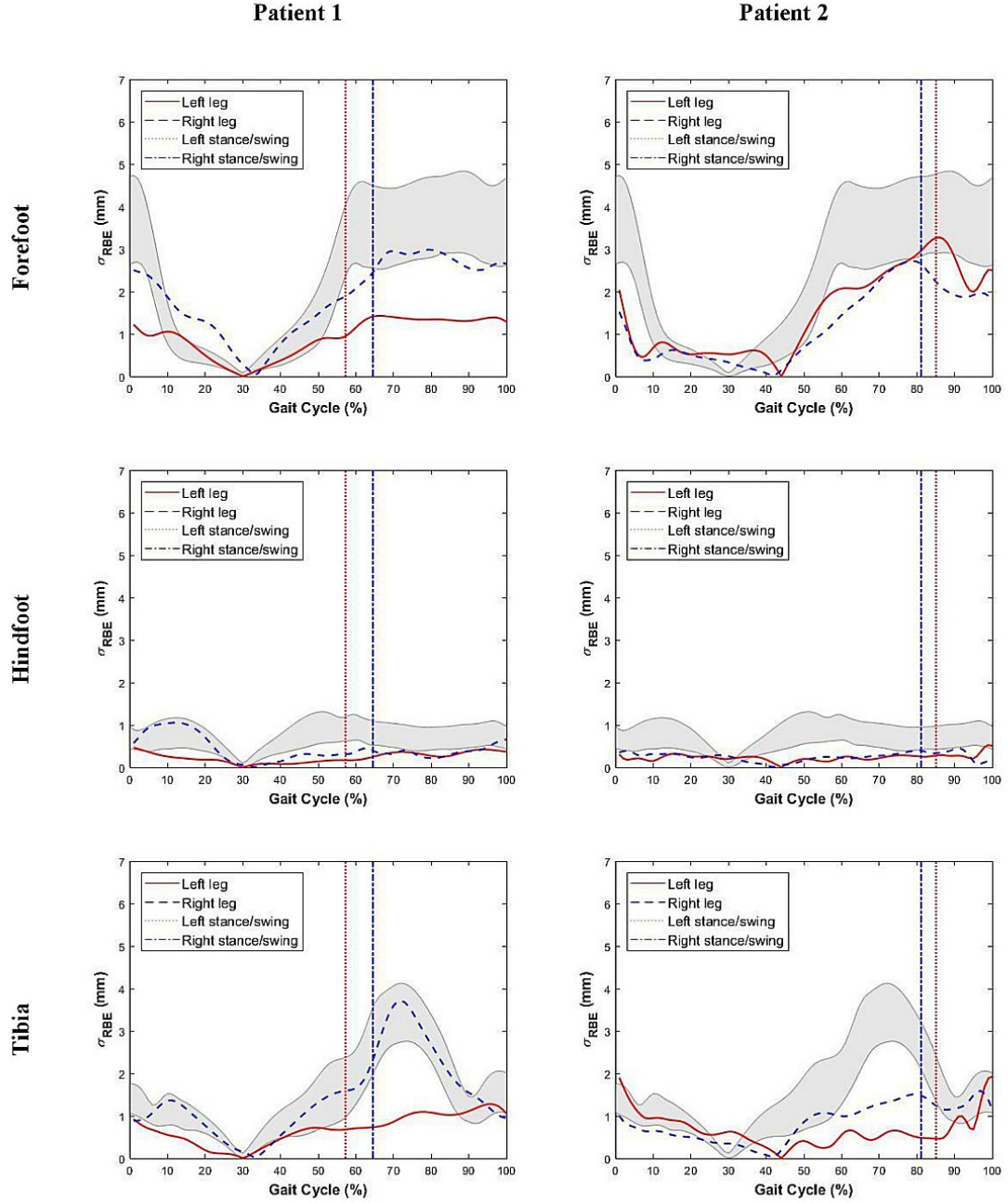


Figure 6.4: Forefoot, hindfoot, and tibia σ_{RBE} of two patients with cerebral palsy, patient 1 without functioning windlass mechanism and patient 2 with equinus deformity. The grey band shows the normal σ_{RBE} values (means ± 1 standard deviation; from Fig. 6.3).

6.4 Discussion

This part of the overall investigation aimed to use σ_{RBE} to quantify rigidity of the forefoot, hindfoot, and tibia segments of a multi-segment foot model widely used in clinical gait analysis. For normal healthy adults, it was confirmed that on average the forefoot was less rigid than the hindfoot and the tibia and that the patterns of deformation of the segments differed considerably over the gait cycle.

From the result of the σ_{av} , none of the segments was perfectly rigid. On average, the forefoot σ_{av} was greater than that of the tibia, followed by the hindfoot segment. This is expected since all hindfoot markers are attached to one bone - the calcaneus, whereas the forefoot markers are spread across the metatarsals. Both soft-tissue movement and segment flexibility are captured in the σ_{RBE} , and although the latter is likely to be more dominant in the forefoot, this study is unable to verify how much each effect contributes to the total σ_{RBE} .

The deformations of the hindfoot and forefoot segments are consistent with our general understanding of how foot shape changes during gait [11, 154]. The forefoot markers most nearly resemble their mid-stance configuration during the single-limb support phase of stance (hence the lower σ_{RBE} in this period); this is when the load is transmitted through the forefoot and the metatarsals are most separated from each other. The forefoot deforms rapidly (with respect to the single limb stance) at two phases in the gait cycle, during loading response and pre-swing. At loading response, the double support period between the initial contact and opposite toe-off, the foot is lowered to the ground through plantarflexion of the ankle. It is during this period that distance between the first and fifth metatarsal increases [82]. During pre-swing, the forefoot returns to its unloaded configuration.

The tibia marker-set deformed quite substantially throughout the entire gait cycle, with the plot of σ_{RBE} (Figs. 6.3 e, f) mildly resembling a characteristic knee

flexion-extension angle graph. The largest deformation occurred during mid-swing to terminal swing. Since the only source of bone motion is between tibia and fibula, which are known to have a very limited mobility between the two bones [27, 243], any excessive deformation was most likely due to the skin or other soft tissue movements. Although it was hypothesised that the marker causing the largest sliding motion was the one attached to the fibula head, which would make sense given the resemblance of the deformation pattern with the knee flexion-extension angle graph due to the proximity of the knee joint centre, further investigation is required to confirm this.

In this study, we have also compared the segment rigidity between normal subjects and two patients with CP. It was expected that smaller and abnormal deformation patterns would be observed in the forefoot segment of the cerebral palsy patients, where bony deformity, restricted range of motion, and stiffness due to contractures and muscle weakness are often present [177, 236]. Patients with impaired windlass are often associated with a collapsed arch and limited range of motion between the forefoot, midfoot and hindfoot [257], which sometimes requires surgical intervention. Quantifying rigidity may be of aid in objectively indicating the stiffness of the foot. In addition, we have also shown a difference in deformation pattern σ_{av} and a lower σ_{av} for the case of a patient with equinus deformity, which may indicate foot stiffness. However, according to the clinical gait analysis kinematic graphs [250], the range of motion of the forefoot/ hindfoot here observed articulation appears to be normal (See Appendix B). σ_{av} can potentially be used to quantify stiffness that may not be indicated on the joint angle graphs, providing additional information to the gait assessment.

Graphs such as Figure 6.4 could be useful clinically in quantifying the progression over time of a foot condition such as flatfoot and overall foot deformity. Although further investigation is required to confirm the correlation between impaired windlass, foot deformity, and forefoot rigidity, we have demonstrated a possible clinical utility.

Residuals from the segment tracking algorithm have been used to assess the tracking error [245, 246], soft tissue artefact [264], and help minimise rotation error [152]. In this study, we utilised the residual to quantify the rigidity of the individual foot segments of a MSFM (the OFM). The residual of each frame throughout the gait cycle was calculated with reference to the configuration of the mid-stance (50 % gait cycle). Both static (Fig. 6.2) and mid-stance poses (Fig. 6.3) were considered to be used as the reference configuration, but we found that mid-stance was more appropriate and reliable due to the variability of the standing posture. It is also worth noting that different marker configurations on the same segment may yield different results.

In Chapter 5, σ_{RBE} has been justified to quantify rigidity in a MSFM through a global sensitivity analysis. However, the problem of determining the rotation matrix can be ill-conditioned when the marker-cluster distribution are bad, that is, when the markers are almost on a straight line with an associated high value of kA . In this Chapter, the average kA for the OFM forefoot segment was found to be 0.004 ± 0.002 for all the healthy subjects. This implies that the geometrical configurations of the segments analysed in this study are associated with a small rotation error.

6.5 Summary

In summary, the overall rigidity and deformation of marker clusters representing the tibia, hindfoot, and forefoot segment have been successfully quantified and tracked using σ_{RBE} . Using the mid-stance posture as reference, the forefoot cluster was the least rigid, followed by the tibia, then the hindfoot cluster. The potential clinical application in using σ_{RBE} as a metric to quantify the stiffness of the foot was also demonstrated in this study. However, further analysis is needed to establish a connection between σ_{RBE} with clinically meaningful biodynamic parameters, such as joint motion, arch height, or existing gait deviation indexes.

The following Chapter will employ σ_{RBE} to quantify rigidity of the metatarsal pairs in order to investigate the most appropriate way of grouping and modelling the forefoot in a MSFM.

Chapter 7

Forefoot Segmentation and Rigidity of Metatarsal Pairs

The previous Chapter applied the orthogonal Procrustes method to quantify the overall rigidity of marker clusters representing the tibia, hindfoot, and forefoot segments of the traditional Oxford Foot Model (OFM). In this Chapter, the same technique will be applied to quantify the rigidity of individual metatarsal pairs for the purpose of guiding forefoot segmentation in the perspective of a possible extension of the OFM.

7.1 Introduction

Multi-segment foot models (MSFMs) such as the OFM [42, 250], Rizzoli Foot Model [143], and Milwaukee Foot Model [135] have been used extensively for both clinical and research purposes to analyse human locomotion.

Despite a wealth of evidence supporting the use of MSFMs [69, 146], there is currently no consensus on exactly how the forefoot should be divided. While most models consider the forefoot to be a single rigid segment, others have subdivided it into two, three, or even more sub-units [210]. Inconsistencies in foot representation and measurement between gait laboratories make comparing and sharing results difficult. In addition, contradictory results may arise due to the foot being represented differently [209].

This Chapter aims to explore possible groupings of the forefoot bones (metatarsals) based on a rigidity criterion, as quantified using the rigid body error (σ_{RBE}). Due to the exploratory and interpretative nature of this study, a combination of quantitative and qualitative approaches will be used to make recommendations on suitable forefoot groupings for the development of a novel MSFM.

7.2 Methodology

7.2.1 Ethical Considerations

As detailed in Section 4.1, ethical approval was obtained from the local Ethics Committee prior to the commencement of this study.

7.2.2 Subjects and Participants

Walking data were collected from 23 healthy adult volunteers (age 27.6 ± 7.3 years; 12 male and 11 female) with no known foot injuries or deformities using a motion capture system (See Sec. 4.2 for more detail). Visual inspection of the feet confirmed that they were neutral with no deformities, and the following motion tracking using only markers for the OFM (Sec. 4.6.2) showed that joint rotations were normal when compared to the normal adult database of the Oxford Gait Laboratory.

7.2.3 Data Collection and Processing

In order to track the movement of the forefoot, ten retro-reflective passive markers (10 mm in diameter) were attached to the dorsal aspect of the proximal base and distal heads of all five metatarsals (Fig. 7.1). For the first and fifth metatarsals, markers were placed on the medial and lateral aspects, respectively, of the bases and heads of the two bones, as for the OFM. For the other three metatarsals, the distal markers were located on the heads of the bones, and the proximal markers were sited just distal to the bases of the bones. This was done to ensure that metatarsal motion, rather than a combination of metatarsal and mid-foot motion, was tracked. Subjects were instructed to walk barefoot in a straight line along a 10 m walkway at a self-selected speed. See Section 4.4 for the full experimental protocol on the procedure of data collection, Section 4.5 for data processing, and Section 4.3 for equipment used in this study.

A metatarsal pair was defined by the four markers placed at the heads and bases of two metatarsals. All ten metatarsal pairings were considered (I/II, I/III, and so on). Pairs of adjacent metatarsals were thought of as ‘neighbouring pairs’ (for example, III/IV), while non-adjacent metatarsals (such as I/V) were considered to be ‘non-neighbouring pairs’. The rigidity of each pair was quantified using the σ_{RBE} [46, 244,



Figure 7.1: Markers on the bases and heads of all five metatarsals

264] calculated at each time frame of one representative walking cycle per participant (right foot only), with reference to the mid-stance pose (50% stance phase). The full method of computing σ_{RBE} can be found in Section 4.8. The results from the previous Chapter have shown that the σ_{RBE} calculated using the static trial was more variable, likely due to different standing postures. Therefore, the mid-stance pose is used here as the reference to calculate σ_{RBE} .

7.2.4 Statistics

Results from Chapter 6 indicated no significant difference in MSFM segment rigidity between genders (male and female). Therefore, the datasets from both male and female were grouped together for further analysis. With regard to the sample size, similar value of mean and standard deviation are expected from this study, and therefore a minimum number of 14 subjects are required to obtain statistical power at the

recommended 0.8 level [54]. 23 subjects were recruited for this study.

The Kolmogorov-Smirnov test and Shapiro-Wilks normality test showed that all of the data (σ_{av} of a particular metatarsal pair) is significantly different from a normal distribution. Therefore, a non-parametric Kruskal Wallis test was used to test for differences in rigidity, followed by individual Mann-Whitney U tests as post-hoc comparison tests. All statistical calculations were completed using SPSS (v23, IBM Corp., Armonk, USA), with a significance level set at $\alpha = 0.05$.

7.3 Results

The average rigid body error σ_{av} (Fig. 7.2) was significantly different between metatarsal pairs ($p < 0.01$). Of all the pairings, the most rigid was II/III, with a σ_{av} of 1.27 ± 0.39 mm; the most flexible was I/V, with a σ_{av} value of 2.24 ± 0.38 .

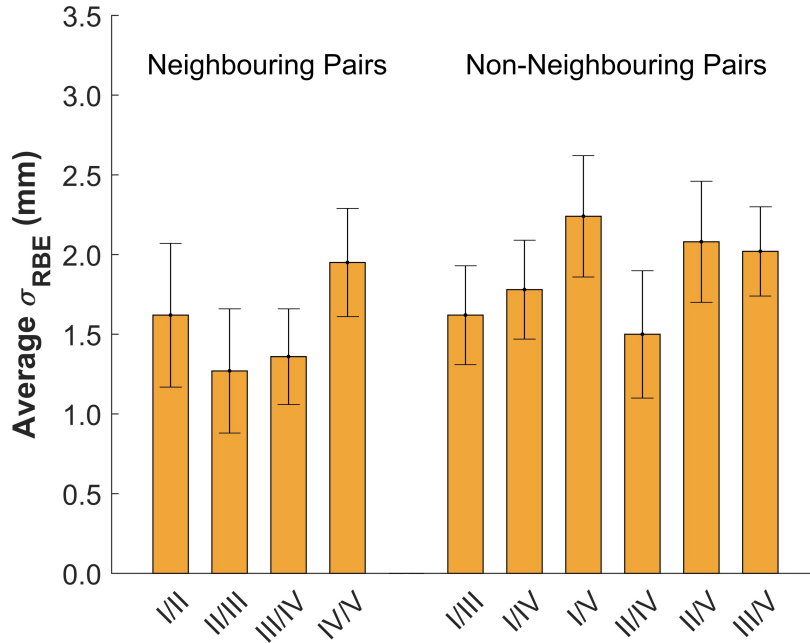


Figure 7.2: Bar chart with σ_{av} for each metatarsal (MT) pair with ± 1 standard deviation.

For neighbouring metatarsal pairs, the outer metatarsal groupings (I/II and IV/V) were significantly less rigid (that is, had higher σ_{av} values) than the inner metatarsal

pairs (II/III and III/IV), with $p < 0.01$ (Table 7.1). There was no difference in rigidity between I/II and IV/V ($p = 0.09$) or between II/III and III/IV ($p = 0.87$).

	I/II	I/III	I/IV	I/V	II/III	II/IV	II/V	III/IV	III/V	IV/V
I/II		0.462	0.199	0.000*	0.009*	0.191	0.009*	0.004*	0.007*	0.089
I/III			0.227	0.001*	0.002*	0.033*	0.016*	0.001*	0.007*	0.124
I/IV				0.006*	0.000*	0.008*	0.055	0.000*	0.048*	0.503
I/V					0.000*	0.000*	0.423	0.000*	0.223	0.024*
II/III						0.147	0.000*	0.869	0.000*	0.000*
II/IV							0.001*	0.099	0.000*	0.003*
II/V								0.000*	0.590	0.150
III/IV									0.000*	0.000*
III/V										0.286
IV/V										

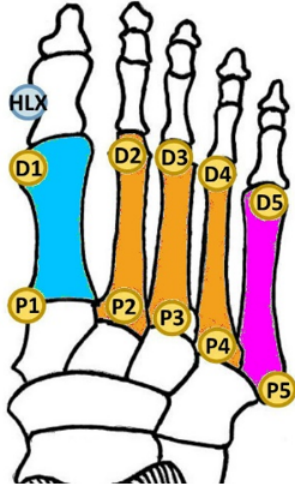
Table 7.1: Pairwise σ_{av} comparison of metatarsal pairs, showing all the p -values of all post-hoc individual Mann-Whitney U tests. Numbers marked with * indicates significant difference

Analysis of non-neighbouring pairs showed that I/III was significantly more rigid than III/V ($p < 0.01$), which indicates that the third metatarsal is more rigidly associated with the first compared to the fifth metatarsal

7.4 Discussion

This study has successfully quantified the rigidity of individual metatarsal pairs. By comparing the rigidity of the outer metatarsal pairs (I/II and IV/V) to that of the inner pairs (II/III and III/IV), the results suggest that the outer pairs show significantly more deformation during walking compared to the pairs at the centre of the forefoot (Fig. 7.2). Therefore, based on the rigidity criteria, it would be appropriate to consider the first and the fifth metatarsals as separate units in a MSFM, with the remaining (middle) metatarsals forming a third unit (Fig. 7.3 a). This recommended forefoot division is consistent with the conclusion of the study of Wolf et al. [271] in which the motion of individual foot bones was measured with intra-cortical pins and motion tracking.

a) Forefoot division 1



b) Forefoot division 2

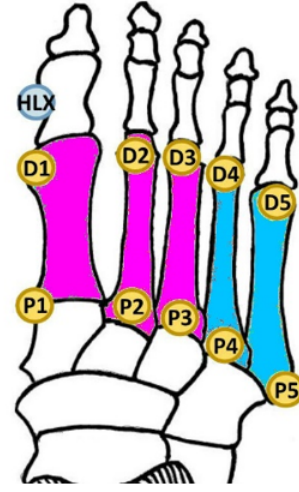


Figure 7.3: Forefoot division based on the rigidity criteria. a) 1-3-1- configuration, with the 1st and 5th metatarsal as individual segments; b) 3-2 configuration, with the 1st through 3rd as the medial and the rest as the lateral segment.

Many existing MSFMs define the forefoot as having a medial and a lateral segment [210]. However, there is no consensus on where the division should be made. Based on the results for non-neighbouring pairs (Fig. 7.2), the third metatarsal should preferentially be grouped with the first metatarsal rather than the fifth metatarsal for a two segment forefoot model (Fig. 7.3 b). This division is consistent with the anatomy of the foot, with the three medial metatarsals connecting to the cuneiform-navicular complex and the two lateral metatarsals articulating with the cuboid bone. This medio-lateral division is, however, in contrast to the results of the study by DiLiberto et al. [76], which used skin-mounted electromagnetic trackers to measure the motion of the first, third, and fifth metatarsal bones. In their control group (age-, gender-, and BMI-matched to patients with diabetes mellitus and peripheral neuropathy), they found the motion of the third metatarsal in the frontal and transverse planes to be more closely correlated to motion of the fifth metatarsal than the first. In the sagittal plane, strong couplings existed both between the third and first metatarsal

and the third and fifth metatarsal. The differing conclusions between their study and ours might be explained by the criteria used for comparison (Euler angles versus σ_{RBE}), and by the different profiles of the study groups. The DiLiberto study population was older (mean age 55.7 years) and included more males (80%) compared to the participants in the current study. Gender, BMI, and age are known to influence dynamic foot morphology [94], although there is no consensus in the literature on the precise links with specific foot measures.

This study has several limitations. It is important to note that σ_{RBE} captures both the soft-tissue movement and the flexibility of the anatomical foot segment. Furthermore, a marker-cluster could remain rigid ($\sigma_{RBE} = 0$) yet still move independently from the underlying bones. It is also acknowledged that justification for a sub-division of a MSFM should also consider the anatomical structure, practicality, repeatability of a marker placement protocol, clinical utility, and the intended application of any proposed model [23].

7.5 Summary

For the first time, σ_{RBE} has been utilised to quantify the rigidity between individual metatarsals. Using rigidity as a criterion, two recommendations have been made for appropriate forefoot subdivision (Fig. 7.3). Based on rigidity, the forefoot can be subdivided into two segments mediolaterally in a 3-2 configuration, where the first through third metatarsals are grouped as the medial forefoot, and the fourth and fifth metatarsals are grouped as the lateral forefoot. Alternatively, the forefoot can also be presented in a 1-3-1 configuration, where the three middle metatarsals (second through fourth) are grouped as a unit and the first metatarsal and fifth metatarsal are treated independently. The result of this study forms the basis of the novel forefoot model design proposed in the following Chapters 8 and 9.

Chapter 8

Novel Forefoot Modelling: An Exploratory Study

The main objective of this thesis is to develop a novel forefoot model in a more anatomically and functionally appropriate manner, that is suitable to be used in a clinical setting. In Chapter 7, post-hoc comparisons were made between the rigidity of different combinations of metatarsal pairs. The analysis suggested two ways in which the forefoot could be divided based on the rigidity criteria. The first option is to divide the forefoot into two segments, with the first through third metatarsals representing the medial segment and the fourth and fifth representing the lateral segment. The other considers the middle three metatarsals (second, third, and fourth) as a single unit and the first and fifth as additional independent segments. This chapter describes an exploratory study, designing and evaluating the differences between several variations of these two types of forefoot models. Variation in joint rotation graphs was expected in each forefoot segment defined using a different collection of forefoot markers.

8.1 Introduction

Various forefoot modelling strategies have been proposed in the past (See Sec. 3.3). Many consider the forefoot as an independent unified three-dimensional segment [42, 135, 212, 250, 280], while others subdivide the forefoot into multiple segments [65, 125, 143, 240]. There is currently no standardised approach in representing and modelling the forefoot in a MSFM. In theory, there can be unlimited ways of representing the forefoot, differing in terms of the segmentation, number of markers, axis definition, calibration method, and kinematic calculation.

Earlier studies utilising intra-cortical pins have shed light on how the forefoot deforms, and therefore how these bones should be grouped in a MSFM based on relative joint motion [189, 188, 271]. In 2008, Wolf et al. [271] suggested a grouping of the medial cuneiform with the first metatarsal, navicular with the cuboid, and the fifth metatarsal as an independent segment. However, representing the suggested groupings to the accuracy reported by the bone-pin studies is impossible using skin-attached markers due to the limited space for marker placement, soft-tissue artefact, and a need to locate anatomical landmarks on the skin in a repeatable manner. In addition, the intricate motion within and between the mid-foot and forefoot bones obtained by the markers on the intra-cortical pins is also not replicable using skin-attached markers, nor necessarily required or interpretable in clinical gait analysis. Therefore, the grouping of the bones as suggested by Wolf et al. [271] and Nester et al. [188] should be used as guidance and reference data for comparison, rather than criteria for forefoot representation in practice.

Thus far, only Rouhani et al. [219] have investigated the appropriateness of forefoot segmentation in a MSFM using skin attached markers, based on a kinematic criterion. “Joint mobility” and “segment flexibility” were computed for each potential segment. Here, “segment flexibility” is synonymous to “segment rigidity” defined

in throughout this thesis. In Rouhani’s paper, joint mobility quantified the amplitude of the motion between two assumed rigid body segments, whereas segment flexibility quantified the angular motion between markers belonging to the same segment (by subdividing the segment into two separate units). It was postulated that the joint is practically relevant only if the joint mobility is significantly larger than the segment flexibility of its constituting markers. Based on this criterion, the authors demonstrated significant mobility between the medial (first and second metatarsals) and lateral (fourth and fifth metatarsals) forefoot segments, and thus recommended a subdivision of the forefoot in the middle. However, Rouhani et al. did not consider different marker placement strategies or the combination of different metatarsals. Furthermore, the third metatarsal was not considered at all in this study.

In Chapter 7, analysis of pairwise metatarsal rigidity revealed that the third metatarsal should be paired with the first and second metatarsals in a two-segment forefoot model using skin attached markers (3-2 configuration). Moreover, this study also recommended a 1-3-1 configuration where the middle three metatarsals are treated as a unified segment while the first and fifth are tracked independently. The objective of this work was to design and evaluate marker placement protocols and the design of a models for the forefoot to be used in MSFM that fulfil criteria described above.

8.2 Methodology

8.2.1 Ethical Considerations

As described in Section 4.1, ethical approval was granted by the local Ethics Committee prior to the commencement of this study.

8.2.2 Subjects and Participants

Walking data were collected from a total of 45 healthy adult volunteers with an average age of 28.2 ± 6.8 years old, of which 25 were male and 20 were female (See Sec. 4.2 for more detail).

8.2.3 Data Collection and Processing

In this exploratory study, novel forefoot models were designed based on the marker set described in Chapter 7, using ten markers placed on the proximal bases and distal heads of all five metatarsals (Oxford forefoot model marker-set, OFFM). OFM was also collected in parallel with the OFFM marker-set to allow the measurement of forefoot motion relative to the OFM hindfoot segment. See Section 4.3 for equipment used in this study, Section 4.4.3 for the full experimental protocol on the procedure of data collection, and Section 4.5 for data processing. One of the six walking trials (right side only) was selected for analysis based on how close its OFM joint rotation patterns were to the average over the six trials for that particular subject.

8.2.4 Biomechanical Models

Novel forefoot models were designed based on the framework of the OFM. The reasons for this choice were:

1. The ability to compare the differences in motion between a subdivided forefoot and a unified forefoot segment (i.e. the OFM forefoot segment).
2. The availability of clinicians and physiotherapists who are familiar with the OFM and can provide expert opinion on the practicality and identify potential clinical utility of the proposed models.
3. The ability to incorporate the backward compatibility feature (to the OFM) into the new forefoot design.
4. OFM is a validated protocol that has been applied to many clinical and methodological studies (See Sec. 3.2.2), and the availability of a large pool of reference

data presented a valuable resource to compare the performance of the novel models. The detailed methods of computing OFM kinematics can be found in Section 4.6.2.

This study aimed to evaluate a number of forefoot models that fulfils the rigidity criteria, as recommended from Chapter 7. Two types of forefoot models were proposed, type A and type B, as defined below. For all the proposed models and their variants, the marker placement was chosen with the design considerations in mind (See Sec. 3.4). Despite the close proximity of some of the landmarks, all markers were visible to the cameras and had no tracking issues.

Type A OFFM

Type A OFFM forefoot segments were designed to capture the medial and lateral movement of the forefoot independently, with the medial segment representing the motion of the first to third metatarsals, and the lateral segment, the fourth and fifth metatarsals. Three different variants for type A segmentation were proposed. OFFM A1 tracks both the medial and lateral forefoot utilising the markers attached directly onto the bones defined by the model. The medial forefoot is tracked using the markers attached to the dorsal aspect of the first and third metatarsals (Fig. 8.1 a), and the lateral forefoot is tracked using the markers attached to the fourth and fifth metatarsals (Fig. 8.1 b). Both medial and lateral forefoot segments utilised a similar marker placement and segment definition to the OFM.

OFFM A2 adopted a similar concept to the Kinfoot model [161], where the medial and lateral segments are defined using shared markers to economise the space for marker placement (Figs. 8.1 c and d). The segment definition follows the same convention as A1, each segment being defined using four markers. In this model, the shared virtual marker between the medial and lateral forefoot segment were estimated using the mid-point of P3M and P4M for the location of P3/4M, and the mid-point of D3M and D4M for D3/4M. One obvious advantage for this marker set is that

the OFM TOE marker does not have to be removed, therefore achieving backward compatibility.

OFFM A3 was designed to be very similar to the OFFM A1 model, with the exception that D3M is replaced by TOE, thus achieving backward compatibility to the original OFM (Figs. 8.1 e and f). The lateral forefoot in A3 is defined in the same way as A1. The segment definitions for type A OFFMs are summarised in Table 8.1.

Type B OFFM

Type B models were designed to track the motion of the first and fifth metatarsals separately (excluding rotation) and the intermediate forefoot representing the second through the fourth metatarsals. These models were designed to satisfy the other type of forefoot subdivision as recommended by the rigidity criteria (Fig. 7.3). Model B1 defines the intermediate forefoot segment based on a similar strategy to the OFM forefoot and the OFFM A1 segments (Fig. 8.2 a). The medial and lateral segments are defined as the vectors of the first and fifth metatarsals respectively, with the first metatarsal defined as the vector from P1M to D1M, and the fifth metatarsal from P5M to D5M (Fig. 8.2 b). To avoid confusion, the medial and lateral forefoot segments of the type B models will be referred to as the first metatarsal and fifth metatarsal segments respectively.

OFFM B2 is a variant of the B1 model (Fig. 8.2 c). The B2 intermediate forefoot segment is defined using TOE instead of D2M to achieve backward compatibility with the OFM. OFFM B3 further economises the marker placement by using only three landmarks to define the intermediate forefoot segment (Fig 8.2 d). The definitions for the medial and lateral segments in B2 and B3 are the same as B1. The segment definitions for type B models are summarised in Table 8.2.

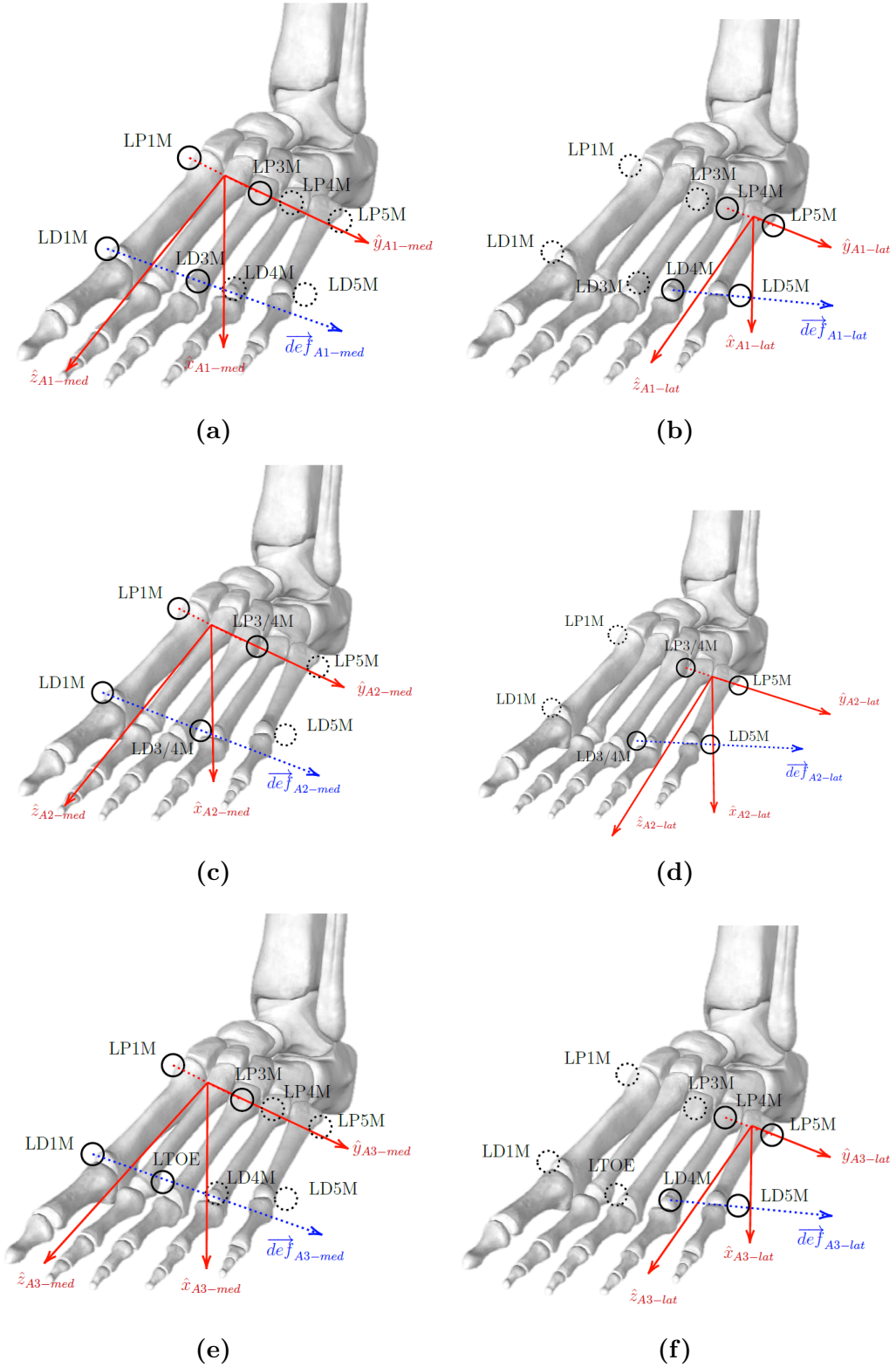


Figure 8.1: Diagram of type A OFFM forefoot segments: a) A1 medial segment; b) A1 lateral segment; c) A2 medial segment; d) A2 lateral segment; e) A3 medial segment; f) A3 lateral segment.

	A1	A2	A3
Medial forefoot anatomical frame	Origin: Mid-point of P1M and P3M; Def. line: Vector from D1M to D3M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P1M and P3M to the mid-point of D1M and D3M.	Origin: Mid-point of P1M and P3/4M; Def. line: Vector from D1M to D3/4M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P1M and P3/4M to the mid-point of D1M and D3/4M.	Origin: Mid-point of P1M and P3M; Def. line: Vector from D1M to TOE, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P1M and P3M to the mid-point of D1M and TOE.
Lateral forefoot anatomical frame	Origin: Mid-point of P4M and P5M; Def. line: Vector from D4M to D5M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P4M and P5M to the mid-point of D4M and D5M.	Origin: Mid-point of P3/4M and P5M; Def. line: Vector from D3/4M to D5M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P3/4M and P5M to the mid-point of D3/4M and D5M.	Origin: Mid-point of P4M and P5M; Def. line: Vector from D4M to D5M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P4M and P5M to the mid-point of D4M and D5M.
Extra markers	P3M, P4M, D3M, D4M.	P3/4 M (virtual), D3/4M (virtual).	P3M, P4M, P5M.

Table 8.1: Type A OFFM segment definition and list of extra markers required to the OFM marker set. A list of extra markers above the existing ones required by the OFM are also listed to compare backward compatibility between the models with the OFM. Within the listed extra markers, virtual markers reconstructed for this study are highlighted. Abbreviations: Def. Line, Defining-line; Perp, Perpendicular.

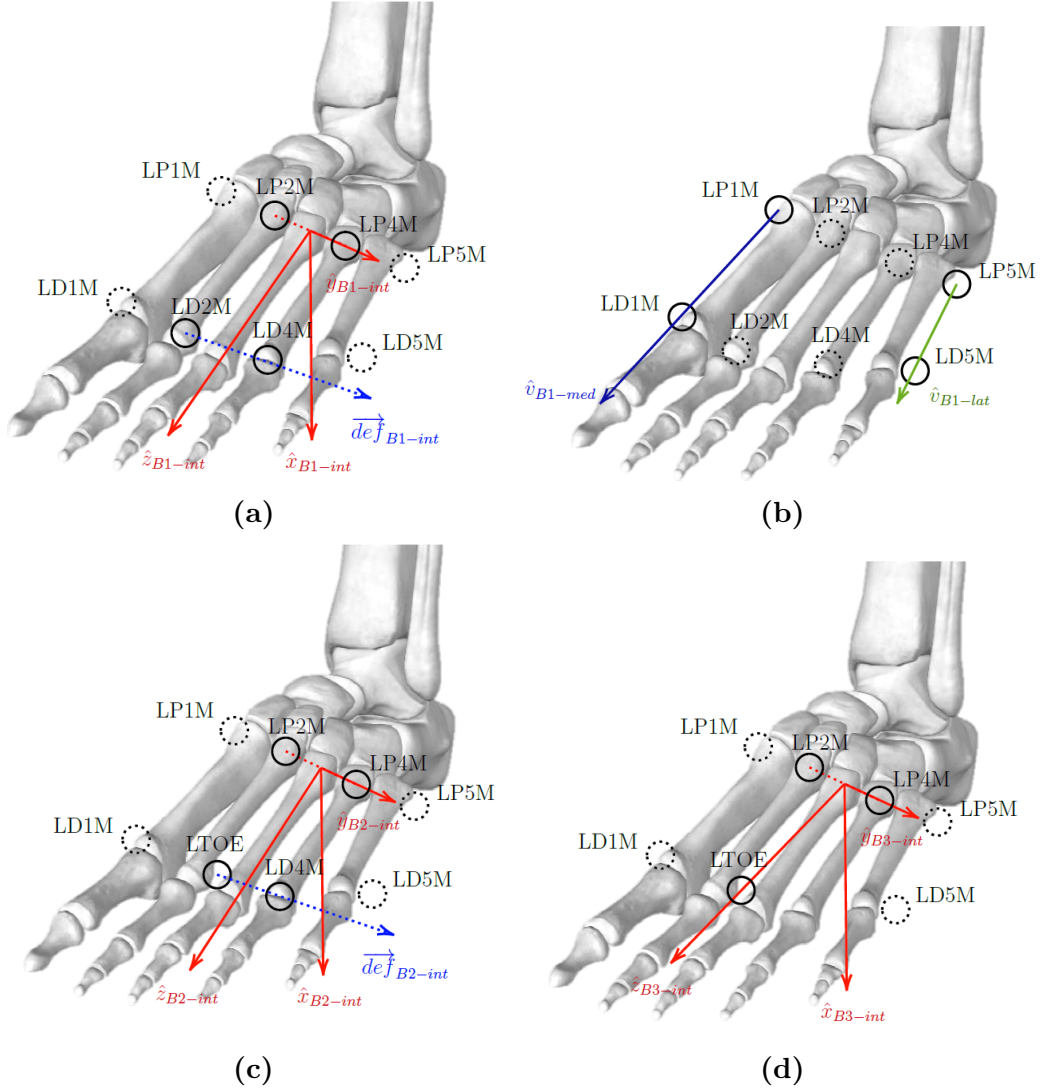


Figure 8.2: Diagram of type B OFFM forefoot segments: a) B1 intermediate segment; b) Type B medial (first metatarsal) and lateral (fifth metatarsal) segments; c) B2 intermediate segment; d) B3 intermediate segment.

