

College of Sport and Exercise Science

VICTORIA UNIVERSITY

**EXPLORING ADAPTABILITY IN LONG-DISTANCE
RUNNERS: EFFECT OF FOOT STRIKE PATTERN ON LOWER
LIMB NEURO-MUSCULAR-SKELETAL CAPACITY**

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DOCTOR OF PHILOSOPHY

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ABSTRACT

This thesis is an exploration of the controversial hypothesis that a runner's foot strike pattern defines the demands on the lower extremity, and hence we expect to observe adaptations to its anatomical, mechanical, and neurological function.

First, we review the state of the literature to find that long-distance running seems to have an osteogenic and myogenic effect on the foot; however, studies often do not control for foot strike or footwear worn, leading to circumstantial evidence. We therefore aim to determine structural differences between two groups of runners with an antithetical foot strike pattern (habitual rearfoot versus habitual forefoot strikers). We find groups to have similar foot muscle size and to have similar toe flexor strength. Further, we find the trabecular bone volume to be larger in the first metatarsal bone in forefoot strikers; however, the calcaneus reveals no differences between groups in bone density or trabecular structure.

We then explore the function of the ankle, in isolation and in coordination with the knee and hip. It appears that habitual forefoot strikers may have access to a wider physiological range of ankle torque and ankle joint angle. This increased potential may allow forefoot strikers to adapt to different footwear by regulating ankle stiffness depending upon motor task. The inter-joint coupling investigation reveals knee-hip coordination pattern of runners to be the most consistent, while ankle-knee couple was the most variable. Forefoot strikers have more variable coordinative patterns than rearfoot strikers irrespective of the footwear worn.

We then asked a neuro-mechanical question: Is the control of running kinematics and kinetics influenced by the foot strike type? Using analysis of persistence in time series and analysis of motor redundancy in human movement, we show that rearfoot strikers employ higher active control over critical variables such as limb posture at initial ground contact and leg stiffness. The results suggest that forefoot strikers achieve control of these parameters through exploitation of the abundant degrees of freedom available in the system.

Finally, we conclude the thesis with indications for short-term objectives in-line with the research that begun in this thesis.

DECLARATION

“I, Alessandro Garofolini, declare that the PhD thesis entitled ‘*Exploring adaptability in long-distance runners: effect of foot strike pattern on lower limb neuro-muscular-skeletal capacity*’ is no more than 100,000 words in length including quotes and exclusive of tables, figures, appendices, bibliography, references and footnotes. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. Except where otherwise indicated, this thesis is my own work”.



15 March 2019

DEDICATION

To the persons I have lost.

To the moments I have missed along the way.

To my family and friends.

That this piece of work may give sense to my absence.

PUBLICATIONS AND PRESENTATIONS

This thesis includes chapters that have been published as the following journal articles:

Chapter 2:

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Chapter 8:

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OTHER PUBLICATIONS

- **Garofolini, A., Taylor, S., McLaughlin, P., Vaughan, B., & Wittich, E.** (2017). Foot strike classification: a comparison of methodologies. *Footwear Science*, 9(sup1), S129-S130.

- **Garofolini, A., Taylor, S., McLaughlin, P., Vaughan, B., & Wittich, E.** (2017). Acute adaptability to barefoot running among professional AFL players. *Footwear Science*, 9(sup1), S44-S45.

- **Garofolini, A., Taylor, S., & McLaughlin, P.** (2017). A proposed experiment: to assess the effect of football boot design on agile movement. *Footwear Science*, 9(sup1), S74-S75.

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- Peacock, J., **Garofolini, A.**, Oppici, L., Serpiello, F., & Ball, K. (2017). Differences in kicking dynamics of futsal and soccer ball. *ISBS Proceedings Archive*, 35(1), 48.

CONFERENCE PRESENTATIONS

- **Garofolini, A.**, Taylor, S., & McLaughlin, P. (2017). Principal component analysis to classify adaptability to barefoot running. ISB Proceedings Archive.
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TABLE OF CONTENTS

ABSTRACT	i
DECLARATION	ii
DEDICATION	iii
PUBLICATIONS AND PRESENTATIONS	iv
ACKNOWLEDGEMENTS	vi
TABLE OF CONTENTS	vii
LIST OF FIGURES	xiii
LIST OF TABLES	xviii
1 Introduction	1
1.1 Preamble.....	1
1.2 The problem	2
1.3 Adaptability and system entropy.....	6
1.4 Context of research design	9
1.4.1 Sample size calculation	9
1.5 Aims	10
1.6 Significance.....	12
1.7 Glossary	13
1.8 References	15
2 The effect of running on foot muscles and bones: A systematic review	19
2.1 Abstract	19
2.2 Introduction.....	20
2.3 Methods.....	23
2.3.1 Search Strategy.....	23
2.3.2 Eligibility criteria.....	23
2.3.3 Coding of studies.....	23
2.3.4 Methodological Quality.....	23
2.4 Results	24
2.4.1 Search results.....	24
2.4.2 Study characteristics.....	25
2.4.3 Sample characteristics	26
2.4.4 Measuring Techniques characteristics.....	28
2.4.5 Methodological quality.....	38

2.5	Discussion	40
2.5.1	Effect on muscles	42
2.5.2	Effect on bones	43
2.5.3	Research limitations	44
2.6	Conclusion	45
2.7	References	47
3 Effect of habitual foot strike on foot musculoskeletal anatomy in long-distance runners		53
3.1	Abstract	53
3.2	Introduction	54
3.3	Methods.....	57
3.3.1	Participants	57
3.3.2	Ultrasound	58
3.3.3	Toe strength test	58
3.3.4	High-Resolution peripheral Quantitative Computed Tomography (HR-pQCT)	58
3.3.5	Data analysis.....	59
3.3.6	Statistical analysis	61
3.4	Results	62
3.4.1	Muscle	62
3.4.2	Bone.....	63
3.5	Discussion	66
3.6	Conclusion	68
3.7	References	69
4 Ankle joint dynamic stiffness in long-distance runners: effect of foot strike and shoe features		73
4.1	Abstract	73
4.2	Introduction.....	74
4.3	Methods.....	77
4.3.1	Participants	77
4.3.2	Experimental protocol	77
4.3.3	Biomechanical Model.....	78
4.3.4	Data analysis.....	80

4.3.5	Statistical analysis	81
4.4	Results	81
4.5	Discussion	93
4.6	Conclusion	95
4.7	References	96
5	The preferred leg joints coordination path in long-distance runners: effect of foot strike and shoe features	100
5.1	Abstract	100
5.2	Introduction	101
5.3	Methods	104
5.3.1	Data Analysis	104
5.3.2	Statistical analysis	106
5.4	Results	107
5.5	Discussion	114
5.6	Conclusion	117
5.7	References	118
5.8	Supplementary A	121
5.9	Supplementary B	122
6	Leg stiffness control in long-distance runners: effect of foot strike and shoe features	123
6.1	Abstract	123
6.2	Introduction	124
6.3	Methods	130
6.3.1	Data Analysis	130
6.3.2	Detrended Fluctuation Analysis	131
6.3.3	Statistical analysis	132
6.4	Results	133
6.4.1	Reconciling control system responsibility for causes of low DFA α	133
6.4.2	Effect of Group and Phase	134
6.4.3	Effect of Shoe	138
6.5	Discussion	140

6.5.1	The DFA-CV results support the first hypothesis that control regulation of leg stiffness involves the interaction of two control systems and this varies with the time-course of stance.	140
6.5.2	The DFA-CV results support the second hypothesis that control regulation of leg stiffness is phase and group dependent.	140
6.5.3	The DFA-CV results do not support the third hypothesis that control regulation of leg stiffness is shoe-dependent.	144
6.5.4	Study Limitations	146
6.6	Conclusion	147
6.7	References	148
7	Limb effector control during the landing phase of running: effect of foot strike and shoe features	154
7.1	Abstract	154
7.2	Introduction	155
7.2.1	Can the measure of GID(par) be used as an indicator of motor abundance and system flexibility?	158
7.3	Methods	161
7.3.1	Data processing and analysis	161
7.3.2	Uncontrolled Manifold formulation	162
7.3.2.3	Step 3: Projecting the joint configuration	164
7.4	Results	167
7.4.1	Variance parallel to the UCM, V_{UCM}	172
7.4.2	Variance orthogonal to the UCM, V_{ORTH}	173
7.4.3	Ratio of variances perpendicular and orthogonal to the UCM, V_{RATIO}	173
7.5	Discussion	174
7.5.1	Redundancy is exploited for leg length and orientation stabilisation	174
7.5.2	Effect of foot strike on GID and GRD	176
7.5.3	Effect of shoes on GID and GRD	177
7.6	Conclusion	179
7.7	References	180
8	Repeatability and accuracy of a foot muscle strength dynamometer	184

8.1	Abstract	184
8.2	Introduction	185
8.3	Methods.....	186
8.3.1	Hardware and software.....	186
8.3.2	Accuracy.....	191
8.3.3	Repeatability and Reliability	192
8.4	Results	193
8.4.1	Accuracy.....	193
8.4.2	Repeatability and reliability	194
8.5	Discussion	195
8.6	Conclusion	196
8.7	References	197
8.8	Supplementary Figure 1	199
9 Evaluating dynamic error of an instrumented treadmill and the effect on measured kinetic gait parameters: implications and possible solutions		200
9.1	Abstract	200
9.2	Introduction.....	201
9.3	Methods.....	202
9.3.1	Stage 1	203
9.3.2	Stage 2	206
9.3.3	Stage 3	207
9.4	Results.....	208
9.4.1	Treadmill frequency response	208
9.4.2	Effect of improved treadmill stiffness.....	210
9.5	Discussion	213
9.6	References	217
10 Conclusions		219
10.1	Summary of results	219
10.1.1	Rearfoot strikers have reduced foot bone density and simpler structural organisation	219
10.1.2	Rearfoot strikers have reduced foot muscle size, tendon thickness, and foot strength.....	219

10.1.3	Rearfoot strikers have reduced ankle stiffness and joint coupling variability.....	220
10.1.4	Rearfoot strikers have reduced control of leg length-force dynamics during stance.....	220
10.1.5	Rearfoot strikers have reduced kinematic synergies of leg length and orientation during impact.....	220
10.2	Executive summary.....	221
10.3	Potential queries for future work.....	222
10.3.1	Does the difference in bone architecture between RFS and FFS result in a different stress distribution along the metatarsus?.....	222
10.3.2	Does the flight phase of running reveal adaptive strategies?	222
10.3.3	Is there a compensatory control between dominant and non-dominant limbs?	223
10.3.4	Can DFA be used to distinguish between the two hierarchical levels of control?	223
10.3.5	Can control of leg stiffness be trained?	223
10.3.6	Can the model used for motor control be linked to physiological processes?	224
10.4	References.....	226
APPENDIX A Published manuscript chapter 2		227
APPENDIX B Published manuscript chapter 8		241
APPENDIX C Published manuscript chapter 9		248
APPENDIX D Questionnaire		256

LIST OF FIGURES

- Figure 1-1** The hierarchical control model 7
- Figure 1-2** Schematic representation of the thesis structure 11
- Figure 2-1** Flow chart of the search strategy. 25
- Figure 2-2** (A) Sample age by weight distribution for all studies but Zhang et al., (2018) who did not report weight but body mass index; (B) training load for studies reporting load as km per week. Solid lines represent the mean of the group. Dotted line is the grand mean. 27
- Figure 2-3** Results summary of the effect of running on foot bones (A) and foot muscles (B). BMC bone mineral content; SOS speed of sound; BUA broadband ultrasound attenuation; BMD bone mineral density; Tb.Th trabecular thickness; Stiff bone stiffness. CSA cross-sectional area; MV muscle volume; Th thickness; PW power; ADM abductor digiti minimi; FDB flexor digitorum brevis; Abd Hal abductor hallucis; IFM intrinsic foot muscles. 41
- Figure 3-1** (A) High resolution peripheral quantitative computed tomography; (B) Example of D_{tot} (average bone density), D_{trab} (trabecular bone density), D_{inn} (inner trabecular bone density), D_{meta} (meta trabecular bone density), D_{comp} (compact bone density); image adopted from Griffith & Genant (2008). (C) Region of interest for calcaneus and first metatarsus. Sequence of 2-dimensional slices are segmented to reconstruct a 3-dimensional model. 60
- Figure 3-2** Cross-sectional area (A) and thickness (B) of the abductor hallucis (ABH), flexor digitorum brevis (FDB), flexor hallucis brevis (FHB), quadratus plantae (QP), gastrocnemius (GAS), soleus (SOL), Achilles tendon (ACH), plantar fascia calcaneal portion (PF1), and plantar fascia middle portion (PF2). 62
- Figure 3-3** Mean and standard deviation of toe flexor strength (normalized to body weight). Comparison between rearfoot strikers (RFS) and forefoot strikers (FFS). Results from individuals are also reported. 63
- Figure 3-4** Results for calcaneus. (A) Exemplar RFS (B) Exemplar FFS (C) Results for density measurements: TV (total volume), BV (bone volume), and BV/TV (bone volume with respect to total volume). For structure measurements: Tb.N (number of trabeculae), Tb.Th (thickness of trabeculae), Tb.Sp (space between trabeculae), and AI (anisotropy index). 64

Figure 3-5 Results for first metatarsal (A) Exemplar RFS (B) Exemplar FFS (C) Results for density measurements: D_{tot} (average bone density), D_{trab} (trabecular bone density), D_{meta} (meta trabecular bone density), D_{inn} (inner trabecular bone density), and D_{comp} (compact bone density). For structure measurements: BV/TV (trabecular bone volume with respect to tissue volume), $Tb.N$ (number of trabeculae), $Tb.Th$ (thickness of trabeculae), $Tb.Sp$ (separation of trabeculae), $Tb.1/N.SD$ (StDev of $Tb.1/N$: Inhomogeneity of trabecular network), and $Ct.Th$ (cortical thickness). 65

Figure 4-1 Example of moment-angle loop for the ankle joint. Adapted from Hamill, Gruber, & Derrick (2014) 76

Figure 4-2 (A) Magnets glued to bony landmarks; (B) Schematic representation of magnets interaction; (C) markers placed over the sock maintaining the same position; (D-F) markers position in the three shoe conditions: Vibram® Five fingers (D), Mizuno® Wave Sonic (E), Mizuno® Wave Rider 21 (F). 79

Figure 4-3 Example of ankle moment-angle relationship for a FFS subject (top) and a RFS subject (bottom) for the normalized stance phase from initial contact (IC) to toe-off (TO). The values for the quasi-stiffness is defined for the three phases of the moment-angle plot: early rising (ERP), late rising (LRP), and descending phase (DP). Thresholds are set to 0.2 ascending moment (Thr.1); 0.95 ascending/descending moments (Thr.2), and to 0.2 descending moment (Thr.3) 82

Figure 4-4 Mean and SD values for ankle joint dynamic stiffness of FFS and RFS for the three phases of stance, in the three shoe conditions. ERP early rising phase, LRP late rising phase, DP descending phase. Shoes conditions are termed as low MI (LOW), medium MI (MED), and high MI (HIGH). 85

Figure 4-5 Ankle moment-angle plot. Group mean profiles comparison for low MI, medium MI and high MI shoes. Insets report linear regression lines between early rising phase (ERP), late rising phase (LRP), and descending phase (DP). 88

Figure 4-6 Mean and SD of ankle plantar flexors work for the three footwear conditions. Values are shown for positive and negative work for FFS and RFS. Dashed line indicates W_{net} , and solid lines signify a statistically significant ($p < .05$) difference. 90

Figure 5-1 Analysis of the spatial variability in one-dimensional (1-D), and multidimensional spaces (2-D, 3-D). 106

Figure 5-2 (A) Three dimensional plot of the mean preferred coordination path for FFS and RFS. Comparison is made between the three footwear conditions: low MI, med MI, and high MI. FC = foot contact; TO = toe off. **(B)** Comparison of mean group within each footwear condition. 108

Figure 5-3 Spatial variance quantification expressed as a function of the stance phase (foot contact – FC to toe off – TO). Results for the one-dimensional analysis **(A)** and for the multidimensional analysis **(B)** are reported. Comparisons are made among the three footwear conditions. 112

Figure 6-1 (A) Schematic virtual leg-spring model used to simulate running with a rearfoot strike pattern, and **(B)** with forefoot striker pattern (Adapted from Birn-Jeffery et al., 2014). Centre of pressure trajectory beneath the shoe is also displayed. **(C)** Comparison of rearfoot loading (solid line) and forefoot loading (broken line) landing types and their ground reaction force changes as a function of leg length. Curves are divided into three task-relevant sub-phases: impact control, loading, unloading. The slope and area features of the graph represent leg stiffness and energy respectively. Leg stiffness is largest during the first sub-phase. The area under the curves represent the potential energy, produced energy, and lost energy during the stance phase. 131

Figure 6-2 Group mean and SD of DFA α values averaged across shoe types for each group, and over the three task-relevant sub-phases of the stance phase. Bar graphs show between-group (FFS vs RFS) differences for average DFA α and average CV across sub-phases and shoe type. * represents significance level $p < .05$; for group \times phase interaction effects, and pairwise comparisons for between group and between phase. 135

Figure 6-3 Group mean and SD represented for each task-relevant phase (K1-K3) and shoe type (LOW, MED, HIGH) for dependent variables: **(A)** DFA α of leg stiffness, **(B)** mean leg stiffness, and **(C)** CV of leg stiffness. 137

Figure 6-4 Conceptual control diagram. Active intervention from the high level controller will cause the DFA α to increase toward anti-persistence if the cost policy is not meet (i.e. too high, too low leg stiffness). If cost policy is meet, despite high movement variability, the high level controller will not intervene but

rather leave the low level controller to exercise its allometric control over the biomechanical state. This will make the $DFA\alpha$ to decrease toward persistence.

145

Figure 7-1 (A) Multi-dimensional manifold represented in 2D space, showing two elemental variables (EV1-2) and one performance variable (UCM, projected as a line). (B) Expanded V_{UCM} , (C) constricted V_{ORTH} , (D) constricted both V_{UCM} and V_{ORTH} .

157

Figure 7-2 Geometric model used to estimate performance variables and joint angles.

162

Figure 7-3 Mean \pm SE ratio values for RFS and FFS groups. Time has been divided in two phases: PRE from 10 frames before foot contact (FC-10) to foot contact (FC); and POST from FC to 10 frames after foot contact (FC+10). Solid lines indicate a statistically significant difference between groups ($p < .05$). * indicates statistically significant difference from zero ($V_{UCM} > V_{ORTH}$). Note: frames correspond to absolute time (mmsec); 1frame = 4mmsec. FC+10 is ~ 15% of stance.

168

Figure 7-4 Mean \pm SE of Variance components parallel (solid lines) and orthogonal (dashed lines) to the linearized UCM. Note: frames correspond to absolute time (mmsec); 1frame = 4mmsec. FC+10 is ~ 15% of stance. Solid lines indicate a statistically significant difference between groups ($p < .05$).

171

Figure 8-1 Overview of the toe flexors strength device: **a** knee-thigh clamping mechanism, **b** carrier, and **c** pulley arrangement

187

Figure 8-2 Schematic of the main foot and phalanges plates. **a** rotary encoder, **b** torsion strain cylinder, and **c** millimetre linear scales

188

Figure 8-3 Labview software interface (**a**) and block diagram (**b**)

190

Figure 9-1 Response of a linear time-invariant system to a sinusoidal input (right). The steady state output (left) depends on the characteristics of the system (FRF).

203

Figure 9-2 GRF archetypal signals with different impact transient properties. The intensity of the loading is low (**A**), moderate (**B**) and high (**C**); IT indicates the Impact Transient.

207

Figure 9-3 Structural components of the instrumented treadmill. Wooden supports were added underneath the lateral sides of the treadmill frame to improve overall

stiffness of the device. Treadmill was resting on the wooden supports instead of on the four legs during the experiment. 208

Figure 9-4 Frequency Response Function test displayed in the Amplitude (**A**) and Phase (**B**) domain. FRF outcomes of the three hammer tests are over-ground sensor (GFS, blue), treadmill sensor (TFS, orange), and treadmill with wood sensor (TWFS, purple). 209

Figure 9-5 Archetypal VGRF signals from over-ground running with low loading (**A**), medium loading (**B**), and high loading (**C**). Archetypal VGRF signal (green) is compared against over-ground model-prediction (GFS blue), treadmill model-prediction (TFS orange), and new treadmill configuration (with wood bearers) model-prediction (TWFS purple). Error for each model is reported for low loading (**D**), medium loading (**E**), and high loading (**F**). 212

LIST OF TABLES

Table 2-1 Characteristics of the included studies.	29
Table 2-2 Methodological quality evaluation using (A) the Downs and Black methodological quality assessment, and (B) the adapted Newcastle-Ottawa Scale.	39
Table 4-1 Primary statistical results for differences between <i>Groups</i> , <i>Shoes</i> , and <i>Slopes</i> for mean ankle stiffness (K_{ankle}), work produced (W_{prod}), work absorbed (W_{abs}), work net (W_{net}), and work ratio (W_{ratio}). ANOVA results are given for main effects and interactions. Statistically significant findings are in bold.	84
Table 4-2 Mean and (SD) for <i>Groups</i> , <i>Shoes</i> , and <i>Slopes</i> for mean ankle stiffness, work [$\text{Nm/kg} \cdot \text{degree} \cdot 100$] produced (W_{prod}), work absorbed (W_{abs}), work net (W_{net}), and work ratio (W_{ratio})	86
Table 4-3 Correlations between moment-angle loop parameters (Spearman correlation coefficient r_s). * represents statistically significant correlations ($p < .05$); ** represents statistically significant correlations ($p < .01$).	92
Table 5-1 Main effects for group, shoe type, and joint coupling, and interaction effects for the coefficient of correspondence (ACC), mean sum of variance and the square root of the sum of squared distances (SSD). For SSD, main effect for shoe comparison instead of shoe is reported. Statistically significant results ($p < .05$) are reported in bold.	109
Table 5-2 Mean \pm standard deviation for the coefficient of correspondence (ACC), and sum of variance [mm^2]. Group comparison for the three joint couples: ankle-knee (AK), ankle-hip (AH), and knee-hip (KH), in each footwear condition.	110
Table 5-3 Mean \pm SD squared root of the sum of squared distances (SSD) group comparison.	111
Table 6-1 Primary statistical results for differences between <i>Groups</i> , <i>Shoes</i> , and <i>Phase</i> for mean leg stiffness, standard deviation (SD), coefficient of variation (CV), and mean DFA α values. ANOVA results are given for main effects and interactions. Statistically significant findings are in bold.	136
Table 6-2 Group mean and (SD) for leg stiffness mean, SD, CV and DFA α values in the three functional phases of impact (K1), loading (K2), and unloading (K3). Comparisons are made among the three shoe type (LOW, MED, HIGH) and pooled data.	139

Table 7-1 Primary statistical results for differences between Groups, Shoes, and Phase for variance parallel to the UCM (V_{UCM}), variance orthogonal (V_{ORTH}), and ratio (V_{RATIO}) for the vertical component (Z) and horizontal component (Y). ANOVA results are given for main effects and interactions. Statistically significant findings are in bold. 170

Table 7-2 Mean \pm standard deviation for variance parallel (V_{UCM}), orthogonal (V_{ORTH}), and ratio (V_{RATIO}) across the three footwear conditions for the vertical (Z) component and horizontal (Y) component. 172

Table 8-1 Validity results for the angle and torque measurements. Difference (Diff) between predicted values and measured are reported; Absolute Average Difference (Abs Avg Diff) is also reported as raw and percentage. Typical error and Coefficient of variation (Coeff of var) are reported as raw and percentage respectively. 193

Table 8-2 Mean (\pm SD) torque produced by toe flexor muscles (in a 30° of dorsiflexion at the MPJ joint) for session one (test) and two (retest). Results reported for Interclass Correlation Coefficient (ICC), within-observation and between-observation variance [Nm]², mean bias, and coefficient of repeatability (\pm CR). 194

Table 9-1 Root mean squared error (RMSE) is reported as a measure of bias. The error of over-ground force platform sensor (GFS), treadmill-installed force platform sensor (TFS), and adapted treadmill (TWFS) are reported for low loading (Low), medium loading (Med) and high loading profiles (High). The average (AVG) is also reported. RMSE is reported as raw values [N], percentage of peak force, and percentage of mean force. Average loading rate (ALR) and Impact peak are reported as percentage change from the archetypal VGRF signals. ALR was computed between 20-90% of impact peak. 211

1 INTRODUCTION

1.1 Preamble

When people move, their nervous system organizes large, redundant (Bernstein, 1967) – or more abundant (Latash, 2012a) – sets of elements (limbs, joints, muscles, etc.) in a task-specific way. Such organization (so-called synergies) (Latash, Scholz, & Schoner, 2007) use all available degrees of freedoms to ensure optimal performance. The neurophysiological control of locomotion depends on the intrinsic biomechanical constraints and conditions presented by both the body's biology and the implicit mechanical task (Chang, 2015). In this thesis, the approach to movement synergies will embody two theoretical frameworks. One of them is the task-specific stability of redundant systems developed as the uncontrolled manifold (UCM) hypothesis (Scholz & Schöner, 1999). The other is the concept of complexity of human movement developed as the fractal scaling of time series (Dingwell, John, & Cusumano, 2010; Peng et al., 1994). This thesis will use the concept of entropy, incorporating both the uncontrolled manifold and fractal hypotheses and the idea of neurophysiological adaptations, illustrated by the results of two experimental studies. In these studies, the anatomical constraints of the foot were first determined, then perturbations of a continuous movement – running – and analysis of variance across repetitive trials were used to explore variability. In conclusion the thesis outlines the implications of this approach for future studies.

1.2 The problem

In recent years, running has increased in popularity worldwide and is currently one of the most popular leisure-time physical activities (Lee, Lavie, Sui, & Blair, 2016; Lee, Lavie, & Vedanthan, 2015). Individuals regularly participate in running not only for competitive or social purposes, but for health reasons. Some of the health benefits of running include a lower risk of obesity, hypertension, dyslipidaemia, stroke, osteoarthritis, and even certain types of cancer (Lee et al., 2017; Lee et al., 2015). Despite its health benefit however, running-related injuries among long-distance runners are very common (Messier et al., 2018).

Over the past forty years, the frequency of injuries has been floating between 15% and 85% without showing a specific trend (Nigg, Baltich, Hoerzer, & Enders, 2015) leading researchers in the field to argue about the origin of those injuries.

Apart from the possible change in demographics of the running population, and an evolving definition of what constitutes an injury, two other possible factors have been proposed and highly researched about: foot strike pattern and footwear (Lieberman et al., 2010). Foot strike pattern refers to the orientation of the foot when it touches the ground. Although a consensus does not exist on a proper classification method (Garofolini, Taylor, McLaughlin, Vaughan, & Wittich, 2017), functionally, runners can be classified as either *rearfoot* strikers, those who produce a dorsiflexion internal ankle moment at landing; or *forefoot* strikers, those who produce a plantarflexion internal ankle moment at landing.

The foot strike pattern is important because it defines the lower extremity mechanics at landing and its progression through the stance phase of running, when external forces are acting on the body (Almeida, Davis, & Lopes, 2015). Rearfoot strikers land with a more dorsiflexed ankle, and the foot lands in front of the body's centre of mass; while forefoot strikers land with a more plantarflexed ankle and the foot lands closer to the body's centre of mass. These differences in foot position and orientation produce a distinct loading pattern in the early part of stance (Boyer, Rooney, & Derrick, 2014). High impact loading forces, typical of rearfoot strikers, have been associated with musculoskeletal injuries (Zadpoor & Nikooyan, 2011) and degenerative processes (Pohl, Hamill, & Davis, 2009). However, evidence is based on retrospective studies that makes it difficult to prove a direct cause-effect relation. Forefoot strikers present lower impact loading forces at landing (Hatala, Dingwall,

Wunderlich, & Richmond, 2013), but the number of injuries per year do not differ between rearfoot and forefoot strikers (Warr et al., 2015). This contrasting evidence justified the interest of researchers toward footwear design as a possible mitigating factor for high impact loadings.