	B1	B2	B3
Medial (first metatarsal) segment definition	Origin: P1M Vector: Direction from P1M to D1M.	Origin: P1M Vector: Direction from P1M to D1M.	Origin: P1M Vector: Direction from P1M to D1M.
Intermediate forefoot anatomical frame	Origin: Mid-point of P2M and P4M; Def. line: Vector from D2M to D4M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P2M and P4M to the mid-point of D2M and D4M.	Origin: Mid-point of P2M and P4M; Def. line: Vector from TOE to D4M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P2M and P4M to the mid-point of TOE and D4M.	Origin: Mid-point of P2M and P4M; Def. line: Vector from D2M to D4M, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P2M and P4M to TOE.
Lateral (fifth metatarsal) segment definition	Origin: P5M Vector: Direction from P5M to D5M.	Origin: P5M Vector: Direction from P5M to D5M.	Origin: P5M Vector: Direction from P5M to D5M.
Extra markers	P2M, P4M, D2M, D4M.	P2M, P4M, D4M	P2M, P4M.

Table 8.2: Type B OFFM segment definition and list of extra markers required to the OFM marker set. A list of extra markers above the existing ones required by the OFM are also listed to compare backward compatibility between the models with the OFM. Within the listed extra markers, virtual markers reconstructed for this study are highlighted. Abbreviations: Def. Line, Defining-line; Perp, Perpendicular.

8.2.5 Inter-Segmental Kinematics

Joint rotations were calculated according to the method outlined in Section 4.7.2. The tri-planar motion of the forefoot was measured relative to the OFM hindfoot segment, that is:

- Dorsiflexion / plantarflexion rotation about the OFM hindfoot y-axis (medio-lateral).
- Supination/ Pronation about the OFFM z-axis (longitudinal).
- Abduction/ Adduction about the OFFM x-axis (proximal-distal).

The two-dimensional planar angles of the first and fifth metatarsal bones were also calculated. To investigate the movement of the first and fifth metatarsal segments in type B OFFM, the vectors were projected onto the planes of the OFM hindfoot, and the planar angles were measured with reference to one of the hindfoot axes (See Sec. 4.6.2 for the OFM hindfoot definition) as follows:

- Dorsiflexion/ plantarflexion: Projection onto the hindfoot sagittal plane, rotation about the y-axis (medio-lateral), the angle made with the hindfoot x-axis.
- Abduction/ Adduction: Projection onto the hindfoot transverse plane, rotation about the x-axis, angle made with the hindfoot z-axis.

Lastly, the angle made between the first and fifth metatarsal vectors projected onto the transverse plane of the original OFM forefoot segment, MTI-V, was also calculated to measure the spread of the metatarsals.

8.2.6 Statistics

The average kinematic angles over one full gait cycle were calculated for all the proposed OFFM segments. To compare gait kinematics, descriptive statistics were reported for each angle, including mean, minimum, and maximum angle, and range of motion (ROM). Where relevant, an unpaired t-test with Bonferroni corrections was

conducted to compare the difference in the descriptive statistics between different segments.

One of the objectives was to evaluate the differences in kinematic waveforms between the same type of segments using different marker placement strategies, for example, whether there is a difference between the medial segment kinematic output of A1 and A2. Due to the difference in model definitions, it is not appropriate to perform hypothesis testing to identify variability between the waveforms, such as statistical parametric mapping [200] or bootstrap-based methods to evaluate differences along the curve [153]. Therefore, assessment of kinematic similarity was based on visual identification of changes in gait pattern throughout the gait cycle. To assess which model may be more practical, repeatability indexes were also calculated for the kinematic output for each model. The same procedure outlined in Section 6.2.5 was used to determine the inter-trial error (subject variability), intra-rater, and inter-rater error [234]. Two staff members, one (senior) had 16 years of experience in clinical gait analysis and another (junior) who had one year of experience, participated in this analysis.

8.3 Results

8.3.1 Inter-Segmental Kinematics

Inter-segmental joint motions were calculated for all the three-dimensional OFFM segments relative to the OFM hindfoot (Figs. 8.3 - 8.5). The kinematic patterns within the same configuration type (A-medial, A-lateral, or B-intermediate) were found to be consistent with each other. However, since the kinematic angles were not zeroed to a specific reference pose, there were expected offsets in the mean angle (Tables 8.1 and 8.2). In this analysis, variability in kinematic patterns was identified visually due to the difference in the model definitions and marker placements.

OFFM type A divides the forefoot into a medial and a lateral segment. The major differences between the medial and lateral segments in kinematic patterns were found in the transverse plane, with the most agreement in the sagittal plane. In the transverse plane, the medial forefoot progressively abducted about 6° in pre-swing, from 45 to 60 % gait cycle (GC) (Fig. 8.3), whereas the lateral forefoot first adducted for about 5° , followed by a rapid abduction of 5° (Fig. 8.4). The frontal plane motions in both the medial and lateral forefoot segments were quite similar, with forefoot pronation during early stance and a rapid supination then pronation at 60% GC (Figs. 8.3 and 8.4). However, the lateral forefoot appeared to be more supinated in the swing phase by approximately 7° than in the stance phase, whereas the medial forefoot appeared to be more equal in the two phases. Overall, the lateral forefoot had a better kinematic agreement with the original OFM in comparison to the medial forefoot, based on the differences in the transverse and frontal plane motion.

The intermediate forefoot segment represented in type B models also shared a very similar kinematic pattern with the OFM forefoot (Fig. 8.5). In comparison to type A OFFMs, the intermediate segment shared more similarities to the lateral than the medial forefoot segment. The largest distinction between the intermediate forefoot and the OFM was observed in the transverse plane, where a more exaggerated “double-bump pattern” was observed at 5% and at 53% GC.

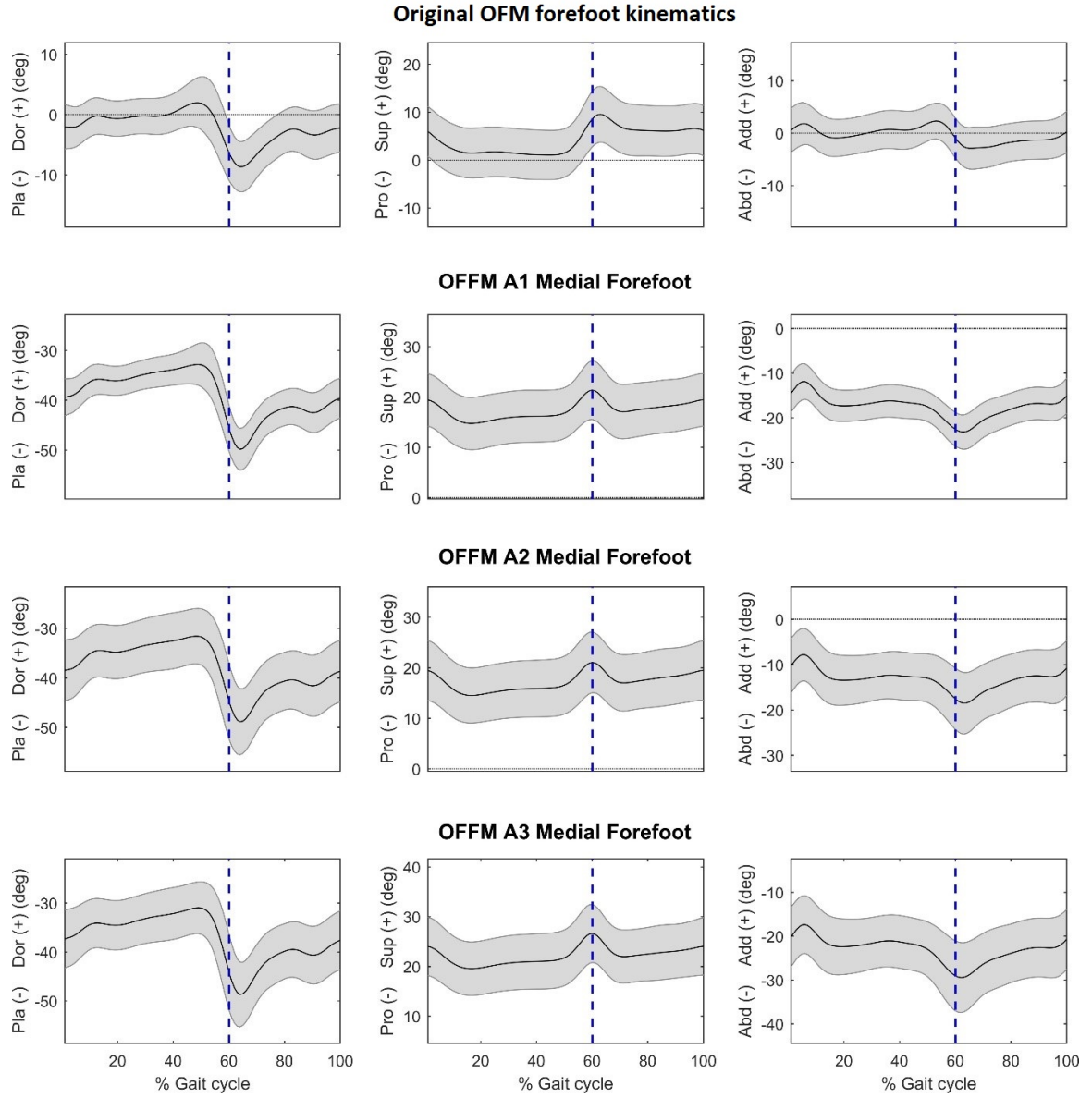


Figure 8.3: Inter-segmental kinematics of the type A medial forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right respectively. Means $\pm$ 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM A1 medial forefoot, OFFM A2 medial forefoot, and OFFM A3 medial forefoot. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. Note the differences in y-axis scaling between the columns.

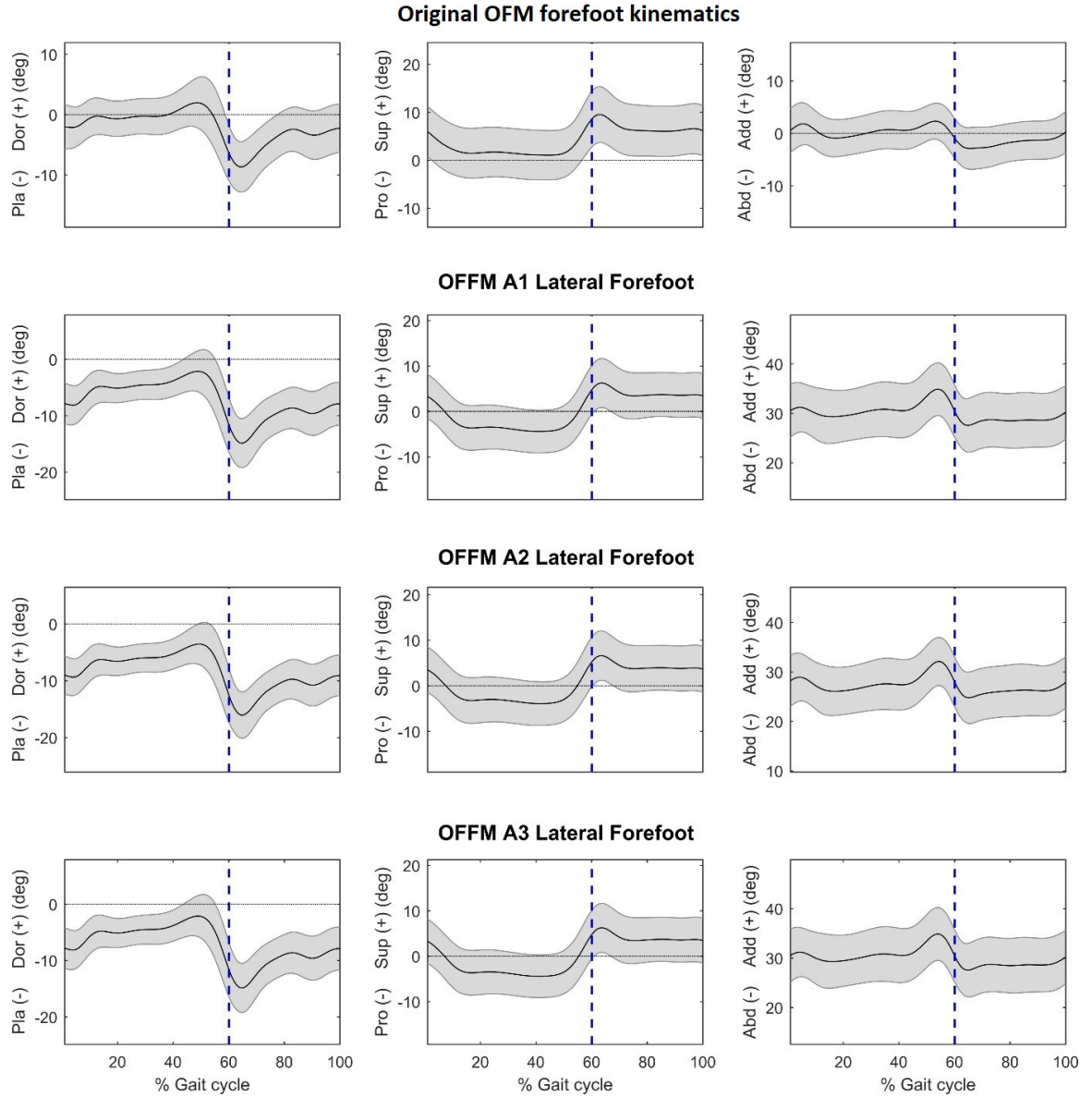


Figure 8.4: Inter-segmental kinematics of the type A lateral forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right respectively. Means $\pm$ 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM A1 lateral forefoot, OFFM A2 lateral forefoot, and OFFM A3 lateral forefoot. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. Note the differences in y-axis scaling between the columns.

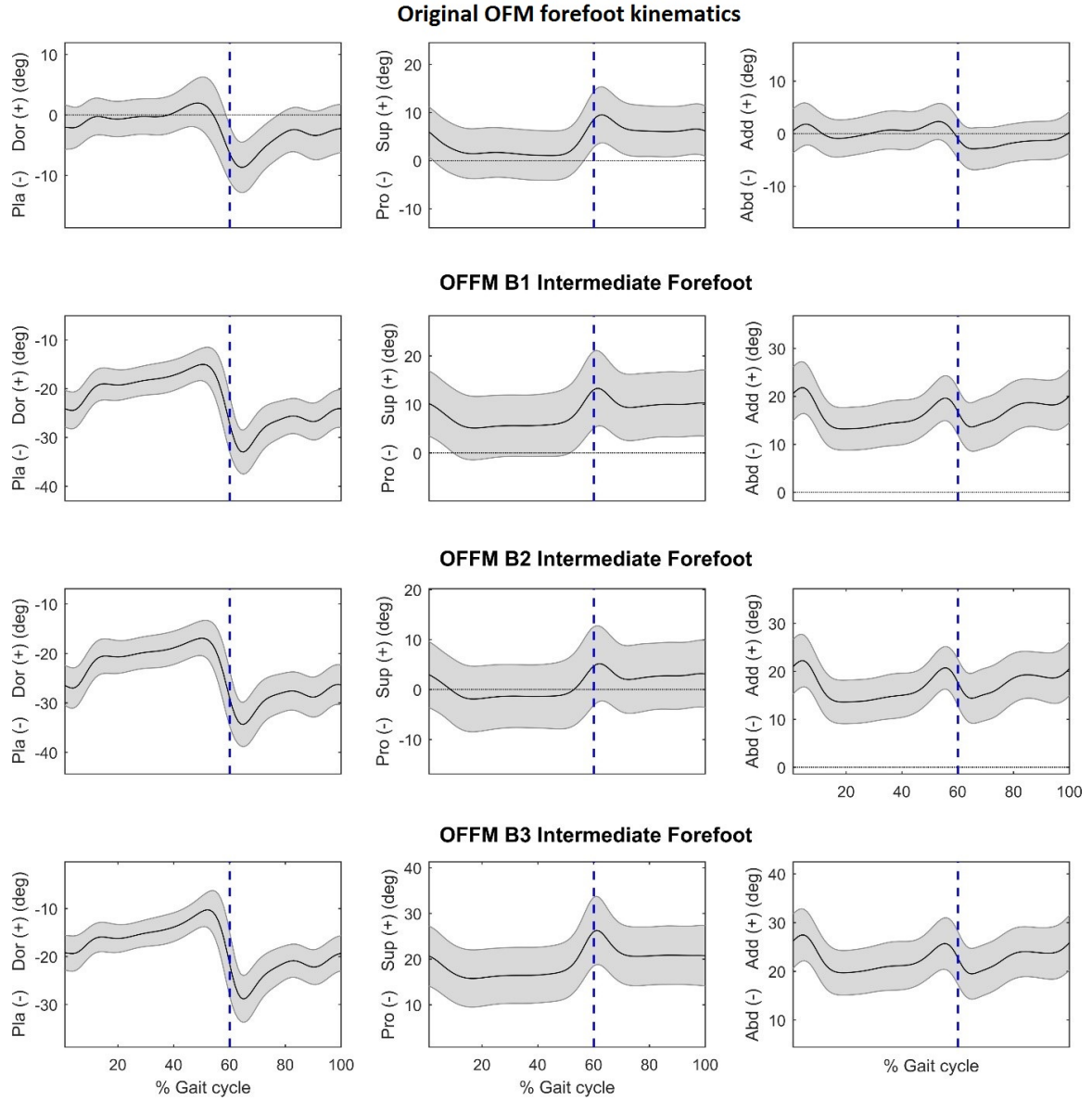


Figure 8.5: Inter-segmental kinematics of the type B intermediate forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right respectively. Means $\pm$ 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM B1 intermediate forefoot, OFFM B2 intermediate forefoot, and OFFM B3 intermediate forefoot. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. Note the differences in y-axis scaling between the columns.

8.3.2 Planar-Angles

Planar angles in the sagittal and transverse planes were calculated for the first (MTI) and fifth metatarsal (MTV) vectors of the type B OFFMs (Figs. 8.6 and 8.7). Similar kinematic patterns were found in the sagittal plane of both metatarsals (Fig. 8.6 - left column). The sagittal plane motion of the two metatarsal vectors was also similar to the OFM forefoot kinematic and other OFFM segments (left column of Figs. 8.3 - 8.5). In comparison, the first metatarsal displayed a larger motion than the fifth metatarsal in the sagittal plane (Table 8.4), with an average value of 17.1° and 11.5° respectively.

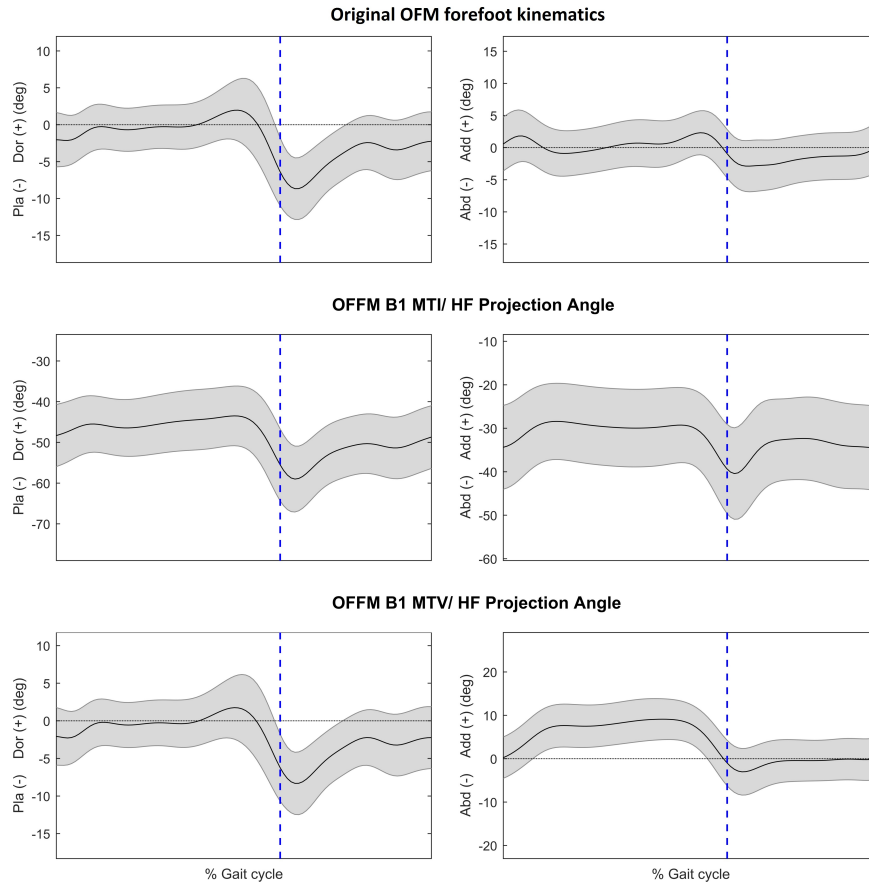


Figure 8.6: Planar angles of the type B medial and lateral forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion (left) and ab/adduction (right) motions. Means ± 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM B medial (first metatarsal) forefoot, and OFFM B lateral (fifth metatarsal) segment. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. Note the differences in y-axis scaling in each column.

The motions of the first and fifth metatarsals in the transverse plane were very similar in the stance phase, both initially adducted in loading response by 6° and 7° respectively. Both metatarsals then stayed adducted throughout the mid and terminal stance. Right before pre-swing, the metatarsals rapidly abducted. In the swing phase, the first metatarsal quickly returned to its adducted position while the fifth metatarsal stayed abducted throughout the entire period. The angle between the first and fifth metatarsal (MT I-V) averaged about 35.3° throughout the full gait cycle. MT I-V initially increased by 4° in the first 15% of the GC during loading response (Fig. 8.7). At 45% GC, the angles progressively reduced by 7° up until 70% GC.

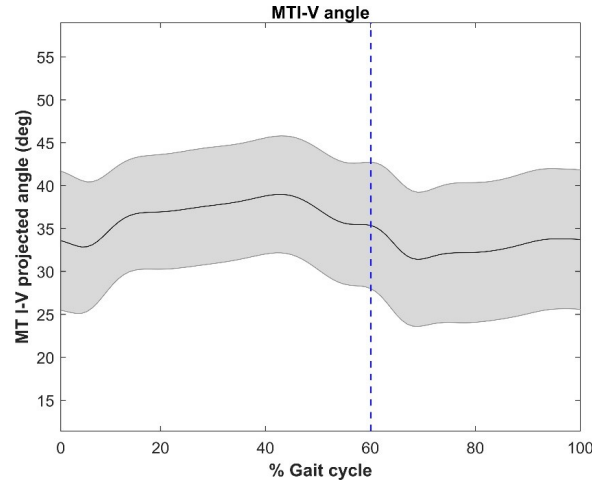


Figure 8.7: The angle between the first and fifth metatarsal segments projected onto the transverse plane of the OFM forefoot segment for the type B model.

8.3.3 Descriptive Statistics

Descriptive statistics for the kinematic outputs from OFM and OFFM segments can be found in Tables 8.3 and 8.4. In this study, the kinematic angles from all the proposed models were not zeroed. Consequently, offsets in patterns within the same model type were found in all three planes due to the differences in marker placement strategy. For type A medial segments, the largest offsets were observed in the transverse plane. This is unsurprising since the change in distal marker placement was along the mediolateral direction. On the contrary, the largest kinematic offsets amongst different variants in the type B intermediate segments were observed in the frontal plane, with a mean value of 8.4° , 1.1° , and 19.4° in B1, B2, and B3 respectively.

In the sagittal plane, the ROM of the type A medial forefoot was significantly larger than the original OFM forefoot ($p < 0.01$ for all A1, A2, and A3). Meanwhile, the ROM of the lateral forefoot was significantly smaller than the OFM forefoot ($p = 0.01$ for A1 and $p = 0.02$ for A2). The result indicates that, in a unified forefoot segment, the medial side contributed a larger portion of its overall sagittal plane motion. In B3, the intermediate forefoot segments also demonstrated a significantly larger sagittal plane motion in comparison to the OFM forefoot segment ($p < 0.01$ for all B1, B2, and B3). On average, the motion of the OFFM segments was larger than the OFM motion in all three planes as indicated by the increased ROM (Tables 8.3 and 8.4). The only exception was the reduced frontal plane motion in the type A segments. This implies that treating the forefoot as an unified segment effectively averages out the ranges for the medial and lateral segment, further demonstrating the need to subdivide the forefoot.

			Descriptive statistics (deg)			Range of motion (deg)
			Min (S.D.)	Max (S.D.)	Mean (S.D.)	Mean (S.D.)
Oxford foot model		Dorsiflexion	-9.1 (4.3)	2.8 (3.4)	-2.0 (3.3)	11.9 (4.4)
		Supination	0.4 (5.1)	10.3 (5.5)	4.4 (5.0)	9.8 (2.7)
		Adduction	-3.9 (3.7)	3.6 (3.6)	-0.3 (3.3)	7.5 (2.3)
OFFM A1	Medial segment	Dorsiflexion	-57.0 (6.4)	-32.4 (5.4)	-39.0 (5.4)	18.3 (4.2)
		Supination	14.1 (5.3)	22.5 (5.8)	17.6 (5.2)	8.4 (2.2)
		Adduction	-23.9 (7.2)	-11.3 (6.0)	-17.5 (6.0)	12.7 (4.1)
	Lateral segment	Dorsiflexion	-15.3 (4.4)	-1.4 (2.9)	-7.2 (3.1)	13.9 (3.7)
		Supination	-4.9 (4.7)	7.0 (5.2)	0.4 (4.7)	11.9 (2.6)
		Adduction	256 (5.3)	35.9 (5.2)	30.0 (4.9)	10.3 (3.3)
OFFM A2	Medial segment	Dorsiflexion	-49.6 (6.6)	-31.1 (5.4)	-37.9 (5.6)	18.5 (4.0)
		Supination	14.0 (5.5)	22.3 (5.9)	17.4 (5.3)	8.3 (2.2)
		Adduction	-19.2 (6.7)	-7.2 (5.3)	-13.2 (5.5)	12.0 (3.9)
	Lateral segment	Dorsiflexion	-16.5 (4.2)	-2.9 (2.9)	-8.5 (2.9)	13.7 (3.6)
		Supination	-4.4 (4.7)	7.3 (5.3)	0.8 (4.7)	11.7 (2.6)
		Adduction	22.9 (5.0)	33.1 (4.7)	27.2 (4.5)	10.1 (3.3)
OFFM A3	Medial segment	Dorsiflexion	-49.5 (6.5)	-30.6 (5.2)	-37.3 (5.3)	19.0 (4.4)
		Supination	19.0 (5.4)	27.7 (5.8)	22.4 (5.2)	8.7 (2.3)
		Adduction	-30.3 (8.0)	-16.6 (6.3)	-23.0 (6.3)	13.7 (4.7)
	Lateral segment	See OFFM A1 lateral segment				

Table 8.3: Comparison of descriptive statistics and range of motion for the fore-foot/hindfoot kinematics for the original OFM and type A OFFM medial and lateral segments. The A3 lateral segment uses the same segment definition as A1 lateral segment, hence the same result.

		Descriptive statistics (deg)			Range of motion (deg)	
		Min (S.D.)	Max (S.D.)	Mean (S.D.)	Mean (S.D.)	
Oxford foot model		Dorsiflexion	-9.1 (4.3)	2.8 (3.4)	-2.0 (3.3)	11.9 (4.4)
		Supination	0.4 (5.1)	10.3 (5.5)	4.4 (5.0)	9.8 (2.7)
		Adduction	-3.9 (3.7)	3.6 (3.6)	-0.3 (3.3)	7.5 (2.3)
OFFM B1	MT I-V	MT I-V angle	30.3 (7.3)	40.4 (7.1)	35.3 (7.1)	10.1 (2.8)
	1st. MT motion	Dorsiflexion	42.7 (7.2)	59.8 (8.1)	48.7 (7.2)	17.1 (4.5)
		Adduction	-41.7 (10.6)	-27.5 (8.7)	-32.4 (9.0)	14.2 (4.9)
	5th. MT motion	Dorsiflexion	-2.7 (3.5)	8.8 (4.2)	2.0 (3.3)	11.5 (3.8)
		Adduction	-3.5 (5.3)	9.6 (4.6)	3.4 (4.7)	13.1 (2.4)
	Interm. segment	Dorsiflexion	-33.7 (4.5)	-14.5 (3.2)	-22.4 (3.2)	19.3 (3.5)
		Supination	4.2 (6.6)	14.5 (7.5)	8.4 (6.5)	10.3 (3.6)
		Adduction	11.1 (4.6)	23.4 (5.1)	16.5 (4.6)	12.3 (3.6)
	OFFM B2	Interm. segment	Dorsiflexion	-35.1 (4.5)	-16.3 (3.3)	-24.2 (3.4)
Supination			-2.9 (6.6)	6.5 (7.3)	1.1 (6.4)	9.4 (3.4)
Adduction			11.3 (4.7)	24.1 (5.1)	17.0 (4.2)	12.7 (3.9)
OFFM B3	Interm. segment	Dorsiflexion	-29.7 (4.8)	-9.6 (3.6)	-18.3 (3.1)	20.0 (4.3)
		Supination	15.1 (6.1)	27.2 (7.2)	19.4 (6.1)	12.1 (3.8)
		Adduction	17.3 (4.9)	29.0 (5.4)	22.6 (4.6)	11.7 (3.5)

Table 8.4: Comparison of descriptive statistics and range of motion for the fore-foot/hindfoot kinematics for the original OFM and type B OFFM medial, lateral, and intermediate segments. The kinematics of the first metatarsal and fifth metatarsal segments are measured as projected planar angles. Abbreviations: Interm, Intermediate. MT, Metatarsal

8.3.4 Repeatability Studies

Repeatability of the OFM and OFFM forefoot kinematics was assessed by average inter-trial, inter-rater, and intra-rater errors (Tables 8.5 and 8.6). This study investigates different marker placement strategies, therefore the inter-rater error was used as the primary index for comparison. The OFM demonstrated a good repeatability in all three planes, with an average inter-rater error of 3.0° , 1.5° , and 3.3° in the sagittal, frontal, and transverse plane respectively. Overall, the kinematics of the type A medial forefoot segments were more repeatable than the OFM in all three planes. Type A lateral forefoot segments were also found to have smaller errors than the OFM in the sagittal plane. The highest variability was reported in the transverse plane of the lateral segment, with inter-rater value of 4.5° and 4.3° for A1 and A2 OFFM respectively.

For type B intermediate forefoot segments, the sagittal plane motion also produced less error compared to the OFM (3.0°), with a value of 1.9° , 1.8° , and 1.8° for B1, B2, and B3 respectively. However, the error was slightly higher in the transverse plane, with a value of 2.5° , 2.0° , and 3.2° for B1, B2, and B3 respectively, compared to a value of 1.5° for the OFM. On average, the planar-angles were less repeatable compared to the inter-segmental kinematics of the three-dimensional segments. This was expected since only two markers were used to define the direction of the vector, and therefore more sensitive to marker placement error.

Overall, changes in the marker placement within the same OFFM type produced minimal changes to the repeatability. Within the same types of OFFM segment, the differences between the repeatability among the same variants were generally less than 1° in every plane. The only cases that exceed 1° difference was the inter-rater error reported between the frontal plane motion of the B2 and B3 intermediate segment, with a value of 2.0° and 3.2° respectively.

Both raters demonstrated a similar level of intra-rater repeatability. However, the junior rater consistently demonstrated a higher error for the motion of the B1 intermediate forefoot segment. This might reflect the importance of the experience in the identification of the proximal base of the metatarsals. Palpation of the proximal bases of the metatarsals located in the middle is known to be more difficult than the distal heads.

			Inter-trial error (deg)			Intra- rater error (deg)		Inter- rater error (deg)	<u>Inter-rater</u> <u>Inter-trial</u>
			Jun.	Sen.	Av.	Jun.	Sen.	Av.	
Oxford foot model		Dor.	0.6	0.5	0.6	2.9	3.0	3.0	5.4
		Sup.	0.5	0.5	0.5	1.5	0.5	1.5	2.9
		Add.	1.0	0.6	0.8	3.4	3.0	3.3	4.1
OFFM A1	Medial segment	Dor.	0.5	0.6	0.6	2.1	3.8	3.1	5.3
		Sup.	0.8	0.5	0.5	0.7	1.4	1.2	2.4
		Add.	0.8	0.9	0.8	1.1	1.1	1.1	1.2
	Lateral segment	Dor.	0.7	0.5	0.6	1.7	2.4	2.2	3.8
		Sup.	0.5	0.5	0.5	3.1	2.5	3.1	5.7
		Add.	1.3	0.8	1.0	5.0	3.8	4.5	4.5
OFFM A2	Medial segment	Dor.	0.5	0.6	0.6	1.8	3.3	2.6	4.7
		Sup.	0.5	0.4	0.5	0.7	0.9	0.9	1.8
		Add.	0.8	0.7	0.8	1.7	1.0	1.4	1.8
	Lateral segment	Dor.	0.6	0.4	0.5	2.2	2.1	2.2	4.2
		Sup.	0.5	0.5	0.5	3.2	2.3	2.9	5.7
		Add.	1.2	0.7	1.0	4.7	3.6	4.3	4.4
OFFM A3	Medial segment	Dor.	0.6	0.7	0.6	1.8	3.6	2.9	4.6
		Sup.	0.6	0.4	0.5	0.6	1.1	0.9	1.9
		Add.	0.8	0.8	0.8	0.9	1.0	1.0	1.3
	Lateral segment	See OFFM A1 lateral segment							

Table 8.5: Comparison of repeatability matrices for the forefoot/hindfoot kinematics for the original OFM and type A OFFM medial and lateral segments. Average inter-trial, intra-rater, inter-rater error and the ratio of inter-rater to inter-trial error were calculated for all kinematic outputs. The A3 lateral segment uses the same segment definition as A1 lateral segment, hence the same result. Abbreviations: Dor, Dorsiflexion; Sup, Supination; Add, Adduction; Jun, Junior; Sen, Senior; Av, Average.

			Inter-trial error (deg)			Intra- rater error (deg)		Inter- rater error (deg)	<u>Inter-rater</u> <u>Inter-trial</u>
			Jun.	Sen.	Av.	Jun.	Sen.	Av.	
Oxford foot model		Dor.	0.6	0.5	0.6	2.9	3.0	3.0	5.4
		Sup.	0.5	0.5	0.5	1.5	0.5	1.5	2.9
		Add.	1.0	0.6	0.8	3.4	3.0	3.3	4.1
OFFM B1	MT I-V	Angle	0.9	0.7	0.8	5.1	5.2	5.2	6.8
	1st. MT motion	Dor.	0.6	0.7	0.7	5.5	6.5	6.1	9.2
		Add.	0.7	0.5	0.6	4.2	5.4	4.9	7.9
	5th. MT motion	Dor.	0.7	0.5	0.6	3.0	3.4	3.1	5.3
		Add.	0.5	0.6	0.5	4.2	3.0	3.6	7.2
	Interm. segment	Dor.	0.5	0.6	0.5	2.0	0.7	1.9	3.5
		Sup.	0.7	0.5	0.6	3.1	1.9	2.5	4.3
		Add.	1.2	1.2	1.2	3.2	2.1	2.7	2.3
	OFFM B2	Interm. segment	Dor.	0.5	0.5	0.5	1.8	0.6	1.8
Sup.			0.7	0.5	0.6	2.4	1.5	2.0	3.3
Add.			1.2	1.2	1.2	3.2	2.3	2.8	2.4
OFFM B3	Interm. segment	Dor.	0.6	0.6	0.6	2.0	0.8	1.8	3.0
		Sup.	0.6	0.5	0.6	4.2	1.4	3.2	5.6
		Add.	1.0	0.7	0.8	4.2	2.0	3.3	4.0

Table 8.6: Comparison of repeatability matrices for the forefoot/hindfoot kinematics for the original OFM and type B OFFM first and fifth metatarsal segments. Average inter-trial, intra-rater, inter-rater error and the ratio of inter-rater to inter-trial error were calculated for all kinematic outputs. Abbreviations: Dor, Dorsiflexion; Sup, Supination; Add, Adduction; Jun, Junior; Sen, Senior; Av, Average. MT, Metatarsal

8.4 Discussion

Single-segment forefoot models have been criticised for oversimplification [65, 125, 161, 187]. However, to date, there are few studies looking at how the forefoot should be divided, in particular, how to represent the groupings of the metatarsals in a MSFM in the most appropriate and optimal manner. This exploratory study is the first to compare different marker placement strategies to represent the MSFM, based on the rigidity criteria from Chapter 7.

Medio-lateral division of the forefoot at the middle is understandably the most popular and a logical way of representing the forefoot as two independent segments (See Sec. 3.3). In this study, both the medial and lateral forefoot segments were tracked using several different marker placement strategies. The largest motion occurred in the sagittal plane, with the medial forefoot demonstrating a significantly larger ROM than the lateral forefoot or a unified forefoot. This indicates that in a unified forefoot, the movement of the medial forefoot contributed towards a larger portion of the overall forefoot motion. This finding concurs with De Mits et al. [65], where a sagittal plane ROM of 12.4° and 10.9° were reported for the medial and lateral forefoot of the Ghent foot model (GFM).