Since their early introduction, running shoes have been designed to address the loading paradigm and to improve stability, but shoe cushioning and stability characteristics have often (although not always, see Malisoux et al. 2016) been proven to be ineffective in lowering running-related injuries (Nielsen et al., 2014; Ryan, Valiant, McDonald, & Taunton, 2011). In the search for an answer, alternative shoe constructs have been proposed which reduces the “material” interface between the foot and the ground to a minimum (Squadrone & Gallozzi, 2009). Minimal shoes have been suggested as promoting a ‘more natural’ foot strike, i.e. forefoot strike (Lieberman, 2012), and in contrast to cushioned shoes that promote a rearfoot strike pattern, minimal shoes are proposed to minimally interfere with one’s “natural” mechanics, and hence promote an optimal way to reduce the risk of injuries in runners (Davis, Rice, & Wearing, 2017). However, the debate is ongoing, and further prospective studies are needed to identify a relationship between injuries and foot strike/footwear characteristics.

Clearly an interaction between foot strike pattern and footwear exists, and in long-distance runners those two elements contribute to the adaptation of the neuromusculoskeletal system, shaping the runners ability to deal with the external environment. Long-term adaptation in running has been widely studied in relation to the adoption of different foot strike patterns and, in parallel, to running with different type of shoes (Bramble & Lieberman, 2004; Hatala, Lieberman, et al., 2013; Lieberman, 2012, 2014; Lieberman et al., 2015; Lieberman et al., 2010; Lieberman, Werbel, & Daoud, 2009; Perl, Daoud, & Lieberman, 2012). However, most of these studies were cross-sectional in nature and focused on metrics and variables related to injury risk and performance without knowing what the body is optimizing (i.e. controlling) and without exploring the inherited complexity of the system controlling those variables.

As the foot is the only part of the human body interacting with the ground, its structure may be the most affected by long-distance running. For instance, an increase in the cross-sectional area of intrinsic foot muscles has been found after 6-months of

running with minimal shoes (Chen, Sze, Davis, & Cheung, 2016). This provides evidences that a certain amount of load is needed in order to tune musculoskeletal tissues during running (Nigg & Wakeling, 2001), but it is unknown how much loading will have an osteogenic and myogenic effect, and how much may become detrimental. For instance, in a cross-sectional study, runners with greater impact magnitudes had fewer injuries compared to a similar group of runners with lower impact magnitudes (Nigg, 1997). The (untested) adoption of a certain foot strike pattern may have explained the different ability to attenuate loading forces expressed by those runners. More recently however, Loundagin, Schmidt, and Edwards (2018) suggested that loading rate has little influence on the mechanical behaviour of foot bones. Despite the increased foot muscle size found in runners after training (Chen et al., 2016) it is not clear if this may have been the result of the adoption of a certain foot strike pattern. Similarly, it is uncertain if running may change foot bone structure. While the external morphology of bones gives important information on function, it is influenced heavily by genetic and ontogenetic factors (Wallace, Demes, & Judex, 2017) that makes interpretation of changes difficult. In contrast, the bone structure (i.e. trabecular architecture) is more sensitive to the applied load (Tsegai et al., 2013), thus it may be more sensitive to a certain foot strike pattern.

As the foot is the first segment in the kinetic chain of the leg, any structural change will translate to a functional adjustment, first at the ankle, then at inter-joint coordination. During landing, ankle joint stiffness is primarily modulated because the moment arm of the ground reaction force is usually larger at the ankle than at the other joints (i.e. knee, and hip) (Farley & Morgenroth, 1999). Habitual rearfoot strikers will experience a different muscle action around the ankle than habitual fore foot strikers (Lieberman et al., 2010). As stabilization of the ankle (joint stiffness control) at landing is critical (Yen & Chang, 2010), a foot strike that is more adaptable will ensure stability. Whether this is achieved through exploitation of elastic structures or via muscle activation may be a function of the foot strike adopted and footwear worn (Fields, Sykes, Walker, & Jackson, 2010).

Along with the ankle, the knee and hip joints work together so that a constant body position is obtained in many joint configurations (Ivanenko, Cappellini, Dominici, Poppele, & Lacquaniti, 2007) – that is, a flexible movement organisation is achieved through intra-limb coordination. Variability is therefore seen as functional to

the task (Bartlett, Wheat, & Robins, 2007) rather than noise (random error) to be minimized (Schmidt, Lee, Winstein, Wulf, & Zelaznik, 2018). The inter-play of multiple joints (coupling) can be explored and explained using spatial measures based on angle-angle plots (Sparrow, Donovan, Van Emmerik, & Barry, 1987) where variability in the cyclograms defines flexibility of the system in organizing the complex and redundant degrees of freedom of the body – called entropy. A distinct foot strike pattern or footwear, will represent constraint at the ankle that will be accounted for by the other joints of the lower limb so that the resultant movement is minimally affected (Nigg, Baltich, Hoerzer, & Enders, 2015). However, how the system organizes (controls) joint coupling is dependent on the cost policy imposed by the control system.

Any anatomical and functional change is inevitably linked to a neural adaptation so that the movements are coordinated and finalized to achieve a task-goal (Latash, 2012b). Two main variables are speculated to be highly controlled during running – leg posture and leg stiffness. While the control of the former has received large attention while walking (Black, Smith, Wu, & Ulrich, 2007; Huang & Kuo, 2014; Kuo, 2007; Verrel, Lovden, & Lindenberger, 2010; Wu, McKay, & Angulo-Barroso, 2009), the latter has only been described through simulations and optimization studies (Bishop, Fiolkowski, Conrad, Brunt, & Horodyski, 2006; Ferris, Liang, & Farley, 1999; Ferris, Louie, & Farley, 1998). However, its control has, as yet, not been quantified. Before presenting the aims of this thesis, it is necessary to clearly define what it is meant by the terms “adaptability” and “system entropy”.

1.3 Adaptability and system entropy

Adaptability can be defined as the complexity (or level of organisation) embodied by the human locomotor control system. Our body is a complex system that has a workspace enabled with an abundance of equivalent solutions (i.e. equifinality) for a given movement problem (Zhou, Solnik, Wu, & Latash, 2014). The complexity of the system can be characterised by its level of entropy; this is a dynamic property that can regress or expand depending upon maturation and experience (Pincus, 1995). For example, it is commonly understood that ageing processes can dissolve many neuro-mechanical properties, functions and interactions that reduce the dimensionality of the system (Lipsitz & Goldberger, 1992; Manor et al., 2010). Alternately, training and experience can preserve and possibly expand system dimensionality through a process of growth adaptation. Hence, the state of entropy can define the expansion, or regression of workspace dimensionality, and this will determine the capacity for neuro-motor abundance. The more adaptive the organism, the more complex the inter-coupled interactions of its highly dimensional constituent components that operate under diverse time scales (Costa, Peng, Goldberger, & Hausdorff, 2003).

The behaviour of the embodied system (neuro-musculoskeletal) is often represented and investigated as a variant of a spring loaded inverted pendulum (SLIP) model (Blickhan, 1989; Ferris et al., 1998). A spring-mass leg with in-series dampener and motor actuator that uses feedforward and feedback information to acquire accurate state estimates of the body and of the environment, in order to plan and select outgoing motor commands required to meet the optimisation policy (i.e. motor goals) of the higher controller – the hierarchical supervisor of the system (**Figure 1-1**).

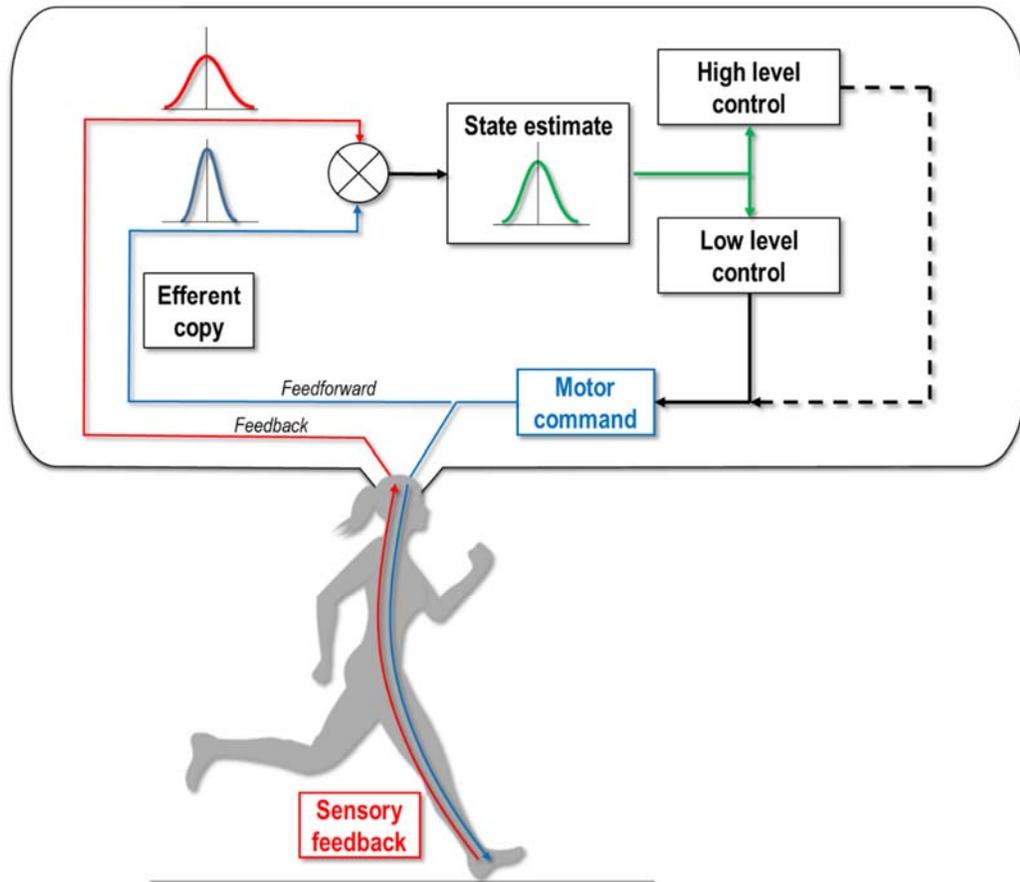


Figure 1-1 The hierarchical control model

The hierarchical control model (**Figure 1-1**) is a combination of the optimal feedback control theory (Todorov & Jordan, 2002) and dynamical system theory (Kelso & Schöner, 1988). The latter deals with the passive organisation of the elemental variables related to the chosen motor command. This low level control allows small variations of the body state away from the attractor state with minimal (if any) intervention because small variations do not destabilize the system. However, continuous variations may accumulate so that the task goal may become compromised. In this case, the high level controller will intervene and actively regulate elemental variables (i.e. constraining segment trajectories) so that the task goal is conserved. The control hierarchy is based on creation of an optimal state estimation combining sensory feedback signals and efferent copy (feedforward) of the motor command. Efferent copy is the prediction of the (un)certainty that the chosen motor command will lead to

(un)stable performance. The cost-policy used by the high level controller is based on the cost-benefit of intervention: it is weighting the energetic cost related with intervention against the cost of allowing variations to happen at that very moment. More complex systems will demonstrate a larger availability of redundant solutions for a given motor task so that intervention from the high control is minimally required. The entropy of that system will therefore be high (Costa, Goldberger, & Peng, 2002). Experienced long distance runners whose lower limb system is subject to frequent forceful impacts, might adapt the entropy of their embodied system by undergoing regression, preservation or expansion. There are two issues related to foot posture and footwear that will influence their state of entropy. First, long distance runners can be categorised into two main groups: those that prefer a rearfoot first foot strike at ground touch down; and those that prefer a forefoot strike (Altman & Davis, 2012; Garofolini et al., 2017; Larson, 2014). This foot strike posture changes the entire biomechanical behaviour of the lower limb system during initial stance phase, and likely influences subsequent tasks through the completion of support phase. Second, the contemporary running shoe is a proposed assistive device that is designed to dissipate impact forces and provide comfort to the runner (Dinato et al., 2015). However, it is unknown how these factors (running pattern and footwear) affect entropy of the neuro-muscular workspace in a long distance runner's embodied system. There are long-term health implications if the system is experiencing regression, rather than preservation or expansion. Therefore, it is important to investigate the effect of footwear and running pattern on system entropy.

The hypothesis of this thesis is that habitual rearfoot strikers running in conventional footwear will show regression of system entropy by evidence of observed adaptations to the following properties of the system:

- Reduced foot bone density and simpler structural organisation
- Reduced foot muscle size, tendon thickness, and foot strength
- Reduced ankle stiffness and joint coupling variability
- Reduced control of leg length-force dynamics during stance
- Reduced kinematic synergies of the leg length and orientation during impact

1.4 Context of research design

This thesis was based on cross-sectional and descriptive research design to compare different groups of runners and the effect of footwear. While claims of cause-effect relationships are avoided, the thesis does use considered language to speculate why differences could exist between groups. The cross-sectional study design tested hypotheses related to differences in neuro-musculoskeletal adaptations between two groups (of ten runners with an antithetical foot strike pattern), and between three different shoe conditions. In this thesis, there are various dependent variables that are used to express neuro-musculoskeletal “adaptation”, and therefore the term is used in a conceptual hypothetical way, and is not empirically proved.

The independent variable of group membership was tightly controlled to enable a degree of confidence when inferring of a cause-effect relationship between foot strike running pattern and expressions of adaptation. Runners were selected based on their training history, running habits, running technique, terrain and habitual footwear conditions (see appendix D). Data collected from the same cohort of runners is used in all the experimental chapters as each chapter investigated adaptability from a unique perspective.

1.4.1 Sample size calculation

Calculations have been based on previous studies (De Wit, De Clercq, & Aerts, 2000; Sinclair, Atkins, & Taylor, 2016) involving experienced long distance runners tested in different footwear conditions (barefoot vs conventional running shoes; minimalistic shoes vs conventional), with reported effect size (f) values of 0.3876 and 0.3905 respectively. For the purpose of this thesis, an a priori power calculation was conducted with the program G*POWER (Faul, Erdfelder, Buchner, & Lang, 2009) using $\alpha=0.05$, and power of 0.8. A total sample size of 20 participants were required to perform ANOVA analysis based on two groups (forefoot loading runners – FFS, and rearfoot loading runners - RFS) and three footwear conditions (high-assisted, medium-assisted, and minimal-assisted).

1.5 Aims

After systematically reviewing the literature to determine the effect of running on foot musculoskeletal properties (Chapter 2), this thesis investigate whether a runner's foot anatomy (bone and muscles) adapts to different foot strike patterns (Chapter 3). Together, chapter 2 and 3 define (i) the philosophical boundaries (what is known) within which we move, and (ii) the biological boundaries – constraints – within which the nervous system is likely to act. At functional levels, this thesis explores how the ankle alone (Chapter 4), or in combination with the knee and hip (Chapter 5) can be affected by footwear and foot strike. The final step is to address the hypothesis that experienced runners with distinct foot strike patterns have developed biomechanical attributes over time that determine their ability to control leg stiffness (Chapter 6) and leg posture (Chapter 7).

This thesis presents findings from a series of studies, divided into four sections: section A determines whether foot structure is affected by long-distance running; section B explores the functional abilities of the lower limb joints; section C examines the control abilities that emerge within the structural and functional constraints defined in the previous sections; and, section D presents the validation of the main instruments used in this study (Chapter 8 and 9). **Figure 1-2** outlines the thesis structure.

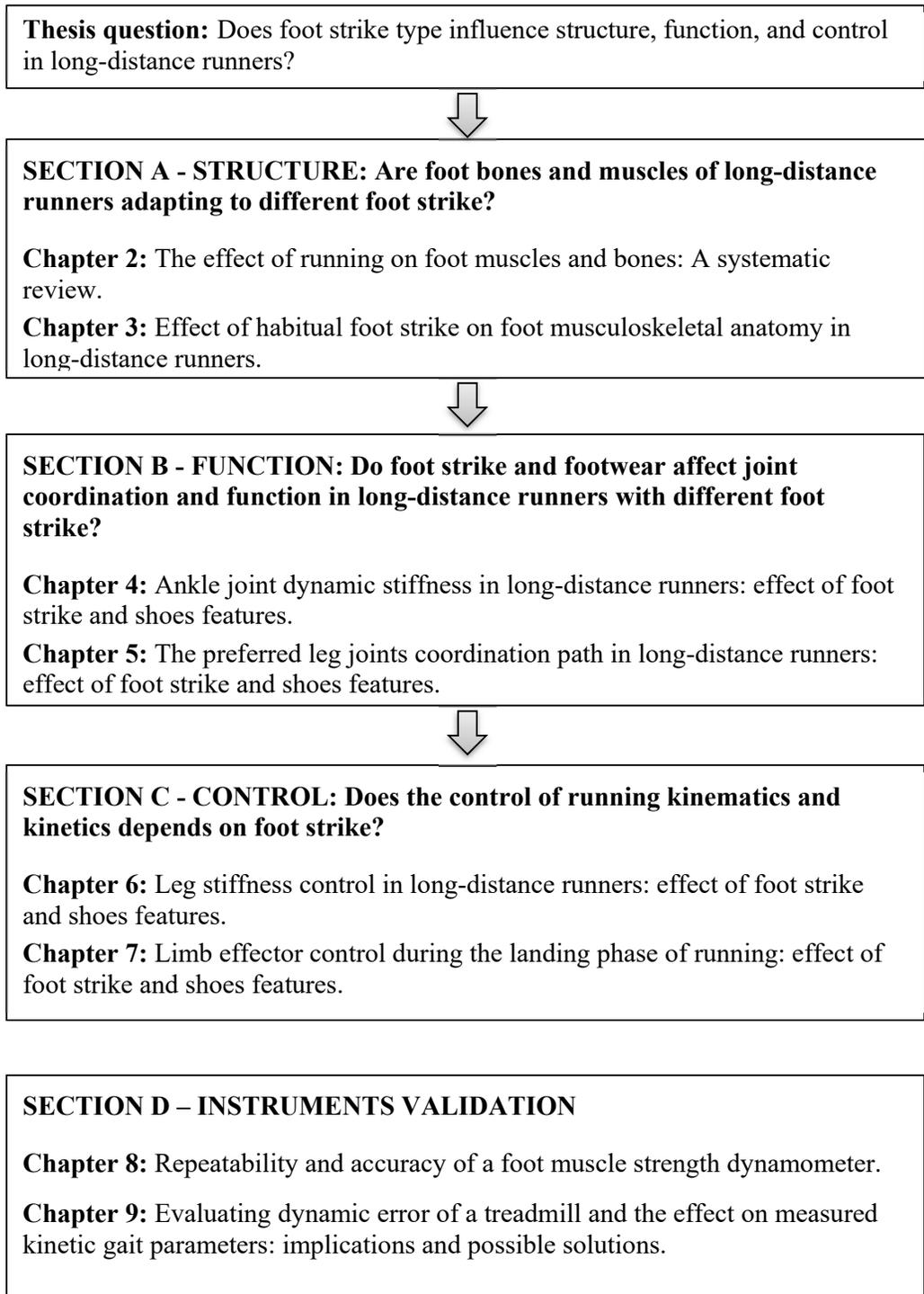


Figure 1-2 Schematic representation of the thesis structure

1.6 Significance

The effects of adopting a consistent foot strike running pattern are not well understood. As running is a world-wide physical activity which millions of people engage in every year (de Almeida, Saragiotto, Yamato, & Lopes, 2015), investigating anatomical and functional adaptations along with the ability of the human body to adapt to different footwear is important to evaluate the long-term effects of running on health, active living and sports performance.

If different shoes constrain foot functions in different ways, movement control is influenced. Impairment in controlling lower limb kinematics and kinetic reflects poor adaptability. From an injury-prevention perspective, defining which combination of foot strike and footwear may enhance adaptability has implications to footwear design, training, and retraining. Similarly, knowing which combination of footwear and foot strike are more likely to be detrimental is also relevant for injury prevention and performance enhancement.

It is hoped that this thesis will be able to explain how running changes the foot's musculoskeletal system, how this may influence (and be influenced by) how running is performed (i.e. foot strike and footwear), and lastly it will help in explaining how anatomical and functional changes affect the control of lower limb kinematics and kinetics, here defined as adaptability.

1.7 Glossary

A list of frequently used, or unfamiliar, terms and their contextual meaning.

Adaptability	The locomotor system embodies a complex level of organised multi-dimensional sub-systems. This enables a rich variation of available motor behaviours that can be selected to accomplish a task-goal with an equivalent outcome.
Complexity	Rich diversity of time-scales among a system's diverse resources.
DFA (detrended fluctuation analysis)	Non-linear time series analysis method used to quantify statistical persistence of a time-varying signal.
Dynamic stiffness	Computed as the slope of the tangent to the moment-angle curve. It can express both: (i) anatomical adaptations that happen in the muscle-tendon units surrounding this joint, and (ii) neural adaptations that control the characteristics of these muscle-tendon units.
Entropy	The change in complexity of the body that can regress or expand depending upon maturation and experience.
FFS (forefoot strike landing pattern)	Runners who tend to land on their forefoot and use internal anatomical properties to control the external impact force.
Functionally relevant phases	Sub-division of the stance phase based on changes in limb or joint stiffness. For dynamic ankle joint stiffness, the phases of stance are divided into early rising (ERP), late rising (LRP), and descending-phase (DP). For effective leg stiffness, the stance phase is divided into impact (K1); loading (K2); and unloading (K3). Functionally, K1-3 refers to the task-goal of stability, safety, and economy respectively.
GID (goal-irrelevant deviations)	An indicator of motor abundance and system flexibility; it represents trials-to-trials fluctuations of the joint configuration that do not cause change to the task-goal (performance).
GRD (goal-relevant deviations)	An indicator of higher-level CNS control over goal-relevant variance behaviour; it represents joint configuration variations consistent with a stable value of the task-goal (performance) variable.

Leg stiffness	Leg force-length dynamics stress-strain property of the leg system components, such as elasticity, hysteresis and energy loss.
Limb effector	A functional system of elements embodied in the limb. A simple kinematic limb effector can be described by a position vector that spans the limb segment components.
Minimalist index	A classification by Esculier et al. (2015) that takes into account structure, flexibility, pronation support, and other footwear features, and ranges from 0% (maximum assistance) to 100% (least interaction with the foot).
Optimal state	When the combined costs of the three major goals of running are minimised (i.e. energy, postural instability and injury risk).
Persistence	An indicator of central nervous system employing a control law leading to the use of a range of equivalent solutions so that deviations of gait parameters are free to persist over time (i.e. repeated trials).
Preferred coordination path	The variable solutions in inter-joint coordination between ankle, knee, and hip that equally satisfy the motor task.
RFS (rearfoot strike landing pattern)	Runners who land on their rearfoot and take advantage of shoe mid-sole material to cushion and control the external impact force.
UCM (uncontrolled manifold theory)	Geometric method used to map the covariance of elemental variables to the performance variable within the same geometric space and units as the performance variable. Variance parallel to the manifold is termed goal-irrelevant, while variance perpendicular to the manifold is termed goal-relevant.

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2 THE EFFECT OF RUNNING ON FOOT MUSCLES AND BONES: A SYSTEMATIC REVIEW

This chapter is an amended version of the manuscript: Garofolini, A., & Taylor, S. (2019). The effect of running on foot muscles and bones: A systematic review. *Human Movement Science*, 64, 75-88. Published version in appendix A.

2.1 Abstract

Despite the widespread evidence of running as a health-preserving exercise, little is known concerning its effect on the foot musculature and bones. While running may influence anatomical foot adaptation, it remains unclear to what extent these adaptations occur. The aim of this paper is to provide a systematic review of the studies that investigated the effects of running and the adaptations that occur in foot muscles and bones. The search was performed following the PRISMA guidelines. Relevant keywords were used for the search through PubMed/MEDLINE, Scopus and SPORTDiscus. The methodological quality of intervention studies was assessed using the Downs and Black checklist. For cross-sectional studies, the Newcastle-Ottawa scale was used. Sixteen studies were found meeting the inclusion criteria. In general, the included studies were deemed to be of moderate methodological quality. Although results of relevant literature are limited and somewhat contradictory, the outcome suggests that running may increase foot muscle volume, muscle cross-sectional area and bone density, but this seems to depend on training volume and experience. Future studies conducted in this area should aim for a standard way of reporting foot muscle/bone characteristics. Also, herein, suggestions for future research are provided.

2.2 Introduction

Running is an important form of exercise because it is inexpensive, accessible, and it provides many health benefits (Lee et al., 2017); however, many of these benefits can only occur through repetitive loading of anatomical structures, and the effect of overload will lead to musculoskeletal injury and non-participation (Nohren, Davis, & Hamill, 2007; Pepper, Akuthota, & McCarty, 2006). Bones and muscles are adaptive tissues that develop in structure and function in response to mechanical load and metabolic demands, which is a demonstration of activity-dependant plasticity (Kiely & Collins, 2016). However, tissue can also be maladaptive. While repetitive load may cause a positive hypertrophic response in bone (J. Chen, Beaupré, & Carter, 2010) and muscles (Seynnes, de Boer, & Narici, 2007); the converse occurs with a reduction (or removal) of load - due to immobilization, physical inactivity, or microgravity exposure – resulting in tissue decay through the process of bone resorption (Holick, 2000; Kiratli, Smith, Nauenberg, Kallfelz, & Perakash, 2000) and muscle atrophy (Powers, Kavazis, & DeRuisseau, 2005). Runners can modulate the nature of the stresses experienced by bone and muscle by altering limb kinematics at impact (Li, Zhang, Gu, & Ren, 2017), or by selecting compliance variations in terrain surface and footwear substrates (Firminger, Fung, Loundagin, & Edwards, 2017); this is because both approaches will effect a change in the direction and magnitude of the external and internal forces applied to the lower limbs. In accordance with activity-dependent plasticity principle, there will exist certain kinematic-substrate combinations that lead to optimal adaptation of foot structure and function and help mitigate injury risk for runners, whereas other combinations will amplify risk. To adequately understand the pathological effect of maladaptive foot structure and function on running injury, a prerequisite step is to first understand the effect of repetitive running load on changes to foot anatomy. The motivation for this review is that this mechanistic effect remains largely unknown due to limited research exploration (Lee et al., 2017).

Repetitive stress injuries are very common among runners, especially stress fractures of the foot (van Gent et al., 2007). Around 55% of these fractures occur in the metatarsals – mostly second and third (Fetzer & Wright, 2006); the calcaneus, talus, navicular and sesamoid account for 6% (Groshar et al., 1997; Pelletier-Galarneau, Martineau, Gaudreault, & Pham, 2015). Long distance runners tend to be afflicted by metatarsal stress fractures more than other athletes (Brukner, Bradshaw,

Khan, White, & Crossley, 1996). This high injury rate might be related to training distance (van Gent et al., 2007), training volume (Hreljac, 2004), and runners' biomechanical adaptations (Davis, Rice, & Wearing, 2017). During running, human locomotor system broadens the distribution of stress that arises from impact forces (Hart et al., 2017) by active modulation of muscle activity (Olin & Gutierrez, 2013) and hence joint torques and rotational energy (Lieberman et al., 2010). Because the foot is the most proximal aspect of the lower limb to the external ground forces, the effect of the stresses will be larger than elsewhere in the lower limb (Lieberman et al., 2010; Daniel E Lieberman, 2012); furthermore, the foot may happen to have the most sensitive anatomy of the lower limb to exhibit activity-dependent plasticity (McKeon, Hertel, Bramble, & Davis, 2014).

Previous studies have shown an increased incidence in bone stress in runners who were transitioning from 'cushioned' footwear to minimal shoes (Johnson, Myrer, Mitchell, Hunter, & Ridge, 2016). The authors found that those who transitioned without negative effects to minimal shoes developed larger adductor hallucis muscles, while those who developed bone stress had smaller foot muscles. Popp et al. (2017) investigated the association between tibial cortical bone density and stress fractures in runners, finding substantially weaker bones in the stress fracture group at the mid-shaft of the tibia. Results from the previous studies (although based on acute interventions) suggest that stronger foot muscles and bones may be protective, while weak feet may be more likely to be injured. However, the long-term effect of the loads generated in the foot bones and muscles during running remains unknown. This knowledge could be used to study the contribution of mechanical load to foot musculoskeletal development and health maintenance, which is essential information for devising methods of injury prevention and treatment.