The majority of the differences between the medial and lateral segments were observed in the coronal and transverse planes. At pre-swing, around 50 – 60% GC, the foot lowers to the ground with increased pressure underneath the metatarsal heads [82]. During this period, the lateral forefoot adducted by 5° while the medial forefoot abducted by 6° , suggesting a tightening of the forefoot that reduces the spread of the first and fifth metatarsals, as indicated by the decreased MT I-V angle towards terminal stance. These results did not agree with the kinematic output of the Salford foot model (SFM) [187] or the GFM, where both models reported adduction of the medial and lateral forefoot towards the end of stance phase. The contradictions can

be explained by the difference in marker placement, segment definition, and that the forefoot motion is measured with reference to the midfoot instead of the hindfoot for both SFM and GFM.

Type B OFFM segments were designed to highlight the movement of the first and fifth metatarsals independently. As expected, the sagittal motion of both vectors displayed a characteristic pattern, similar to the result obtained from the forefoot/hindfoot motion of the OFM and OFFM. Interestingly, transverse plane motion matched surprisingly well with the foot kinematics reported in an in-vivo cadaver study by Nester et al. [189], demonstrating a good anatomical motion representation, and therefore warrants further investigation.

There is currently no MSFM with an independent segment that tracks only the mid-section of the forefoot. This study was a unique opportunity to explore this alternative representation. The intermediate segment presented a very similar kinematic pattern in all three planes with the OFM as well as the type A OFFM lateral forefoot segments. Based on this observation, it is unlikely that tracking of the intermediate forefoot segment will provide any further useful clinical data.

One of the key objectives of this Chapter was to evaluate the marker placement strategies in the tracking of different forefoot segments. First of all, a clinically applicable foot model requires that the technical markers be visible to the cameras at all times. In this study, a set of ten markers were placed on all five proximal bases and distal heads of the metatarsals in order to test different marker placement strategies at the same time. Despite the close proximity of the markers, all of them could be tracked independently without any issue related to occlusion or marker merging. However, only adults over the age of 18 were included in this study, and therefore all the markers were well-spaced (Fig. 7.1). It is acknowledged that a potential problem may emerge if the same marker set were to be applied to adults or children with smaller feet. Nevertheless, all the anatomical landmarks chosen here were easily

palpable (with distal heads being easier than the proximal bases) with a good level of repeatability, as indicated by the similar level of intra-rater error between the junior and the senior rater.

Since the kinematic pattern and the repeatability between the forefoot models of the same type were practically indistinguishable, the optimal choice of marker placement design should be based on simplicity, practicality, and backward compatibility. For the type A OFFM medial segment, A2 is the least complex and therefore preferred over the slightly more complicated A1 model. The fact that using the P3M and TOE in A3 also produced the same kinematic pattern as A2 means that the TOE can be retained to achieve backward compatibility with the OFM model. Based on the same logic, B2 is preferred out of the three type B models due to the reduced number of markers.

8.5 Summary

Overall, this study provided a comprehensive evaluation of different marker placement strategies to track different subdivided forefoot segments as recommended by the rigidity criteria laid out in Chapter 6. The models proposed here produced kinematic outputs consistent with our understanding of foot biomechanics, along with additional information that cannot be observed in a unified forefoot segment. Dividing the forefoot medio-laterally in a 3-2 configuration has revealed the tightening of the forefoot in the distal end towards the end of stance phase, thus allowing more detailed characterisation of in-vivo forefoot biomechanics. The independent motion of the first and fifth metatarsal segments also produced unique patterns that were not observed previously from an existing MSFM, which may be of clinical use. The following Chapter continues to explore different forefoot marker placement strategies that captured the information from both the type A and B models.

Chapter 9

Novel Forefoot Modelling: Analysis of Hybrid Models

The previous Chapter explored different approaches to representing the forefoot as suggested by the rigidity criteria from Chapter 7. Two types of forefoot models (type A and type B, each with three variants) were proposed and their respective kinematic characteristics, repeatability, marker placement strategy and practicality were visually analysed and discussed. This Chapter extends the study by proposing three hybrid models that incorporate the features of type A and B Oxford Forefoot Models (OFFMs). The same kinematic analysis was performed, along with the computation of descriptive statistics that characterises the inter-segmental motion, and a repeatability analysis to assess subject and marker-placement variability. A final model was chosen to be piloted on individual clinical case studies to explore the clinical feasibility, practicality, and potential contribution to the identification of abnormal gait.

9.1 Introduction

In the previous Chapter, visual analysis of the kinematic patterns revealed no discernible differences between the variants of either type A and B OFFMs. Therefore, preference were given to models that are least complex as well as being backward compatible with the Oxford Foot Model (OFM). Model A2 was preferred out of the three type-A model variants, since the segment definition only requires two additional markers to the original OFM marker-set. In type A2 OFFM, both the medial and lateral forefoot segments are defined using two shared landmarks, namely D3/4M and P3/4M, placed in between the proximal base and distal head of the third and fourth metatarsal. In order to economise the number of markers placed on the subject's body, landmarks are often shared amongst the definition of neighbouring segments in a multi-segment foot model (MSFM). In the Rizzoli Foot Model (RFM) [143], a marker placed on the proximal base of the second metatarsal is used in defining the forward axis of the forefoot and mid-foot segment. Kinfoot [161] uses the markers placed on the proximal base and distal head of the metatarsals in the definition of both the medial and lateral forefoot segments. In Jenkyn and Nicol's foot model [125], the landmarks located on the base of the first and fifth metatarsals are used to define the segment coordinate system of both the medial and lateral forefoot. See Tables 3.1 and 3.2 for more information about anatomical frames of the aforementioned models. The results from the previous Chapter also supported the use of shared landmarks, as differences between the kinematic output of the type A variants (Fig. 8.1) were found to be minimal.

This Chapter is an extension of the work of the previous Chapter. Both type A and B models enabled further characterisation of forefoot biomechanics in greater detail, showing features that were previously not observed in a unified forefoot. This aim of this study was to propose and explore the clinical utility of a novel forefoot model.

To achieve this, three hybrid models were proposed, all capable of simultaneously tracking all the segments represented in both type A and B OFFMs, utilising sharing markers for the definition of individual forefoot segments. A combination of kinematic analysis, repeatability studies, and design considerations outlined in Section 3.4 were used to recommend a suitable model to be put forward to the next stage.

9.2 Methodology

9.2.1 Ethical Considerations

This study used the same data set as the previous Chapter, with ethical approval obtained from the local Ethics Committee as described in Section 4.1.

9.2.2 Subjects and Participants

Walking data were collected from a total of 45 healthy adult volunteers with an average age of 28.2 ± 6.8 years old, of which 25 were male and 20 were female (See Sec. 4.2 for more information).

9.2.3 Data Collection and Processing

In this Chapter, type C hybrid models were developed based on the same marker set used in Chapters 7 and 8. For each subject, ten markers were attached to the proximal base and distal heads of all five metatarsals. As before, both the OFM and the OFFM marker-sets were collected simultaneously to allow for the measurement of forefoot motion relative to the OFM hindfoot. See Section 4.3 for equipment used in this study, Section 4.4.3 for the full experimental protocol on the method of data collection, and Section 4.5 for data processing. One of the six walking trials (right foot only) was selected for analysis based on how close its OFM joint angle graphs

were to the average for that particular subject.

9.2.4 Biomechanical Models

In the previous Chapter, unique kinematic patterns that were previously not seen in a unified forefoot were demonstrated in both type A and type B models (See Sec. 8.2.4). where the forefoot was subdivided into multiple segments. Therefore, as a means of exploring the utility of these segments in clinical case studies, type C hybrid models were designed to incorporate the features of both type A and type B models. Marker placement was defined in order to track the motion of the medial forefoot, lateral forefoot, and the intermediate forefoot simultaneously.

The C1 model requires only two additional markers to the original OFM, that is, P2/3M (mid-point between P2M and P3M) and P4M (Fig. 9.1). Its medial forefoot segment is defined based on the concept of A3, where the TOE marker is retained for backward compatibility. Both the lateral and the intermediate segment for the C1 model were designed based on the concept of the B3 intermediate segment (Fig. 8.2). Using only three markers, the kinematic patterns of B3 intermediate forefoot were very similar to that of B1 and B2 (defined using four markers) without compromising repeatability. Therefore, the lateral and intermediate segments for C1 were constructed using only three markers each. Here, the individual motion of the first and fifth metatarsals were also tracked as before.

The C2 model was designed based on the same concept as C1, however, the P4M marker was replaced with D2M (Fig. 9.2). This meant that the y-axis definition for the lateral and intermediate forefoot segment depended on positions of the distal metatarsal heads. It is generally recognised that the distal head of the metatarsal are more easily palpable in comparison to the proximal bases [38]. Therefore, this slight adjustment in marker positioning could potentially improve repeatability. Model C2b is a variant of C2, with the exact same marker placement and segment definition.

However, the D1M marker in C2b is virtually reconstructed using its position from the static trial, based on the geometrical relations with P1M, P2/3M and TOE. The use of a physical D1M marker is not always possible in the case of foot deformity, such as equinovalgus. Therefore, in order to maximise the potential clinical application of this model, C2b was proposed as an alternative design. The segment definition for type C models are summarised in Table 9.1 below.

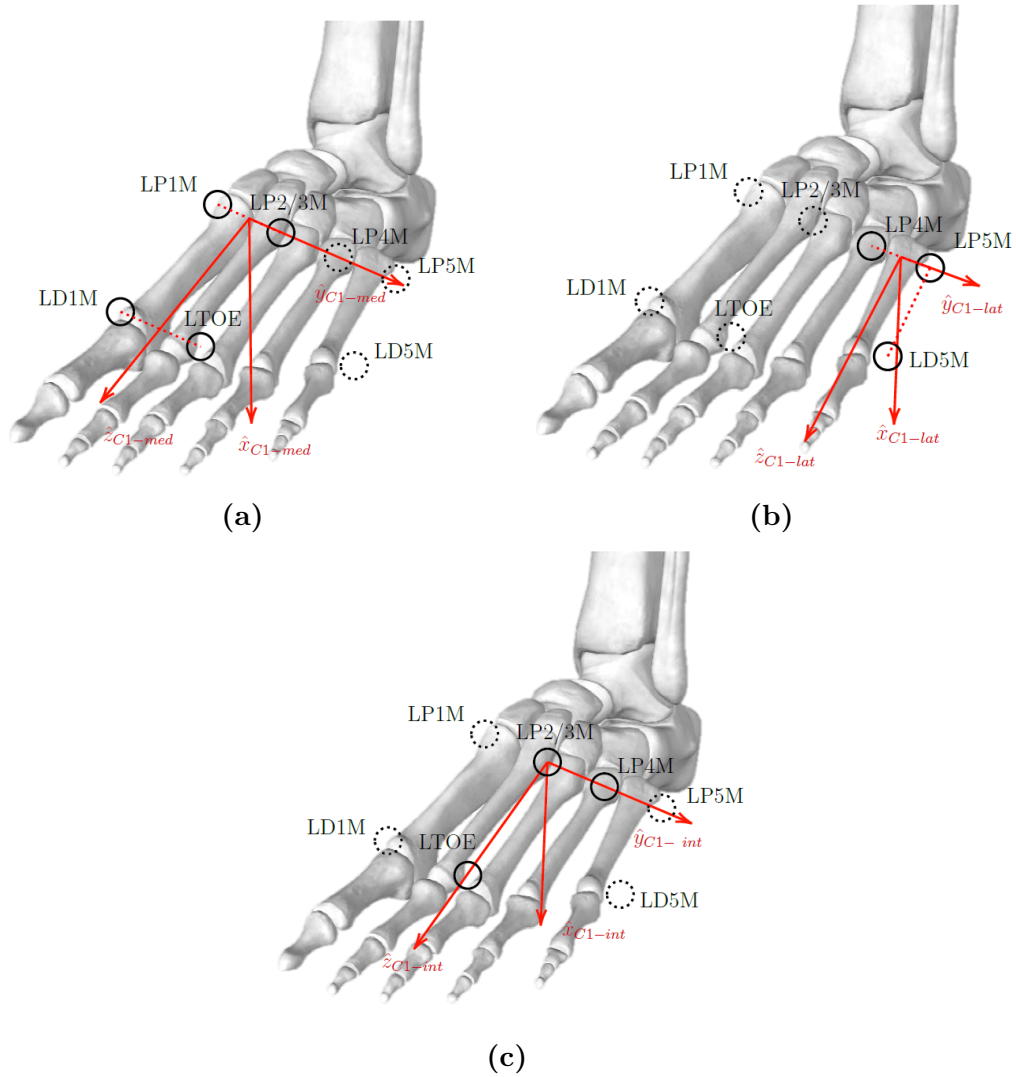


Figure 9.1: Diagram of C1 OFFM forefoot segments: a) medial segment; b) lateral segment; and c) intermediate segment.

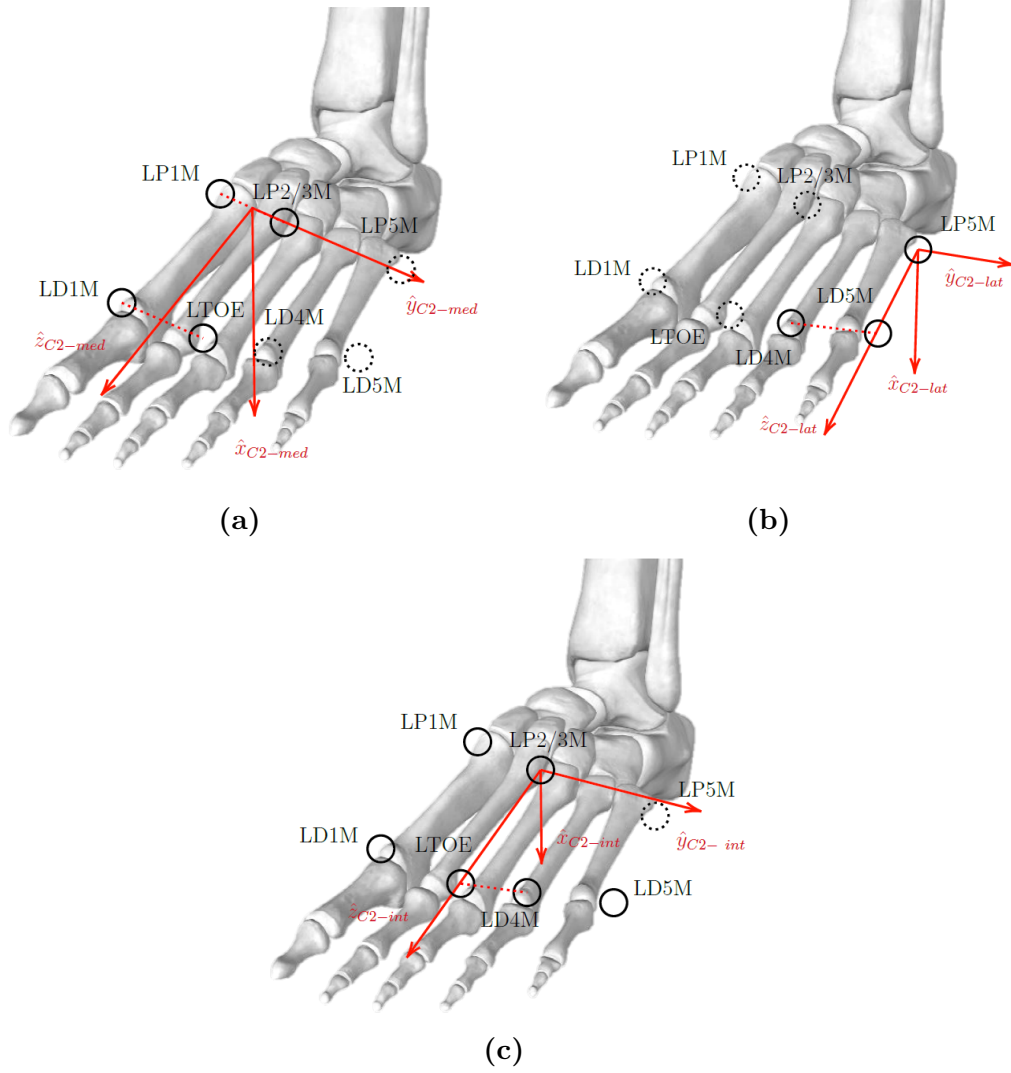


Figure 9.2: Diagram of C2 OFFM forefoot segments: a) medial segment; b) lateral segment; and c) intermediate segment.

	C1	C2	C2b
Medial forefoot anatomical frame	Origin: Mid-point of P1M and P2/3M; Def. Line: Vector from D1M to TOE, negated for the right foot; x-axis: Perp. to the def. line and the z-axis; y-axis: Perp. to the other two axes; z-axis: Vector from mid-point of P1M and P2/3M to the mid-point of D1M and TOE.	Same as C1	Same as C2
Lateral forefoot anatomical frame	Origin: Mid-point of P4M and P5M; x-axis: Perp. to the z-axis and the vector from P5M to P5M, negated for the right foot; y-axis: Perp. to the other two axes; z-axis: Vector from P5M and D5M.	Origin: P5M; x-axis: Perp. to the z-axis and the vector from D4M to D5M, negated for the right foot; y-axis: Perp. to the other two axes; z-axis: Vector from P5M and D5M.	
Intermediate forefoot anatomical frame	Origin: P2/3M; x-axis: Perp. to the z-axis and the vector from P2/3M to P4M, negated for the right foot; y-axis: Perp. to the other two axes; z-axis: Vector from P2/3M and TOE.	Origin: P2/3M; x-axis: Perp. to the z-axis and the vector from TOE to D4M, negated for the right foot; y-axis: Perp. to the other two axes; z-axis: Vector from P2/3M and TOE.	
Extra markers	P2/3M; P4M.	P2/3M; D4M.	D1M (virtual); P2/3M; D4M.

Table 9.1: Type C OFFM segment definition and list of extra markers required to the OFM marker set. A list of extra markers above the existing ones required by the OFM is also listed to compare backward compatibility between the models with the OFM. Within the listed extra markers, virtual markers reconstructed for this study are highlighted. Abbreviations: Def. line, Defining-line; Perp, Perpendicular.

9.2.5 Inter-Segmental Kinematics

Joint kinematics were calculated using the same procedure outlined in the methodology Section 4.7.2, with the forefoot motion measured relative to the OFM hindfoot segment. In summary, the tri-planar motion [101] of the three-dimensional segments was computed as:

- Dorsiflexion / plantarflexion rotation about the hindfoot y-axis (medio-lateral).
- Supination/ Pronation about the forefoot z-axis (longitudinal).
- Abduction/ Adduction about the floating axis perpendicular to the previous two axes..

The instantaneous planar angles of the first and fifth metatarsals were measured with reference to one of the hindfoot axes as:

- Dorsiflexion/ plantarflexion: Projection onto the hindfoot sagittal plane, rotation about the y-axis (medio-lateral), the angle made with the hindfoot x-axis.
- Abduction/ Adduction: Projection onto the hindfoot transverse plane, rotation about the x-axis, angle made with the hindfoot z-axis.

Finally, the angle made between the first and fifth metatarsal vectors projected onto the transverse plane of the OFM forefoot segment, MTI-V, was calculated to measure the spread of the metatarsals. In this Chapter, the planar angle for the fifth metatarsals is not reported since the information is already available in Chapter 8.

9.2.6 Statistics

Average kinematic angles were calculated as before for all the proposed segments, normalised to 100% of the gait cycle. Descriptive statistics were reported for each kinematic output, including the mean, minimum, and maximum angle, and range of motion (ROM). Unpaired t-tests with Bonferroni corrections were used to compare the differences of the group means of the descriptive statistics where relevant.

As with the previous Chapter, assessment of kinematic similarity between different models was based on the identification of visual differences in the kinematic pattern throughout the full gait cycle. Inter-trial, inter-rater, and intra-rater errors were calculated for all the kinematic outputs to assess repeatability of the type C models. Two staff members with a different experience level were involved in this study, the senior rater had 16 years of experience and the junior only had one. The full procedure is outlined in Section 6.2.5.

9.3 Results

9.3.1 Inter-Segmental Kinematics

Figures 9.3, 9.4, and 9.5 shows the inter-segmental kinematics of the medial, lateral, and intermediate forefoot segments relative to the OFM hindfoot segment. These graphs were obtained by averaging the representative trials of all 45 healthy normal subjects, as done in the previous Chapter. All gait kinematics were scaled to 100% gait cycle (GC) with a grey band indicating one standard deviation to the mean angle. In general, the inter-segmental kinematics were found to be similar between different type C variants. Each type C forefoot segment (medial, lateral, and intermediate) exhibited a good match in kinematic patterns with their corresponding segments of the type A and type B OFFMs (Figs. 8.3, 8.4, and 8.5).

In this study, the medial forefoot was defined using the same marker placement strategies in all three type C models (C1, C2, and C2b), with the only difference being the use of a virtually reconstructed D1M marker in C2b instead of a physical tracking marker. While this change did not appear to influence the sagittal plane motion, minor differences were found in the frontal and transverse plane (Fig. 9.3). An offset of approximately 10° and 3° were found in the frontal and transverse plane kinematics. The C2b model also had an increased ROM in the frontal plane ($p <$

0.01). Finally, instead of a gradual forefoot abduction as the foot approaches terminal stance, as seen in the C1 model, a rapid abduction of 10° was observed about 50% GC for the C2b model.

Two approaches were investigated for defining the lateral forefoot segment, the C1 model used P4M to define the x-axis and the C2 (and C2b) used D5M (Fig. 9.4). The z-axis of the lateral forefoot was defined in the exact same way in all type C models, as the direction from P5M to D5M. Therefore, it is not surprising that all models produced nearly the same sagittal plane motion. The frontal plane motion was also very similar, with expected offsets. The overall transverse plane kinematics was consistent in general, with some minor differences in the ROM. Between 0 and 16% GC, the C1 lateral forefoot abducted by 5° whereas the C2 model only abducted by 1° . From 42 – 65% GC, the C2 model displayed a larger ROM by first adducting by 5° followed by abduction of 7° . In the corresponding period, the C1 model only exhibited a ROM of only 4° . There were little observable differences in the kinematics of the intermediate forefoot models (Fig. 9.5), with expected offsets found between the C1 and C2 model, but only the in transverse plane and not the other two planes.

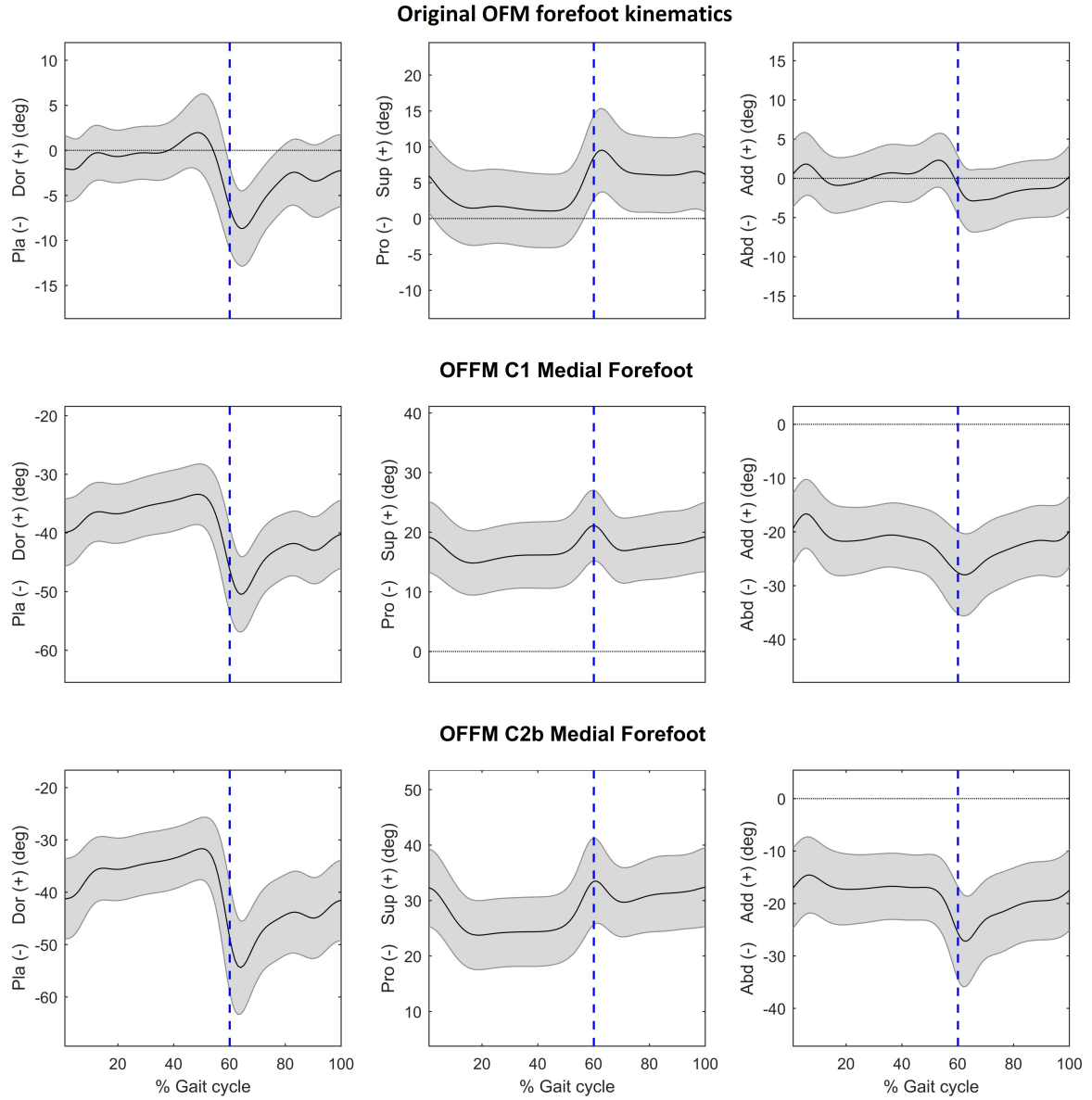


Figure 9.3: Inter-segmental kinematics of the type C medial forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right respectively. Means $\pm$ 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM C1 medial forefoot, and OFFM C2b medial forefoot. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. OFFM C2 medial forefoot segment has the same definition as C1 and hence omitted.

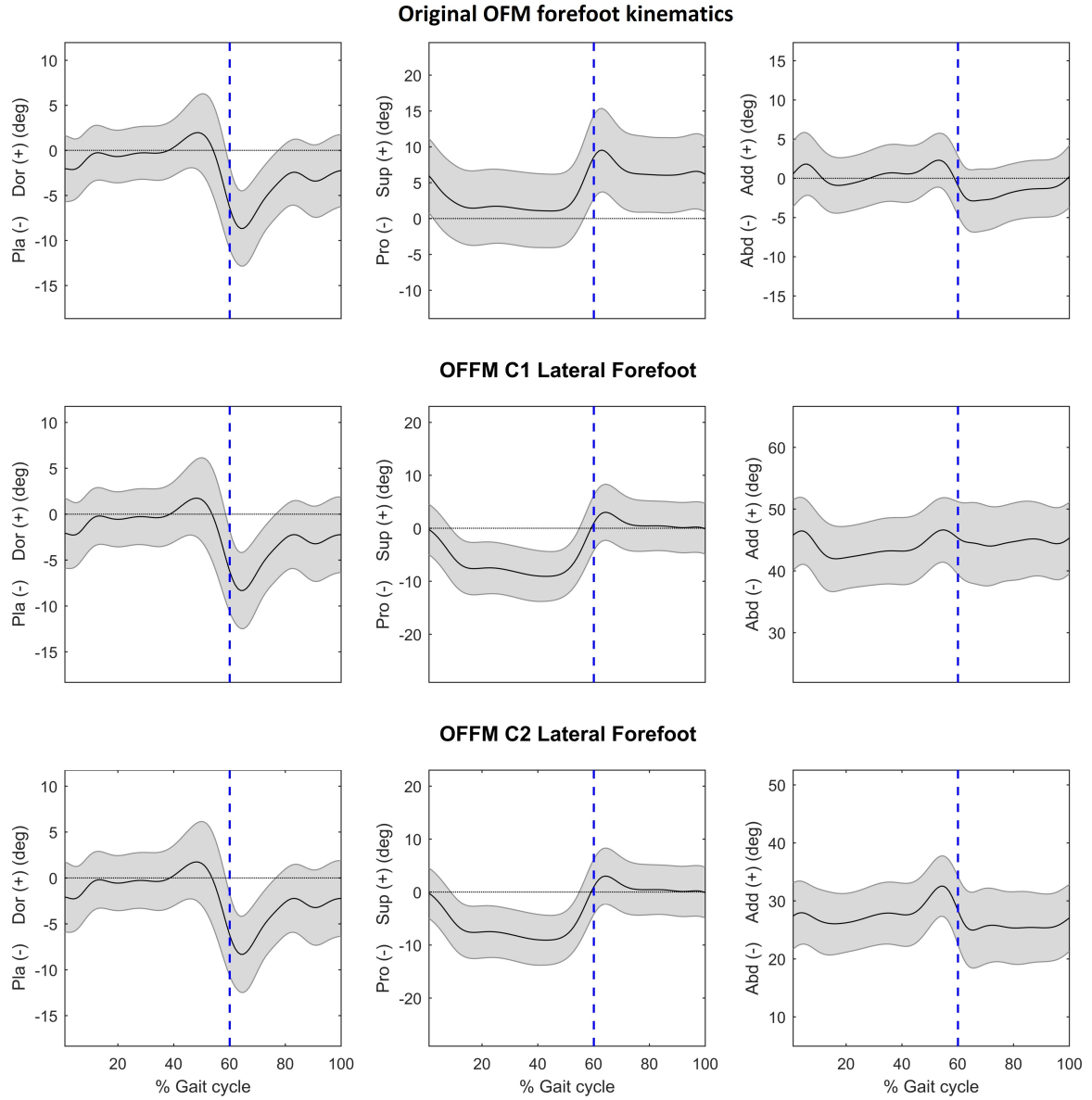


Figure 9.4: Inter-segmental kinematics of the type C lateral forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right respectively. Means $\pm$ 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM C1 lateral forefoot, and OFFM C2 lateral forefoot. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. OFFM C2b lateral forefoot segment has the same definition as C2 and hence omitted.

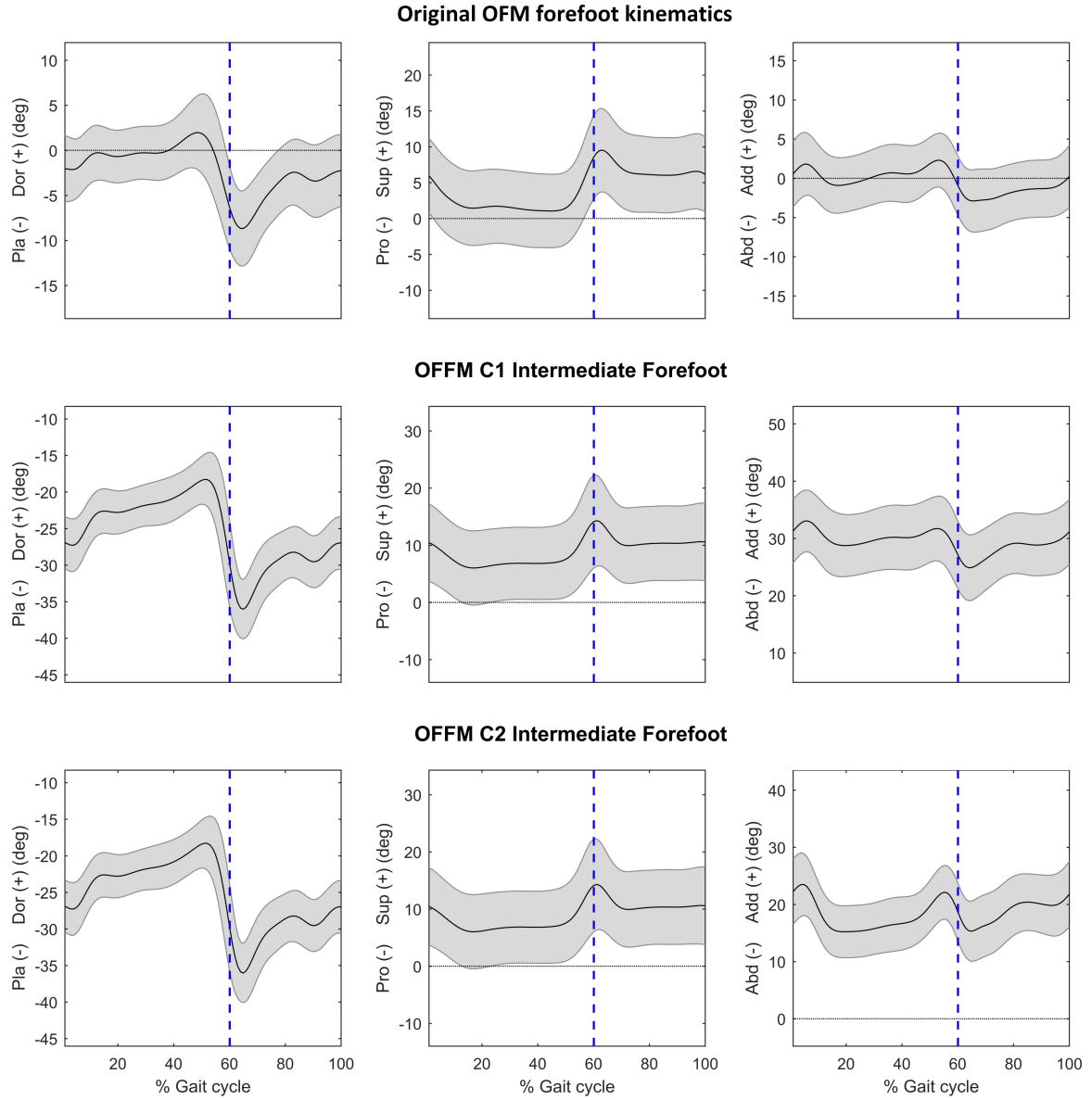


Figure 9.5: Inter-segmental kinematics of the type C lateral forefoot relative to the OFM hindfoot segment. Over the columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right respectively. Means $\pm$ 1 standard deviation (deg). Over the rows are the motions of the OFM forefoot as a reference, OFFM C1 intermediate forefoot, and OFFM C2 intermediate forefoot. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%. OFFM C2b intermediate forefoot segment has the same definition as C2 and hence omitted.

9.3.2 Planar-Angles

The motion of the first and fifth metatarsal segments measured with reference to the hindfoot sagittal and transverse plane were investigated in the previous Chapter (See Sec. 8.3.2). In the present study, the aim was to compare the differences in the kinematic output of the first metatarsal motion (MTI) between C2 and C2b, where the former uses physical markers to track the segment and the latter relies on a virtual D1M marker. Overall, the kinematics of MTI appeared to have a good consistency between the two models (Fig. 9.6). However, the ROM of the C2b first metatarsal segment was significantly larger ($p < 0.01$) than the C2 model by more than 10° (Table 9.2), with a value of 29.3° versus 17.1° in the sagittal plane, and 27.2° versus 14.2° in the transverse plane.

Interestingly, the MTI-V angle between the C2 and the C2b model displayed a very different pattern (Fig. 9.7). An increase in the forefoot spread was expected in early stance, followed by a progressive decrease towards terminal stance. However, the MTI-V for the C2b model stayed relatively undeformed throughout the stance phase, indicating a rigid forefoot configuration. Since the D1M marker is virtually reconstructed in this model, the motion of the D1M in C2b actually represented a combined motion of P1M, P2/3M and TOE. Therefore, this angle instead reflected the spread of the second/ third metatarsal and the fifth metatarsal, where P2/3M and TOE were attached.

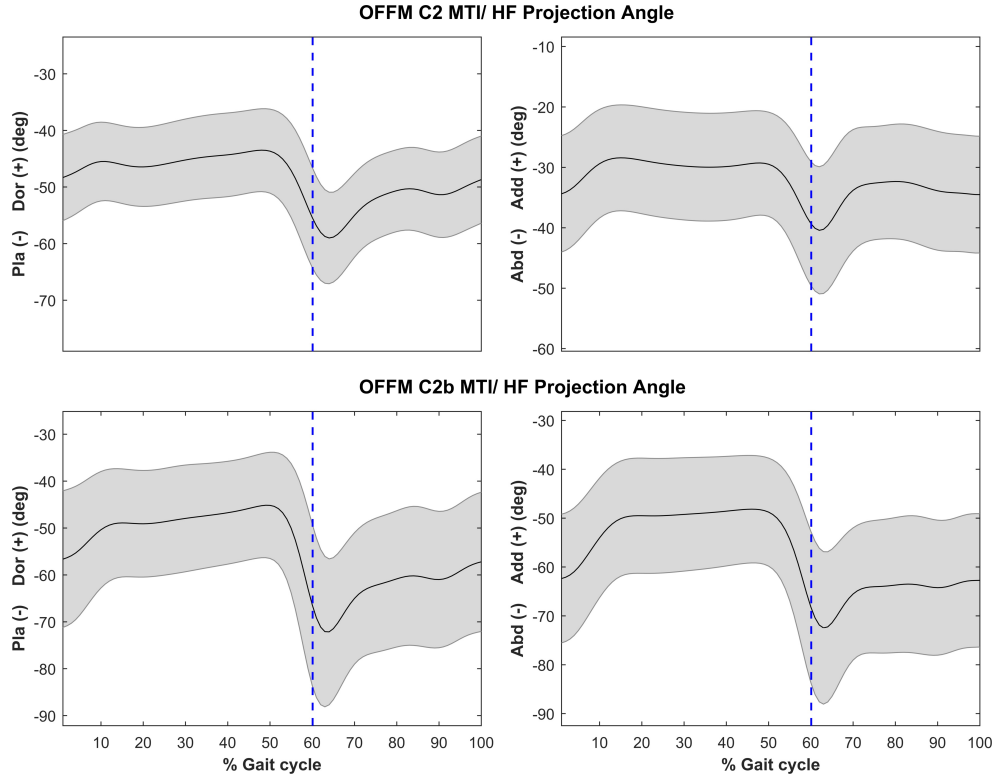


Figure 9.6: Planar angles of the type C first metatarsal segments relative to the OFM hindfoot segments. Over the columns are the dorsi/plantar flexion (left) and ab/adduction (right) motions respectively. The top row shows the C2 (same as C1) model and the bottom row the C2b model. The dotted line indicates the separation between stance and swing phase, here with an average of 60.1%.