Measuring bone and muscle adaptations is difficult *in vivo*. Even if bone strength can be approximated by dual-energy x-ray absorptiometry (DXA) (Cummings, Bates, & Black, 2002) and computed tomography techniques (Norton & Gamble, 2001), the problem remains that bone mineral density (BMD) is not the only determinant of bone strength. Innovative 3D analysis of high-resolution images can now provide an insight into bone microstructure and architecture; this technique has shown to be less dependent on bone density than DXA (Geusens et al., 2014), outperforming ultrasound and previous x-ray scanning techniques in terms of image

resolution (up to 82 μm) and level of radiation exposure ($<3 \mu\text{ Sievert}$) (Cheung et al., 2013). Muscles have been imaged by techniques other than conventional radiography, such as magnetic resonance imaging (MRI), and ultrasound scanning. Compared to the former, ultrasound imaging (US) is widely available and rather inexpensive, allowing valid measure of muscle size through real-time high-resolution imaging (Mickle, Nester, Crofts, & Steele, 2013).

The load-related changes (adaptations) in foot muscle and bone may influence more variable running form and biomechanical solutions (Daniel E Lieberman et al., 2015), resulting in minimisation of an accumulation of repeat stresses, however, solid evidence on the effect of running on the anatomical foot structure is needed to perorate this claim. Several original papers (Bobbert, Yeadon, & Nigg, 1992; Bramble & Lieberman, 2004; Bus, 2003; Davis et al., 2017; Gruber, Davis, & Hamill, 2011; Hasegawa, Yamauchi, & Kraemew, 2007; Hunter, Marshall, & McNair, 2005; Kasmer, Wren, & Hoffman, 2014; Lieberman et al., 2010; Daniel E. Lieberman, 2012; D. E. Lieberman, 2014; Daniel E Lieberman et al., 2015; Benno Maurus Nigg, 2010; B. M. Nigg, De Boer, & Fisher, 1995; Shu et al., 2015; Stefanyshyn & Nigg, 1997), as well as systematic reviews (Almeida, Davis, & Lopes, 2015; Hall, Barton, Jones, & Morrissey, 2013; Hollander, Heidt, Van Der Zwaard, Braumann, & Zech, 2017; Perkins, Hanney, & Rothschild, 2014; Schubert, Kempf, & Heiderscheit, 2014) analysed kinematics and kinetics of runners, with only some (Hollander et al., 2017; Shu et al., 2015) reporting findings on the long-term effect of running on foot morphology. The review by Hollander et al. (2017) concluded that habitual barefoot runners have wider feet and a reduced hallux angle than individuals that habitually wear shoes. However, most of the studies included in their review did not control for likely confounding variables such as body weight or running experience. Indeed, any structural change has also to be related to running volume and the amount of time spent resting between runs. Moreover, although they reported changes in foot morphology, the review by Hollander et al. (2017) focused on the differences between barefoot and shod populations, and they did not address adaptations to intrinsic foot muscle or bone. Therefore, the aim of the present paper is to review the evidence regarding the effect of running on foot musculoskeletal adaptations.

2.3 Methods

2.3.1 Search Strategy

A systematic search of the literature was conducted in accordance with the PRISMA guidelines (Moher, Liberati, Tetzlaff, Altman, & Group, 2009). PubMed/MEDLINE, Scopus, and SPORTDiscus databases were used to search for relevant literature from the inception of indexing up to the 1st November 2018. Combinations of the following keywords were used as search: running AND (“foot muscle” OR “foot muscles” OR “bone density” OR “bone strength” OR “bone composition” OR “muscle cross sectional area” OR “muscle volume” OR “foot morphology” OR “foot muscle morphology” OR “muscle strength” OR “foot strength”). Secondary searches were performed by checking the reference list of included articles as suggested by Greenhalgh and Peacock (2005). Forward citation tracking of the included studies was performed in Google Scholar.

2.3.2 Eligibility criteria

Studies were considered eligible if they met the following inclusion criteria: (1) published in English language; (2) published in a peer-reviewed journal; (3) included human participants; (4) used a randomized controlled trial (RCT), a case-control, a prospective cohort, or a cross-sectional study design; (5) measured foot muscle characteristics and/or foot bone characteristics; (6) at least one of the included groups was comprised of active runners. Exclusion criteria were studies reporting on groups or individuals with pre-existing medical conditions, such as metabolic diseases or foot anatomical deformation.

2.3.3 Coding of studies

The following information was extracted from the included studies: (i) sample size; (ii) groups description; (iii) main findings related to muscle/bone characteristics; and (iv) methods used to measure muscle/bone characteristics.

2.3.4 Methodological Quality

Methodological quality of the included intervention studies was assessed using the validated Downs and Black scale (Downs & Black, 1998). For assessing cross-sectional studies, the modified Newcastle-Ottawa Scale was used (Wells et al., 1999).

For the Downs and Black scale, studies scoring from 0 to 8 points were considered as being of poor methodological quality, studies scoring from 9 to 17 points were considered as being of moderate quality, and studies that scored 18 to 27 points were considered as being of high methodological quality. The maximum score on the Newcastle-Ottawa scale is 10 points. Based on the total score on the Newcastle-Ottawa Scale the studies were defined as either low quality (score ≤ 3 points), moderate quality (4-7 points), or high quality (score > 7 points). The datasets analysed during the current study are available from the corresponding author on reasonable request.

2.4 Results

2.4.1 Search results

The initial search resulted with 5487 search results. After the removal of duplicates, 3677 papers were screened, and excluded based on title, abstract, or in some cases, based on the full-text. In total, 41 full-text papers were read. Thirteen studies met the inclusion criteria (Best, Holt, Troy, & Hamill, 2017; T. L.-W. Chen, Sze, Davis, & Cheung, 2016; Escamilla-Martinez et al., 2016; Fredericson et al., 2007; Fuller et al., 2018; Harber, Webber, Sutton, & MacDougall, 1991; Johnson, Myrer, Mitchell, Hunter, & Ridge, 2015; Kersting & Bruggemann, 1999; Laabes, Vanderjagt, Obadofin, Sendeht, & Glew, 2008; Lara et al., 2016; Miller, Whitcome, Lieberman, Norton, & Dyer, 2014; Senda et al., 1999; Zhang, Delabastita, Lissens, De Beenhouwer, & Vanwanseele, 2018). After screening the reference lists of the included studies, three additional studies were included (Drysdale, Collins, Walters, Bird, & Hinkley, 2007; Williams, Wagner, Wasnich, & Heilbrun, 1984). Forward citation tracking of the included studies did not result in the inclusion of additional studies. Thus, the total number of included studies was 16. **Figure 2-1** reports the flow diagram of the search process.

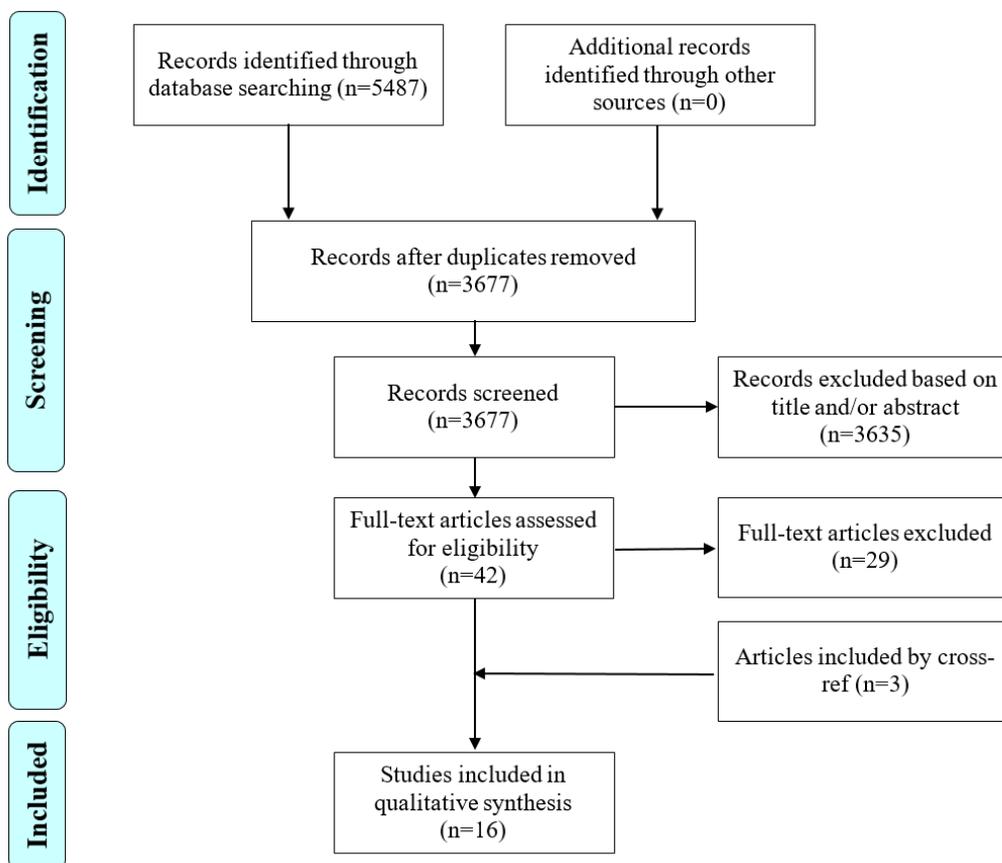


Figure 2-1 Flow chart of the search strategy.

2.4.2 Study characteristics

Ten studies used a cross-sectional design (Best et al., 2017; Drysdale et al., 2007; Escamilla-Martinez et al., 2016; Fredericson et al., 2007; Harber et al., 1991; Kemmler et al., 2006; Laabes et al., 2008; Lara et al., 2016; Senda et al., 1999; Zhang et al., 2018) with a sample size ranged from 11 to 401 (median = 45). Four studies (T. L.-W. Chen et al., 2016; Fuller et al., 2018; Johnson et al., 2015; Miller et al., 2014) used a RCT design, with sample sizes of $n = 20$, $n = 19$, $n = 18$ and $n = 33$, respectively, one study (Kersting & Bruggemann, 1999) used a 20-week long non-randomized intervention ($n = 8$), and one study (Williams et al., 1984) used a 9 month controlled before-and-after study design ($n = 7$). Two of the RCT studies (Johnson et al., 2015;

Miller et al., 2014) were short in duration (10 and 12 weeks, respectively) while the study by Chen et al. (2016) had a 6-month transitioning program.

2.4.3 Sample characteristics

Overall, 624 males and 347 females (mean=39M and 22F; median=20M and 4F) were tested. Eight studies did not include female subjects while two did not include males. Runners ranged on average from 20 to 50 years old (mean=32) and their body weight ranged from 46 to 78 kg (mean= 68) (**Figure 2-2A**). Habitual training volume was quantified as km/week by ten studies (Best et al., 2017; T. L.-W. Chen et al., 2016; Fuller et al., 2018; Johnson et al., 2015; Kemmler et al., 2006; Kersting & Bruggemann, 1999; Laabes et al., 2008; Lara et al., 2016; Miller et al., 2014; Zhang et al., 2018) and was on average 40km/week (ranged from 25 to 69); whilst two studies (Kemmler et al., 2006; Laabes et al., 2008) reported training volume as kcal/kg/day (mean=27±12) and min/week (mean=555±129) respectively, making those studies incomparable with others (**Figure 2-2B**).

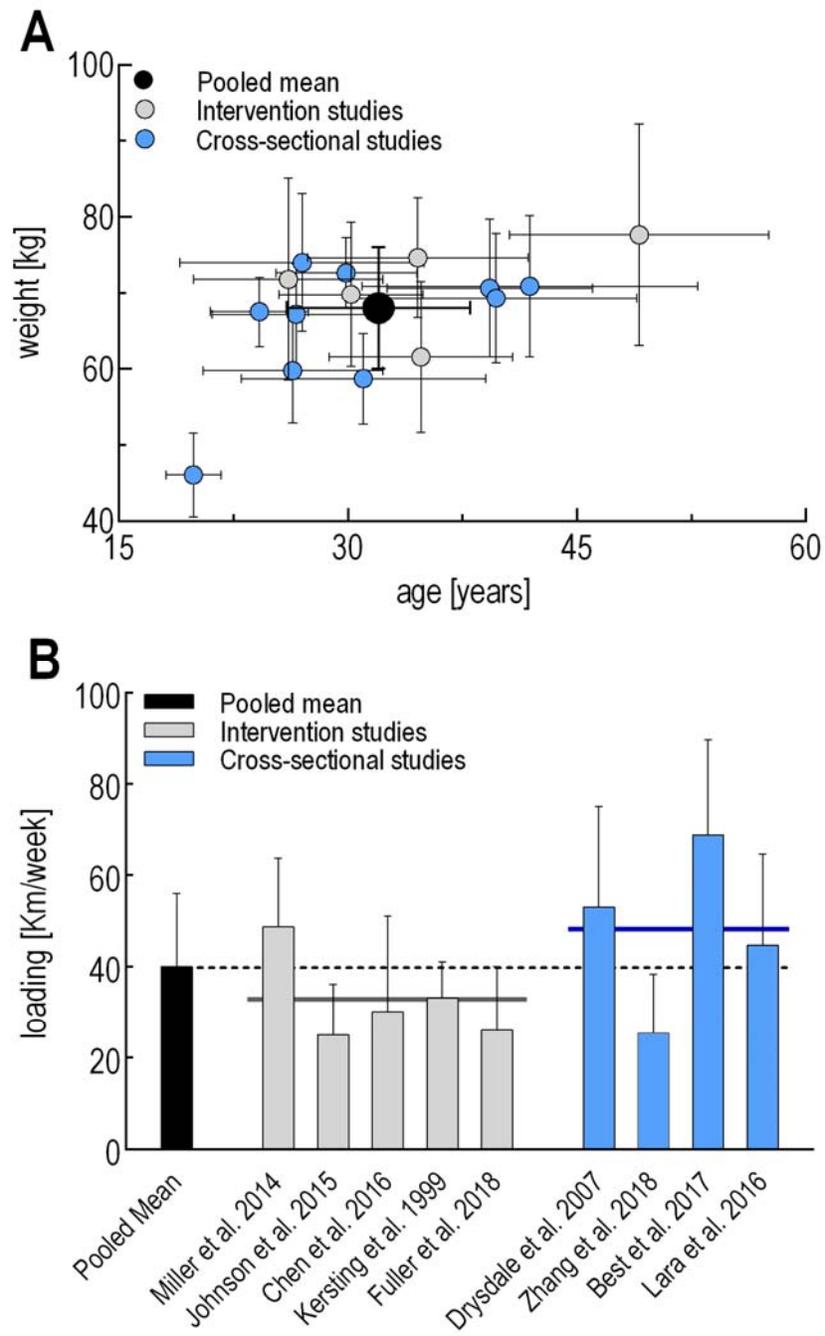


Figure 2-2 (A) Sample age by weight distribution for all studies but Zhang et al., (2018) who did not report weight but body mass index; (B) training load for studies reporting load as km per week. Solid lines represent the mean of the group. Dotted line is the grand mean.

Only three studies (Fredericson et al., 2007; Kemmler et al., 2006; Senda et al., 1999) included elite long distance runners, whose definition was not given by Fredericson et al. (2007); while Senda et al. (1999) defined ‘elite level’ using personal best time for the 3000 m run (mean 9 min and 19 sec.) and Kaup index (14.8-21.9). Kemmler et al. (2006) defined elite runners as those having a running history of at least 5 years and a running volume of 75 km/week and a time of less than 1.15 h for a half-marathon (or <32:30 min for 10,000 m). The other studies involved ‘recreational runners’ whose definition was also inconsistent. For instance, Miller et al. (2014) defined recreational as those who run an average of 30 miles per week (48.3 km) for a minimum of 12 months. Similarly, for Johnson et al. (2015) recreational was defined as an individual who runs an average of 24-48 km/week for the 6 months prior to the start of the study. However, Escamilla-Martinez et al. (2016) defined recreational runners as those who had been distance running as amateurs for at least five years and training at least three times per week with minimum per session duration of one hour.

2.4.4 Measuring Techniques characteristics

Methods used to measure foot muscle or bone characteristics also varied between the studies. Ultrasound-transmission velocity and broadband ultrasound attenuation were the main methods used to quantify bone density. Other techniques reported were photon absorptiometry, compton scattering technique, and peripheral instantaneous x-ray imaging. Only one study, (Best et al., 2017) used high resolution peripheral computed tomography to analyse trabecula characteristics of the calcaneus. For muscle measures, ultrasound and magnetic resonance imaging were most commonly used along with a custom toe dynamometer. **Table 2-1** summarize the details of studies included in the analysis.

Table 2-1 Characteristics of the included studies.

Muscle												
Study	Total sbj	Design	Grouping	Age(y) – BW (kg)	Footwear Foot-strike	Training volume	Intervention duration	Muscle measures		Method	Findings	
Senda et al., (1999)	49	cross-sectional	12 top level marathon runners - 37 healthy control	19.9±1.8y – 46.1±5.5kg	//	//	//	total toe flexors power, abductor power of 1st and 5 th	TD	Running (in conventional running shoes) decreases total flexor power.		
Miller et al., (2014)	33	randomized control study	control (recreational runners; n=16)-recreational runners+intervention (n=17)	30.2±4.7y – 69.8±9.5kg	TRS	48.7±15 km/week	12-week training regime	MV and CSA of the FDB, abductor digiti minimi (ADM), and ABDH	MRI	Running in minimal shoes (with 4 mm offset or less) strengthen the foot.		
Johnson et al., (2015)	37	randomized control study	sex-blocked randomization . 19 control (recreational runners)- 18 recreational	26.1±6.2y – 71.8±13.3kg	TRS	25±11 km/week	10-week transition period	ABDH CSA (cm ²)-FDB CSA (cm ²)-FHB thickness (cm)-EDB	MRI, USI	Significant 10.6 % increase in abductor hallucis cross-sectional area in the Vibram		

		runners+intervention					thickness (cm)		FiveFingers™ group compared with the control group (p = 0.01).
Chen et al., (2016)	38 Randomized, single-blinded control study	control (training program in TRS; n=18)-intervention (training program in MRS+ transition exercises+ transitioning tips; n=20)	34.8±6y – 61.6±9.9kg	TRS (heel-toe drop >5mm)	30.4±21.3 km/week	6-month transition period	IFM volume	MRI	MRS group had significantly larger foot (p = 0.01, Cohen's d = 0.62) muscles after transition. The forefoot mainly contributed to foot muscle growth.
Zhang et al., (2018)	38 cross-sectional	Neutral shoes (n=11); motion control shoes (n=10); minimalistic shoe (n=7); insole (n=10)	26.3±6.9y – 22±2.1 BMI	Mixed shoe models	25.4±13 // km/week		ABDH CSA (mm ²) and thickness (mm)-FDB CSA (mm ²) and thickness (mm)-FHB thickness (mm)	US	Runners in minimal shoes had the thickest abductor hallucis.

Bone										
Study	Total subj	Design	Grouping				Intervention duration	Bone measures	Method	Findings
Williams et al., (1984)	30	controlled before-and- after study	consistent runners (n=7); inconsistent runners (n=13); control (n=10)	49.1±8.5y –	//	//	9 months	Calcaneal bone mineral content	PA	Calcaneal bone mineral content is dependent on training volume. Post intervention, subject training more than 16km per month has significantly (p<0.05) higher bone mineral content than control.

Harber et al., (1991)	42	cross-sectional	Group A (eumenorrheic normoactive females) n=14 subjects who reported 9 or more months per year and who exercised fewer than 3 times per week but did not participate in any formal exercise;	26.4±5.9y	-	//	//	//	Calcaneal density	CST	Amenorrhea in athletes is not associated with any reduction in heel bone density. However, bone turnover rate is significantly greater in athletes.
			Group B (eumenorrheic athletes) n=17 runners who reported 9 or more menses per year and who trained 7—12 times per week.								
			Group C (amenorrheic athletes) n=11 runners who								

reported no
menses in the
last 12 months
and who
trained 7—12
times per
week.

Kersting et al., (1999)	26	non-randomized intervention	3 groups. running shoes of similar construction but different midsole hardness: 45°(n=9), 53°(n=9) and 61°(n=8)	34.6±7.2y – 74.7±7.9kg	RFS	33.8±8.2 km/week	20-week training regime	Calcaneal density	SOS, MRI	No relationship between midsole hardness and external or in-shoe impacts. Bone parameters showed specific differences for all groups which are pronounced in runners with intermediate impacts.
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Kemmler et al., (2006)	31	cross-sectional	Endurance trained male runners (n = 20), BMI-matched control (n = 11) aged 20–35 years.	26.6±5.5y – 67.2±6.7kg	//	555±129 min/week	//	Calcaneal density	SOS, BUA	Runners displayed significantly higher SOS and BUA than control.
Drysdale et al., (2007)	401	cross-sectional	Marathon runners (n = 401; 217 M, 184 F), control group from previous studies (n = 601; 267 M, 334 F).	41.9±11y – 70.9±9.3kg	//	53.8±22.3 km/week	//	Calcaneal density	BUA	The rate of decline of BMD appeared to be reduced significantly in marathon runners compared with the normative group.
Fredericson et al., (2007)	45	cross-sectional	Elite male soccer players (n = 15), elite male long-distance runners (n = 15) and sedentary male controls (n =	24.2±3.2y – 67.5±4.6kg	//	//	//	total regional bone mineral density	and DXA	Running is associated with higher BMD at directly loaded sites (the calcaneus) but not at relatively unloaded sites (the spine).

15) aged 20–30 years.

Laabes et al., (2008)	102	cross-sectional	football (n = 68), running (n = 15), handball (n = 7), taekwondo (n = 6), cycling (n = 2), judo (n=1), badminton (n=1) and high jump (n=1)	31±8y – 58.7±6kg	//	27±12 kcal/kg/d (runners only)	//	calcaneal bone stiffness index	BUA	Repetitive skeletal loading at the heel has the potential to improve bone density in black male athletes. The magnitude of increase may be higher in medium impact sports such as soccer and running compared with low or non-impact sports.
Escamilla et al., (2016)	95	cross-sectional	amateur runners (n=33); control (n=62)	39.3±6.7y – 70.7±9.1kg	RFS	//	//	Calcaneal density	BUA	Distance running seems to have a negative effect on calcaneal bone mass density during the course of a 700-km training season.

Fuller et al., (2018)	39	randomized control study	Minimal shoes (n=19); conventional shoes (n=20)	27±8y – 74±9.1kg	TRS and MRS	26±14k m/week	20-week training regime	Calcaneal and metatarsal (1 st to 5 th) mineral density (g cm ²)	DXA	Minimalist shoes did not affect bone mineral density after 20 weeks follow-up
Best et al., (2017)	18	cross-sectional	FFS (n=6); RFS (n=6); control (n=6)	29.9±4.6y – 72.7±4.6kg	TRS and MRS	68.8±20.9km/week	//	Calcaneal volumetric density, trabecular thickness, number, distance between; DA	HRpQC T	trabecular thickness and mineral density were greatest in forefoot runners with strong effect sizes (<0.80). Trabecular thickness was positively correlated with weekly running distance (r ² = 0.417, p<0.05) and years running (r ² = 0.339, p<0.05). individuals with the greatest summative loading stimulus had, after body

Lara et al., 278 (2016)	cross-sectional	Long-distance runners (n=122); short distance runners (n=81); control (n=75)	39.7±9.2y – 69.3±8.5kg	//	44.7±20k // m/week	Calcaneal bone stiffness	BUA, SOS	mass adjustment, the thickest trabeculae. long distance runners and short distance runners presented higher values than sedentary counterparts in SOS (P < 0.05), and calcaneus stiffness (P < 0.05). However, there were no significant differences between longer distance and shorter distance runners.
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MV muscle volume, CSA cross-sectional area, FDB flexor digitorum brevis, ADM abductor digiti minimi, ABDH abductor hallucis, FHB flexor hallucis brevis, EDB extensor digitorum brevis, TD toe dynamometer, PA photon absorptiometry, CST Compton scattering technique, MRI magnetic resonance imaging, USI ultrasound imaging, TRS traditional running shoes, RFS rear foot strike, FFS fore foot strike, MRS minimalist running shoes. IFM intrinsic foot muscles, SOS speed of sound, BUA broadband ultrasound attenuation, DXA dual-energy x-ray absorptiometry, HRpQCT high resolution peripheral computed tomography, DA degree of anisotropy.

2.4.5 *Methodological quality*

Quality scores for the Downs and Black scale and the modified Newcastle-Ottawa Scale are reported in **Table 2-2**. The RCTs (T. L.-W. Chen et al., 2016; Fuller et al., 2018; Johnson et al., 2015; Miller et al., 2014) had a score ≥ 18 points and were classified as being of high methodological quality. The non-randomized studies (Kersting & Bruggemann, 1999; Williams et al., 1984) scored 10 points and were classified as being of moderate methodological quality (**Table 2-2A**). Eight of the ten cross-sectional studies (Best et al., 2017; Escamilla-Martinez et al., 2016; Fredericson et al., 2007; Harber et al., 1991; Kemmler et al., 2006; Laabes et al., 2008; Lara et al., 2016; Senda et al., 1999) scored between 4 and 7 points on the Newcastle-Ottawa Scale, and, therefore, they were all classified as being of moderate quality (**Table 2-2B**). Only the Drysdale et al. (2007) and Zhang et al. (2018) studies were classified as of high quality (8 points).

Table 2-2 Methodological quality evaluation using (A) the Downs and Black methodological quality assessment, and (B) the adapted Newcastle-Ottawa Scale.

Study	Scale items																										Total	
	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24	25	26		27
Williams et al., (1984)	1	1	0	1	0	1	1	0	0	1	0	0	0 ^a	0	0	1	0 ^a	1	0	0	0 ^a	1	0	0	1	0	0	10
Kersting et al., (1999)	1	1	1	1	1	1	1	0	0	1	0 ^a	0	0 ^a	1	0 ^a	1	0 ^a	0 ^a	0	0	0 ^a	0 ^a	0	10				
Miller et al., (2014)	1	1	1	1	1	1	1	1	0	1	1	0	0 ^a	0	1	0	0 ^a	1	1	1	1	1	1	0 ^a	0	1	0	18
Johnson et al., (2015)	1	1	1	1	1	1	1	1	1	1	1	0	0	0	1	1	1	1	1	0	1	1	1	0 ^a	1	1	1	22
Chen et al., (2016)	1	1	1	1	1	1	1	0	1	1	1	0	0 ^a	1	1	0	0 ^a	1	1	1	1	1	1	1	0	1	0	20
Fuller et al., (2018)	1	1	1	1	1	1	1	1	1	1	1	0 ^a	0 ^a	0	0 ^a	1	1	1	1	1	1	0	1	0 ^a	1	1	1	21

Items 1-10 are related to reporting, items 11-13 are related to external validity, items 14-26 are related to internal validity, item 27 is related to statistical power.