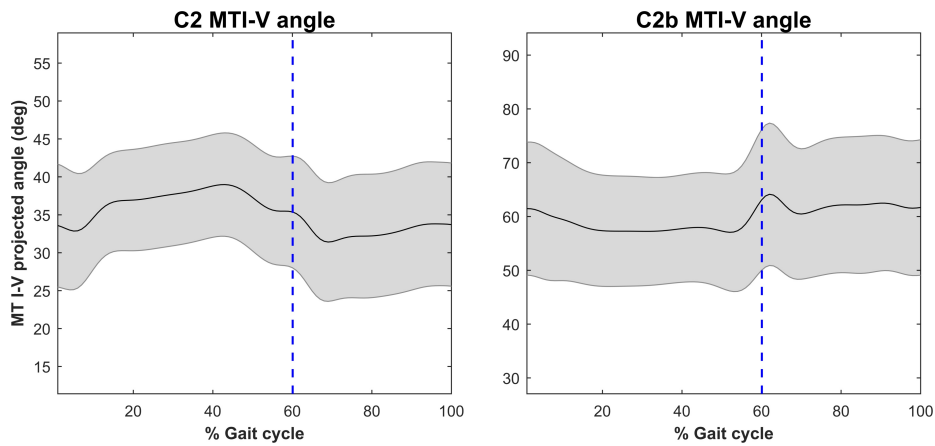


Figure 9.7: The angle between the first and fifth metatarsal segments projected onto the transverse plane of the OFM forefoot segment, for the C2 model (left) and the C2b model (left).

9.3.3 Descriptive Statistics

The average minimum, maximum, and mean angles, and the average ROM for the inter-segmental kinematics of the type C OFFM segments are reported in Table 9.2. The corresponding values for the OFM, i.e. the single rigid forefoot segment, are also included for reference. While the kinematic patterns were found to be consistent between the same forefoot segments (medial, lateral, and intermediate), offsets were found in all three anatomical planes likely due to the differences in segment definitions.

As noted previously, the C2b medial forefoot segment and the first metatarsal segment demonstrated a higher ROM than the other type C variants, as well as the OFM. On average, the motion of the OFFM segments was greater than the OFM motion in all three planes. The only exceptions were the frontal plane motion of the C1 medial forefoot (8.9°) and the sagittal motion of the C2 lateral forefoot segment (11.5°), however, both by less than 1° . The largest difference in ROM within the same segment was the first metatarsal segment motion between the C2 (same as C1) and the C2b model. The ROM observed in C2b almost doubled the values of C1. The standard deviation for the kinematic outputs of the C2b model were also higher in comparison to the C2 model, as indicated by the increased standard deviation in the motion across all three planes.

			Descriptive statistics (deg)			ROM (deg)
			Min (S.D.)	Max (S.D.)	Mean (S.D.)	Mean (S.D.)
Oxford Foot Model		Dorsiflexion	-9.1 (4.3)	2.8 (3.4)	-2.0 (3.3)	11.9 (4.4)
		Supination	0.4 (5.1)	10.3 (5.5)	4.4 (5.0)	9.8 (2.7)
		Adduction	-3.9 (3.7)	3.6 (3.6)	-0.3 (3.3)	7.5 (2.3)
OFFM C1	Medial Segment	Dorsiflexion	-51.3 (6.4)	-33.0 (5.0)	-39.6 (5.2)	18.3 (4.3)
		Supination	14.2 (5.4)	22.2 (5.9)	17.4 (5.3)	8.9 (2.3)
		Adduction	-28.8 (7.7)	-16.0 (6.2)	-22.2 (6.3)	12.8 (4.4)
	Lateral Segment	Dorsiflexion	-8.8 (4.2)	2.7 (3.5)	-2.0 (3.3)	11.5 (3.8)
		Supination	-9.6 (4.6)	3.5 (5.2)	-3.4 (4.6)	13.0 (2.4)
		Adduction	40.1 (5.9)	48.4 (5.2)	44.2 (5.3)	8.3 (2.8)
	Interm. Segment	Dorsiflexion	-36.9 (4.0)	-17.7 (3.5)	-25.6 (3.0)	19.1 (3.7)
		Supination	5.2 (6.4)	15.5 (7.6)	9.1 (6.3)	10.3 (3.6)
		Adduction	23.7 (5.3)	34.6 (5.6)	29.4 (5.1)	10.9 (3.3)
OFFM C2	Medial Segment	See OFFM C1 medial segment				
	Lateral Segment	Dorsiflexion	-8.8 (4.2)	2.7 (3.5)	-2.0 (3.3)	11.5 (3.8)
		Supination	-9.6 (4.6)	3.5 (5.2)	-3.4 (4.6)	13.0 (2.4)
		Adduction	22.3 (5.2)	33.6 (5.1)	27.0 (4.8)	11.3 (3.4)
	Interm. Segment	Dorsiflexion	-36.9 (4.0)	-17.7 (3.5)	-25.6 (3.0)	19.1 (3.7)
		Supination	5.2 (6.4)	15.5 (7.6)	9.1 (6.3)	10.3 (3.6)
		Adduction	12.7 (4.7)	25.4 (5.2)	18.4 (4.3)	12.7 (3.8)
OFFM C2b	Medial Segment	Dorsiflexion	-55.6 (8.8)	-31.2 (6.0)	-40.1 (6.7)	24.4 (5.1)
		Supination	22.9 (6.2)	35.7 (7.1)	28.4 (6.2)	12.9 (3.3)
		Adduction	-27.9 (8.6)	-13.4 (7.1)	-18.9 (6.9)	14.6 (4.0)
	Lateral Segment	See OFFM C2 lateral segment				
	Interm. Segment	See OFFM C2 intermediate segment				
	OFFM C1; C2	Dorsiflexion	42.7 (7.2)	59.8 (8.1)	48.7 (7.2)	17.1 (4.5)
		Adduction	-41.7 (10.6)	-27.5 (8.7)	-32.4 (9.0)	14.2 (4.9)
1st. MT motion	OFFM C2b	Dorsiflexion	44.3 (11.5)	73.6 (15.9)	55.0 (12.9)	29.3 (7.7)
		Adduction	-73.9 (15.5)	-46.8 (11.0)	-57.6 (12.3)	27.2 (9.0)
5th. MT motion	All type C models	Same as type B model, see Table 8.4				
MT I-V angle	OFFM C1; C2	Angle	30.3 (7.3)	40.4 (7.1)	35.3 (7.1)	10.1 (2.8)
	OFFM C2b	Angle	54.0 (10.4)	66.6 (12.7)	59.9 (11.2)	12.7 (5.4)

Table 9.2: Comparison of descriptive statistics and range of motion for the fore-foot/hindfoot kinematics for the original OFM and type C OFFM medial, lateral, and intermediate segments. The kinematics of the first and fifth metatarsal segments are measured as projected planar angles. Abbreviations: Interm, Intermediate; MT, Metatarsal; ROM, Range of motion.

9.3.4 Repeatability Analysis

The repeatability of type C OFFMs and OFM kinematics was reported using the average inter-trial, inter-rater, and intra-rater errors as before (Table 9.3). Overall, the OFM demonstrated a good marker-placement repeatability in all three planes, with an average inter-rater error of less than 3.3° in all three planes. The repeatability of the type C OFFM forefoot segments was more variable than the OFM, with the three-dimensional segments being more repeatable than the first metatarsal segment in general. For three-dimensional segments, the sagittal motion was the most repeatable and the transverse motion the least repeatable. The highest variability was reported for the C1 intermediate forefoot and C2 lateral forefoot segment, with an inter-rater error of 5.5° and 5.2° respectively.

For the planar angles, the highest error was observed for the first metatarsal motion of the C2b model, with an inter-rater error of 8.5° and 6.4° in the sagittal and transverse plane respectively. In all three planes, the variability of the C2b exceeded the C2 model. In terms of the experience level, both raters demonstrated a similar level of intra-rater repeatability.

			Inter-trial error (deg)			Intra- rater error (deg)		Inter- rater error (deg)	Inter-rater Inter-trial
			Jun.	Sen.	Av.	Jun.	Sen.	Av.	
Oxford foot model		Dor.	0.6	0.5	0.6	2.9	3.0	3.0	5.4
		Sup.	0.5	0.5	0.5	1.5	0.5	1.5	2.9
		Add.	1.0	0.6	0.8	3.4	3.0	3.3	4.1
OFFM C1	Medial segment	Dor.	0.5	0.7	0.6	2.3	3.8	3.2	5.3
		Sup.	0.6	0.4	0.5	0.7	1.1	1.0	2.0
		Add.	0.7	0.8	0.7	0.8	1.0	1.0	1.3
	Lateral segment	Dor.	0.7	0.5	0.6	3.0	3.1	3.1	5.3
		Sup.	0.5	0.6	0.5	4.1	3.0	3.6	7.2
		Add.	0.9	0.5	0.7	3.8	1.8	3.8	5.2
	Interm. segment	Dor.	0.5	0.7	0.6	0.9	1.2	1.1	1.7
		Sup.	0.7	0.5	0.6	3.3	1.2	2.6	4.4
		Add.	0.7	0.6	0.6	7.0	3.3	5.5	8.6
OFFM C2	Medial segment	See OFFM C1 medial segment							
	Lateral segment	Dor.	0.7	0.5	0.6	3.0	3.1	3.1	5.3
		Sup.	0.5	0.6	0.5	4.1	3.0	3.6	7.2
		Add.	1.3	0.9	1.1	6.0	4.1	5.2	4.8
	Interm. segment	Dor.	0.5	0.7	0.6	0.9	1.2	1.1	1.7
		Sup.	0.7	0.5	0.6	3.3	1.2	2.6	4.4
Add.		1.1	1.0	1.0	3.8	2.8	3.3	3.2	
OFFM C2b	Medial segment	Dor.	0.7	0.9	0.8	3.6	3.6	3.7	4.7
		Sup.	0.7	0.6	0.7	1.9	1.4	1.7	2.5
		Add.	0.8	0.7	0.8	1.4	0.9	1.5	1.9
	Lateral segment	See OFFM C2 lateral segment							
	Interm. segment	See OFFM C2 intermediate segment							
	1st. MT motion	C1; C2	Dor.	0.6	0.7	0.7	5.5	6.5	6.1
Add.			0.7	0.5	0.6	4.2	5.4	4.9	7.9
C2b		Dor.	0.9	1.2	1.0	9.2	7.5	8.5	8.3
		Add.	1.0	1.1	1.0	5.2	7.2	6.4	6.2
5th. MT motion	All type C models	Same as type B models, see Table 8.6							
MT I-V angle	C1; C2	Angle	0.9	0.7	0.8	5.1	5.2	5.2	6.7
	C2b	Agnle	1.1	1.1	1.1	6.2	7.2	6.8	6.4

Table 9.3: Comparison of repeatability matrices for the forefoot/hindfoot kinematics for the original OFM and type C OFFM medial and lateral segments. Average inter-trial, intra-rater, inter-rater error and the ratio of inter-rater to inter-trial error were calculated for all kinematic outputs. Abbreviations: Dor, Dorsiflexion; Sup, Supination; Add, Adduction; Jun, Junior; Sen, Senior; Av, Average.

9.4 Discussion

The present study investigated three hybrid models that combined the features of both type A and B models proposed in Chapter 8. The goal was to design and validate a clinically applicable model that is able to measure the motion of all the proposed forefoot segments, including the medial forefoot, intermediate forefoot, lateral forefoot, and the first and fifth metatarsal segments simultaneously. To achieve this, three hybrid models were proposed and evaluated.

All three type C OFFM models were designed and evaluated using the same marker-set and data as before, with ten markers simultaneously attached to all proximal bases and distal heads of all five metatarsals. As commented in the previous Chapter (Sec. 8.4), all the models fulfilled the criteria of being visible to the cameras at all times. All three type C models required two additional markers compared to the OFM, and therefore were equal in terms of model complexity. The only differences between C1 and C2/C2b was the choice of D4M or P4M marker in the definition of the x-axis for the intermediate and lateral segment (Table 9.1). However, the distal heads of the metatarsals are more easily identifiable by manual palpation than the proximal bases [38]. Therefore, based on practicality, model C2 (and C2b) is preferred over C1.

Overall, the gait patterns of the medial, intermediate, and lateral forefoot segment between the C1, C2, and C2b models showed a good agreement. However, higher variability was observed for the C2b model in the medial forefoot, the first metatarsal, and the MTI-V angle, as indicated by a higher standard deviation in the range of motion as well as a larger inter-trial, intra-rater, and inter-rater error. The increase in variability can be explained by the use of a virtually reconstructed D1M instead of a physical marker. In addition, the MTI-V angle measured from model C2b was different from the C1/C2, in contrast to what was expected of the forefoot deformation

pattern according to the literature [82, 240]. This finding suggests that the motion of the distal head of the first metatarsal is unique and that recreating the movement using a virtually reconstructed D1M is inappropriate. However, it is acknowledged that the current study only investigated rigid body reconstruction using neighbouring markers (P1M, P2/3M, and TOE). Using different marker combinations or calibration techniques, such as CAST [38], may yield different results. Alternatively, other non-invasive approaches such as directly attaching multiple markers to the first metatarsal (e.g. Ghent foot model [65]) or by using a single sensor with multiple degrees of freedom [75] may be more appropriate in tracking the first metatarsal segment.

9.5 Summary

This Chapter presented the second part of a two-part study (including Chapter 8) to explore different marker placement and segment definition strategies to track different forefoot segments. Several unique gait patterns have been identified that were previously not available if the forefoot were to be represented as a unified single segment. In particular, the convergence of the medial and lateral forefoot in the transverse plane during late stance and the patterns of the first and fifth metatarsal segments were interesting findings. Although the intermediate forefoot did not provide new information, with a similar gait pattern of the lateral forefoot, tracking this segment does not require additional effort. In addition, it may be useful in the event of a deformity that affects the first ray, such as hallux valgus [34].

In consideration of design limitations, including marker visibility, practicality, model complexity, backward compatibility, and methodological factors such as repeatability, consistency with our knowledge of foot biomechanics, and the identification of unique kinematic patterns, model C2 is preferred and recommended over the other models. In order to further explore the clinical utility of a subdivided forefoot

model, five individual clinical case studies were performed using this model in the next Chapter.

Chapter 10

Clinical Case Studies

Through a two-part study investigating various forefoot marker placement and segment definition strategies, a final protocol (Oxford Forefoot Model C2) has been proposed to track a number of forefoot segments, that is, the medial forefoot, lateral forefoot, intermediate forefoot, and the independent motion of the first and fifth metatarsals. The information obtained from this protocol is consistent with the general knowledge of foot biomechanics. Furthermore, the model is also backward compatible with the original Oxford Foot Model (OFM) with easy to identify landmarks. This Chapter presents the results of five clinical case studies which explored the potential utility and efficacy of this model.

10.1 Introduction

The rationale for proposing a MSFM with a sub-divided forefoot arises from the inadequate representation of the complex motion of the forefoot in most MSFMs [210, 188]. Through extensive analysis and testing of a variety of marker placement strategies in the previous two Chapters, a novel forefoot model (type C2) based on the rigid body criteria set out in Chapter 6 was proposed to explore the kinematics of different forefoot segments. The hybrid model was developed to track three independent forefoot segments simultaneously as well as the planar motion of the first and fifth metatarsals.

The core value of instrumented clinical gait analysis (CGA) is to be able to objectively identify and understand abnormal gait characteristic that eventually lead to recommendations that positively influence the outcome of a treatment plan [12]. Therefore, one of the objectives of this thesis is to explore the potential clinical applications of a subdivided forefoot kinematics model.

10.2 Methodology

10.2.1 Ethical Considerations

To explore potential clinical utility of the proposed forefoot model, this study used the data collected from patients who were referred to the Oxford Gait Laboratory for routine CGA. The data was collected with the approval of the National Research Ethics Service, as outlined in Section 4.1.

10.2.2 Subjects and Participants

Five subjects with a range of conditions were involved in this study, summarised in Table 10.1 below. Figures 10.1 - 10.5 shows the structures of the patient's foot from three different angles, revealing the presence of relevant foot deformity and the position of marker placement.

Subject code	Age	Sex	Condition	Treatment history
A	13	M	Cerebral palsy/ spastic diplegia	Right-sided multi-level surgery one year ago which consisted of adductor release, hamstring lengthening, gastrocnemius lengthening, and femoral derotation with titanium elastic nails (TENS). The patient had removal of end cap from TENS nails and compression of osteotomy one month after surgery due to distraction at femoral osteotomy site.
B	11	F	Rotational malalignment and hip pain	No treatment.
C	10	F	Bilateral clubfoot	Treated with Ponsetti casting as an infant.
D	10	F	Torsional deformity	No treatment.
E	12	F	Clubfoot (L)	Treated with Ponsetti casting as an infant.

Table 10.1: Clinical case subject information and clinical background.

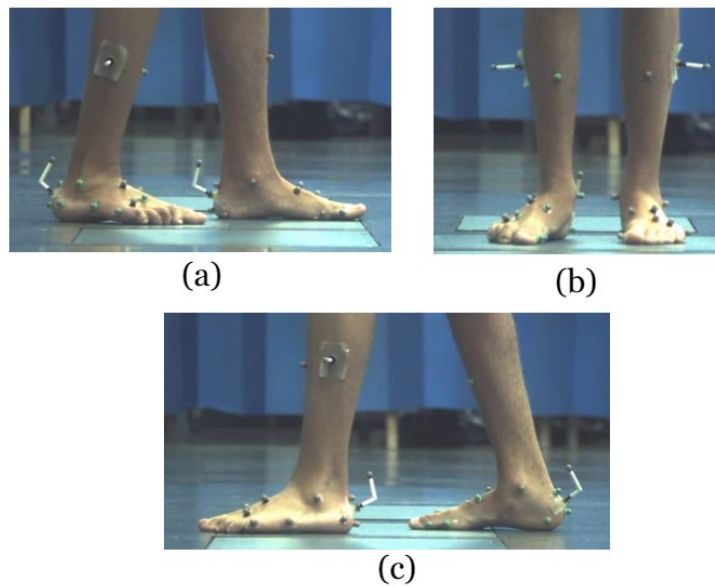


Figure 10.1: Static foot posture of Subject A, a 13 year old boy diagnosed with cerebral palsy. The subfigures reveal the foot structure and marker placement in the a) right-view, b) frontal-view, and c) left-view.

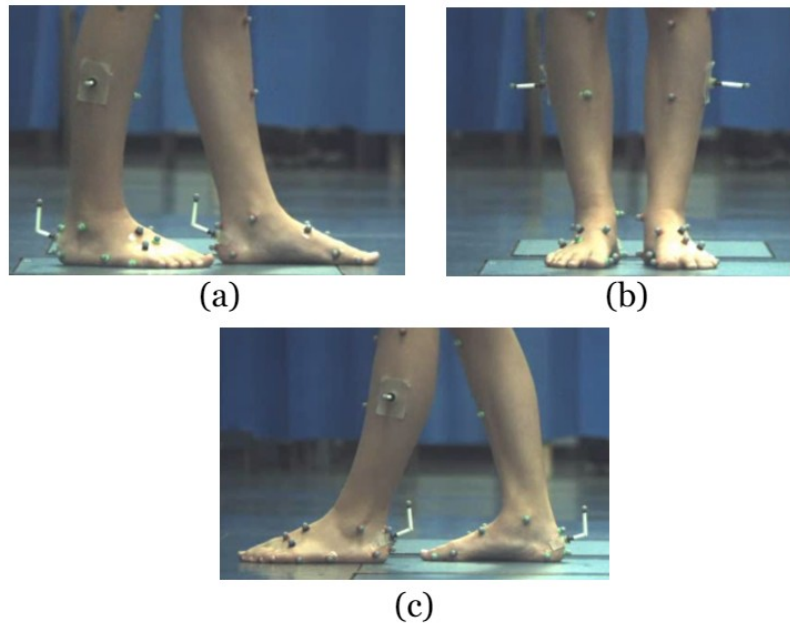


Figure 10.2: Static foot posture of Subject B, a 11 year old girl diagnosed with rotational malalignment and hip pain. The subfigures reveal the foot structure and marker placement in the a) right-view, b) frontal-view, and c) left-view.

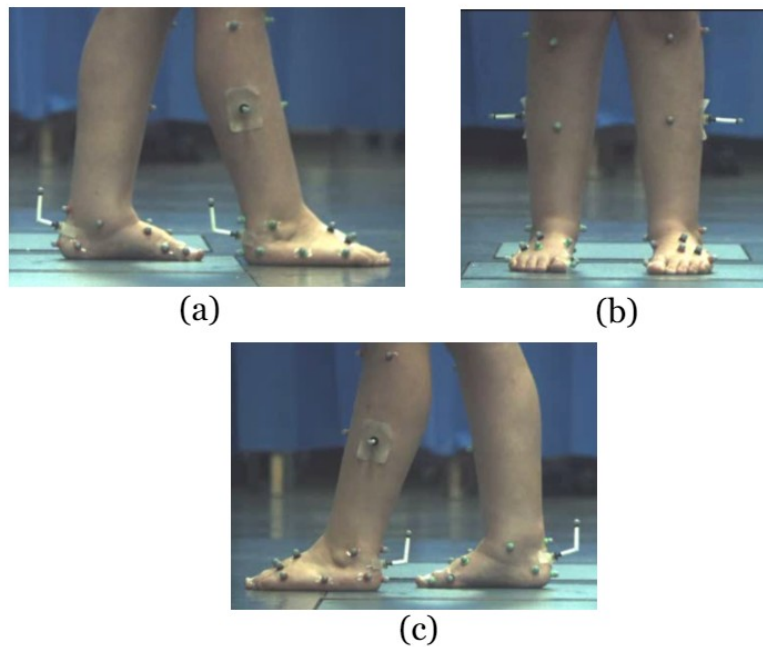


Figure 10.3: Static foot posture of Subject C, a 10 year old girl diagnosed with bilateral clubfoot. The subfigures reveal the foot structure and marker placement in the a) right-view, b) front-view, and c) left-view.

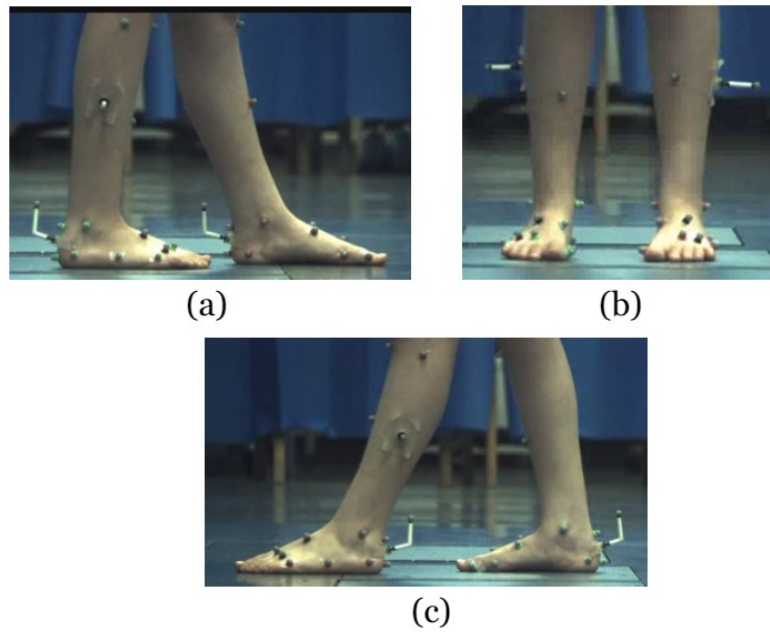


Figure 10.4: Static foot posture of Subject D, a 10 year old girl diagnosed with torsional deformity. The subfigures reveal the foot structure and marker placement in the a) right-view, b) front-view, and c) left-view.

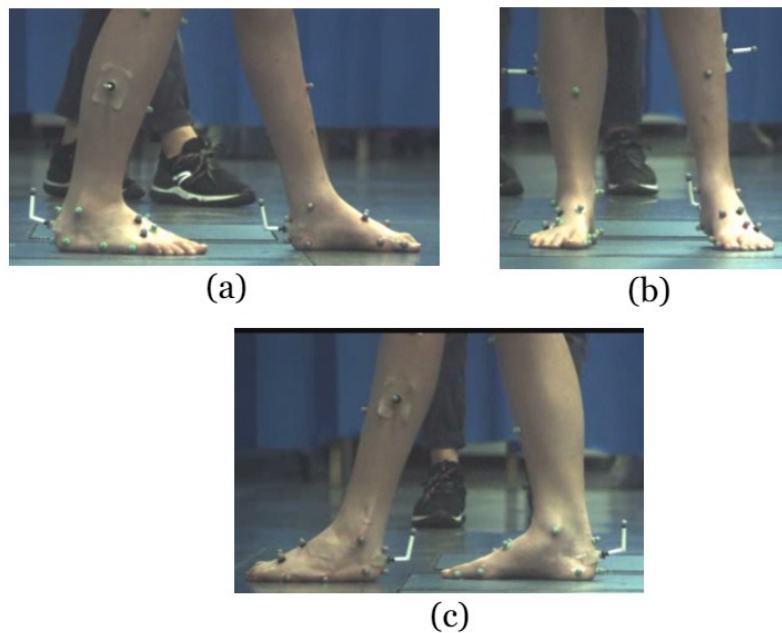


Figure 10.5: Static foot posture of Subject E, a 12 year old girl diagnosed with left-sided clubfoot. The subfigures reveal the foot structure and marker placement in the a) right-view, b) front-view, and c) left-view.

10.2.3 Data Collection and Processing

To explore the potential clinical utility of the OFFM, gait data were analysed from five patients who were referred to the OGL. Since OFFM C2 is backward compatible with the OFM, requiring only two additional markers to be attached to the patient's foot, the data were collected during a routine clinical analysis with no alteration to the existing protocol. Since the data collection protocol designed for this research was based on the OGL protocols, this study used the same data collection and processing procedure as all the previous Chapters. See Sections 4.5 and 4.4 for more information on data collection and processing, and Section 4.3 for the equipment used.

Forefoot kinematics were calculated using OFFM type C2 (Fig. 9.2). The kinematics of the medial forefoot, intermediate forefoot, and lateral forefoot were measured with reference to the OFM hindfoot using the same kinematic convention (Sec. 4.7.2). The motion of the individual first and fifth metatarsals (MTI and MTV) and the angle between the first and fifth metatarsals projected onto the transverse plane (MTIV) were also calculated. For each subject, at least one static trial and four walking trials were collected during a single gait analysis session. To avoid the loss of information due to averaging, a representative trial was selected for further analysis based on how close its OFM joint kinematic graphs were to the average for the particular subject. OFM kinematics were also computed for each subject to aid interpretation and to serve as a reference for comparing OFFM to a unified forefoot segment (See Appendix B, Figs. B.3 - B.7).

10.2.4 Abnormal Gait Identification

The OFM kinematics calculated for each subject were compared to the mean values of healthy normal children who were age and gender matched. However, since this novel forefoot model is currently still at a developmental stage, the OFFM kinematics were compared against the mean values of normal healthy adults established in Chapter 9. See Section 9.3 for more detail.

10.3 Results

10.3.1 Case A

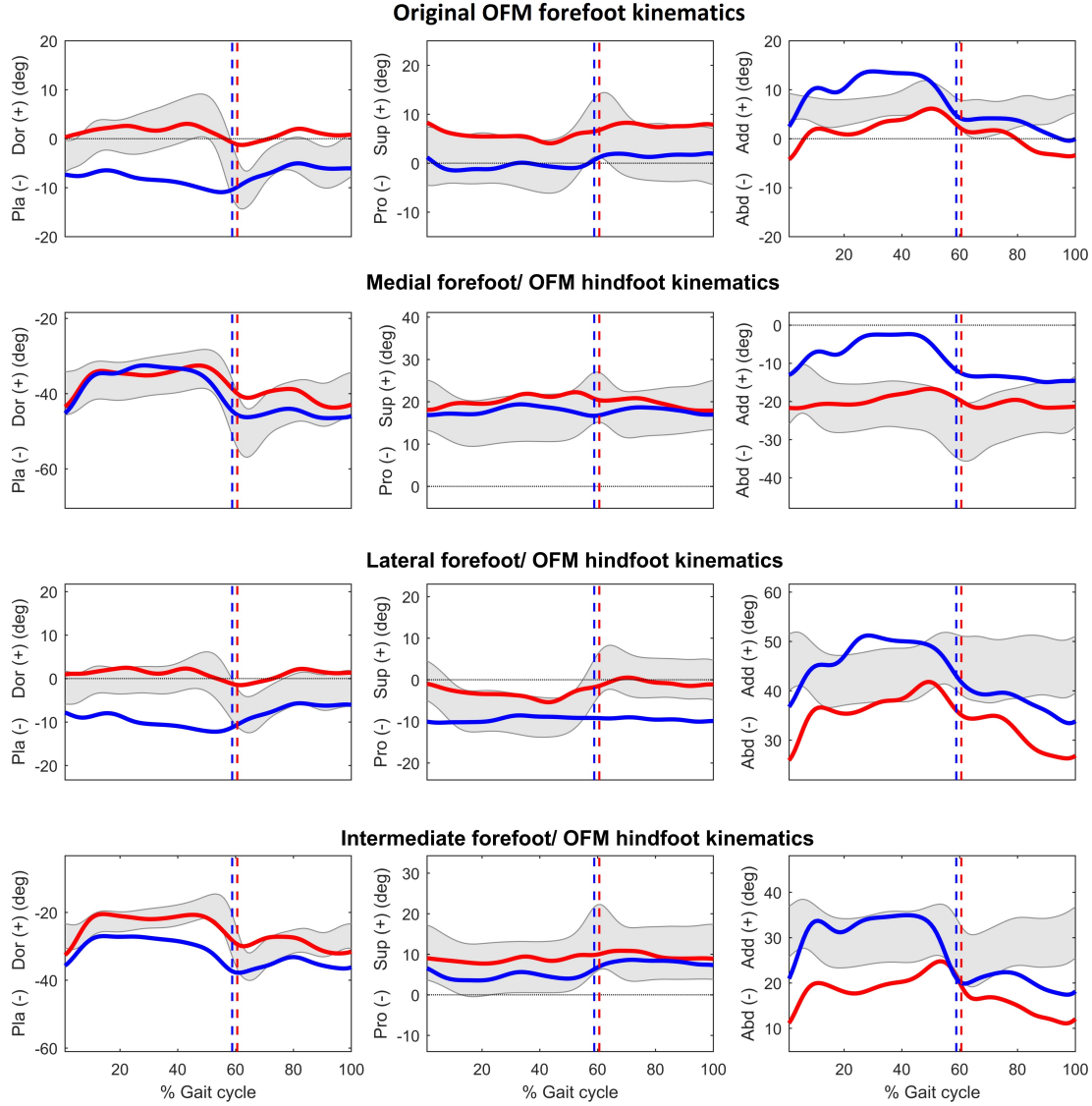


Figure 10.6: Inter-segmental kinematics of the OFM forefoot (top row), type C2 OFFM medial forefoot (second row), OFFM lateral forefoot (third row), and OFFM intermediate forefoot (bottom row) relative to the OFM hindfoot segment, for the subject A. Over the three columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults (except for the top row). The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phase.

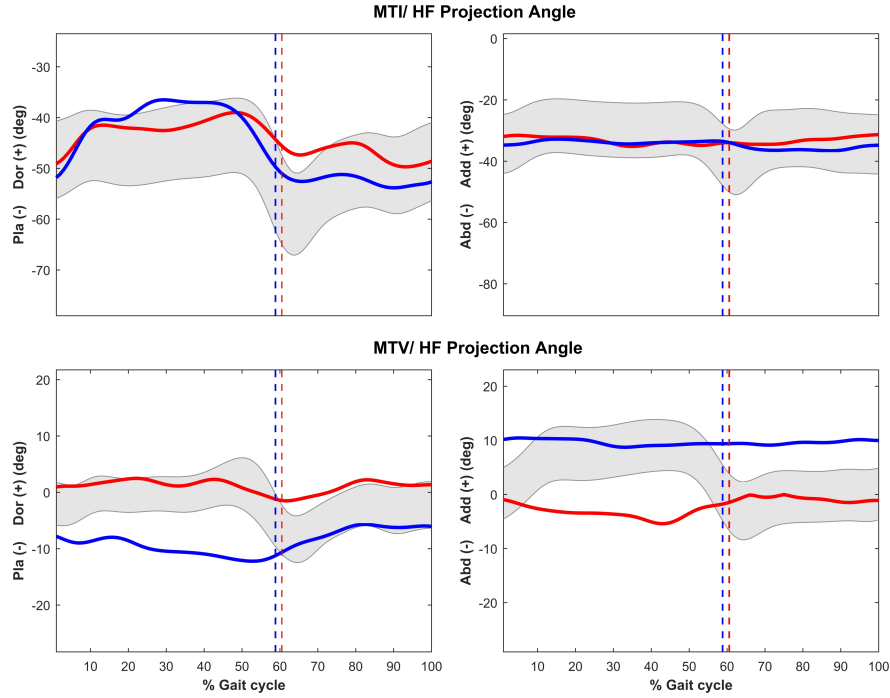


Figure 10.7: Planar angles of the first (top row) and fifth (bottom row) metatarsal relative to the OFM hindfoot segment, for the subject A. Over the columns are the dorsi/plantar flexion and ab/adduction motions from left to right. The grey band represents the mean values of 45 healthy adults. The blue line is one full gait cycle for the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

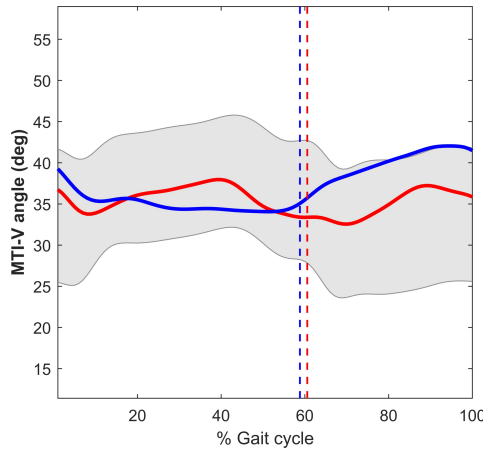


Figure 10.8: The angle between the first and fifth metatarsal vectors projected onto the OFM forefoot transverse plane, for the subject A. The grey band represents the mean values of 45 healthy adults. The blue line is one full gait cycle from the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

The first case study presents the kinematic analysis of a 13 year old male subject with spastic diplegia cerebral palsy. A reduced range of motion (ROM) in the sagittal and frontal planes were observed in the OFM forefoot kinematics (Fig. 10.6). By subdividing the forefoot into two segments, medial and lateral, it is apparent that the reduced ROM mainly occurred on the lateral side of the forefoot. This is confirmed by the loss of fifth metatarsal motion as shown in Figure 10.7. While there was still a clear reduction in the ROM in both the sagittal and frontal plane for the medial forefoot, the kinematic pattern appeared to be within the normal range. For the lateral forefoot segment, the kinematic pattern showed a good agreement with the OFM. Since the OFM mainly uses the markers on the lateral side for tracking (the D1M marker on the distal head of the first metatarsal is virtually reconstructed), the resemblance of the kinematic pattern between the lateral forefoot and the OFM is not surprising.

Theoretically, the intermediate forefoot should represent the average kinematic of the medial and lateral forefoot. However, the intermediate forefoot showed a better agreement with the medial forefoot, with no new information gained. Due to the lack of transverse plane movement of the first and fifth metatarsal (Fig. 10.7), the MTI-V angle stayed relatively undeformed throughout the stance phase, followed by a progressive increase throughout the swing phase (Fig. 10.8).

10.3.2 Case B

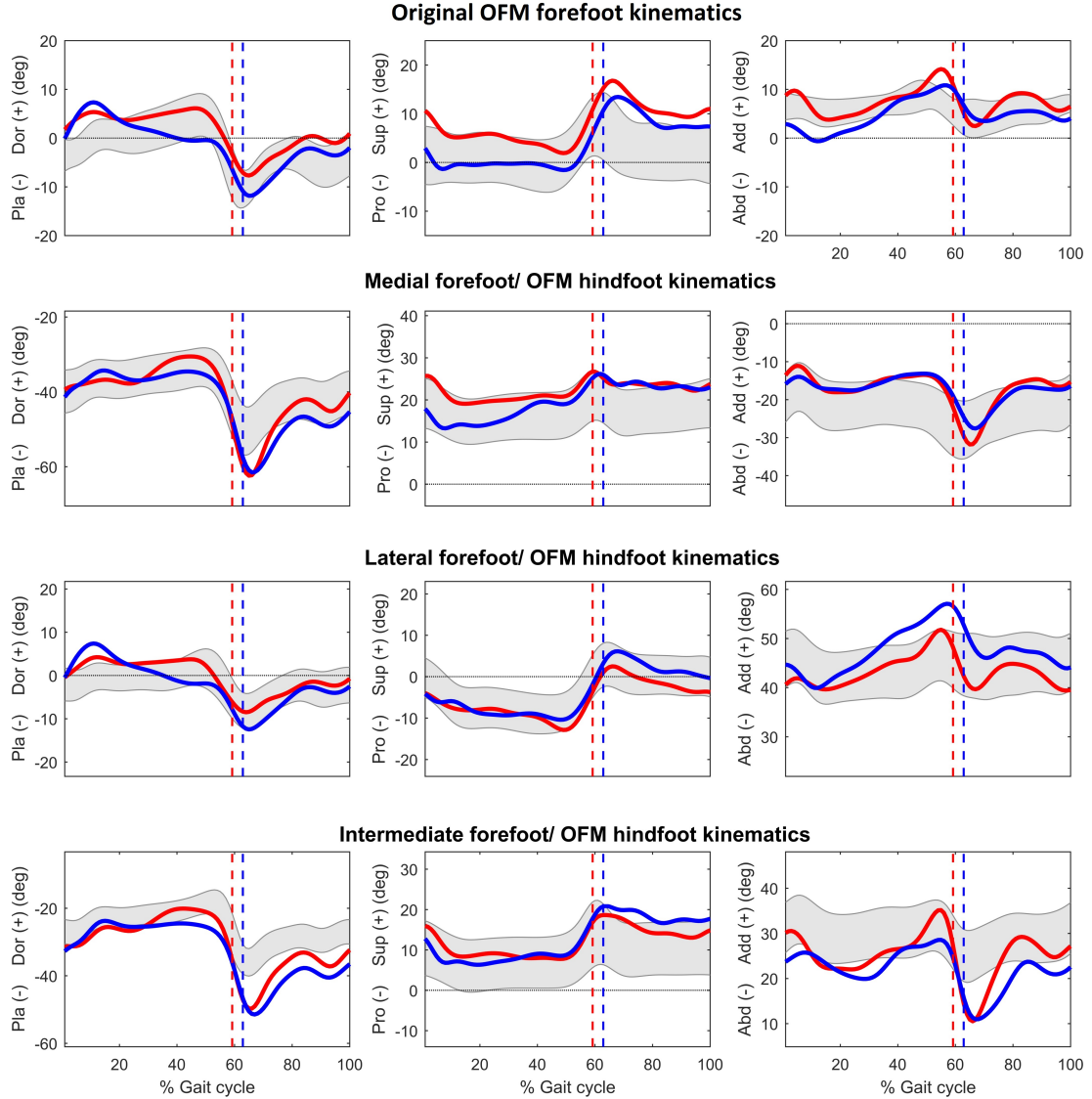


Figure 10.9: Inter-segmental kinematics of the OFM forefoot (top row), type C2 OFFM medial forefoot (second row), OFFM lateral forefoot (third row), and OFFM intermediate forefoot (bottom row) relative to the OFM hindfoot segment, for the subject B. Over the three columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults (except for the top row). The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phase.