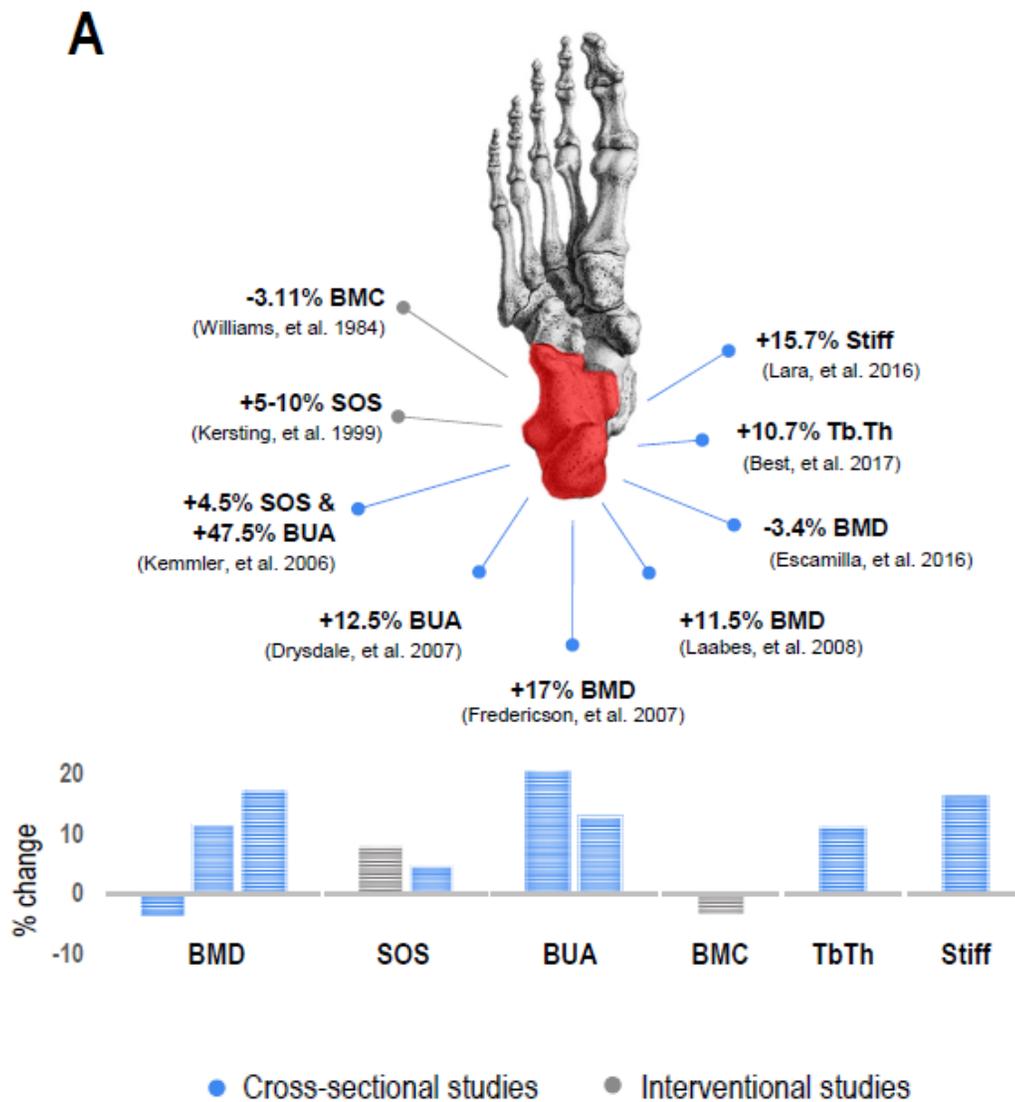
1 criteria met, 0 criteria not met

^a Item was unable to be determined, scored 0

Study	Selection				Comparability	Outcome		Total
	1	2	3	4	1	1	2	
Harber et al., (1991)	0	0	0	2	1	2	1	6
Senda et al., (1999)	0	0	0	1	1	2	0	4
Kemmler et al., (2006)	0	0	0	2	2	2	1	7
Drysdale et al., (2007)	1	0	1	1	2	2	1	8
Fredericson et al., (2007)	0	0	1	1	1	2	1	6
Laabes et al., (2008)	0	0	0	1	1	2	1	5
Escamilla et al., (2016)	0	0	0	1	2	2	1	6
Lara et al., (2016)	1	0	0	1	2	2	0	6
Zhang et al., (2018)	1	1	0	1	2	2	1	8
Best et al., (2017)	1	0	0	1	2	2	1	7

2.5 Discussion

This systematic review summarises findings related to the effect of running on foot muscle and bone characteristics from 16 studies. The current body of evidence on this topic is limited, which highlights the need for future studies. In the next sections, we discuss the most significant findings and provide recommendations for future research in this area. **Figure 2-3** depicts the main findings of this review.



B

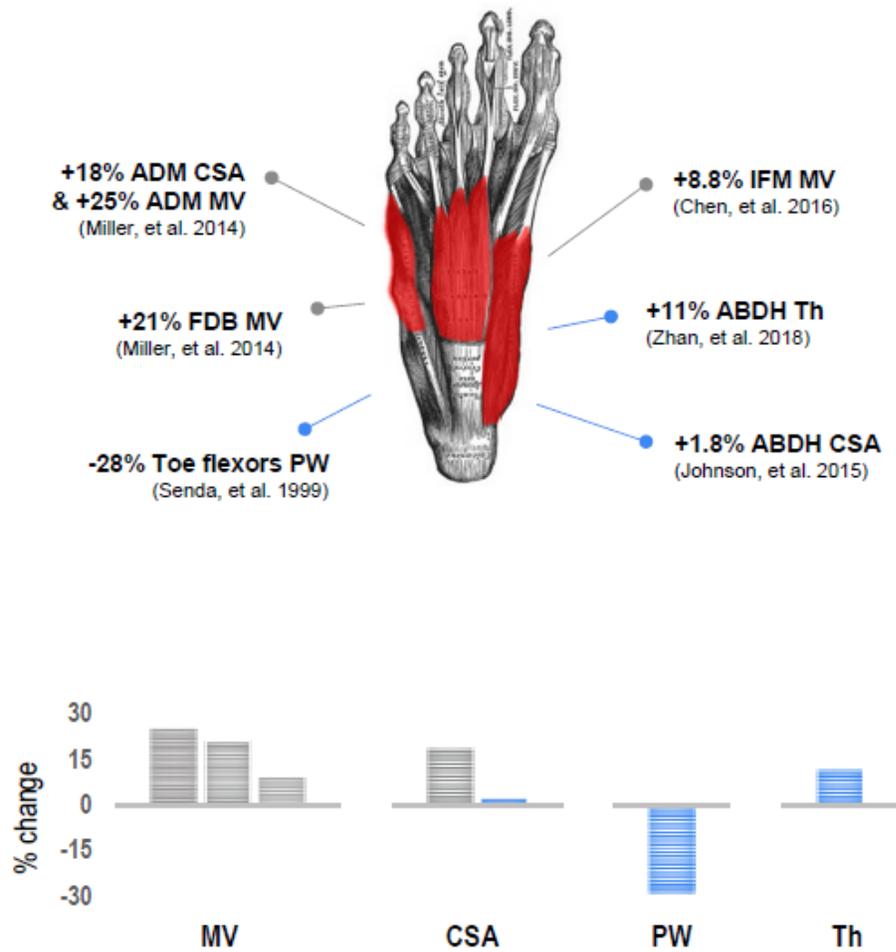


Figure 2-3 Results summary of the effect of running on foot bones (A) and foot muscles (B). BMC bone mineral content; SOS speed of sound; BUA broadband ultrasound attenuation; BMD bone mineral density; Tb.Th trabecular thickness; Stiff bone stiffness. CSA cross-sectional area; MV muscle volume; Th thickness; PW power; ADM abductor digiti minimi; FDB flexor digitorum brevis; Abd Hal abductor hallucis; IFM intrinsic foot muscles.

2.5.1 *Effect on muscles*

Very limited evidence exists indicating that running is associated with increased foot muscle size. T. L.-W. Chen et al. (2016) found a muscle growth (+8.8%, $p = .01$) in intrinsic foot muscles (measured as a whole) after a 6-month transitioning program to minimal shoes. However, a muscle-strengthening program was also part of the intervention, which may partially explain the change in muscle volume. The control group running in traditional shoes showed no change in foot muscle volume after the program.

Short training intervention may be more effective in increasing muscle size. Johnson et al. (2015) reported a significant increase (+10.6%, $p = .01$) in abductor hallucis cross-sectional area after 10 weeks of training in minimal running shoes compared with the change (pre-post) in the control group (+1.8%) who were using traditional running shoes; however, no significant differences were found among all the other intrinsic muscles that were examined. Similarly, after a 12 weeks transitioning period, a +24.7% increase was found in the abductor digiti minimi muscle volume ($p = .009$) and a +18.0% increase in the abductor digiti minimi muscle cross-sectional area ($p = .007$) of recreational runners (Miller et al., 2014). For the other tested muscles no significant differences were found, and furthermore, no statistically significant differences were found between pre-and post-training in the control group running in traditional shoes.

Based on the limited evidence available, there is an indication that intrinsic muscle strength and muscle size may increase with running but this is dependent on type of footwear and the associated biomechanical changes (Davis et al., 2017; Daniel E. Lieberman, 2012). A stronger foot may better control loading redistribution at each step (McKeon et al., 2014) while reduced strength may limit the ability to control inter-joint movements resulting in increased soft tissue strain; therefore, greater foot strength may be a beneficial adaptation in response to the repetitive loading imposed on the foot during running, which may contribute to a decreased incidence of injuries (McKeon & Fouchet, 2015). When controlling for the shoe worn, loading seems to have less of an effect in stimulating muscle growth: while comparing 4 type of running shoes (neutral, motion control, minimalistic, and neutral with insoles), Zhang et al. (2018) found that among all intrinsic foot muscles selected, only abductor hallucis showed a significant difference between groups. Runners using minimalistic shoes had

the thickest abductor hallucis. More cushioning and restrictive design of traditional shoes may neutralize the action of the intrinsic foot muscles making runners relying more on extrinsic foot muscles for loading redistribution (Murley, Landorf, Menz, & Bird, 2009). Muscle imbalance could explain the lower (-28%) global foot power recorded in marathoners compared against a control group (Senda et al., 1999). Long-term, muscle imbalance may cause foot deformity (Kwon, Tuttle, Johnson, & Mueller, 2009) and increase risk of injury (Nigg et al., 2017; Page, Frank, & Lardner, 2010).

2.5.2 *Effect on bones*

A number of studies (Pocock, Eisman, Yeates, Sambrook, & Eberl, 1986; Strobe et al., 2015; Whitfield, Kohrt, Gabriel, Rahbar, & Kohl III, 2015) suggest that increased physical activity can result in an increase in bone mineral density (BMD) in common skeletal loading sites. In long-distance runners the calcaneus showed greater (+17%, $p = .002$) BMD compared with sedentary controls (Fredericson et al., 2007), greater (+3.1%) mineral content in ‘consistent’ (>16 km/month) runners compared with a control group ($p < .05$) (Williams et al., 1984), and greater (+12%) stiffness compared to sedentary counterparts (Lara et al., 2016). Greater (+11.5%) calcaneus BMD was also reported in male runners (sprinters, middle distance and marathoners) when compared with athletes from low or no-impact disciplines; running was a significant ($p < .001$) determinant of BMD and independent of age and body weight (Laabes et al., 2008).

The repetitive high forces generated during running should theoretically increase foot bone density (Hart et al., 2017); Kersting and Bruggemann (1999) speculated that impact forces are constantly, and directly, regulating calcaneal bone adaptations. For example, Kemmler et al. (2006) compared high volume runners (>75 km/week) with BMI-matched controls (≤ 2 h exercise/week) and reported that runners display a significantly higher calcaneal density. Similarly, in a large cross-sectional study involving marathon runners ($n = 401$; 217 men and 184 women) the rate of decline of BMD appeared to be reduced significantly in marathon runners compared with a normative group (Drysdale et al., 2007).

Overall, runners have higher calcaneus BMD than sedentary population; however, due to their continued practice the accelerated bone turnover (Harber et al., 1991) would inevitably decrease bone mass (Hetland, Haarbo, & Christiansen, 1993).

For instance, Escamilla-Martinez et al. (2016) reported distance running to have a negative effect on calcaneal BMD during a 700-km training season in amateur runners ($n = 33$); similarly, Fuller et al. (2018) found no differences ($p \geq .319$) at the 20-week follow-up of a minimalist training intervention. Regular high volume of running may therefore decrease foot bone strength, increasing the risk of osteopenia and/or stress fracture.

2.5.3 *Research limitations*

The main limitations of the included studies are (i) the inconsistency on the dependent variable chosen as a proxy for foot muscles strength, (ii) primarily only one site (the calcaneus) was chosen to investigate foot bone characteristics, (iii) the inconsistency on the methodology used to measure muscles and bone properties, and (iv) the incomplete information regarding the footwear, pattern of foot strike (heel vs. fore foot), physical activity background (training volume) of participants of the studies.

Experimental devices have been designed to measure foot muscles strength (Goldmann & Brüggemann, 2012; Senda et al., 1999); however, no device is able to distinguish between intrinsic and extrinsic muscles. Moreover, other biomechanical factors such as the moment arms of intrinsic foot muscles and muscle-tendon length may also influence the capacity of these muscles to generate force. An accurate measure of intrinsic foot muscles may provide valuable insight into their ability to produce force; however, such a technology still needs to be developed.

Although the calcaneus is considered an important peripheral site for osteoporosis assessment (Frost, Blake, & Fogelman, 2000; Glüer et al., 2004), prediction of the risk of hip fracture (Ross et al., 2000), and often used as a representation of skeletal status (Baroncelli, 2008; Langton & Langton, 2000), foot accounts for 26 bones with a unique shape that varies the magnitude and direction of the load they are subjected to. The choice of the calcaneus as an indicator of bone characteristics is questionable as this bone seems to be less affected by stress fractures than others. For example, the evidence indicates that sites of high risk stress fractures include the tarsal navicular, base of the fifth metatarsal, talus, base of the second metatarsal, sesamoids, and medial malleolus (Boden & Osbahr, 2000). While low-risk fractures in the foot and ankle include the calcaneus, and the second through fifth metatarsals (Boden, Osbahr, & Jimenez, 2001).

Moreover, bone density is only a proxy of bone strength that also depends on bone geometry, bone quality (metabolism and collagen cross-linking), cortical and trabecular morphology (Ammann & Rizzoli, 2003; Saito et al., 2010; Seeman, 2008). Only one study (Best et al., 2017) investigated trabecular characteristics using high resolution peripheral quantitative computed tomography – HR-pQCT; they found trabecular thickness to be positively correlated to weekly running distance ($r^2 = 0.417$, $p < .05$) and experience ($r^2 = 0.339$, $p < .05$). Clearly, more study of other foot bones and their specifics, other than density, may unveil new perspective on the effect of running on foot bones. Furthermore, bone density is not only influenced by mechanical external stresses (i.e. physical activity level), but also by age, diet, hormonal characteristics and genotype (Herbert et al., 2018), these internal physiological mechanisms together are suggested to explain around 50–85% of bone density; it is therefore important for future studies to consider those possible confounding variables when seeking to explain the effect of exercise (i.e. running) on bone density.