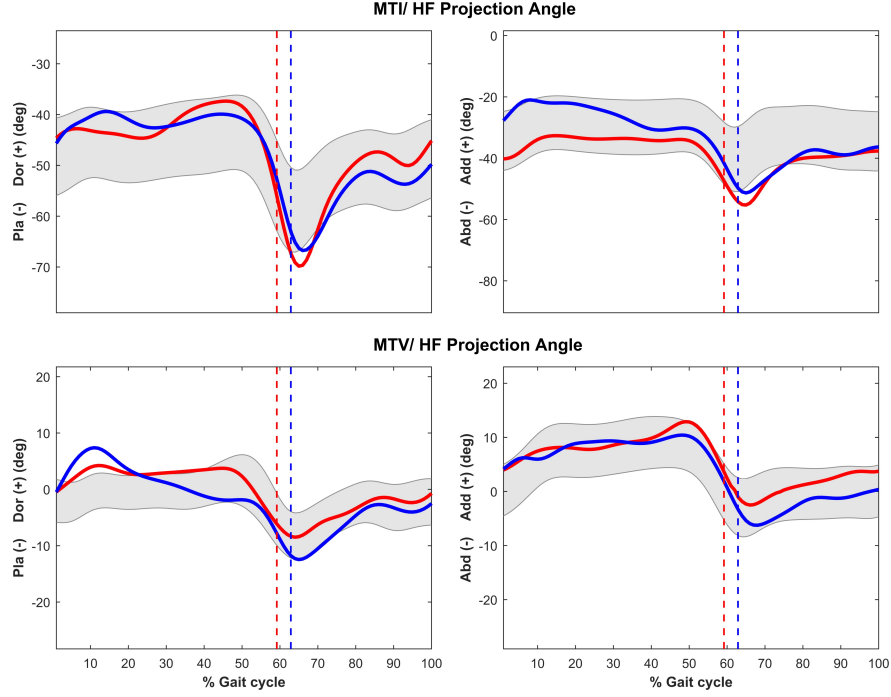


Figure 10.10: Planar angles of the first (top row) and fifth (bottom row) metatarsal relative to the OFM hindfoot segment, for the subject B. Over the two columns are the dorsi/plantar flexion and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults. The blue line is one full gait cycle for the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

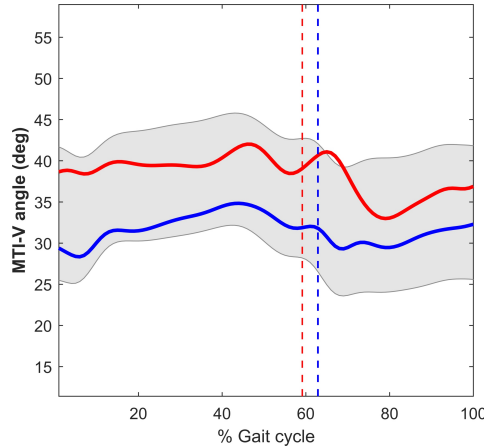


Figure 10.11: The angle between the first and fifth metatarsal vectors projected onto the OFM forefoot transverse plane, for the subject B. The grey band represents the mean values of 45 healthy adults. The blue line is one full gait cycle from the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

The second case investigates the gait of 11 year old girl with a lower-limb rotational malalignment and hip pain. Fig. B.4 in Appendix B shows an external rotation of the hindfoot relative to the tibia (i.e. ankle joint complex), and a supinated forefoot relative to the tibia, which may have reflected the diagnosis. However, there was no obvious gait deviation in the forefoot motion (Fig. 10.9), nor was any expected from this type of disorder. The only difference was the increased ROM observed in the transverse plane of the lateral forefoot and the intermediate forefoot.

Interestingly, the left intermediate forefoot abducted by 24° from 56 – 66 % GC, while the medial and lateral forefoot only abducted by 15° and 12° respectively. The intermediate forefoot was expected to represent the average of the medial and lateral forefoot, which was not the case here.

While the transverse plane motion of the OFM forefoot, OFFM medial and OFFM intermediate forefoot remained in their normal range throughout the stance phase, the right lateral forefoot exhibited a larger adduction throughout the stance phase by 16° from 12% to 60% GC, exceeding the normal range at approximately 34% GC.

10.3.3 Case C

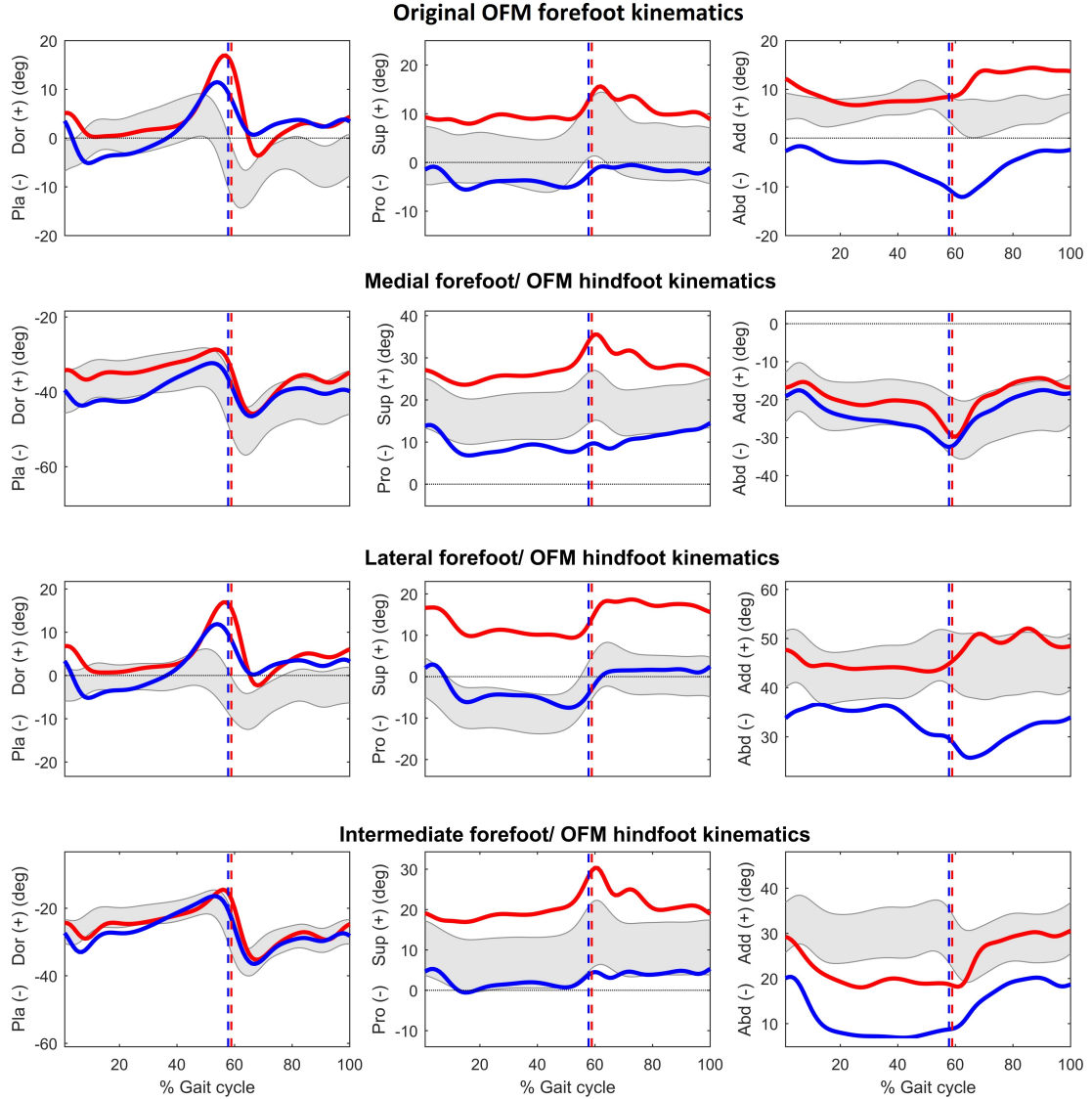


Figure 10.12: Inter-segmental kinematics of the OFM forefoot (top row), type C2 OFFM medial forefoot (second row), OFFM lateral forefoot (third row), and intermediate forefoot (bottom row) relative to the OFM hindfoot segment, for the subject C. Over the three columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults (except for the top row). The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phase.

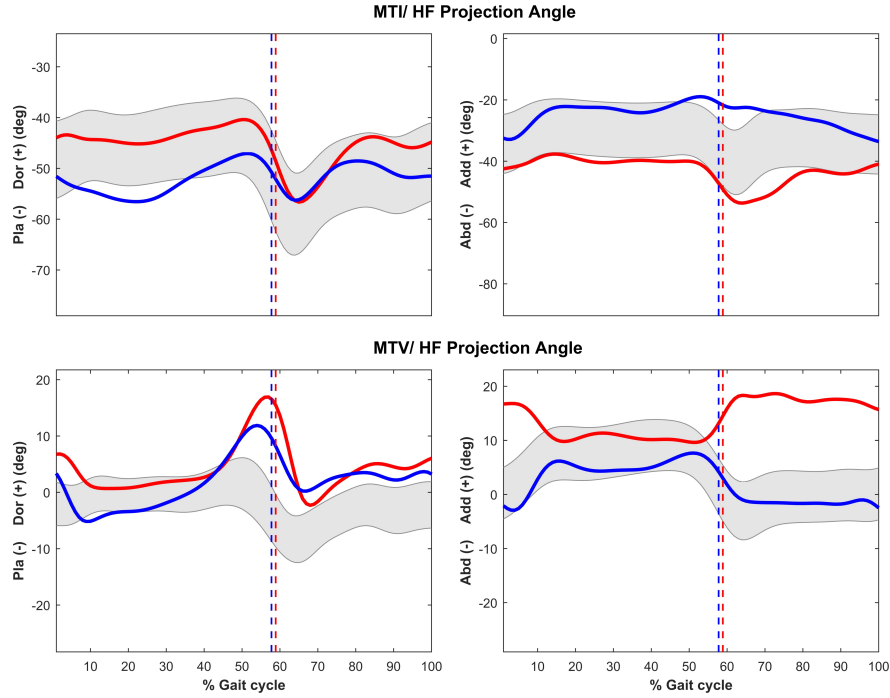


Figure 10.13: Planar angles of the first (top row) and fifth (bottom row) metatarsal relative to the OFM hindfoot segment, for the subject C. Over the two columns are the dorsi/plantar flexion and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults. The blue line is one full gait cycle for the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

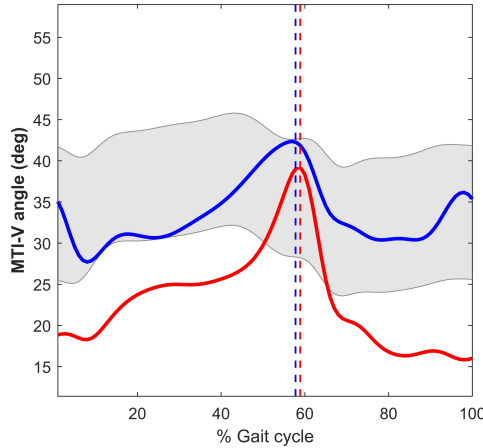


Figure 10.14: The angle between the first and fifth metatarsal vectors projected onto the OFM forefoot transverse plane, for the subject C. The grey band represents the mean values of 45 healthy adults. The blue line is one full gait cycle from the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

The third clinical case study involved a 10 years old girl with a bilateral clubfoot, who had been treated with Ponseti casting as a baby. According to the OFM, the right foot appeared to be in an abducted position, and the left foot relatively more supinated, bordering the normal range (Fig. 10.12). Similar to what was observed in Case A, the lateral forefoot showed a more consistent pattern with the OFM in comparison to the medial forefoot.

The motion of the right lateral forefoot in the transverse plane acted opposite to the expected normal motion, which was confirmed by the opposite right fifth metatarsal motion seen in Figure 10.13. During loading response (0 – 10 % GC), the metatarsals normally spread as the forefoot lowers to the ground and re-tightens in pre-swing (50 – 60 % GC). However, in this case study the right forefoot became more compact in loading response and spread in initial swing.

The left forefoot appeared to be more normal according to the inter-segmental kinematics in Figure 10.12. However, the fifth metatarsal of the left foot displayed an opposite motion to what was expected normally (Fig. 10.13). The abduction of the fifth metatarsal in loading response followed by adduction in pre-swing resulted in a continuous spread of the forefoot during stance phase (Fig. 10.14).

10.3.4 Case D

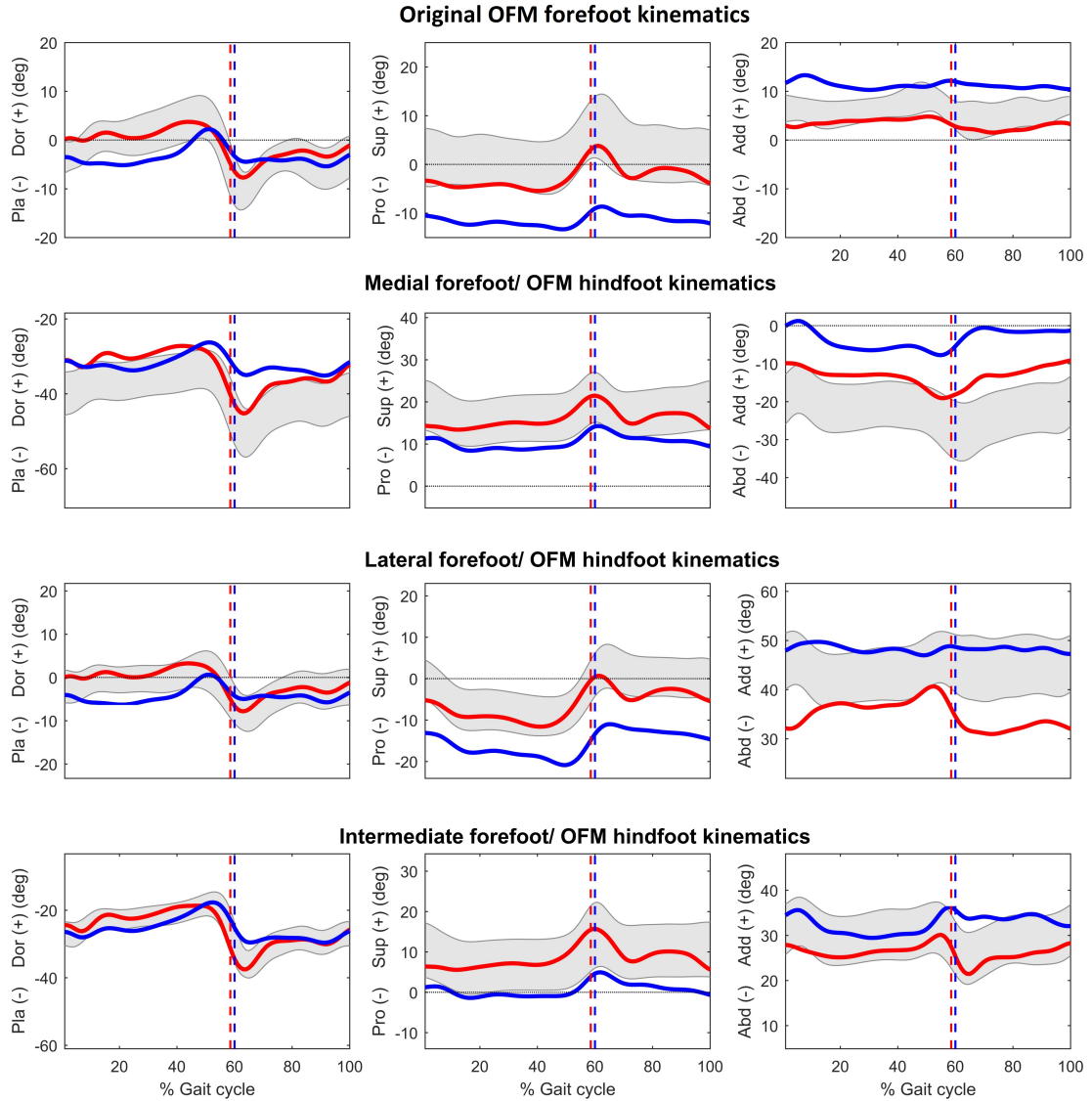


Figure 10.15: Inter-segmental kinematics of the OFM forefoot (top row), type C2 OFFM medial forefoot (second row), lateral forefoot (third row), and intermediate forefoot (bottom row) relative to the OFM hindfoot segment, for the subject D. Over the three columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right. The grey band represents corresponding the mean values of 45 healthy adults (except for the top row). The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phase.

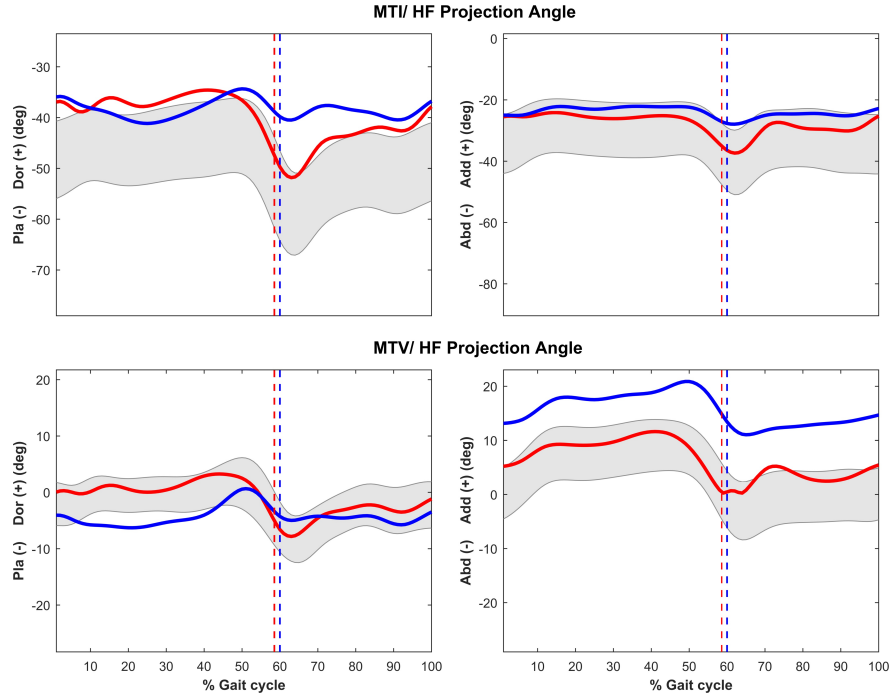


Figure 10.16: Planar angles of the first (top row) and fifth (bottom row) metatarsal relative to the OFM hindfoot segment, for the subject D. Over the two columns are the dorsi/plantar flexion and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults. The blue line is one full gait cycle for the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

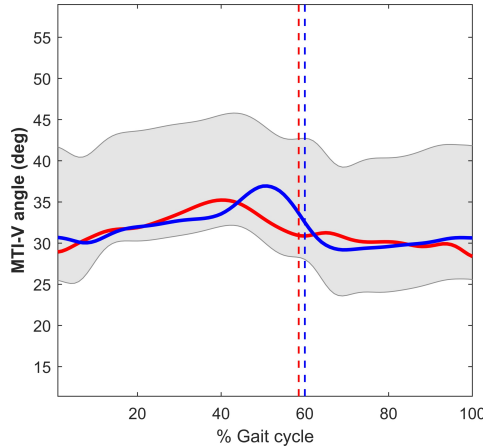


Figure 10.17: The angle between the first and fifth metatarsal vectors projected onto the OFM forefoot transverse plane, for the subject D. The grey band represents the mean values of 45 healthy adults. The blue line is one full gait cycle from the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

Subject D is an 11 years old girl with a torsional deformity. According to the OFM, the right forefoot had a supination/pronation offset of approximately -10° from the normal range (Fig. 10.15), which was also observed in the OFFM lateral forefoot. In the transverse plane, the right medial forefoot had an adduction offset of roughly 10° from the normal range, while the left lateral forefoot had an abduction offset bordering normal range. However, the intermediate forefoot appeared to be within the normal range, as with the OFM forefoot.

In this case, the largest variation between the forefoot models was the transverse plane. The motion of the first metatarsal, fifth metatarsal (Fig. 10.16) and the angle between them appeared to be within the normal range (Fig. 10.17), with the exception of the MTV angle, which appeared to be in an adducted position.

10.3.5 Case E

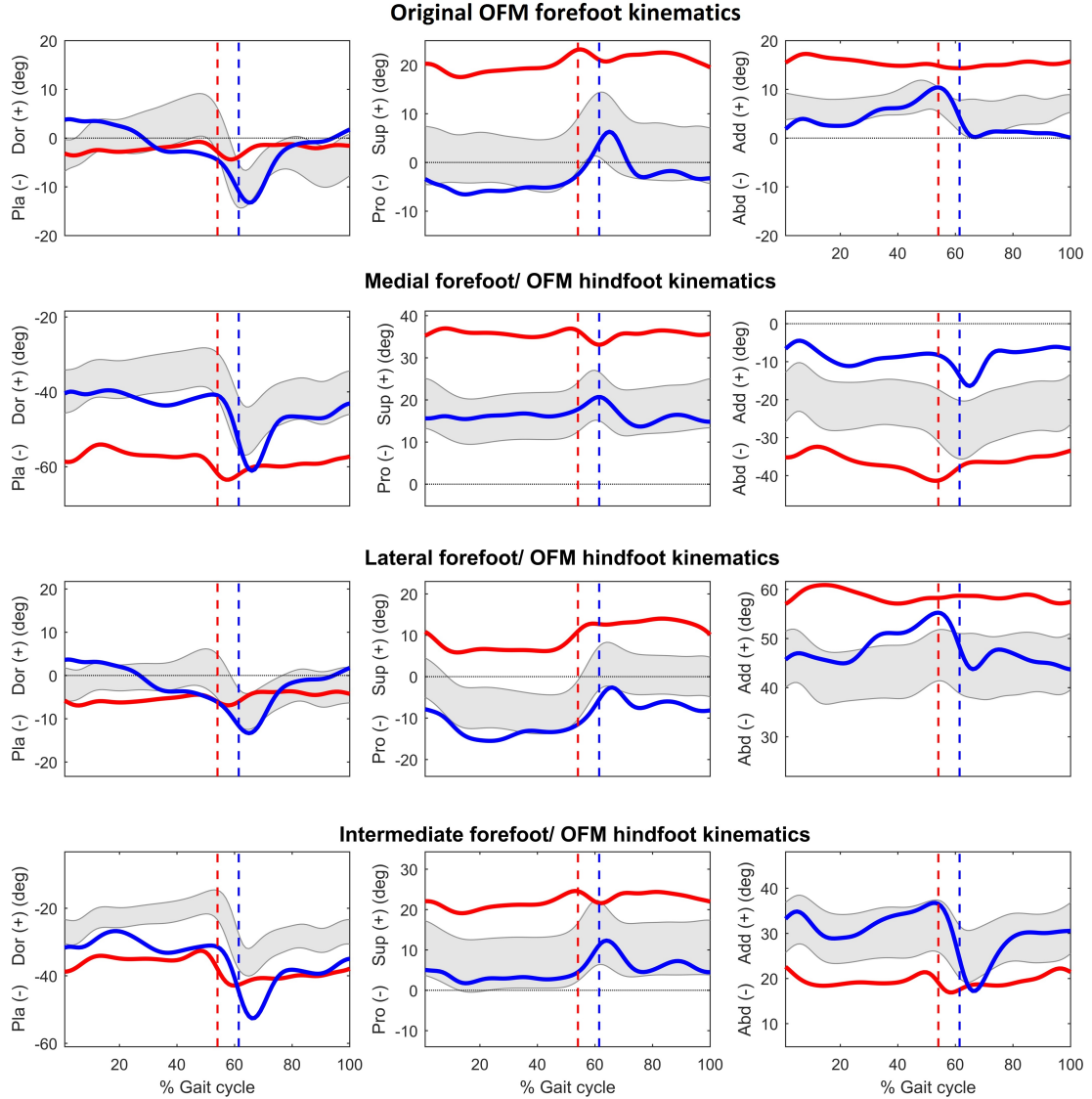


Figure 10.18: Inter-segmental kinematics of the OFM forefoot (top row), type C2 OFFM medial forefoot (second row), OFFM lateral forefoot (third row), and intermediate forefoot (bottom row) relative to the OFM hindfoot segment, for the subject E. Over the three columns are the dorsi/plantar flexion, pronation/supination, and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults (except for the top row). The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phase.

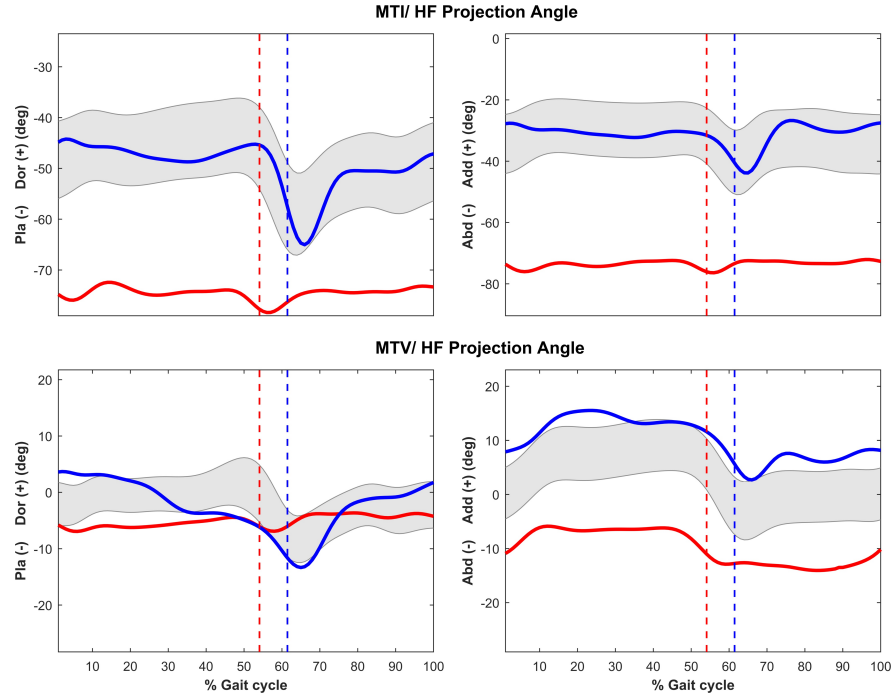


Figure 10.19: Planar angles of the first (top row) and fifth (bottom row) metatarsal relative to the OFM hindfoot segment, for the subject E. Over the two columns are the dorsi/plantar flexion and ab/adduction motions from left to right. The grey band represents the corresponding mean values of 45 healthy adults. The blue line is one full gait cycle for the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

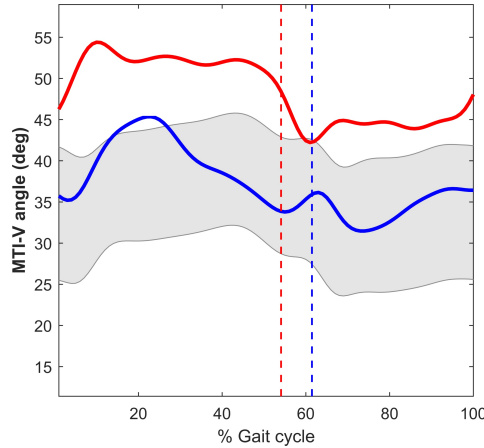


Figure 10.20: The angle between the first and fifth metatarsal vectors projected onto the OFM forefoot transverse plane, for the subject E. The grey band represents the mean values of 45 healthy adults. The blue line is one full gait cycle from the right foot, and red line is for the left foot. The vertical dotted line divides the stance and swing phase.

The final case study involved a 12 year old girl with left-sided clubfoot. Based on the result from the OFM, the right foot motion appeared to be within the normal range in all three planes, but the forefoot failed to dorsiflex throughout the stance phase. Due to foot deformity, a loss of range of motion was observed in all three planes and the foot was in a supinated and adducted pose (Figs. 10.18 & 10.19). Similar to Case A, the lateral forefoot showed a more consistent pattern with the OFM. The plantarflexed first ray was characterised by a minus 20° offset in the medial forefoot sagittal plane motion, which was consistent with the observation of the foot deformity. The medial forefoot also appeared to be in an abducted pose, as confirmed by the offset first metatarsal angle in Figure 10.19.

10.4 Discussion

To explore the potential clinical utility of the OFFM, case studies involving five patients have been carried out. The kinematic outputs produced by the OFFM can be used to further characterise the deformation of the forefoot during gait, as well as highlighting regional problems within the forefoot. Overall, the sagittal plane kinematics showed similar patterns of motion between the OFFM segments, as well as with the original OFM forefoot. The main differences occurred in the transverse plane, with the lateral forefoot displaying the most similarity to the OFM. This is not surprising since the OFM uses markers attached to the lateral forefoot for tracking.

The subjects involved in this study represented the typical patients referred for clinical gait analysis [12]. Although specific inter-segmental motions with gait deviation varied from patient to patient, a reduced sagittal motion of the hindfoot and forefoot were commonly observed [255], as with the first case study. Clubfoot is another disorder that has benefited from the use of CGA [227, 255, 256]. To date, there have been many clinical studies aimed at characterising pathological gait due

to foot deformity using MSFM [146]. However, none have tracked the movement of the medial and lateral forefoot independently. By dividing the forefoot into two segments, it is possible to further identify the location of the abnormality and allowed for a more detailed characterisation of foot deformity and functional deficits. In this study, two cases have demonstrated such potential. For case study A, the reduction of the sagittal plane motion was observed in both the OFM and the lateral forefoot, but not the medial forefoot. This information may lead to potential treatments that target correcting the motion of the lateral forefoot. In case study E, the plantarflexed first ray was identified from the sagittal motion offset of the medial forefoot, but not the original OFM. This information is consistent with the clinical observation (cavus foot), and can be used to provide quantitative assessment of the foot deformity.

Another important goal of this study was to evaluate the clinical practicality and feasibility of implementing this model in a clinical setting. According to the feedback from the members of the OGL team, anatomical landmarks for the marker placement were easily identifiable. Since only two additional markers were added to the original OFM protocol, there was minimal interference to the procedure. In addition, the full marker set can be tracked by the motion capture system without any problem. With the addition of two markers, this new forefoot model allows the calculation of three additional three-dimensional segments. The planar angles has been measured with the existing marker set, however, with a physical first metatarsal marker instead of a reconstructed virtually landmark, as the previous Chapter found that a virtually reconstructed D1M does not represent the movement of the distal first metatarsal head.

There are several limitations to this study. It should be noted that in this exploratory study, no statistical tests were carried out due to the insufficient sample size as well as the feasibility in carrying out a comprehensive clinical study within the limited time frame. Therefore, the kinematic comparisons made between differ-

ent forefoot segments in the result section involved visual analysis only. It is also acknowledged that comparing kinematic data that is matched for gender and age group is more appropriate [21, 165]. The small sample size and lack of comparison to a matching age group implies that these results require further investigation. Nevertheless, this exploratory study serves as an essential first step to identify possible clinical applications of this novel forefoot model.

10.5 Summary

This study compared the forefoot gait patterns of a selection of patients to those of healthy normal individuals. Differences in kinematic patterns were observed, and in some cases, have shown potential clinical utility of the novel forefoot model. By dividing the forefoot into a medial and a lateral segment, biomechanics of the forefoot can be characterised in greater detail, as well as providing quantitative assessment of foot deformities such as in Case E. Future studies should include a larger sample of patients, a more detailed analysis of both kinetics, muscle-activity, and plantar pressure in order to fully realise the clinical potential of this model.

This study has only investigated the forefoot motion using technical frames, that is, the coordinate system constructed using the tracking skin markers. Here, the technical frames of the OFFM forefoot segments were also assumed to be their anatomical frame. Different ways of improving the anatomical accuracy and clinical relevance of the model, including anatomical alignment, static pose angle off-set, design of a technical frame and calibration algorithm, should also be explored.

The next Chapter summarises the finding of this thesis and outlines its limitations, as well as recommending future research directions.

Chapter 11

Conclusions

To date, 39 multi-segment foot models (MSFMs) have been proposed [146], each with its own unique segment definition, marker-placement, kinematic convention, and reporting method. While there are many different segmentation strategies, there is currently very little agreement on how the forefoot is modelled in an MSFM [210]. This research project was undertaken to determine the most appropriate way of representing the forefoot in a MSFM for CGA and to propose and evaluate the clinical applicability such novel forefoot model. This Chapter summarises the findings of the research carried out, providing details of the key contributions, study limitations, and recommendations for future research in this field.

11.1 Key Contributions

11.1.1 Quantifying Segment Rigidity Using σ_{RBE}

While many literature sources [80, 244, 245, 264] have shown that residuals from a transformation matrix (hence, σ_{RBE}) can be used as a direct indication of the magnitude of deformation, there is currently no study looking at the precise relationship between residual and the magnitude of deformation. The research carried out in Chapter 5 investigated the effectiveness of using σ_{RBE} to quantify segment rigidity in an MSFM. This is the first time that a Monte-Carlo based global sensitivity analysis has been used to explore the relationships between several parameters that could potentially affect the output (deviation magnitude, rotational angle, segment size, and the geometrical configuration – condition number) of the σ_{RBE} .

The result of this study showed that the deformation magnitude accounts for 86.7% of the total variation in σ_{RBE} . A linear regression model suggested that σ_{RBE} is, in fact, an effective measure of rigidity. The use of σ_{RBE} to quantify rigidity has several practical applications, including exposing potential violation of the rigid body assumption and quantifying soft-tissue artefacts, with a possible clinical utility in quantifying joint flexibility during gait. Importantly, the result of this Chapter also validates the use of σ_{RBE} in Chapters 6 and 7.

11.1.2 Segment Rigidity and Forefoot Representation

The present review of the literature (Chapter 2) exposed the lack of consensus on forefoot segmentation among existing MSFMs. Several studies have suggested that the forefoot should not be considered as a single segment rigid body and recommended suitable metatarsal groupings based on their own criteria [76, 188, 219, 271]. By demonstrating that the forefoot is the least rigid of all the Oxford Foot Model (OFM)

segments, the result of Chapter 6 adds to a growing body of evidence which suggested the forefoot should not be treated as a single rigid segment. This is the first study that used σ_{RBE} to quantify the rigidity of a MSFM segment over the full gait cycle, and showed that it is possible to track the deformation of a segment using this approach.

Chapter 7 extended the application of σ_{RBE} to investigate forefoot subdivision by quantifying and comparing the rigidity of all combinations of metatarsal pairs. To date, most of the studies that recommended forefoot groupings utilised intra-cortical pins [186, 188, 271]. While the kinematics measured using bone pins is more anatomically accurate than the use of skin-attached markers, the metatarsal groupings suggested by these studies may not be translatable, or appropriate, to be adopted to the design of an MSFM for clinical use. Based on this concern, Chapter 7 utilised standard passive tracking markers attached to anatomical landmarks that are easily replicable, that is, the proximal bases and distal heads of all five metatarsals, to identify suitable metatarsal groupings. This is the first time that segment rigidity has been measured for all metatarsal combinations, which served as a base for the design of novel forefoot models in Chapters 8 and 9.

11.1.3 Novel Forefoot Modelling

Following the rigidity analysis, Chapters 8 and 9 investigated several ways of representing the forefoot based on the rigidity criteria recommended from Chapter 7. Inter-segmental kinematics and repeatability indexes were calculated for each of the proposed novel forefoot segments. This exploratory research is the largest study ($n = 45$) so far to simultaneously track the motion of multiple forefoot segments. We have successfully evaluated the kinematics of the medial forefoot, lateral forefoot, intermediate forefoot, and the individual first and fifth metatarsal segments, and compared their motion to a unified forefoot (as defined using the OFM forefoot segment). Compared to previous studies that utilised different motion tracking techniques, such as

intra-cortical pins [188] and EM-sensors [76], the present study utilising standard skin-attached markers offered a more practical approach.

On average, the range of motion of the OFFM segments was found to be larger than the OFM forefoot segment in all three planes. This suggests that by analysing the forefoot as a unified segment, the ranges of the medial and lateral segments are effectively being averaged out, thus masking unique movement profiles that could be relevant to the clinical interpretations. Segmenting the forefoot also revealed different motions that were not observed previously in the OFM, which could be used to characterise the biomechanics of the forefoot in greater detail.

Chapter 9 evaluated the efficacy of three different hybrid forefoot representations that can simultaneously track all the forefoot segments proposed in Chapter 8. This study showed that the motion of the first metatarsal is unique and therefore should not be reconstructed virtually from its static position. Furthermore, differences in the kinematics between different hybrid models were indiscernible, hence a model comprising of a medial forefoot segment, an intermediate forefoot segment, a lateral forefoot segment, and individual first and fifth metatarsal segments (OFFM C2 - See Fig. 9.2) was selected to be piloted on clinical cases in part for the fact that this marker-set is still compatible with the traditional OFM single forefoot model.

11.1.4 Clinical Case Studies

The final Chapter explored the potential clinical utility of the novel forefoot model proposed from the investigation of Chapter 9. Although the current study was based on a small number of patients, the findings confirmed the practical applicability of the novel model in the clinical context. Unique gait patterns have been identified that were previously not recognisable from the unified forefoot (here, represented using OFM forefoot), in particular, the difference in the motion between the medial and lateral forefoot segments.

One of the key advantages of the final proposed model is that it is backward compatible with the original validated OFM model, only requiring two additional markers placed on easily palpable landmarks without any interference with the tracking of the other markers. Many existing MSFMs are also capable of measuring the OFFM on top of their current marker-set protocol. Therefore, this initial work encourages further exploration of clinical utility using this novel forefoot model.

11.2 Limitations

11.2.1 Quantifying Segment Rigidity Using σ_{RBE}

In this thesis, a Monte-Carlo based GSA successfully showed that σ_{RBE} is appropriate to quantify rigidity. However, the study design and several assumptions made in the first place have limited the scope of this study. Firstly, a random deformation ($||[\delta D]||$) was applied to only one single marker of the cluster in order to generate a uniformly distributed deviation magnitude. Further work is required to fully understand the behaviour of σ_{RBE} when all markers are deformed simultaneously, which represents a more realistic (but mathematically more complicated) scenario. Moreover, the GSA was only performed on a four-marker cluster. Although adding more markers will only improve the accuracy of determining the transformation matrix [245], further analysis is required to fully understand whether the σ_{RBE} are comparable between segments with a different number of markers.

To show that the σ_{RBE} can be used to predict the deformation magnitude, a linear regression model was proposed in Chapter 5. It is acknowledged that the residuals were still negatively skewed after the (natural) log-log data transformation, which means that the prediction accuracy decreases as the σ_{RBE} increases. In our study, we have utilised a simple data-transformed linear regression model in order to demonstrate the correlation between the σ_{RBE} and $||[\delta D]||$. If the goal were to accurately

predict $\|[\delta D]\|$ with a given σ_{RBE} value, other methods such as weighted least-squares regression or regression with “heteroscedasticity-consistent standard error estimators” (regression model that estimates standard errors without assuming homoscedasticity) [106] are both viable solutions. However, using complex data-transformation and weighted standard error could limit the interpretability of the model.

11.2.2 Segment Rigidity and Forefoot Representation

Several other limitations of the research carried out have been recognised. In Chapters (6 and 7), σ_{RBE} has been used to quantify the rigidity of MSFM segments. However, it must be noted that both soft-tissue movement and the segment flexibility (i.e. relative bone movements) were captured simultaneously. Although the latter is likely to be more dominant, further analysis is required to confirm this. A study by Maslen & Ackland [169] on the soft-tissue artefact (STA) of the foot and ankle using radiography reported the largest STA displacement of 3 mm at the fifth metatarsal, which can be translated to a σ_{RBE} value of 0.83 mm (the worst-case scenario) according to the regression equation (Eq. 5.1). Although this is unlikely to affect the result of Chapter 6, in Chapter 7 the differences between the rigidity of the metatarsal pairs were much smaller and demand more caution on the interpretation of results. Future analysis, including the simultaneous measurement of segment deformation using intra-cortical pins and skin marker data may help distinguish STA from the bone movement [164].

Another major limitation that reduces the generalisability of the result from Chapter 6 is that this study only examined the segment rigidity as quantified by the tracking markers of the OFM. This means that testing of other marker configurations would be required to extend the application to other MSFMs. This limitation is also applicable to Chapter 7. The markers on the first and fifth metatarsals were placed according to the OFM protocol, on the medial and lateral side of the forefoot. This choice gave us the greatest confidence that the markers were located on the correct

bones. However, it has been suggested by the Oxford Gait Laboratory team that the marker placement for the first and fifth metatarsal should be moved to the dorsal aspect of the foot to achieve consistency with the marker placement of the three middle metatarsals. Future studies should assess the impact of using this recommended marker placement.

11.2.3 Novel Forefoot Modelling

The OFFM forefoot segments were in part designed to be compatible to the description of the widely-used MSFMs in order to promote adaptation into the existing models. These are, the Oxford Foot Model [42], Milwaukee Foot Model [135], Rizzoli Foot Model [143], and Kinfoot [161]. However, it must be realised that there are other viable options to track and represent the forefoot than the marker-placement protocol outlined in Chapter 8 and 9. This may include the use of flexible [187] or rigid marker-clusters [125, 194], and other modelling strategies.

Several studies recognised the need to define and track an independent midfoot segment [65, 125, 143]. The current study has only examined the inter-segmental motion of the forefoot segments with reference to the OFM hindfoot, leaving the motion with other segments such as the tibia, hallux, and the potentially more clinically relevant mid-foot unexplored.

Finally, the kinematic output of a MSFM depends on multiple factors. The current research only explored the kinematics of the technical frame, also assumed to be the anatomical frame, of the model by defining the segment coordinate system using the physical location of the tracking markers. Defining an anatomical frame along with an appropriate calibration strategy could potentially improve the repeatability and clinical relevance of the kinematic angles [38].

11.2.4 Clinical Case Studies

The final Chapter (10) explored the potential clinical utility of the novel forefoot model proposed from the investigation of Chapter 9. However, the findings from this research were limited by the number of patients involved, and therefore caution must be applied in the interpretation of the results. Future clinical studies involving more patients are required to understand the clinical utility of the proposed forefoot model. In addition, the comparison of joint rotations was made between children against adult's normal data. The inherent difference between the gait of children and adult could further limit the interpretation of this study.

11.3 Future Directions

This thesis has made several noteworthy contributions to suggest how the forefoot should be modelled in a MSFM based on a rigidity criteria, along with the proposal of a clinically relevant forefoot model. However, the novel forefoot model proposed in the final Chapter still requires further validation and clinical pilot studies in order to justify the incorporation of this model into routine CGA. The first step would be to include more patients with a wider range of foot conditions in order to further explore the clinical utility of the forefoot model. Testing of different segment definitions, calibration techniques, anatomical alignment methods, and the ways in which joint rotation motion is reported would provide valuable information that contributes to the development of a better forefoot model.

In addition to addressing the limitations discussed in all the chapters, a number of studies can be undertaken to provide further insight into the understanding of forefoot biomechanics in general. The present study mainly dealt with the kinematic modelling of the forefoot, which represents a significant part of any MSFM. However, kinetics measurement using MSFMs have provided valuable information regarding forefoot

biomechanics, in particular, the importance of the power generated at midfoot during propulsion [77, 161, 72]. Therefore, future research should also explore the impact of forefoot segmentation on the kinetic outputs of a MSFM.

Several possible forefoot segmentation and modelling strategies were tested through the exploratory kinematics analysis in Chapters 8 and 9. Model C2 were selected for clinical case studies in Chapter 10 due to its practical and kinematic advantages. However, future studies should explore different marker-placement and segment definition strategies. The selection of a “optimal” MSFM could be made quantifiable. For example, through an expert panel to assess the quality of a model in terms of practicality, simplicity, repeatability, compatibility to the existing model, and other indicators.

Finally, it is acknowledged that the design of a MSFM should be focused on gathering information relevant to the specific pathology under interest and its treatment. Although it is an attractive proposition to develop a generic model that can be used to in both the healthy and deformed foot, it is unlikely that one model is suitable for all applications. Having a collection of different models designed specifically to target different foot conditions may lead to more personalised treatment.

Appendix A

Data Collection Documentations

A.1 Subject Measurement Sheet

Investigator	
Date	
Start time	
End time	

	Measurement	Notes
Subject ID		
Subject Name	Anonymised	
Subject Code (trial)		
Mass (kg)		<i>obtain from force plate</i>
Height (mm)		
Gender (M/F/prefer not to say/others)		
Age (years and months)		
Inter-ASIS (mm)		<i>Measure with digital caliper</i>
Leg Length (mm)	(L) (R)	ASIS-FE-MMA
Knee Width (mm)	(L) (R)	
Ankle Width (mm)	(L) (R)	ANK-MMA
Foot Length (mm)	(L) (R)	
Shoe Size		
Dominant Leg		
Notes		
<i>(Specific condition or healthy normal volunteer with no known previous injuries; Specific time of data collection, dominant leg)</i>		

Appendix B

Additional Kinematic Graphs

This Appendix contains additional kinematic graphs generated for all the clinical case studies involved in this research.

Appendices B.1 to B.2 contains the OFM kinematic outputs for the two patients involved in the study of Chapter 6.

Appendices B.3 to B.7 contains the OFM kinematic outputs for the five patients involved in the study of Chapter 10.

B.1 Chapter 6 Pathological Gait - Patient 1

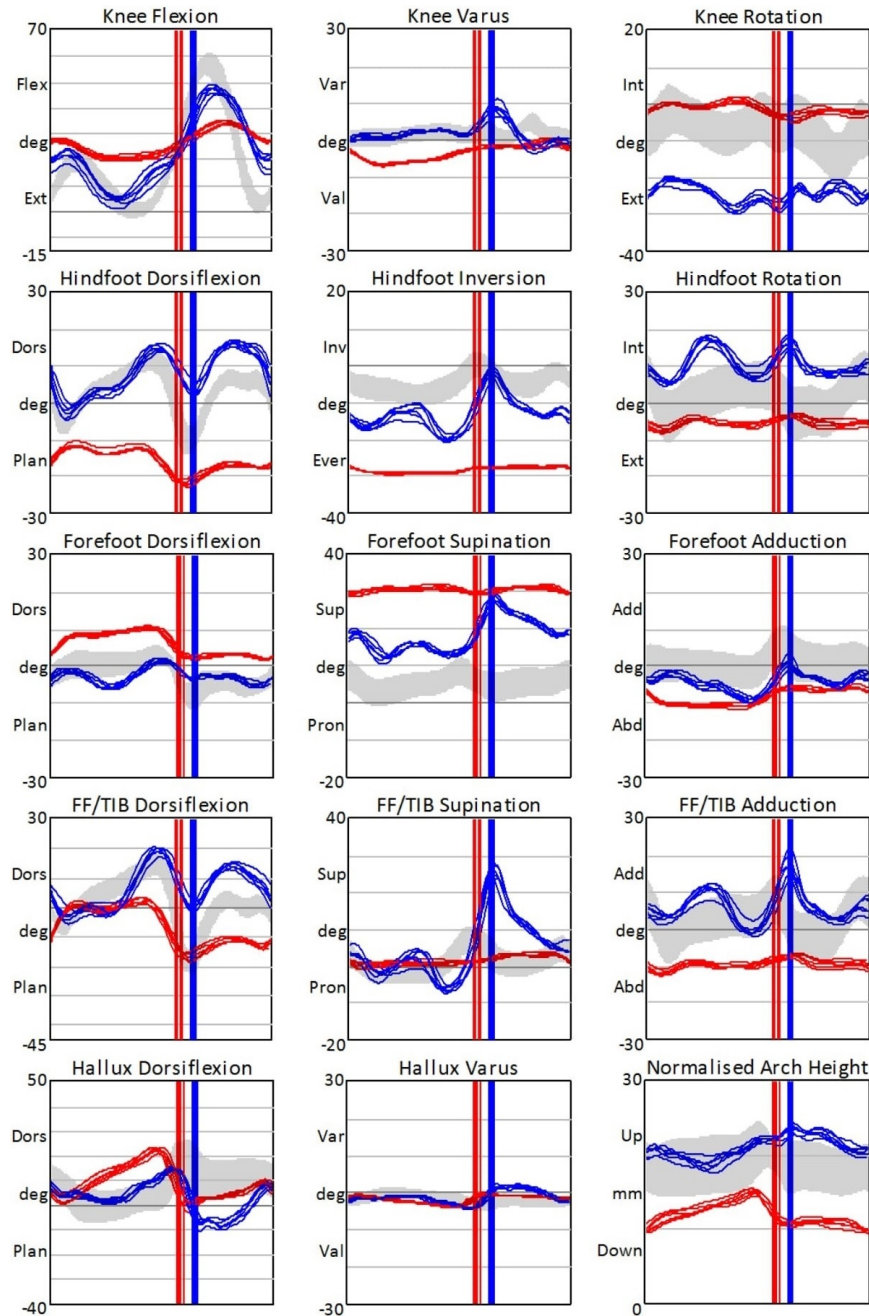


Figure B.1: Kinematic information for patient 1, who had a negative windlass mechanism (arch not re-gained upon toe-raise) on his left foot. Joint angles (6 walking trials) calculated using the OFM. Red line indicates left foot and blue right foot, and the vertical line indicates the separation of stance and swing phase. Grey band indicates the normal values.

B.2 Chapter 6 Pathological Gait - Patient 2

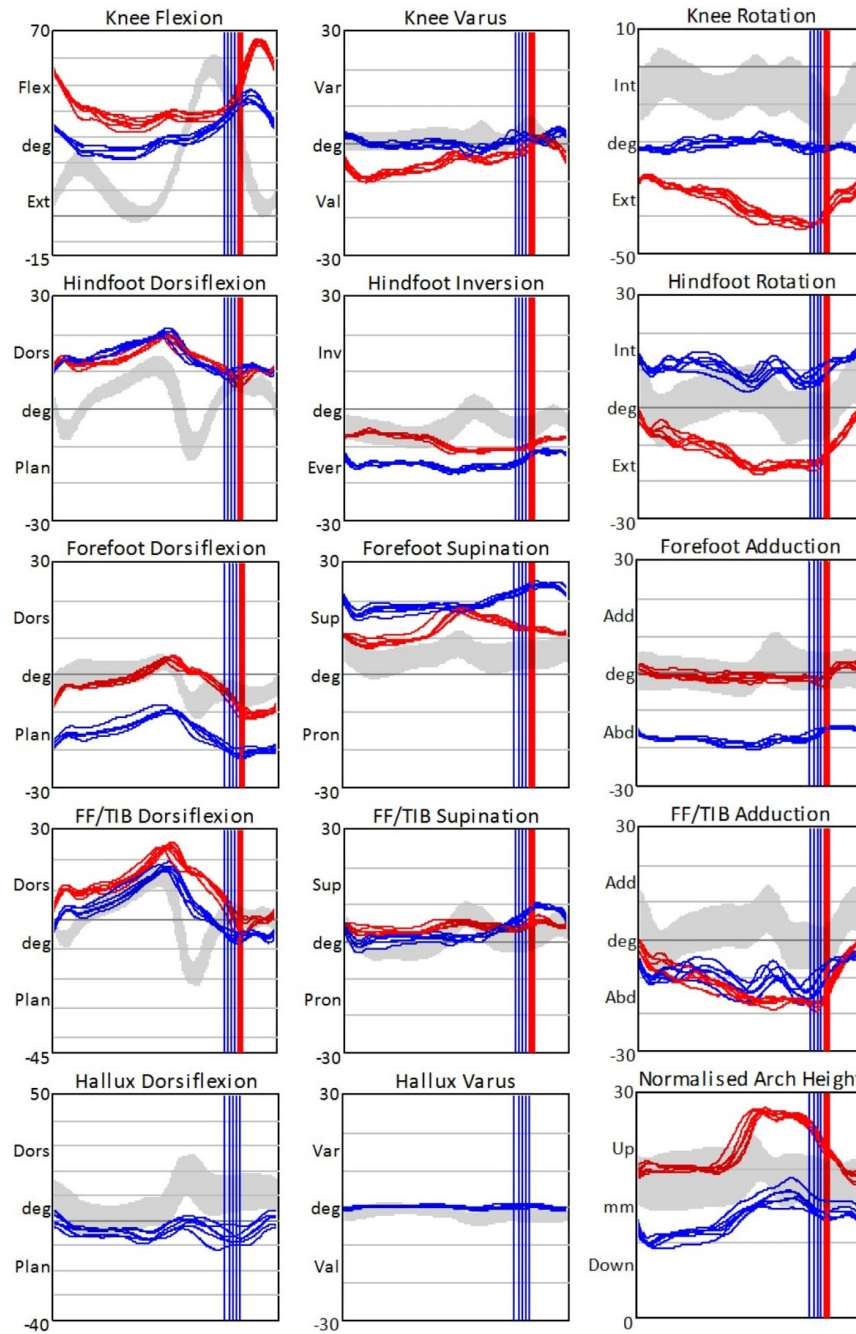


Figure B.2: Kinematic information for patient 2, who had a bilateral fixed ankle equinus and a midfoot break bilaterally. Joint angles (6 walking trials) calculated using the OFM. Red line indicates left foot and blue right foot, and the vertical line indicates the separation of stance and swing phase. Grey band indicates the normal values.

B.3 Chapter 10 Clinical Case Study - Subject A

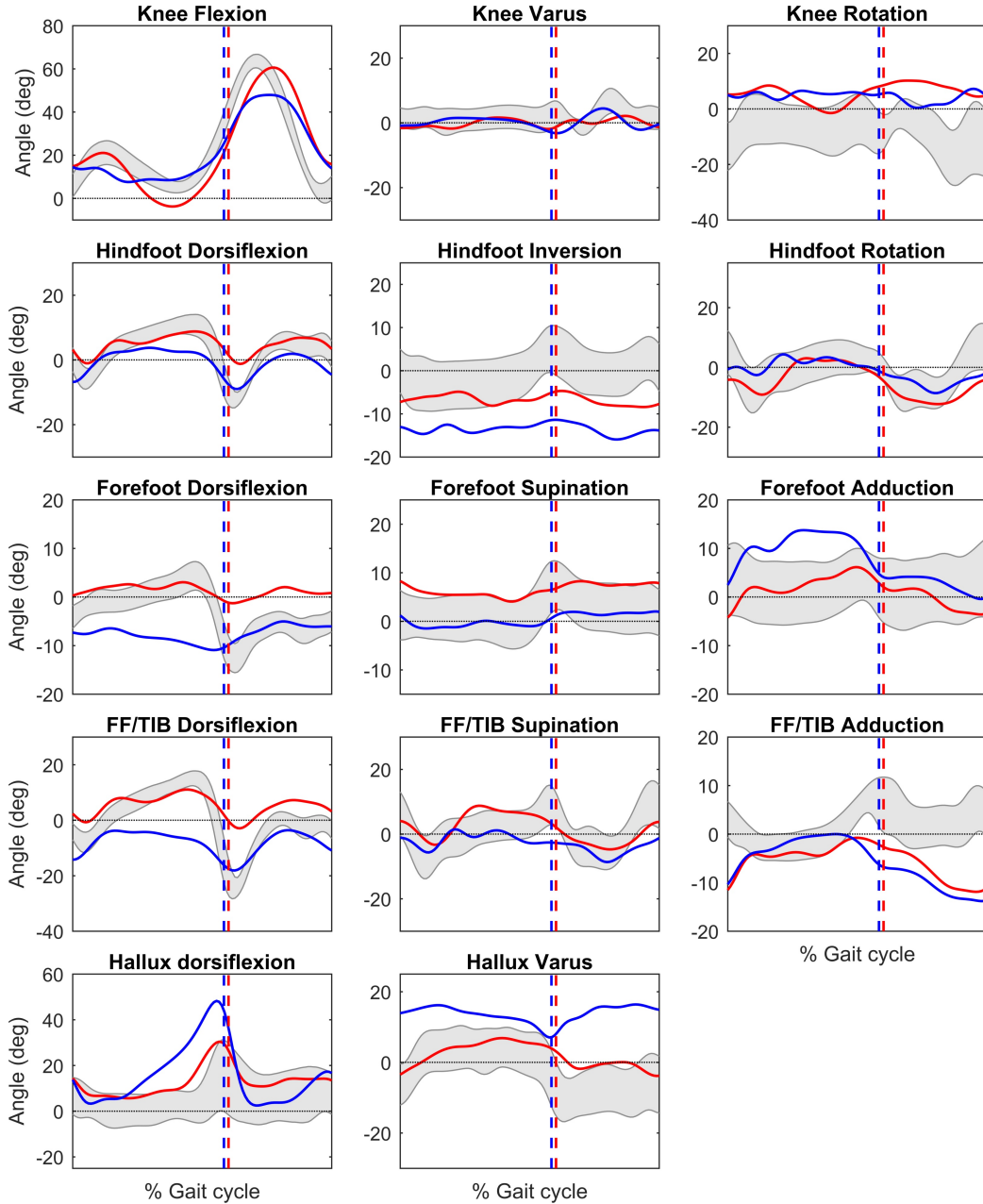


Figure B.3: Oxford foot model kinematic output for subject A, a 13 year old boy diagnosed with cerebral palsy – spastic diplegia, demonstrating the difference between normal and pathological gait. The grey band represents mean values of 11 healthy normal boys between the age of 13 – 16. The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phases.

B.4 Chapter 10 Clinical Case Study - Subject B

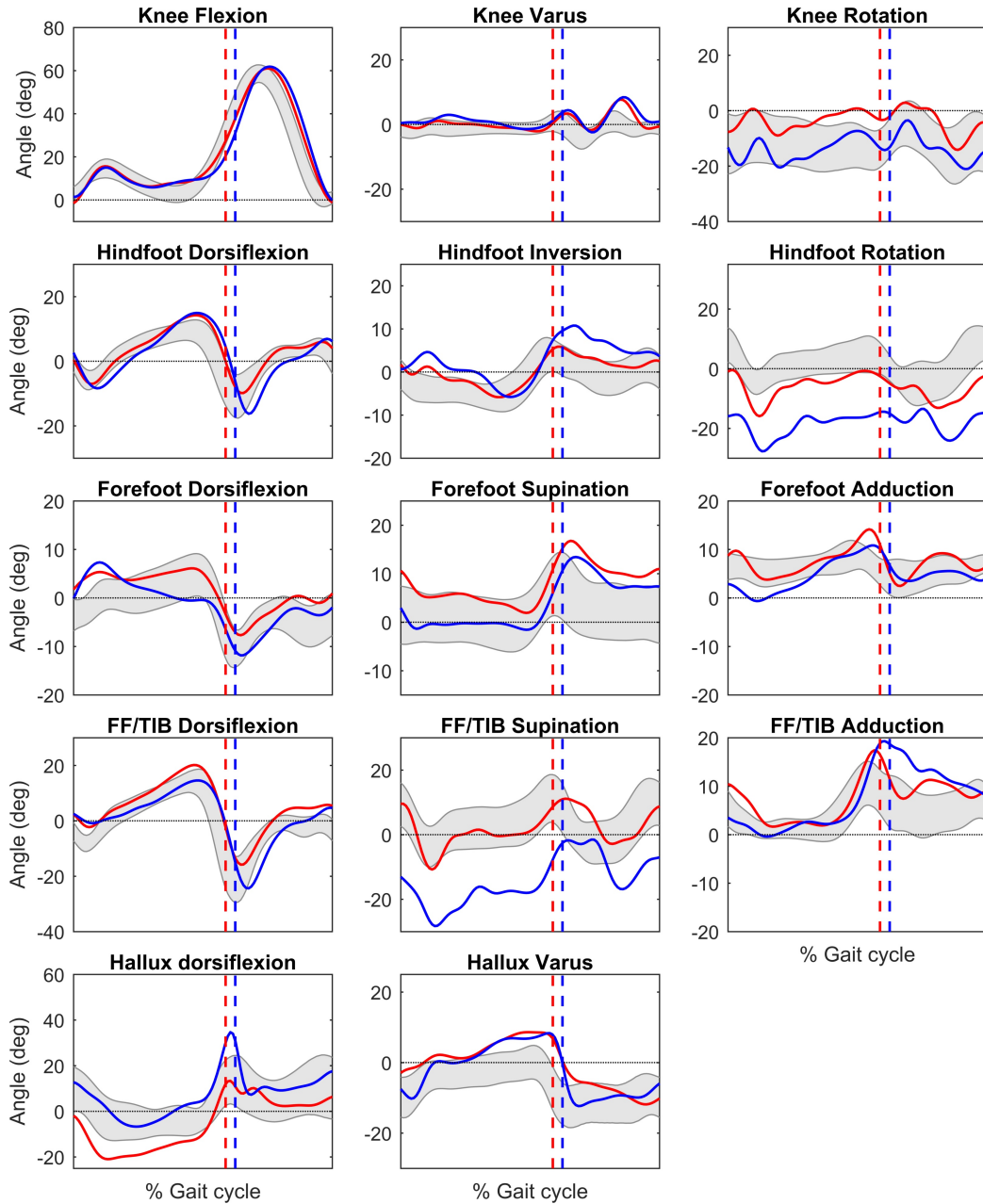


Figure B.4: Oxford foot model kinematic output for subject B, a 11 year old girl who was diagnosed with rotational malalignment and hip pain, demonstrating the difference between normal and pathological gait. The grey band represents mean values of 11 healthy normal girls between the age of 10 – 12. The blue line is one full gait cycle for the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phases.

B.5 Chapter 10 Clinical Case Study - Subject C

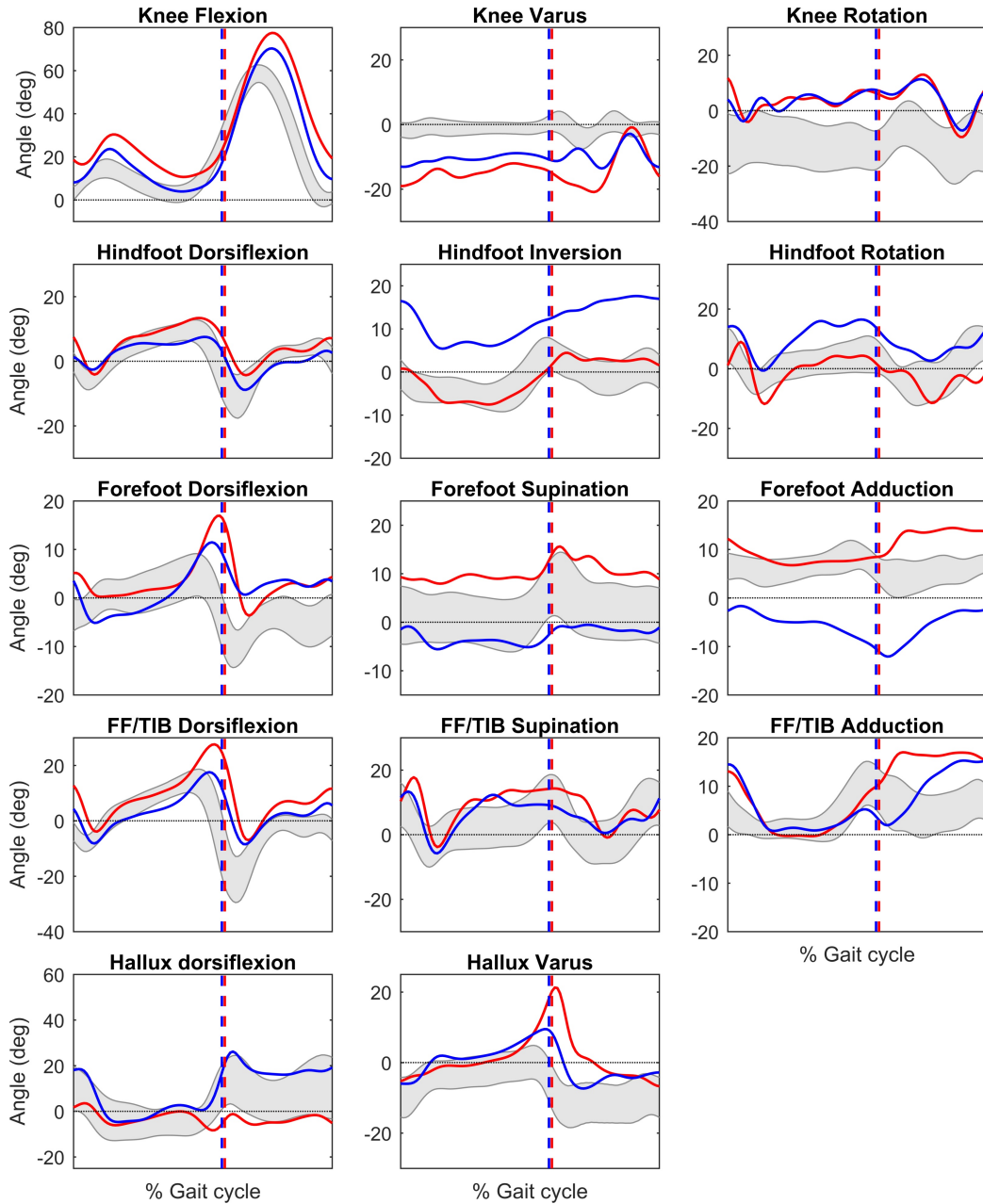


Figure B.5: Oxford foot model kinematic output for subject C, a 10 year old girl who was diagnosed with bilateral clubfoot, demonstrating the difference between normal and pathological gait. The grey band represents mean values of 11 healthy normal girls between the age of 10 – 12. The blue line is one full gait cycle from the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phases.

B.6 Chapter 10 Clinical Case Study - Subject D

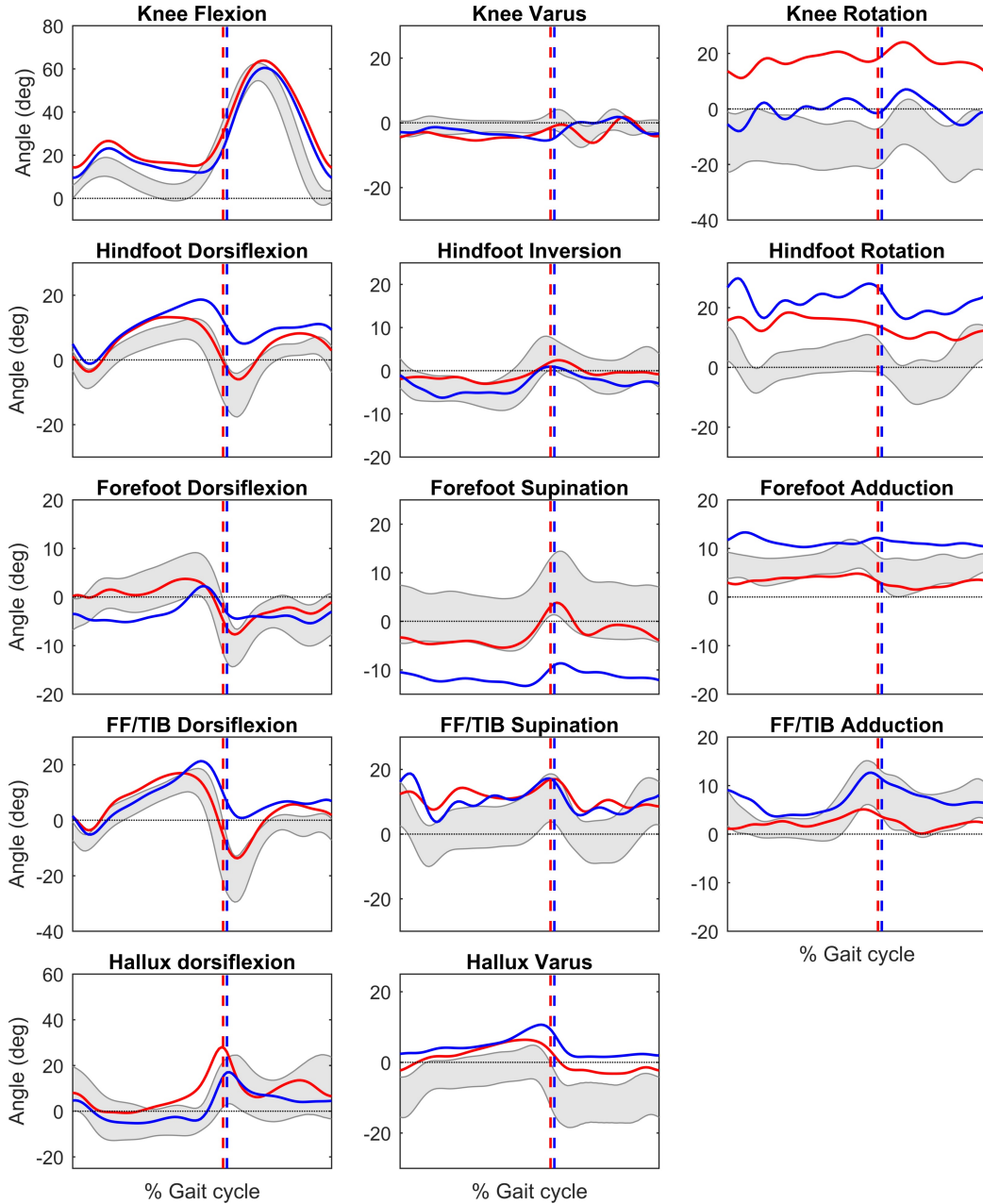


Figure B.6: Oxford foot model kinematic output for subject D, a 10 year old girl who was diagnosed with torsional deformity, demonstrating the difference between normal and pathological gait. The grey band represents mean values of 11 healthy normal girls between the age of 10 – 12. The blue line is one full gait cycle from the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phases.

B.7 Chapter 10 Clinical Case Study - Subject E

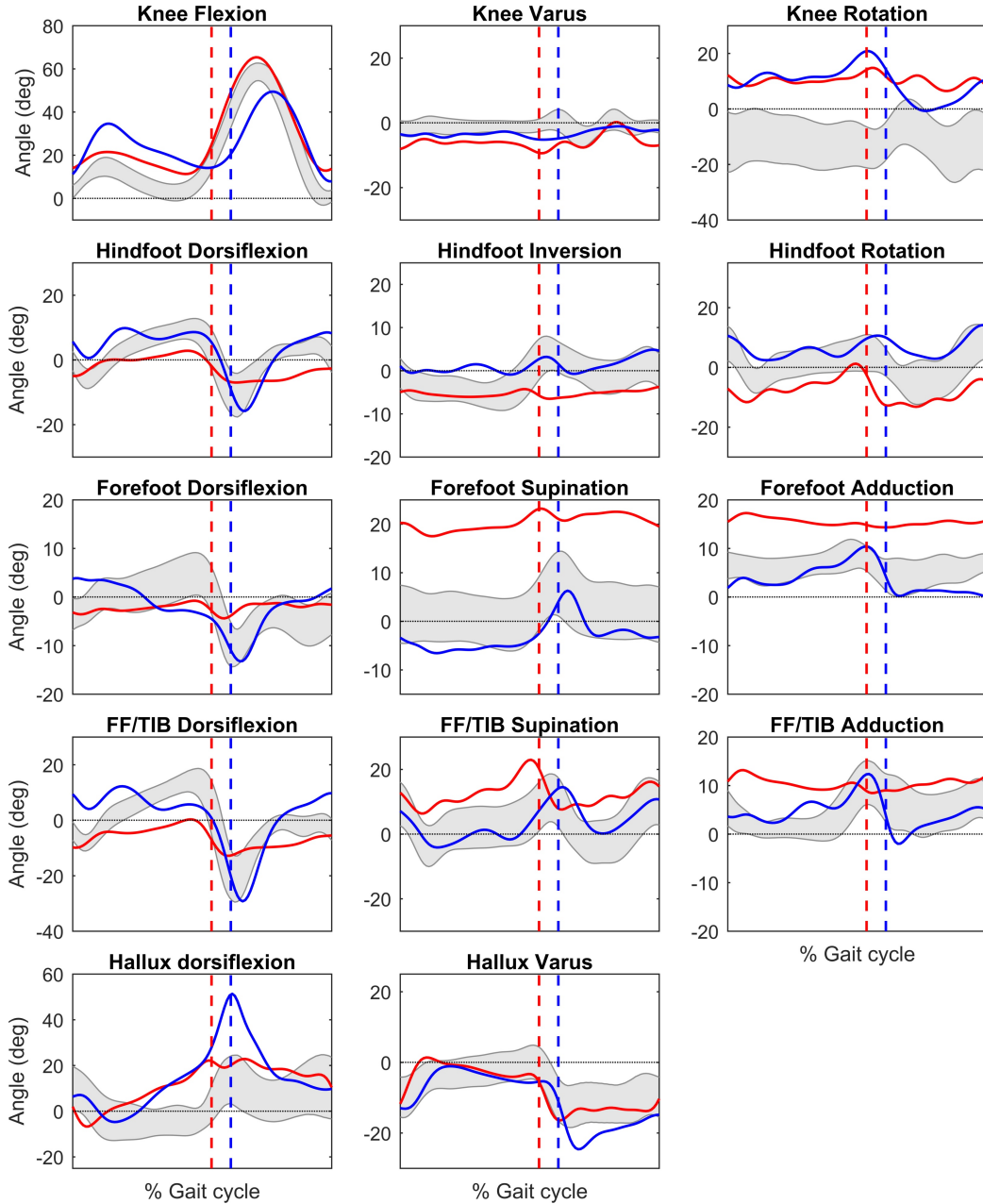


Figure B.7: Oxford foot model kinematic output for subject E, a 12 year old girl who was diagnosed with clubfoot (left side), demonstrating the difference between normal and pathological gait. The grey band represents mean values of 11 healthy normal girls between the age of 10 – 12. The blue line is one full gait cycle from the right foot, and the red line is for the left foot. The vertical dotted line divides the stance and swing phase.

Appendix C

Derivations

C.1 Orthogonal Procrustes Problem

The solution to problem outlined in Section 4.8 is presented here. The problem can be stated as follows:

N landmarks are initially positioned at $x_i \in \mathbb{R}^n$ (here, $n = 3$), and are moved to position $y_i \in \mathbb{R}^n$ for $i = 1, \dots, N$. For a rigid body motion, the transformation can be expressed as:

$$y = Rx + d \quad (\text{C.1})$$

for some translation vector $d \in \mathbb{R}^n$ and some orthogonal rotation matrix $R \in \mathbb{R}^n$ (hence $A^T A = I$). Since R is comprised of only rotations - no reflections, $\det A = 1$. Due to some measurement errors or deformation, the mapping is not exact, formulating the following problem [99]:

$$\min \sum_{i=1}^n ||Rx_i + d - y_i||^2 \quad (\text{C.2})$$

Problem (Eq. C.2) can be solved by using the method of singular value decomposition (SVD) described by Hanson and Norris [105]. We first define the mean position of the landmarks as:

$$\bar{x} = \frac{1}{n} \sum_{i=1}^n x_i \quad \text{and} \quad \bar{y} = \frac{1}{n} \sum_{i=1}^n y_i \quad (\text{C.3})$$

The modified positions of the initial and final landmark position with reference to its mean can be stated as:

C.1. ORTHOGONAL PROCRUSTES PROBLEM

$$[A] = [x_1 - \bar{x}, \dots, x_n - \bar{x}] \quad \text{and} \quad [B] = [y_1 - \bar{y}, \dots, y_n - \bar{y}] \quad (\text{C.4})$$

Now, consider the SVD of the matrix of the matrix $C = \sum_{i=1}^n AB^T$, which can be written as $C = USV^T$ for $n \times n$ orthogonal matrices U and V , and diagonal matrix of singular values $\sigma_1 \geq \dots \geq \sigma_n \geq 0$. The solution to (Eq. C.2) is:

$$R = UWV^T \quad \text{and} \quad d = \bar{y} - R\bar{x} \quad (\text{C.5})$$

where $W = \text{diag}(1, 1, \dots, \det U^T V)$. For the full proof, see [105].

C.2 Random Rotation Matrix

The simplest and most intuitive method of generating a random rotation matrix that covers all possible angles is by extracting three random Euler angles α , β , and γ each from a uniform distribution, followed by the computation of the rotation matrix through multiplying the three elemental rotation matrices. However, this method will not result in a rotation matrix that is uniformly distributed, as shown in Figure C.1 a).

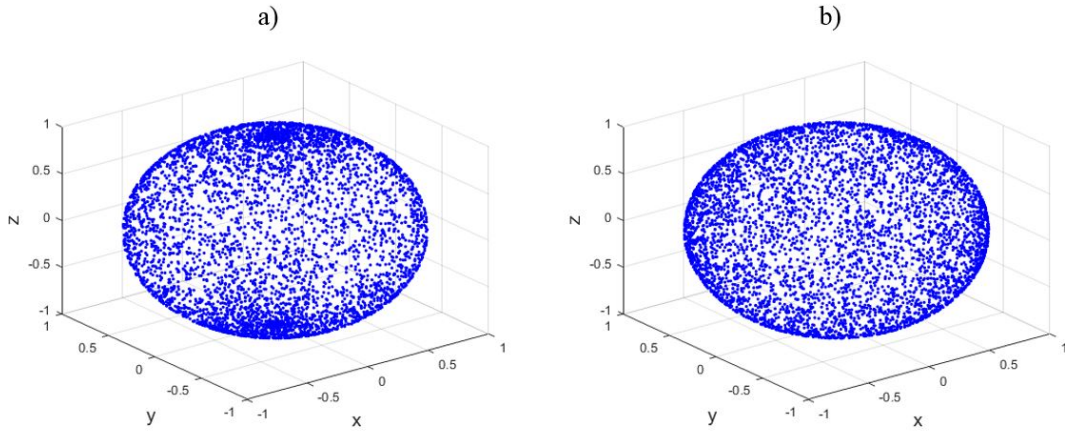


Figure C.1: Random rotation matrix distribution using a) sequential elemental rotation matrices and b) solution proposed by [8]

In Figure C.1 a), a vector $k = (1, 0, 0)$ is rotated 5000 times using a rotation matrix (Euler Z-Y-X convention) generated by the sequential elemental rotation matrices, where individual Euler angles are sampled from a uniform distribution over $[0, 2\pi]$. A higher density of points around the poles (where the z -coordinate takes values closer to either 1 or -1) can be observed, demonstrating non-uniformity of the rotation being generated. This effect can be seen more clearly when we limit the range of rotation angle as shown in Figure C.2. For the vector $k = (1, 0, 0)$, the latitude position of the rotated point depends on the Z angle whereas the position along the longitude depends on the Y angle. For Y angles near the poles, the points tend to be more

C.2. RANDOM ROTATION MATRIX

concentrated as the circumference of the latitude becomes smaller.

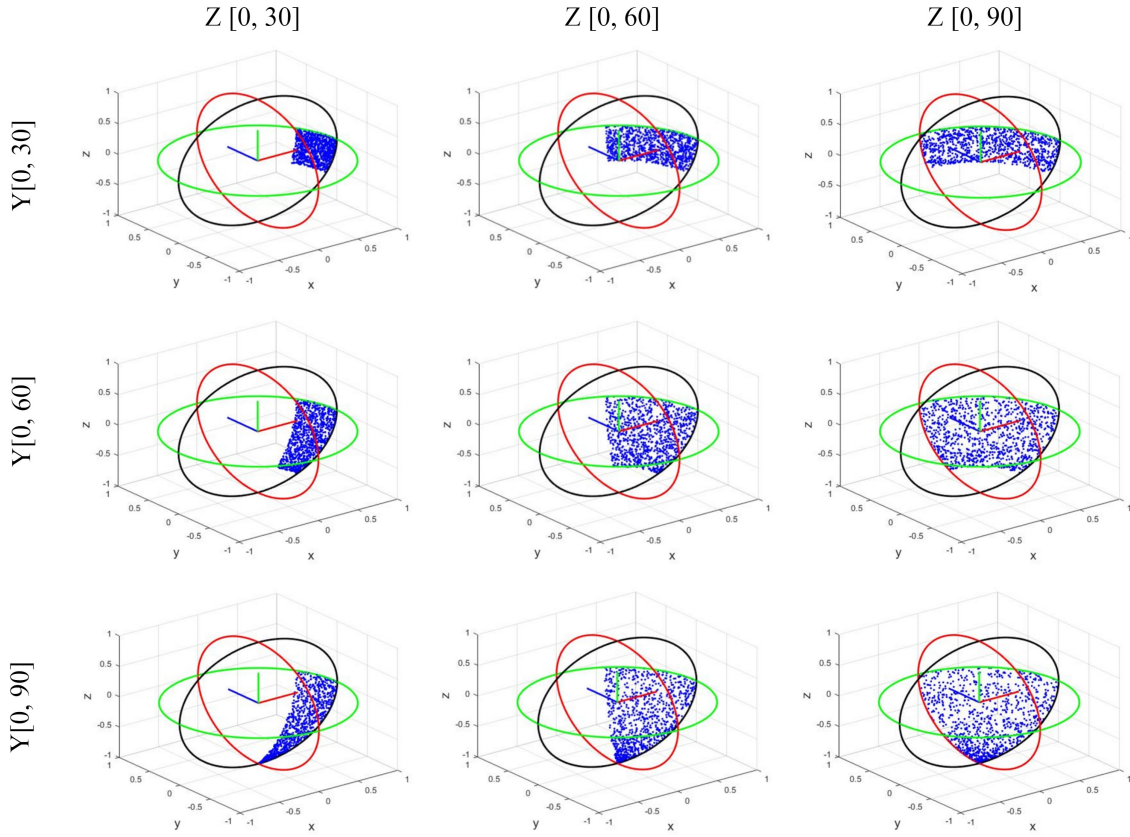


Figure C.2: Random rotation matrix distribution using sequential elemental rotation matrices in Z-Y-X Euler convention, where both Z and Y angles are generated based on uniform distribution. From left to right column: Increasing Z angle range; From top to bottom: Increasing Y angle range.

Instead of generating rotation by three consecutive random angles, Arvo [8] utilises a random reflection after the first rotation to produce a genuinely random rotation (Fig. C.2 b). The first step involves rotation about the z-axis:

$$R = \begin{bmatrix} \cos(\alpha_N) & \sin(\alpha_N) & 0 \\ -\sin(\alpha_N) & \cos(\alpha_N) & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad (\text{C.6})$$

The second step involves a calculation of a Householder reflection (i.e. a linear trans-

C.2. RANDOM ROTATION MATRIX

formation that describes a reflection about a plane containing the origin):

$$H = I - 2vv^T \quad (\text{C.7})$$

The reflection is able to move point k to any coordinates p on the sphere via a reflection through the hyperplane that is orthogonal to the line $\overline{zp}$ and containing the origin, where v is a unit vector parallel to the line $\overline{zp}$. The vector v is defined as:

$$v = \begin{bmatrix} \cos(\beta_N)\sqrt{x_N} \\ \sin(\beta_N)\sqrt{x_N} \\ \sqrt{1-x_N} \end{bmatrix} \quad (\text{C.8})$$

where $\beta_N = [0, 2\pi]$ and $x_N = [0, 1]$. To show that this is the solution to the problem of generating a uniformly distribution of points across the surface of the sphere, if we take the initial point to be $k = (1, 0, 0)$ the rotated point can be calculated as $p = Hz$:

$$p = (1 - 2vv^T)k = \begin{bmatrix} -2\cos(\beta_N)\sqrt{x_N(1-x_N)} \\ -2\sin(\beta_N)\sqrt{x_N(1-x_N)} \\ 2x_N - 1 \end{bmatrix} \quad (\text{C.9})$$

The z-component of p is uniformly distributed over $[-1, 1]$, and the x and y -component of p is uniformly distributed over $[0, 2\pi]$, as required. In summary, Arvo's random rotation matrix can be generated using three separate uniformly generated random inputs, two angles (α_B, β_N) and one constant (x_N) to describe the orientation of the randomly generated hyperplane that reflects the point anywhere on the sphere.

Appendix D

Additional Analysis and Assessments

D.1 Convergence Assessment for Sensitivity Analysis

Monte-Carlo based sensitivity analysis (SA) is a numerical procedure that estimates the descriptive statistics and the sensitivity parameters. Naturally, the accuracy and stability of the estimates depend on the sample size or the stopping rule imposed on each Monte-Carlo iteration, and there are various methods to achieve this. Similar to how a convergence assessment is usually conducted in a finite-element analysis to determine mesh-size [20], the assessment can be conducted iteratively within a refine loop that increases the sample size until the output stabilises. Alternatively, the rate of convergence (i.e. sample size) can be pre-determined using a confidence interval [141, 142]. In this analysis, the latter method was chosen to fix the number of samples to be generated for the primary analysis (Chapter 5) and any further analysis for consistency. This assessment was carried out in MATLAB (v2017a, MathWorks Inc., Natick, USA).

The confidence interval was determined using the method proposed by Ata [9], which is composed of a convergence band-length and band-width that acts as an exit condition for the loop that determines the accuracy of the steady-state mean. This is also termed the acceptable shifting convergence band rule (ASCBR) and can be easily embedded within the Monte-Carlo algorithm as a stopping rule for the while loop.

In ASCBR, the convergence of a parameter is detected when its mean stabilises

D.1. CONVERGENCE ASSESSMENT FOR SENSITIVITY ANALYSIS

within a defined convergence band (CB) with $(2\varepsilon^*)$ over a number of samples (CD length ξ^*) generated by the Monte-Carlo algorithm. At the start of the code, the algorithm requires the definition of the convergence band half-width (ε^*) and length (ξ^*). Let X be a random variable (e.g. $|\delta B|_i$) with the distribution $\sigma_{RBE}(X)$ and x_i the random sample from $F(X)$. This algorithm begins by calculating the mean of the Monte-Carlo samples consisting all the observations from the 1st to the j^{th} in the current trial:

$$\bar{x}_j = \frac{1}{j} \sum_{i=1}^j x_i \quad (\text{D.1})$$

For each loop, the algorithm assesses whether adding one extra sample x_j will result in a significant change in the mean such that it falls outside the current CB width, where the upper bound $U(\bar{x}_{j-1})$ and lower bound $L(\bar{x}_{j-1})$ are defined as:

$$U(\bar{x}_{j-1}) = \bar{x}_{j-1} + \varepsilon^* \quad \text{and} \quad L(\bar{x}_{j-1}) = \bar{x}_{j-1} - \varepsilon^* \quad (\text{D.2})$$

If the new mean ($\bar{x}_j$) falls outside this interval, the CB width will be redefined, otherwise the band remains the same. The condition for the new CB band can be described as follows:

$$U(\bar{x}_j) = \begin{cases} U(\bar{x}_{j-1}), & \delta_j = 0 \\ \bar{x}_j + \varepsilon^*, & \delta_j = 1 \end{cases} \quad (\text{D.3})$$

and

$$L(\bar{x}_j) = \begin{cases} L(\bar{x}_{j-1}), & \delta_j = 0 \\ \bar{x}_j - \varepsilon^*, & \delta_j = 1 \end{cases} \quad (\text{D.4})$$

where

$$\delta_j = \begin{cases} 0, & L(\bar{x}_{j-1}) < \bar{x}_j < U(\bar{x}_{j-1}) \\ 1, & \{\bar{x}_j \leq L(\bar{x}_j)\} \vee \{\bar{x}_j \geq U(\bar{x}_j)\} \end{cases} \quad (\text{D.5})$$

D.1. CONVERGENCE ASSESSMENT FOR SENSITIVITY ANALYSIS

The number of adjacent sample means, which share the same upper and lower band limits as defined in Eqs. D.3 and D.4, i.e. $\delta_j = 0$ for all of them, is recorded as:

$$Z_j = \begin{cases} Z_j + 1, & \delta_j = 0 \\ 1, & \{\bar{x}_j \leq L(\bar{x}_j)\} \vee \{\bar{x}_j \geq U(\bar{x}_j)\} \end{cases} \quad (\text{D.6})$$

for every loop. Once Z_j reaches the required CB length (ξ^*), the algorithm terminates and concludes that convergence is reached. Figure D.1 shows an example of a section of the ASCBR coding embedded within the Monte-Carlo simulation conducted for the Chapter 5 SA. When the mean falls outside the CB width, the algorithm redefines a new CB. The loop will only end when the mean stays within the CB over a predefined length.

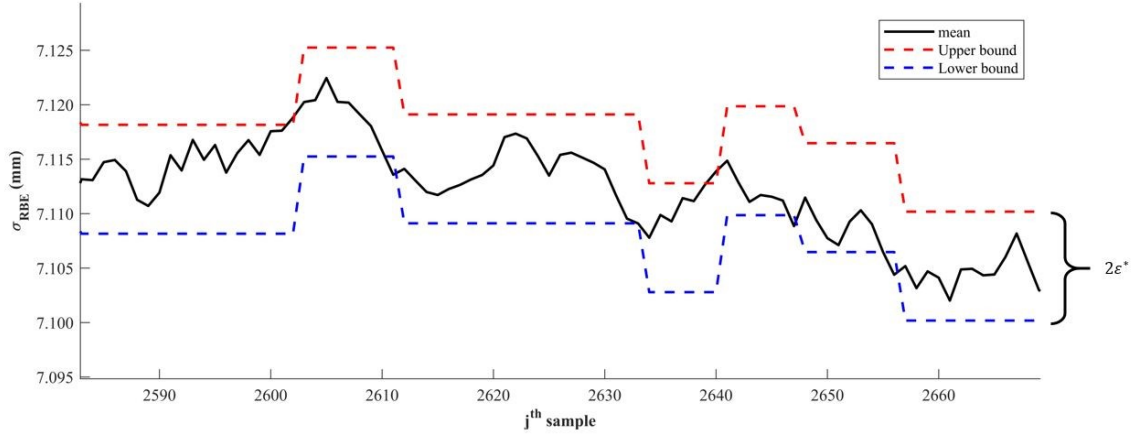


Figure D.1: Example of a shifting convergence band

In this assessment, the CB width (half the resolution) was defined according to the resolution specified in Table D.1. In this assessment, the CB length (ξ^*) is set to 50 iterations, as suggested in the literature as the optimal length [9].

The required resolution and the number of samples required to meet the target for each parameter are listed in Table D.2. The result of this assessment suggests that the sample size for the Monte-Carlo based global sensitivity analysis (Chapter

D.1. CONVERGENCE ASSESSMENT FOR SENSITIVITY ANALYSIS

5) should be a minimum of 3197.

Notation	Type	Description	Unit	Resolution	CB half width (ε^*)
σ_{RBE}	Output	Rigid body error	mm	0.01	0.005
S	Output	Segment size	mm	0.1	0.05
kA	Output	Condition number	1/mm	0.0001	0.00005
R_{err}	Output	Rotation error	-	0.001	0.0005
$ \delta D $	Input	Deformation magnitude	mm	0.1	0.05
β_N	Input	Input for rotation matrix	Rad	0.01	0.005
γ_N	Input	Input for rotation matrix	Rad	0.01	0.005
x_N	input	Input for rotation matrix	-	0.01	0.005

Table D.1: Parameters tested in the Monte-Carlo convergence assessment and CB half width

Notation	Unit	Required Resolution	Converged mean (standard deviation)	Convergence sample size
σ_{RBE}	mm	0.01	7.32 (4.42)	2747
S	mm	0.1	203.9 (39.2)	3197
kA	1/mm	0.0001	0.0047 (0.0016)	191
R_{err}	-	0.01	0.096 (0.069)	468
$ \delta D $	mm	0.1	25.1 (14.4)	1349
β_N	Rad	0.01	3.14 (1.81)	370
γ_N	Rad	0.01	3.17 (1.81)	302
x_N	-	0.01	0.51 (0.29)	350

Table D.2: Converged size obtained by the ASCBR algorithm

Bibliography

- [1] Human Anatomy: The foot. Retrieved from <http://humananatomybody.info/>.
- [2] R. J. Abboud. Relevant foot biomechanics. *Current Orthopaedics*, 16:165–179, 2002.
- [3] M. Akbarshahi, A. G. Schache, J. W. Fernandez, R. Baker, S. Banks, and M. G. Pandy. Non-invasive assessment of soft-tissue artifact and its effect on knee joint kinematics during functional activity. *Journal of Biomechanics*, 43(7):1292–1301, 2010.
- [4] M. S. Andersen, M. Damsgaard, J. Rasmussen, D. K. Ramsey, and D. L. Benoit. A linear soft tissue artefact model for human movement analysis: Proof of concept using in vivo data. *Gait & Posture*, 35(4):606–611, 2012.
- [5] T. P. Andriacchi, E. J. Alexander, M. K. Toney, C. Dyrby, and J. Sum. A point cluster method for in vivo motion analysis: Applied to a study of knee kinematics. *Journal of Biomechanical Engineering*, 120(6):743–749, dec 1998.
- [6] A. Arndt, P. Wolf, A. Liu, C. Nester, A. Stacoff, R. Jones, P. Lundgren, and A. Lundberg. Intrinsic foot kinematics measured in vivo during the stance phase of slow running. *Journal of Biomechanics*, 40(12):2672–2678, 2007.
- [7] J. B. Arnold, S. Mackintosh, S. Jones, and D. Thewlis. Repeatability of stance phase kinematics from a multi-segment foot model in people aged 50 years and older. *Gait & Posture*, 38(2):349–351, 2013.
- [8] J. Arvo. Fast Random Rotation Matrices. In *Graphics Gems III*, pages 117 – 120. Academic Press, 2nd edition, 1992.
- [9] M. Y. Ata. A convergence criterion for the Monte Carlo estimates. *Simulation Modelling Practice and Theory*, 15(3):237–246, 2007.
- [10] R. Baker. The history of gait analysis before the advent of modern computers. *Gait & Posture*, 26(3):331–342, 2007.
- [11] R. Baker. *Measuring Walking: A Handbook of Clinical Gait Analysis*. Mac Keith Press, 2013.

- [12] R. Baker, A. Esquenazi, M. G. Benedetti, and K. Desloovere. Gait analysis: clinical facts. *European Journal of Physical and Rehabilitation Medicine*, 52(4):560–74, 2016.
- [13] R. Baker, L. Finney, and J. Orr. A new approach to determine the hip rotation profile from clinical gait analysis data. *Human Movement Science*, 18(5):655–667, 1999.
- [14] R. Baker, F. Leboeuf, J. Reay, and M. Sangeux. The conventional gait model - success and limitations. In *Handbook of Human Motion*, pages 1–19. Springer International Publishing, 2017.
- [15] M. E. R. Balsdon and C. E. Dombroski. Reliability of a multi-segment foot model in a neutral cushioning shoe during treadmill walking. *Journal of Foot and Ankle Research*, 11(1):60, 2018.
- [16] C. H. Barnett and J. R. Napier. The axis of rotation at the ankle joint in man; its influence upon the form of the talus and the mobility of the fibula. *Journal of Anatomy*, 86(1):1–9, 1952.
- [17] A. Barre and S. Armand. Biomechanical ToolKit: Open-source framework to visualize and process biomechanical data. *Computer Methods and Programs in Biomedicine*, 114(1):80–87, 2014.
- [18] J. W. Bartlett and C. Frost. Reliability, repeatability and reproducibility: Analysis of measurement errors in continuous variables. *Ultrasound in Obstetrics and Gynecology*, 31(4):466–475, 2008.
- [19] M. S. Bartlett. The Use of Transformations. *Biometrics*, 3(1):39, 1947.
- [20] K. Bathe. *Finite Element Procedures*. Prentice Hall, Pearson Education, Inc, 2nd edition, 2014.
- [21] M. G. Benedetti, E. Beghi, A. De Tanti, A. Cappozzo, N. Basaglia, A. G. Cutti, A. Cereatti, R. Stagni, F. Verдини, M. Manca, S. Fantozzi, C. Mazzà, V. Camomilla, I. Campanini, A. Castagna, L. Cavazzuti, M. Del Maestro, U. D. Croce, M. Gasperi, T. Leo, P. Marchi, M. Petrarca, L. Piccinini, M. Rabuffetti, A. Ravaschio, Z. Sawacha, F. Spolaor, L. Tesio, G. Vannozzi, I. Visintin, and M. Ferrarin. SIAMOC position paper on gait analysis in clinical practice: General requirements, methods and appropriateness. Results of an Italian consensus conference. *Gait & Posture*, 58:252–260, 2017.
- [22] D. L. Benoit, M. Damsgaard, and M. S. Andersen. Surface marker cluster translation, rotation, scaling and deformation: Their contribution to soft tissue artefact and impact on knee joint kinematics. *Journal of Biomechanics*, 48(10):2124–2129, 2015.
- [23] C. Bishop, G. Paul, and D. Thewlis. Recommendations for the reporting of foot and ankle models. *Journal of Biomechanics*, 45(13):2185–2194, 2012.

- [24] C. B. Blackwood, T. J. Yuen, B. J. Sangeorzan, and W. R. Ledoux. The midtarsal joint locking mechanism. *Foot & Ankle International*, 26(12):1074–1080, 2005.
- [25] Y. Blanc, C. Balmer, T. Landis, and F. Vingerhoets. Temporal parameters and patterns of the foot roll over during walking: normative data for healthy adults. *Gait & Posture*, 10(2):97–108, 1999.
- [26] L. Blankevoort, R. Huiskes, and A. Delange. The Envelope of Passive Knee-Joint Motion. *Journal of Biomechanics*, 21(9):705–720, 1988.
- [27] C. L. Brockett and G. J. Chapman. Biomechanics of the ankle. *Orthopaedics and Trauma*, 30(3):232–238, 2016.
- [28] J. W. Brodsky, D. a. Charlick, S. C. Coleman, F. E. Pollo, and C. T. Royer. Hindfoot motion following reconstruction for posterior tibial tendon dysfunction. *Foot & Ankle International*, 30(7):613–618, 2009.
- [29] D. A. Bruening, K. M. Cooney, and F. L. Buczek. Analysis of a kinetic multi-segment foot model. Part I: Model repeatability and kinematic validity. *Gait & Posture*, 35(4):529–534, 2012.
- [30] D. C. Montgomery, E. A. Peck, and G. G. Vining. *Introduction to Linear Regression Analysis*. John Wiley & Sons, Ltd, 5th edition, 2012.
- [31] R. Caldas, M. Mundt, W. Potthast, F. Buarque de Lima Neto, and B. Markert. A systematic review of gait analysis methods based on inertial sensors and adaptive algorithms. *Gait & Posture*, 57:204–210, 2017.
- [32] V. Camomilla, R. Dumas, and A. Cappozzo. Human movement analysis: the soft tissue artefact issue. *Journal of Biomechanics*, 2017.
- [33] K. Canseco, J. Long, R. Marks, M. Khazzam, and G. Harris. Quantitative motion analysis in patients with hallux rigidus before and after cheilectomy. *Journal of Orthopaedic Research*, 27(1):128–134, 2009.
- [34] K. Canseco, L. Rankine, J. Long, T. Smedberg, R. M. Marks, and G. F. Harris. Motion of the multisegmental foot in hallux valgus. *Foot & Ankle International*, 31(2):146–152, 2010.
- [35] A. Cappello, A. Cappozzo, P. F. La Palombara, L. Lucchetti, and A. Leardini. Multiple anatomical landmark calibration for optimal bone pose estimation. *Human Movement Science*, 16(2-3):259–274, 1997.
- [36] A. Cappello, P. F. La Palombara, and A. Leardini. Optimization and smoothing techniques in movement analysis. *International Journal of Bio-Medical Computing*, 41(3):137–151, 1996.

- [37] A. Cappozzo. Gait analysis methodology. *Human Movement Science*, 3(1-2):27–50, 1984.
- [38] A. Cappozzo, F. Catani, U. D. Croce, and A. Leardini. Position and orientation in space of bones during movement: anatomical frame definition and determination. *Clinical Biomechanics*, 10(4):171–178, 1995.
- [39] P. Caravaggi, M. G. Benedetti, L. Berti, and A. Leardini. Repeatability of a multi-segment foot protocol in adult subjects. *Gait & Posture*, 33(1):133–135, 2011.
- [40] P. Caravaggi, A. B. Matias, U. T. Taddei, M. Ortolani, A. Leardini, and I. C. Sacco. Reliability of medial-longitudinal-arch measures for skin-markers based kinematic analysis. *Journal of Biomechanics*, 88:180–185, may 2019.
- [41] P. Caravaggi, T. Pataky, J. Y. Goulermas, R. Savage, and R. Crompton. A dynamic model of the windlass mechanism of the foot: evidence for early stance phase preloading of the plantar aponeurosis. *Journal of Experimental Biology*, 212(15):2491–2499, aug 2009.
- [42] M. Carson, M. Harrington, N. Thompson, J. O’Connor, and T. Theologis. Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis. *Journal of Biomechanics*, 34(10):1299–1307, 2001.
- [43] S. L. Carter, N. Sato, and L. S. Hopper. Kinematic repeatability of a multi-segment foot model for dance. *Sports Biomechanics*, 17(1):48–66, 2018.
- [44] A. Cereatti, T. Bonci, M. Akbarshahi, K. Aminian, A. Barré, M. Begon, D. L. Benoit, C. Charbonnier, F. Dal Maso, S. Fantozzi, C.-C. Lin, T.-W. Lu, M. G. Pandy, R. Stagni, A. J. van den Bogert, and V. Camomilla. Standardization proposal of soft tissue artefact description for data sharing in human motion measurements. *Journal of Biomechanics*, 62:5–13, 2017.
- [45] E. Ceseracciu, Z. Sawacha, and C. Cobelli. Comparison of markerless and marker-based motion capture technologies through simultaneous data collection during gait: Proof of concept. *PLoS ONE*, 9(3):1–7, 2014.
- [46] P.-H. Chan, J. Stebbins, and A. B. Zavatsky. Marker cluster rigidity in a multi-segment foot model. *Journal of Biomechanics*, 84:284–289, 2019.
- [47] F. M. Chang, A. J. Seidl, K. Muthusamy, A. K. Meininger, and J. J. Carollo. Effectiveness of instrumented gait analysis in children with cerebral palsy - Comparison of outcomes. *Journal of Pediatric Orthopaedics*, 26(5):612–616, 2006.
- [48] R. Chang, P. A. Rodrigues, R. E. A. Van Emmerik, and J. Hamill. Multi-segment foot kinematics and ground reaction forces during gait of individuals with plantar fasciitis. *Journal of Biomechanics*, 47(11):2571–2577, 2014.

- [49] E. Y. Chao. Justification of triaxial goniometer for the measurement of joint rotation. *Journal of Biomechanics*, 13(12):989–1006, 1980.
- [50] L. Cheze, B. J. Fregly, and J. Dimnet. A solidification procedure to facilitate kinematic analyses based on video system data. *Journal of Biomechanics*, 28(7):879–884, 1995.
- [51] K. Chia and M. Sangeux. Quantifying sources of variability in gait analysis. *Gait & Posture*, 56:68–75, 2017.
- [52] L. Chiari, U. D. Croce, A. Leardini, and A. Cappozzo. Human movement analysis using stereophotogrammetry Part 2: Instrumental errors. *Gait & Posture*, 21(2):197–211, 2005.
- [53] L. Chiari, U. D. Croce, A. Leardini, A. Cappozzo, L. Chiari, U. D. Croce, A. Cappozzo, U. Della Croce, A. Leardini, L. Chiari, and A. Cappozzo. Human movement analysis using stereophotogrammetry Part 1: theoretical background. *Gait & Posture*, 21(2):226–237, 2005.
- [54] J. Cohen. *Statistical power analysis for the behavioral sciences*. Lawrence Erlbaum Associates, New York, NY, 2nd edition, 1988.
- [55] G. K. Cole, B. M. Nigg, J. L. Ronsky, and M. R. Yeadon. Application of the joint coordinate system to three-dimensional joint attitude and movement representation: A standardization proposal. *Journal of Biomechanical Engineering*, 115(4A):344–349, 1993.
- [56] D. Cook and S. Weisberg. *Residuals and Influence in Regression*. Chapman and Hall, 1982.
- [57] R. E. Cook, I. Schneider, M. E. Hazlewood, S. J. Hillman, and J. E. Robb. Gait analysis alters decision-making in cerebral palsy. *Journal of Pediatric Orthopaedics*, 23(3):292–295, 2003.
- [58] M. W. Cornwall and T. G. McPoil. Three-dimensional movement of the foot during the stance phase of walking. *Journal of the American Podiatric Medical Association*, 89(2):56–66, 1999.
- [59] M. W. Cornwall and T. G. McPoil. Motion of the calcaneus, navicular, and first metatarsal during the stance phase of walking. *Journal of the American Podiatric Medical Association*, 92(2):67–76, 2002.
- [60] N. J. Cronin, T. Rantalainen, J. P. Ahtiainen, E. Hynynen, and B. Waller. Markerless 2D kinematic analysis of underwater running: A deep learning approach. *Journal of Biomechanics*, 87:75–82, 2019.
- [61] D. J. Curtis, J. Bencke, J. A. Stebbins, and B. Stansfield. Intra-rater repeatability of the Oxford foot model in healthy children in different stages of the foot roll over process during gait. *Gait & Posture*, 30(1):118–121, 2009.

- [62] R. B. Davis. Reflections on clinical gait analysis. *Journal of Electromyography and Kinesiology*, 7(4):251–257, 1997.
- [63] R. B. Davis, S. Öunpuu, D. Tyburski, and J. R. Gage. A gait analysis data collection and reduction technique. *Human Movement Science*, 10(5):575–587, 1991.
- [64] A. de Lange, R. Huiskes, and J. M. G. Kauer. Measurement errors in roentgen-stereophotogrammetric joint-motion analysis. *Journal of Biomechanics*, 23(3):259–269, 1990.
- [65] S. De Mits, V. Segers, J. Woodburn, D. Elewaut, D. De Clercq, and P. Roosen. A clinically applicable six-segmented foot model. *Journal of Orthopaedic Research*, 30(4):655–661, 2012.
- [66] U. Della Croce, A. Cappozzo, and D. C. Kerrigan. Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. *Medical & Biological Engineering & Computing*, 37(2):155–161, 1999.
- [67] U. Della Croce, A. Leardini, L. Chiari, and A. Cappozzo. Human movement analysis using stereophotogrammetry Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics. *Gait & Posture*, 21(2):226–237, 2005.
- [68] P. A. DeLuca, R. B. Davis, S. Ounpuu, S. Rose, and R. Sirkin. Alterations in surgical decision making in patients with cerebral palsy based on three-dimensional gait analysis. *Journal of Pediatric Orthopedics*, 17(5):608–14, 1997.
- [69] K. Deschamps, I. Birch, K. Desloovere, and G. A. Matricali. The impact of hallux valgus on foot kinematics: A cross-sectional, comparative study. *Gait & Posture*, 32(1):102–106, 2010.
- [70] K. Deschamps, F. Staes, H. Bruyninckx, E. Busschots, E. Jaspers, A. Atre, and K. Desloovere. Repeatability in the assessment of multi-segment foot kinematics. *Gait & Posture*, 35(2):255–260, 2012.
- [71] K. Deschamps, F. Staes, H. Bruyninckx, E. Busschots, G. A. Matricali, P. Spaepen, C. Meyer, and K. Desloovere. Repeatability of a 3D multi-segment foot model protocol in presence of foot deformities. *Gait & Posture*, 36(3):635–638, 2012.
- [72] K. Deschamps, F. Staes, K. Peerlinck, C. Van Geet, C. Hermans, G. A. Matricali, and S. Lobet. 3D Multi-segment foot kinematics in children: A developmental study in typically developing boys. *Gait & Posture*, 52:40–44, 2017.

- [73] K. Deschamps, F. Staes, P. Roosen, F. Nobels, K. Desloovere, H. Bruyninckx, and G. A. Matricali. Body of evidence supporting the clinical use of 3D multi-segment foot models: A systematic review. *Gait & Posture*, 33(3):338–349, 2011.
- [74] R. Di Marco, S. Rossi, V. Racic, P. Cappa, and C. Mazza. Concurrent repeatability and reproducibility analyses of four marker placement protocols for the foot-ankle complex. *Journal of Biomechanics*, 49(14):3168–3176, 2016.
- [75] F. E. DiLiberto, J. Tome, J. F. Baumhauer, J. Houck, and D. A. Nawoczenski. Individual metatarsal and forefoot kinematics during walking in people with diabetes mellitus and peripheral neuropathy. *Gait & Posture*, 42(4):435–441, 2015.
- [76] F. E. DiLiberto, J. Tome, J. F. Baumhauer, J. R. Quinn, J. Houck, and D. A. Nawoczenski. Multi-joint foot kinetics during walking in people with Diabetes Mellitus and peripheral neuropathy. *Journal of Biomechanics*, 48(13):3688–3693, 2015.
- [77] P. C. Dixon, H. Böhm, and L. Döderlein. Ankle and midfoot kinetics during normal gait: A multi-segment approach. *Journal of Biomechanics*, 45(6):1011–1016, 2012.
- [78] P. C. Dixon, M. Bowtell, and J. Stebbins. The use of regression and normalisation for the comparison of spatio-temporal gait data in children. *Gait & Posture*, 40(4):521–525, 2014.
- [79] P. C. Dixon, J. J. Loh, Y. Michaud-Paquette, and D. J. Pearsall. biomechZoo: An open-source toolbox for the processing, analysis, and visualization of biomechanical movement data. *Computer Methods and Programs in Biomedicine*, 140:1–10, 2017.
- [80] L. Dorst. First order error propagation of the procrustes method for 3D attitude estimation. *IEEE Transactions on Pattern Analysis and Machine Intelligence*, 27(2):221–229, 2005.
- [81] R. Dubbeldam, J. H. Buurke, C. Simons, C. G. M. Groothuis-Oudshoorn, H. Baan, A. V. Nene, and H. J. Hermens. The effects of walking speed on forefoot, hindfoot and ankle joint motion. *Clinical Biomechanics*, 25(8):796–801, 2010.
- [82] S. Duerinck, F. Hagman, I. Jonkers, P. Van Roy, and P. Vaes. Forefoot deformation during stance: Does the forefoot collapse during loading? *Gait & Posture*, 39(1):40–47, 2014.
- [83] R. Dumas, V. Camomilla, T. Bonci, L. Cheze, and A. Cappozzo. Generalized mathematical representation of the soft tissue artefact. *Journal of Biomechanics*, 47(2):476–481, 2014.

- [84] J. Durbin and G. S. Watson. Testing for serial correlation in least squares regression. II. *Biometrika*, 38(1-2):159–78, 1951.
- [85] M. N. Eek, R. Zügner, I. Stefansdottir, and R. Tranberg. Kinematic gait pattern in children with cerebral palsy and leg length discrepancy: Effects of an extra sole. *Gait & Posture*, 55:150–156, 2017.
- [86] L. E.H., G. J.C.H., and B. K. Value of gait analysis in the assessment of surgery in cerebral palsy. *Archives of Physical Medicine and Rehabilitation*, 73(7):642–646, 1992.
- [87] Y. Ehara, H. Fujimoto, S. Miyazaki, M. Mochimaru, S. Tanaka, and S. Yamamoto. Comparison of the performance of 3D camera systems II. *Gait & Posture*, 5(3):251–255, 1997.
- [88] Y. Ehara, H. Fujimoto, S. Miyazaki, S. Tanaka, and S. Yamamoto. Comparison of the performance of 3D camera systems. *Gait & Posture*, 3(3):166–169, 1995.
- [89] P. Eichelberger, M. Ferraro, U. Minder, T. Denton, A. Blasimann, F. Krause, and H. Baur. Analysis of accuracy in optical motion capture – A protocol for laboratory setup evaluation. *Journal of Biomechanics*, 49(10):2085–2088, 2016.
- [90] H. Elftman. The measurement of the external force in walking. *Science*, 88(2276):152–153, 1938.
- [91] H. Elftman. The transverse tarsal joint and its control. *Clinical Prthopaedics*, 16:41–6, 1960.
- [92] H. Elftman and J. Manter. The evolution of the human foot, with especial reference to the joints. *Journal of Anatomy*, 70:56–67, 1935.
- [93] F. Ferretti, A. Saltelli, and S. Tarantola. Trends in sensitivity analysis practice in the last decade. *Science of the Total Environment*, 568:666–670, 2016.
- [94] B. Fritz, T. Schmeltzpfenning, C. Plank, T. Hein, and S. Grau. Anthropometric influences on dynamic foot shape: Measurements of plantar three-dimensional foot deformation. *Footwear Science*, 5(2):121–129, 2013.
- [95] J. R. Gage, J. Perry, R. R. Hicks, S. Koop, and J. R. Werntz. Rectus femoris transfer to improve knee function of children with cerebral palsy. *Developmental Medicine & Child Neurology*, 29(2):159–166, 1987.
- [96] R. H. Gardner, R. V. O’Neill, J. B. Mankin, and J. H. Carney. A comparison of sensitivity analysis and error analysis based on a stream ecosystem model. *Ecological Modelling*, 12(3):173–190, 1981.
- [97] G. E. Gorton, D. A. Hebert, and M. E. Gannotti. Assessment of the kinematic variability among 12 motion analysis laboratories. *Gait & Posture*, 29(3):398–402, 2009.

- [98] C. M. Goss. On movement of muscles by galen of pergamon. *American Journal of Anatomy*, 123(1):1–25, 1968.
- [99] B. F. Green. The orthogonal approximation of an oblique structure in factor analysis. *ETS Research Bulletin Series*, 1951(2), 1951.
- [100] E. Grimpampi, V. Camomilla, A. Cereatti, P. De Leva, and A. Cappozzo. Metrics for describing soft-tissue artefact and its effect on pose, size, and shape of marker clusters. *IEEE Transactions on Biomedical Engineering*, 61(2):362–367, 2014.
- [101] E. S. Grood and W. J. Suntay. A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. *Journal of Biomechanical Engineering*, 105(2):136, 1983.
- [102] E. Halilaj, A. Rajagopal, M. Fiterau, J. L. Hicks, T. J. Hastie, and S. L. Delp. Machine learning in human movement biomechanics: Best practices, common pitfalls, and new opportunities. *Journal of Biomechanics*, 81:1–11, 2018.
- [103] D. M. Hamby. A review of techniques for parameter sensitivity analysis of environmental models. *Environmental Monitoring and Assessment*, 32(2):135–154, 1994.
- [104] J. Hamill, K. Knutzen, and T. Derrick. *Biomechanical basis of human movement*. Lippincott Williams and Wilkins, 4th edition, 2014.
- [105] R. J. Hanson and M. J. Norris. Analysis of measurements based on the singular value decomposition. *SIAM Journal on Scientific and Statistical Computing*, 2(3):363–373, 1981.
- [106] A. F. Hayes and L. Cai. Using heteroskedasticity-consistent standard error estimators in OLS regression: An introduction and software implementation. *Behavior Research Methods*, 39(4):709–722, 2007.
- [107] P. Heiselberg, H. Brohus, A. Hesselholt, H. Rasmussen, E. Seinre, and S. Thomas. Application of sensitivity analysis in design of sustainable buildings. *Renewable Energy*, 34(9):2030–2036, 2009.
- [108] J. C. Helton. Uncertainty and sensitivity analysis techniques for use in performance assessment for radioactive waste disposal. *Reliability Engineering & System Safety*, 42(2-3):327–367, 1993.
- [109] J. Henley, D. Hudson, S. Coleman, F. Miller, C. Church, L. Kersetter, and J. Richards. Reliability of a Clinically Practical Multisegment Foot Marker Set/Model. In *Foot and Ankle Motion Analysis: Clinical Treatment and Technology*, pages 445–463. CRC Press, 2007.
- [110] J. Hicks. The foot as a support. *Cells Tissues Organs*, 25(1):34–45, 1955.

- [111] J. H. Hicks. The mechanics of the foot. I. The joints. *Journal of Anatomy*, 87(4):345–57, 1953.
- [112] J. H. Hicks. The mechanics of the foot: II. The plantar aponeurosis and the arch. *Journal of Anatomy*, 88:25–30, 1954.
- [113] B. Hintermann. Biomechanics of the unstable ankle joint and clinical implications. *Medicine and Science in Sports and Exercise*, 31(7):459–469, 1999.
- [114] A. Hof and W. Zijlstra. Comment on “normalization of temporal-distance parameters in pediatric gait”. *Journal of Biomechanics*, 30(3):299, 1997.
- [115] A. L. Hof. Scaling gait data to body size. *Gait & Posture*, 4(3):222–223, 1996.
- [116] M. Hopson, T. McPoil, and M. Cornwall. Motion of the first metatarsophalangeal joint. Reliability and validity of four measurement techniques. *Journal of the American Podiatric Medical Association*, 85(4):198–204, 1995.
- [117] M. Hösl, H. Böhm, C. Multerer, and L. Döderlein. Does excessive flatfoot deformity affect function? A comparison between symptomatic and asymptomatic flatfeet using the Oxford Foot Model. *Gait & Posture*, 39(1):23–28, 2014.
- [118] J. R. Houck, J. M. Tome, and D. A. Nawoczenski. Subtalar neutral position as an offset for a kinematic model of the foot during walking. *Gait & Posture*, 28(1):29–37, 2008.
- [119] C. K. Huang, H. B. Kitaoka, K. N. An, and E. Y. Chao. Biomechanical evaluation of longitudinal arch stability. *Foot & Ankle International*, 14(6):353–7, 1993.
- [120] J. Hughes, P. Clark, and L. Klenerman. The importance of the toes in walking. *Journal of Bone and Joint Surgery*, 72(2):245–251, 1990.
- [121] V. T. Inman, H. Ralston, J. De C.M. Saunders, M. Bertram Feinstein, and E. W. Wright. Relation of human electromyogram to muscular tension. *Electroencephalography and Clinical Neurophysiology*, 4(2):187–194, 1952.
- [122] B. Iooss and P. Lemaître. A review on global sensitivity analysis methods. In *Uncertainty Management in Simulation-Optimization of Complex Systems*, pages 101–122. Springer, 2015.
- [123] R. Isman and V. T. Inman. *Anthropometric Studies of the Human Foot and Ankle*, volume 58. Biomechanics Laboratory, University of California, 1968.
- [124] H. A. C. Jacob. *Biomechanics of the forefoot*. PhD thesis, University of Strathclyde, 1989.
- [125] T. Jenkyn and A. Nicol. A multi-segment kinematic model of the foot with a novel definition of forefoot motion for use in clinical gait analysis during walking. *Journal of Biomechanics*, 40(14):3271–3278, 2007.

- [126] M. P. Kadaba, H. K. Ramakrishnan, and M. E. Wootten. Measurement of lower extremity kinematics during level walking. *Journal of Orthopaedic Research*, 8(3):383–392, 1990.
- [127] M. P. Kadaba, H. K. Ramakrishnan, M. E. Wootten, J. Gainey, G. Gorton, and G. V. B. Cochran. Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *Journal of Orthopaedic Research*, 7(6):849–860, 1989.
- [128] I. A. Kapandji. The Plantar Vault: Volume Two, The lower limb. In *Physiology of the Joints*, pages 216–239. Churchill Livingstone, 6th edition, 2010.
- [129] R. M. Kay, S. Dennis, S. Rethlefsen, R. A. K. Reynolds, D. L. Skaggs, and V. T. Tolo. The Effect of Preoperative Gait Analysis on Orthopaedic Decision Making. *Clinical Orthopaedics and Related Research*, 372:217–222, 2000.
- [130] T. M. Kepple, S. J. Stanhope, K. N. Lohmann, and N. L. Roman. A video-based technique for measuring ankle-subtalar motion during stance. *Journal of Biomedical Engineering*, 12(4):273–280, 1990.
- [131] T. W. Kernozek and J. D. Willson. Clinicalgate: Gait cycle. Retrieved from <http://clinicalgate.com>, 2015.
- [132] M. Khazzam, J. T. Long, R. M. Marks, and G. F. Harris. Preoperative gait characterization of patients with ankle arthrosis. *Gait & Posture*, 24(1):85–93, 2006.
- [133] M. Khazzam, J. T. Long, R. M. Marks, and G. F. Harris. Kinematic changes of the foot and ankle in patients with systemic rheumatoid arthritis and forefoot deformity. *Journal of Orthopaedic Research*, 25(3):319–29, 2007.
- [134] T. W. Kibble and F. H. Berkshire. *Classical Mechanics*. Imperial College Press, 5th edition, 2012.
- [135] S. M. Kidder, F. S. Abuzzahab, G. F. Harris, and J. E. Johnson. A system for the analysis of foot and ankle kinematics during gait. *IEEE Transactions on Rehabilitation Engineering*, 4(1):25–32, 1996.
- [136] C. Kirtley, M. W. Whittle, and R. J. Jefferson. Influence of walking speed on gait parameters. *Journal of Biomedical Engineering*, 7(4):282–288, 1985.
- [137] H. B. Kitaoka, X. M. Crevoisier, D. Hansen, B. Katajarvi, K. Harbst, and K. R. Kaufman. Foot and ankle kinematics and ground reaction forces during ambulation. *Foot & Ankle International*, 27(10):808–813, 2006.
- [138] A. Kothari, S. Bhuvu, J. Stebbins, A. B. Zavatsky, and T. Theologis. An investigation into the aetiology of flexible flat feet: the role of subtalar joint morphology. *The Bone & Joint Journal*, 98-B(4):564–568, 2016.

- [139] L. M. Lang. The anatomy of the foot. *Bailliere's Clinical Rheumatology*, 1(2):215–40, 1987.
- [140] P. W. Lapidus. Kinesiology and mechanical anatomy of the tarsal joints. *Clinical Orthopaedics and Related Research*, 30:20–36, 1963.
- [141] A. M. Law and W. D. Kelton. Confidence intervals for steady-state simulations II: A survey of sequential procedures. *Management Science*, 28(5):550–562, 1982.
- [142] A. M. Law and W. D. Kelton. Confidence intervals for steady-state simulations: I. A survey of fixed sample size procedures. *Operations Research*, 32(6):1221–1239, 1984.
- [143] A. Leardini, M. Benedetti, L. Berti, D. Bettinelli, R. Nativo, and S. Giannini. Rear-foot, mid-foot and fore-foot motion during the stance phase of gait. *Gait & Posture*, 25(3):453–462, 2007.
- [144] A. Leardini, M. G. Benedetti, F. Catani, L. Simoncini, and S. Giannini. An anatomically based protocol for the description of foot segment kinematics during gait. *Clinical Biomechanics*, 14(8):528–536, 1999.
- [145] A. Leardini and P. Caravaggi. Kinematic foot models for instrumented gait analysis. In *Handbook of Human Motion*. Springer International Publishing, 2018.
- [146] A. Leardini, P. Caravaggi, T. Theologis, and J. Stebbins. Multi-segment foot models and their use in clinical populations. *Gait & Posture*, 69:50–59, 2019.
- [147] A. Leardini, L. Chiari, U. Della Croce, and A. Cappozzo. Human movement analysis using stereophotogrammetry Part 3. Soft tissue artifact assessment and compensation. *Gait & Posture*, 21(2):212–225, 2005.
- [148] A. Leardini, J. J. O'Connor, F. Catani, and S. Giannini. Kinematics of the human ankle complex in passive flexion; A single degree of freedom system. *Journal of Biomechanics*, 32:111–118, 1999.
- [149] A. Leardini, J. J. O'Connor, F. Catani, and S. Giannini. The role of the passive structures in the mobility and stability of the human ankle joint: A literature review. *Foot & Ankle International*, 21(7):602–15, 2000.
- [150] A. Leardini, R. Stagni, and J. J. O'Connor. Mobility of the subtalar joint in the intact ankle complex. *Journal of Biomechanics*, 34(6):805–809, 2001.
- [151] M. S. Lee, J. V. Vanore, J. L. Thomas, A. R. Catanzariti, G. Kogler, S. R. Kravitz, S. J. Miller, and S. C. Gassen. Diagnosis and treatment of adult flatfoot. *The Journal of Foot and Ankle Surgery*, 44(2):78–113, 2005.

- [152] J. Leitch, J. Stebbins, and A. B. Zavatsky. Subject-specific axes of the ankle joint complex. *Journal of Biomechanics*, 43(15):2923–2928, 2010.
- [153] M. W. Lenhoff, T. J. Santner, J. C. Otis, M. G. Peterson, B. J. Williams, and S. I. Backus. Bootstrap prediction and confidence bands: a superior statistical method for analysis of gait data. *Gait & Posture*, 9(1):10–7, mar 1999.
- [154] D. Levine, J. Richards, and M. W. Whittle. *Whittle’s Gait Analysis*. Churchill Livingstone, 5th edition, 2012.
- [155] J. L. Lewis and W. D. Lew. A note on the description of articulating joint motion. *Journal of Biomechanics*, 10(10):675–678, 1977.
- [156] W. Liu, S. Siegler, H. Hillstrom, and K. Whitney. Three-dimensional, six-degrees-of-freedom kinematics of the human hindfoot during the stance phase of level walking. *Human Movement Science*, 16:283–298, 1997.
- [157] P. R. G. Lucareli, L. B. G. Contani, B. Lima, N. D. d. A. Rabelo, C. L. Ferreira, F. P. S. Lima, J. C. F. Correa, and F. Politti. Repeatability of a 3D multi-segment foot model during anterior and lateral step down tests. *Gait & Posture*, 43:9–16, 2016.
- [158] A. Lundberg. Kinematics of the ankle and foot: In vivo roentgen stereophotogrammetry. *Acta Orthopaedica Scandinavica*, 60:1–26, 1989.
- [159] A. Lundberg and O. Svensson. The axes of rotation of the talocalcaneal and talonavicular joints. *The Foot*, 3(2):65–70, 1993.
- [160] P. Lundgren, C. Nester, A. Liu, A. Arndt, R. Jones, A. Stacoff, P. Wolf, and A. Lundberg. Invasive in vivo measurement of rear-, mid- and forefoot motion during walking. *Gait & Posture*, 28(1):93–100, 2008.
- [161] B. A. MacWilliams, M. Cowley, and D. E. Nicholson. Foot kinematics and kinetics during adolescent gait. *Gait & Posture*, 17(3):214–224, 2003.
- [162] R. Mahaffey. *Biomechanics of the paediatric foot and lower limb: Associations with adiposity*. PhD thesis, University of East London, 2013.
- [163] R. Mahaffey, S. C. Morrison, W. I. Drechsler, and M. C. Cramp. Evaluation of multi-segmental kinematic modelling in the paediatric foot using three concurrent foot models. *Journal of Foot and Ankle Research*, 6(1):43, 2013.
- [164] C. Maiwald, A. Arndt, C. Nester, R. Jones, A. Lundberg, and P. Wolf. The effect of intracortical bone pin application on kinetics and tibio-calcaneal kinematics of walking gait. *Gait & Posture*, 52:129–134, 2017.
- [165] J. Majernik. Normative Human Gait Databases. *Statistics Research Letters*, 2(3):69–74, 2013.

- [166] L. P. Maletsky, J. Sun, and N. A. Morton. Accuracy of an optical active-marker system to track the relative motion of rigid bodies. *Journal of Biomechanics*, 40(3):682–685, 2007.
- [167] J. T. Manter. Movements of the subtalar and transverse tarsal joints. *The Anatomical Record*, 80(4):397–410, 1941.
- [168] R. M. Marks, J. T. Long, M. E. Ness, M. Khazzam, and G. F. Harris. Surgical reconstruction of posterior tibial tendon dysfunction: Prospective comparison of flexor digitorum longus substitution combined with lateral column lengthening or medial displacement calcaneal osteotomy. *Gait & Posture*, 29(1):17–22, 2009.
- [169] B. Maslen and T. Ackland. Radiographic study of skin displacement errors in the foot and ankle during standing. *Clinical Biomechanics*, 9(5):291–296, 1994.
- [170] J. D. Maurer, V. Ward, T. A. Mayson, K. R. Davies, C. M. Alvarez, R. D. Beauchamp, and A. H. Black. Classification of midfoot break using multi-segment foot kinematics and pedobarography. *Gait & Posture*, 39(1):1–6, 2014.
- [171] J. McCahill, J. Stebbins, B. Koning, J. Harlaar, and T. Theologis. Repeatability of the Oxford Foot Model in children with foot deformity. *Gait & Posture*, 61:86–89, 2018.
- [172] T. G. McPoil and H. G. Knecht. Biomechanics of the foot in walking: a function approach. *The Journal of Orthopaedic and Sports Physical Therapy*, 7(2):69–72, 1985.
- [173] G. J. McRae, J. W. Tilden, and J. H. Seinfeld. Global sensitivity analysis-a computational implementation of the Fourier Amplitude Sensitivity Test (FAST). *Computers and Chemical Engineering*, 6(1):15–25, 1982.
- [174] K. C. Moisio, D. R. Sumner, S. Shott, and D. E. Hurwitz. Normalization of joint moments during gait: A comparison of two techniques. *Journal of Biomechanics*, 36(4):599–603, 2003.
- [175] M. Molloy, J. Salazar-Torres, C. Kerr, B. C. McDowell, and A. P. Cosgrove. The effects of industry standard averaging and filtering techniques in kinematic gait analysis. *Gait & Posture*, 28(4):559–562, 2008.
- [176] K. L. Moore, A. M. Agur, and A. F. Dally. *Essential Clinical Anatomy*. Lippincott Williams and Wilkins, 2014.
- [177] D. S. Morrell, J. M. Pearson, and D. D. Sauser. Progressive bone and joint abnormalities of the spine and lower extremities in cerebral palsy. *Radiographics*, 22(2):257–68, 2002.
- [178] L. Moseley, R. Smith, a. Hunt, and R. Gant. Three-dimensional kinematics of the rearfoot during the stance phase of walking in normal young adult males. *Clinical Biomechanics*, 11(1):39–45, 1996.

- [179] Motion Lab Systems. The C3D File Format User Manual, 2002.
- [180] A. Muro-de-la Herran, B. García-Zapirain, and A. Méndez-Zorrilla. Gait analysis methods: An overview of wearable and non-wearable systems, highlighting clinical applications. *Sensors*, 14(2):3362–3394, 2014.
- [181] G. A. Murphy, S. G. Pneumáticos, E. Kamaric, P. C. Noble, S. G. Trevino, and D. E. Baxter. Biomechanical consequences of sequential plantar fascia release. *Foot & Ankle International*, 19(3):149–152, 1998.
- [182] E. Muybridge. The horse in motion. Retrieved from <http://www.eadweardmuybridge.co.uk>.
- [183] K. A. Myers, M. Wang, R. M. Marks, and G. F. Harris. Validation of a multi-segment foot and ankle kinematic model for pediatric gait. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 12(1):122–130, 2004.
- [184] D. A. Nawoczinski, J. Ketz, and J. F. Baumhauer. Dynamic Kinematic and Plantar Pressure Changes Following Cheilectomy for Hallux Rigidus: A Mid-Term Followup. *Foot & Ankle International*, 29(3):265–272, mar 2008.
- [185] A. Nene, R. Mayagoitia, and P. Veltink. Assessment of rectus femoris function during initial swing phase. *Gait & Posture*, 9(1):1–9, 1999.
- [186] C. Nester, R. Jones, A. Liu, D. Howard, A. Lundberg, A. Arndt, P. Lundgren, A. Stacoff, and P. Wolf. Foot kinematics during walking measured using bone and surface mounted markers. *Journal of Biomechanics*, 40(15):3412–3423, 2007.
- [187] C. J. Nester, H. L. Jarvis, R. K. Jones, P. D. Bowden, and A. Liu. Movement of the human foot in 100 pain free individuals aged 18-45: Implications for understanding normal foot function. *Journal of Foot and Ankle Research*, 7(1):1–10, 2014.
- [188] C. J. Nester, A. M. Liu, E. Ward, D. Howard, J. Cocheba, and T. Derrick. Error in the description of foot kinematics due to violation of rigid body assumptions. *Journal of Biomechanics*, 43(4):666–672, 2010.
- [189] C. J. Nester, a. M. Liu, E. Ward, D. Howard, J. Cocheba, T. Derrick, and P. Patterson. In vitro study of foot kinematics using a dynamic walking cadaver model. *Journal of biomechanics*, 40(9):1927–37, 2007.
- [190] K. Nicholson, C. Church, C. Takata, T. Nüiler, B. P.-J. J. Chen, N. Lennon, J. P. Sees, J. Henley, and F. Miller. Comparison of three-dimensional multi-segmental foot models used in clinical gait laboratories. *Gait & Posture*, 63:236–241, 2018.
- [191] B. Nigg and W. Herzog. *Biomechanics of the Musculo-skeletal System*. Wiley, 3rd edition, 2007.

- [192] M. Nordin and V. Frankel. *Basic Biomechanics of the Musculoskeletal System*. Lippincott Williams and Wilkins, 4th edition, 2012.
- [193] A. C. Novak, D. J. Mayich, S. D. Perry, T. R. Daniels, and J. W. Brodsky. Gait Analysis for Foot and Ankle Surgeons— Topical Review, Part 2. *Foot & Ankle International*, 35(2):178–191, 2014.
- [194] N. Okita, S. A. Meyers, J. H. Challis, and N. A. Sharkey. An objective evaluation of a segmented foot model. *Gait & Posture*, 30(1):27–34, 2009.
- [195] N. Okita, S. A. Meyers, J. H. Challis, and N. A. Sharkey. Midtarsal joint locking: New perspectives on an old paradigm. *Journal of Orthopaedic Research*, 32(1):110–115, 2014.
- [196] M. J. O’Malley. Normalization of temporal-distance parameters in pediatric gait. *Journal of Biomechanics*, 29(5):619–625, 1996.
- [197] T. Ota, T. Nagura, T. Kokubo, M. Kitashiro, N. Ogihara, K. Takeshima, H. Seki, Y. Suda, M. Matsumoto, and M. Nakamura. Etiological factors in hallux valgus, a three-dimensional analysis of the first metatarsal. *Journal of Foot and Ankle Research*, 10(1):1–6, 2017.
- [198] T. J. Ouzounian and M. J. Shereff. In vitro determination of midfoot motion. *Foot & Ankle International*, 10(3):140–146, 1989.
- [199] D. J. Pannell. Sensitivity Analysis of Normative Economic Models: Theoretical Framework and Practical Strategies. *Journal of Agricultural Economics*, 16:139–152, 1997.
- [200] T. C. Pataky. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. *Journal of Biomechanics*, 43(10):1976–1982, 2010.
- [201] J. Pek, O. Wong, and C. M. Wong. Data transformations for inference with linear regression: Clarifications and recommendations. *Practical Assessment, Research & Evaluation*, 22(9), 2017.
- [202] J. Perry. *Gait Analysis: Normal and Pathological Function*. SLACK, illustrate edition, 1992.
- [203] J. Perry and M. Hoffer. Preoperative and postoperative dynamic electromyography as an aid in planning tendon transfers in children with cerebral palsy. *The Journal of Bone and Joint Surgery*, 59(4):531–537, 1977.
- [204] A. Peters, B. Galna, M. Sangeux, M. Morris, and R. Baker. Quantification of soft tissue artifact in lower limb human motion analysis: A systematic review. *Gait & Posture*, 31(1):1–8, 2010.

- [205] O. Pinzone, M. H. Schwartz, and R. Baker. Comprehensive non-dimensional normalization of gait data. *Gait & Posture*, 44:68–73, 2016.
- [206] M. B. Pohl, N. Messenger, and J. G. Buckley. Changes in foot and lower limb coupling due to systematic variations in step width. *Clinical Biomechanics*, 21(2):175–183, 2006.
- [207] N. Portinaro, A. Leardini, A. Panou, V. Monzani, and P. Caravaggi. Modifying the Rizzoli foot model to improve the diagnosis of pes-planus: Application to kinematics of feet in teenagers. *Journal of Foot and Ankle Research*, 7(1):1–7, 2014.
- [208] C. Pothrat, G. Authier, E. Viehweger, E. Berton, and G. Rao. One- and multi-segment foot models lead to opposite results on ankle joint kinematics during gait: Implications for clinical assessment. *Clinical Biomechanics*, 30(5):493–499, 2015.
- [209] C. Pothrat, B. Goislard de Monsabert, L. Vigouroux, E. Viehweger, E. Berton, and G. Rao. Quantifying foot deformation using finite helical angle. *Journal of Biomechanics*, 48(13):3725–3728, 2015.
- [210] L. Rankine, J. Long, K. Canseco, and G. Harris. Multisegmental foot modeling: a review. *Critical Reviews in Biomedical Engineering*, 36(2-3):127–181, 2008.
- [211] S. Rao, C. Saltzman, and H. J. Yack. Segmental foot mobility in individuals with and without diabetes and neuropathy. *Clinical Biomechanics*, 22(4):464–471, 2007.
- [212] U. Rattanaprasert, R. Smith, M. Sullivan, and W. Gillear. Three-dimensional kinematics of the forefoot, rearfoot, and leg without the function of tibialis posterior in comparison with normals during stance phase of walking. *Clinical Biomechanics*, 14(1):14–23, 1999.
- [213] J. O. Rawlings, S. G. Pantula, and D. A. Dickey. *Applied Regression Analysis*. Springer, 2nd edition, 1998.
- [214] M. Roberts, D. Mongeon, and F. Prince. Biomechanical parameters for gait analysis: a systematic review of healthy human gait. *Physical Therapy and Rehabilitation*, 4(1):6, 2017.
- [215] J. L. Robinson and G. L. Smidt. Quantitative gait evaluation in the clinic. *Physical Therapy*, 61(3):351–353, 1981.
- [216] P. A. Rockar. The Subtalar Joint: Anatomy and Joint Motion. *Journal of Orthopaedic and Sports Physical Therapy*, 21(6):361–372, 1995.
- [217] M. M. Rodgers. Dynamic foot biomechanics. *The Journal of Orthopaedic and Sports Physical Therapy*, 21(6):306–16, 1995.

- [218] H. Rouhani, J. Favre, X. Crevoisier, and K. Aminian. A wearable system for multi-segment foot kinetics measurement. *Journal of Biomechanics*, 47(7):1704–1711, 2014.
- [219] H. Rouhani, J. Favre, X. Crevoisier, B. M. Jolles, and K. Aminian. Segmentation of foot and ankle complex based on kinematic criteria. *Computer Methods in Biomechanics and Biomedical Engineering*, 14:773–781, 2011.
- [220] A. Saltelli, T. H. Andres, and T. Homma. Sensitivity analysis of model output: An investigation of new techniques. *Computational Statistics & Data Analysis*, 15(2):211–238, 1993.
- [221] A. Saltelli and P. Annoni. How to avoid a perfunctory sensitivity analysis. *Environmental Modelling and Software*, 25(12):1508–1517, 2010.
- [222] A. Saltelli, M. Ratto, T. Andres, F. Campolongo, J. Cariboni, D. Gatelli, M. Saisana, and S. Tarantola. *Global Sensitivity Analysis. The Primer*. John Wiley & Sons, Ltd, Chichester, UK, 2007.
- [223] A. Saltelli, S. Tarantola, and F. Campolongo. Sensitivity analysis as an ingredient of modeling. *Statistical Science*, 15(4):377–395, 2000.
- [224] C. L. Saltzman, D. A. Nawoczenski, and K. D. Talbot. Measurement of the medial longitudinal arch. *Archives of Physical Medicine and Rehabilitation*, 76(1):45–49, jan 1995.
- [225] M. Sandau, H. Koblauch, T. B. Moeslund, H. Aanæs, T. Alkjær, and E. B. Simonsen. Markerless motion capture can provide reliable 3D gait kinematics in the sagittal and frontal plane. *Medical Engineering & Physics*, 36(9):1168–1175, 2014.
- [226] M. Sangeux, A. Barré, and K. Aminian. Evaluation of knee functional calibration with and without the effect of soft tissue artefact. *Journal of Biomechanics*, 62:53–59, 2017.
- [227] W. N. Sankar, S. A. Rethlefsen, J. Weiss, and R. M. Kay. The recurrent clubfoot: Can gait analysis help us make better preoperative decisions? *Clinical Orthopaedics and Related Research*, 467(5):1214–1222, 2009.
- [228] P. Saraswat, B. A. MacWilliams, and R. B. Davis. A multi-segment foot model based on anatomically registered technical coordinate systems: Method repeatability in pediatric feet. *Gait & Posture*, 35(4):547–555, 2012.
- [229] S. K. Sarrafian. Functional characteristics of the foot and plantar aponeurosis under tibiotalar loading. *Foot & Ankle International*, 8(1):4–18, 1987.
- [230] Z. Sawacha, G. Cristoferi, G. Guarneri, S. Corazza, G. Donà, P. Denti, A. Facchinetti, A. Avogaro, and C. Cobelli. Characterizing multisegment foot kinematics during gait in diabetic foot patients. *Journal of Neuroengineering and Rehabilitation*, 6:37, 2009.

- [231] Z. Sawacha, G. Gabriella, G. Cristoferi, A. Guiotto, A. Avogaro, and C. Cobelli. Diabetic gait and posture abnormalities: A biomechanical investigation through three dimensional gait analysis. *Clinical Biomechanics*, 24(9):722–728, 2009.
- [232] P. H. Schönemann. A generalized solution of the orthogonal procrustes problem. *Psychometrika*, 31(1):1–10, 1966.
- [233] M. Schwartz, S. Koop, J. Bourke, and R. Baker. A nondimensional normalization scheme for oxygen utilization data. *Gait & Posture*, 24(1):14–22, 2006.
- [234] M. H. Schwartz, J. P. Trost, and R. A. Wurvey. Measurement and management of errors in quantitative gait data. *Gait & Posture*, 20(2):196–203, 2004.
- [235] S. H. Scott and D. A. Winter. Talocrural and talocalcaneal joint kinematics and kinetics during the stance phase of walking. *Journal of Biomechanics*, 24(8):743–752, 1991.
- [236] J. P. Sees and F. Miller. Overview of foot deformity management in children with cerebral palsy. *Journal of Children’s Orthopaedics*, 7(5):373–377, 2013.
- [237] G. Selvik. Roentgen stereophotogrammetry - A method for the study of the kinematics of the skeletal system. *Acta Orthopaedica*, 60(s232):1–51, 1984.
- [238] M. J. Shereff, F. J. Bejjani, and F. J. Kummer. Kinematics of the first metatarsophalangeal joint. *The Journal of Bone and Joint Surgery*, 68(3):392–398, 1986.
- [239] K. L. Siegel, T. M. Kepple, P. G. O’Connell, L. H. Gerber, and S. J. Stanhope. A technique to evaluate foot function during the stance phase of gait. *Foot & Ankle International*, 16(12):764–70, 1995.
- [240] J. Simon, L. Doederlein, A. McIntosh, D. Metaxiotis, H. Bock, and S. Wolf. The Heidelberg foot measurement method: Development, description and assessment. *Gait & Posture*, 23(4):411–424, 2006.
- [241] S. R. Simon. Quantification of human motion: Gait analysis - Benefits and limitations to its application to clinical problems. *Journal of Biomechanics*, 37(12):1869–1880, 2004.
- [242] J. Sinclair, P. J. Taylor, and S. J. Hobbs. Digital filtering of three-dimensional lower extremity kinematics: An assessment. *Journal of Human Kinetics*, 39(1):25–36, 2013.
- [243] R. Soavi, M. Girolami, I. Loreti, L. Bragonzoni, C. Monti, A. Visani, and M. Marcacci. The mobility of the proximal tibio-fibular joint. A Roentgen Stereophotogrammetric Analysis on six cadaver specimens. *Foot & Ankle International*, 21(4):336–342, 2000.

- [244] I. Söderkvist and P. Å. Wedin. Determining the movements of the skeleton using well-configured markers. *Journal of Biomechanics*, 26(12):1473–1477, 1993.
- [245] I. Söderkvist and P. Å. Wedin. On condition numbers and algorithms for determining a rigid body movement. *Bit*, 34(3):424–436, 1994.
- [246] C. W. Spoor and F. E. Veldpaus. Rigid body motion calculated from spatial co-ordinates of markers. *Journal of Biomechanics*, 13(4):391–393, 1980.
- [247] R. Stagni, S. Fantozzi, A. Cappello, and A. Leardini. Quantification of soft tissue artefact in motion analysis by combining 3D fluoroscopy and stereophotogrammetry: a study on two subjects. *Clinical Biomechanics*, 20(3):320–329, 2005.
- [248] R. Stagni, A. Leardini, J. J. O’Connor, and S. Giannini. Role of passive structures in the mobility and stability of the human subtalar joint: A literature review. *Foot & Ankle International*, 24(5):402–409, 2003.
- [249] S. M. Stearne, K. A. McDonald, J. A. Alderson, I. North, C. E. Oxnard, and J. Rubenson. The Foot’s Arch and the Energetics of Human Locomotion. *Scientific Reports*, 6(1), 2016.
- [250] J. Stebbins, M. Harrington, N. Thompson, A. Zavatsky, and T. Theologis. Repeatability of a model for measuring multi-segment foot kinematics in children. *Gait & Posture*, 23(4):401–410, 2006.
- [251] D. H. Sutherland. The evolution of clinical gait analysis part 1: Kinesiological EMG. *Gait & Posture*, 14(1):61–70, 2001.
- [252] D. H. Sutherland. The evolution of clinical gait analysis Part II kinematics. *Gait & Posture*, 16(2):159–179, 2002.
- [253] D. H. Sutherland. The evolution of clinical gait analysis part III - Kinetics and energy assessment. *Gait & Posture*, 21(4):447–461, 2005.
- [254] S. Tadano, R. Takeda, and H. Miyagawa. Three dimensional gait analysis using wearable acceleration and gyro sensors based on quaternion calculations. *Sensors*, 13(7):9321–9343, 2013.
- [255] T. Theologis and J. Stebbins. The use of gait analysis in the treatment of pediatric foot and ankle disorders. *Foot and Ankle Clinics*, 15(2):365–382, 2010.
- [256] T. N. Theologis, M. E. Harrington, N. Thompson, and M. K. D. Benson. Dynamic foot movement in children treated for congenital talipes equinovarus. *The Journal of Bone and Joint Surgery*, 85:572–577, 2003.
- [257] K. Thomas Haendlmayer and N. John Harris. (ii) Flatfoot deformity: an overview. *Orthopaedics and Trauma*, 23(6):395–403, 2009.

- [258] D. B. Thordarson, P. J. Kumar, T. P. Hedman, and E. Ebrahimzadeh. Effect of Partial Versus Complete Plantar Fasciotomy on the Windlass Mechanism. *Foot & Ankle International*, 18(1):16–20, 1997.
- [259] J. Tome, D. a. Nawoczenski, A. Flemister, and J. Houck. Comparison of foot kinematics between subjects with posterior tibialis tendon dysfunction and healthy controls. *The Journal of Orthopaedic and Sports Physical Therapy*, 36(9):635–644, 2006.
- [260] R. Tranberg and D. Karlsson. The relative skin movement of the foot: a 2-D roentgen photogrammetry study. *Clinical Biomechanics*, 13(1):71–76, 1998.
- [261] J. W. Tukey. On the Comparative Anatomy of Transformations. *The Annals of Mathematical Statistics*, 28(3):602–632, 1957.
- [262] D. E. Turner, P. S. Helliwell, P. Emery, and J. Woodburn. The impact of rheumatoid arthritis on foot function in the early stages of disease: A clinical case series. *BMC Musculoskeletal Disorders*, 7:102, 2006.
- [263] D. E. Turner and J. Woodburn. Characterising the clinical and biomechanical features of severely deformed feet in rheumatoid arthritis. *Gait & Posture*, 28(4):574–580, 2008.
- [264] A. J. van den Bogert, G. D. Smith, and B. M. Nigg. In vivo determination of the anatomical axes of the ankle joint complex: An optimization approach. *Journal of Biomechanics*, 27(12):1477–1488, 1994.
- [265] F. E. Veldpaus, H. J. Woltring, and J. M. G. Dortmans, L. A least-squares algorithm for the equiform transformation. *Journal of Biomechanics*, 21(I):45–54, 1988.
- [266] Vicon Motion Systems. Plug-In Gait Manual, 2012.
- [267] W. Weber and E. Weber. Ueber die Mechanik der menschlichen Gehwerkzeuge, nebst der Beschreibung eines Versuchs über das Herausfallen des Schenkelkopfs aus der Pfanne im luftverdünnten Raume. *Annalen der Physik und Chemie*, 116(1):1–13, 1837.
- [268] J. Weidow, R. Tranberg, T. Saari, and J. Kärrholm. Hip and knee joint rotations differ between patients with medial and lateral knee osteoarthritis: Gait analysis of 30 patients and 15 controls. *Journal of Orthopaedic Research*, 24(9):1890–1899, 2006.
- [269] P. Westblad, T. Hashimoto, I. Winson, A. Lundberg, and A. Arndt. Differences in ankle-joint complex motion during the stance phase of walking as measured by superficial and bone-anchored markers. *Foot & Ankle International*, 23(9):856–863, 2002.

- [270] D. Williams and I. McClay. Measurements used to characterize the foot and the medial longitudinal arch : Reliability and Validity. *Physical Therapy*, 80(9):864–871, 2000.
- [271] P. Wolf, A. Stacoff, A. Liu, C. Nester, A. Arndt, A. Lundberg, and E. Stuessi. Functional units of the human foot. *Gait & Posture*, 28(3):434–441, 2008.
- [272] H. J. Woltring. A Fortran package for generalized, cross-validatory spline smoothing and differentiation. *Advances in Engineering Software (1978)*, 8(2):104–113, 1986.
- [273] H. J. Woltring. Representation and calculation of 3-D joint movement. *Human Movement Science*, 10(5):603–616, 1991.
- [274] H. J. Woltring, R. Huiskes, A. de Lange, and F. E. Veldpaus. Finite centroid and helical axis estimation from noisy landmark measurements in the study of human joint kinematics. *Journal of Biomechanics*, 18(5):379–389, 1985.
- [275] T. A. Wren, G. E. Gorton, S. Öunpuu, and C. A. Tucker. Efficacy of clinical gait analysis: A systematic review. *Gait & Posture*, 34(2):149–153, 2011.
- [276] C. J. Wright, B. L. Arnold, T. G. Coffey, and P. E. Pidcoe. Repeatability of the modified Oxford foot model during gait in healthy adults. *Gait & Posture*, 33(1):108–112, 2011.
- [277] D. G. Wright, S. M. Desai, and W. H. Henderson. Action of the subtalar and ankle-joint complex during the stance phase of walking. *The Journal of Bone and Joint Surgery*, 46(2):361–82, 1964.
- [278] G. Wu and P. R. Cavanagh. ISB recommendations in the reporting for standardization of kinematic data. *Journal of Biomechanics*, 28(10):1257–1261, 1995.
- [279] G. Wu, S. Siegler, P. Allard, C. Kirtley, A. Leardini, D. Rosenbaum, M. Whittle, D. D. D’Lima, L. Cristofolini, H. Witte, O. Schmid, and I. Stokes. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *Journal of Biomechanics*, 35(4):543–548, 2002.
- [280] W. L. Wu, F. C. Su, Y. M. Cheng, P. J. Huang, Y. L. Chou, and C. K. Chou. Gait analysis after ankle arthrodesis. *Gait & Posture*, 11(1):54–61, 2000.
- [281] J. A. Zeni and J. S. Higginson. Differences in gait parameters between healthy subjects and persons with moderate and severe knee osteoarthritis: A result of altered walking speed? *Clinical Biomechanics*, 24(4):372–378, 2009.