Finally, no standard protocols to investigate foot muscles and bones characteristics have been developed that would allow comparison between studies. These limitations could be addressed in future. Besides the comparison of runners and nonrunners, it would be interesting to compare foot anatomical characteristics in individuals with similar running experiences (i.e. weekly mileage and years of running) but different footwear choices. Despite the generalized perception that running is good for health, there are still questions that need to be answered: what is the impact of running on foot health? Do the shoes worn affect the potential benefits associated with running?

2.6 Conclusion

The present review systematically appraises the current level of knowledge on the effect of running on foot anatomical structures. Due to the moderate-quality and small sample size (and possible low statistical power) of the majority of the included studies, caution must be used when attempting to generalize their results to the wider population. The limited body of evidence suggests that running may increase foot muscles size and calcaneal BMD, but this seems to depend on training volume, running experience, and footwear.

The lack of details on the shoes worn by participants involved does not allow any inference on the contribution of footwear (and the associated biomechanical changes)

on foot anatomical adaptations. It is evident that the role of footwear in ‘modelling’ the foot has not received enough attention and further experimental investigations are warranted. Future research should therefore, more closely, examine the links between running and foot musculoskeletal adaptations.

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3 EFFECT OF HABITUAL FOOT STRIKE ON FOOT MUSCULOSKELETAL ANATOMY IN LONG-DISTANCE RUNNERS

3.1 Abstract

There is an ongoing debate about whether, or not, running with a rearfoot strike pattern may increase the risk of injury while a forefoot strike pattern may prevent them. Although a large body of evidence exists on biomechanical differences between foot strike patterns, whether adopting one or the other foot strike pattern may prevent or enhance long-term anatomical foot adaptations is still unknown. Using ultrasound imaging and a novel toe flexor strength dynamometer, we quantified differences in intrinsic foot muscle size (cross-sectional area and thickness), and toe flexor strength in two groups of runners with an antithetical foot strike pattern – rearfoot strikers (n = 11) versus forefoot strikers (n = 12). We found no differences in muscles size and toe flexor strength, indicating that habitual foot strike does not affect the size of intrinsic foot muscles and their ability to produce flexion force around the metatarsophalangeal joint. We also investigated foot bone microstructure using a high resolution peripheral tomography in a subset of participants (n = 10). Results suggest rear foot strikers have a lower trabecular area (-67%; p = .003) but similar cortical area (-7%; p = .30) in the first metatarsal compared to forefoot strikers, while no differences between groups were found in the calcaneus. This suggests habitual rearfoot strikers have similar bone strength but lower bone elasticity in the metatarsals compared to forefoot strikers. Our findings add to the current knowledge on the effect of running on health and adaptation of the human foot, and footwear companies, as well as coaches may benefit from implementing this evidence into their practice in order to improve runners' health and performance.

3.2 Introduction

Although running is a popular physical activity with well-established health benefits (Lee et al., 2014), repeated high-magnitude forces are exchanged between the ground and the foot during each step (Cavanagh & LaFortune, 1980). The nature of the loading experienced by the foot at impact may vary substantially depending on the kind of footwear worn (Lieberman et al., 2015), and the type of foot-to-ground strike adopted (Shih, Lin, & Shiang, 2013). Variations in loading of the foot has implications on the evolution of foot function and structure when subjected to extended periods, or high volume, of running. However, our understanding of the long-term effects of loading on the physiological function of foot bones and muscles is rather limited (Canciani et al., 2015), and there is a clear paucity of comprehensive studies on the effect of running on the musculoskeletal health of a runner's foot (Garofolini & Taylor, 2019); see Chapter 2.

Foot strike influences the capacity of foot muscles to produce torque between bone segments of the foot (Kelly, Farris, Lichtwark, & Cresswell, 2018), changing the direction and magnitude of stress applied to the foot bones. The majority of runners take advantage of mid-sole shoe structure and cushioning by adopting a rearfoot strike pattern (RFS) – landing on the heel – to control foot impact forces (de Almeida, Saragiotto, Yamato, & Lopes, 2015; Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, et al., 2010). Over time, a rearfoot strike runner is likely to develop a reliance upon the extrinsic mechanical properties that the shoe mid-sole provides (Davis, Rice, & Wearing, 2017) and subsequently undergo anatomical adaptations relevant to these loading conditions. In contrast, runners who tend to land on the ball of the foot – forefoot strikers (FFS) – rely less on the shoe properties (Davis et al., 2017) and utilise foot biological properties to control impact forces (Hashizume & Yanagiya, 2015). Over a long period of time, it is hypothesized that the foot muscle-tendon units will alter their ability to exert contractile osteogenic force onto the bone depending on the foot strike adopted (Cianferotti & Brandi, 2014; Hart et al., 2017), and this will in turn redefine the structural strength of the bone (Hart et al., 2017).

The difference in the nature of forces applied to the foot between a rearfoot and a forefoot strike at impact is well established (Lieberman et al., 2015; Yong, Silder, & Delp, 2014). When a runner shifts from a reliance on intrinsic anatomical structures to an extrinsic device (the shoe) (and vice versa), the property of bone and muscle will

remodel itself (Ireland, Rittweger, & Degens, 2014; Ireland, 2015). However, we do not know the precise extent of this adaptation (Hamill & Gruber, 2017). Given the importance of foot health (Mickle, Munro, Lord, Menz, & Steele, 2009; Mickle, Munro, Lord, Menz, & Steele, 2011), it is surprising that the effect of landing technique on the foot anatomy is still unknown.

The magnitude and location of the external ground reaction force may change the recruitment of muscles around a joint (Dorn, Schache, & Pandy, 2012). In addition, the position of the foot at landing may affect the ratio between the moment arm of the resultant ground reaction force and the moment arm of the intrinsic foot muscle force (gear ratio) (Carrier, Heglund, & Earls, 1994). While rearfoot strikers do not rely on intrinsic foot muscles to control impact forces, forefoot strikers, by comparison, may recruit intrinsic foot muscles earlier and to a greater extent (Riddick, Farris, & Kelly, 2019). In a recent study, sprinters (known to adopt a forefoot strike pattern) (Wood, 1987), were found to have more developed foot muscles than non-sprinters (Tanaka et al., 2018) arguably due to greater muscle activity during sprinting. However, sprinting is only one mode of running. At the other end of the spectrum there are millions of people engaging in long-distance running (Running-USA, 2016). It is necessary to be able to distinguish whether foot intrinsic muscles develop because of running or whether how running is performed influences foot intrinsic muscles.

If a type of foot strike pattern induces anatomical maladaptation in foot bones and muscles, this may affect the ability of foot bones to resist fracture. Although bone is designed to meet the mechanical loading we face in everyday life and in athletic contexts, high volume of running may prevent proper development of foot structures leading to increased risk of injury (Hart, Nimphius, Weber, Dobbin, & Newton, 2013). With aging, foot muscle weakness will increase the risk of falls (Mickle et al., 2009), reduce mobility and thus quality of life (Moreland, Richardson, Goldsmith, & Clase, 2004). It is important therefore to be able to evaluate foot muscle morphology and bone mechanical properties such as bone density, and structural organisation (i.e. trabeculae number, thickness, and anisotropy).

The aim of this study was to investigate differences in foot muscles and bone characteristics between forefoot and rearfoot strikers. We expected RFS to have smaller cross-sectional area and thickness of intrinsic foot muscles, and consequently they will be able to produce less flexion force compared to FFS. Based on previous

findings that reported bone structure to change in response to different loading conditions (Wallace, Kwaczala, Judex, Demes, & Carlson, 2013; Wallace, Demes, & Judex, 2017), we expected RFS to have a lower bone mineral density and a less organised bone structure in the metatarsus than FFS; while both groups will have similar bone characteristics at the calcaneus.

3.3 Methods

3.3.1 Participants

Forty male long-distance runners volunteered to take part in this study. Participants were excluded if they had not been running for at least 5 years, with an average of at least 40 km/week, and had not been free of neurological, cardiovascular, or musculoskeletal problems within the previous six months. After passing the exclusion criteria, 23 runners (age: 31.2 ± 6.9 yrs, height: 1.77 ± 0.07 cm, weight: 73.4 ± 7.9 kg) were eligible to participate and provided informed consent prior to data collection. Participants were classified as rearfoot strikers (RFS, $n=11$) or forefoot strikers (FFS, $n=12$) based on their habitual foot strike tested on an instrumented treadmill (AMTI Pty, Watertown, MA, USA) at their preferred running speed wearing their habitual running shoes. After a standardized 7-minute progressive warm-up and accommodation period, participants ran for 3 minutes at their preferred running speed identified using a similar approach as Jordan, Challis, and Newell (2007). In brief, starting at low speed, the investigator gradually increased the speed until the participant reported they were running at a speed that was no longer comfortable (too fast) if running continuously for 1 hour. The speed was then gradually decreased until the participant reported they were running at a speed that was no longer comfortable (too slow) if running continuously for 1 hour. This procedure was then repeated (maximum three times) until reaching stable high and low speeds. Then the average speed was computed and reported as preferred running speed.

Habitual foot strike was based on data collected in the last minute of 3-min running by computing the time integral of the joint ankle moment during initial impact ($0.2 - 1$ body weight - BW) on the vertical component (GRF_v) of the ground reaction force. Runners who displayed a positive (dorsiflexor) moment for at least 90% of the analysed period were classified as rearfoot strikers (RFS); conversely, runner who displayed a negative (plantarflexor) moment for at least 90% of the analysed period were classified as forefoot strikers (FFS). This classification method has been proposed to be more closely aligned with the function of the ankle muscles compared to conventional methods (Garofolini, Taylor, McLaughlin, Vaughan, & Wittich, 2017).

3.3.2 *Ultrasound*

Scans were performed on the dominant stance limb (i.e. best performing leg on a single-leg dynamic balance test (Plisky et al., 2009)) using a B-mode ultrasound (Philips CX50, Netherlands) with a 12-3 MHz linear array transducer (38 mm aperture). An experienced examiner (KJM) took all scans and was blinded from participants' group assignment. A standardised protocol (Mickle, Nester, Crofts, & Steele, 2013) was used to measure cross-sectional area (CSA) and thickness of the following intrinsic toe flexors muscles: abductor hallucis, flexor hallucis brevis, flexor digitorum brevis, quadratus plantae. In addition, we measured the thickness of the plantar fascia (proximal and mid portions), Achilles tendon, gastrocnemius (medial head), and soleus. Depth and gain of scans were adjusted to obtain satisfactory definition of muscle contour. Three measurements were taken at each site.

3.3.3 *Toe strength test*

Toe strength was measured using a custom-made dynamometer that we previously validated (Chapter 9). The test-retest reliability (ICC, bias, repeatability coefficient) was determined using data from 10 young subjects (7 men, and 3 women) tested twice within a week (at least one day apart) for maximal toe flexor strength (0.99, -1.13 Nm, 3.9). In brief, participants sat on a chair with their knee and ankle fixed at 90 degrees on the dynamometer. After a pre warm-up period of 1 min, the metatarsal-phalangeal joints (MPJs) were fixed at 30 degrees of dorsiflexion. In this position, participants performed a series of submaximal isometric contractions with incremental exertion up to maximal contraction. After a rest period, three 5 second-maximal contractions were performed.

3.3.4 *High-Resolution peripheral Quantitative Computed Tomography (HR-pQCT)*

A sub-set of participants (5 x RFS, 5 x FFS) underwent a HR-pQCT scan of the foot (calcaneus and first metatarsal) on the dominant leg only. The participants were selected based on their habitual foot strike angle. The most extreme subjects were selected for comparison. Scans occurred at the Department of Medicine, Austin Health (Melbourne). Participants sat in a chair with their foot positioned in the carbon fibre foot cast normally used for a distal tibia scan. The foot was positioned with the ankle maximally plantar-flexed for the scan of both the calcaneal bone and the first

metatarsal bone. Scans of the calcaneus were obtained between the posterior part of the calcaneal tuberosity (Achilles tendon attachment) and the distal part of the plantar fascia attachment as suggested by Metcalf et al. (2017). Scans of the first metatarsal bone were obtained between the proximal end (base) and the distal end (head) of the metatarsus.

Adaptation of trabecular bone to different force directions is described in terms of the extent to which trabeculae are aligned into one or more direction (anisotropy index - AI) (Hildebrand & Rüegsegger, 1997), the number of trabeculae present, and their thickness (Dougherty & Kunzelmann, 2007). The geometry (shape) of trabeculae was also investigated because plate-shaped trabeculae have been shown to develop primarily in joint regions that sustain high mechanical loads, whereas rod-shaped trabeculae tend to develop in regions that experience lower magnitude loads (Ding, Odgaard, Linde, & Hvid, 2002).

3.3.5 Data analysis

Ultrasound images were stored and transferred to a computer for measurement. Cross-sectional area (cm^2) and muscle thickness (cm) were measured using Image J software (National Institute for Health, Bethesda, MD, USA). The mean values (three images) of each site were used for data analysis.

For the toe strength test, raw data were filtered using a 101-point (2s) moving average and the highest torque value among the three maximal exertion trials was used for analysis.

Bone structure was evaluated using high-resolution peripheral quantitative CT (Xtreme CT, Scanco Medical AG, Brüttisellen, Switzerland) (Figure 1A), which had an isotropic voxel size of $82\mu\text{m}$. Attenuation data were converted to equivalent hydroxyapatite densities. For the calcaneus, a volume of interest of 160 mm^3 (50 slices X 0.0082 mm X 400 mm^2) was selected starting from the inner cortical border of the posterior border (i.e. Achilles tendon attachment) going forward (**Figure 3-1**). The volume of interest for the metatarsal was selected from mid shaft going longitudinally forward (24 slices) and backward (25 slices) (**Figure 3-1**). Bone volume was then separated into cortical and trabecular regions with a threshold-based algorithm (Laib, Häuselmann, & Rüegsegger, 1998), so that from the total bone density (D_{tot}), the compact bone density (D_{comp}) and the trabecular bone density (D_{trab}) can be

separated (**Figure 3-1**). Dtrab was then sub-divided into meta-trabecular bone density (Dmeta) and inner trabecular bone density (Dinn).

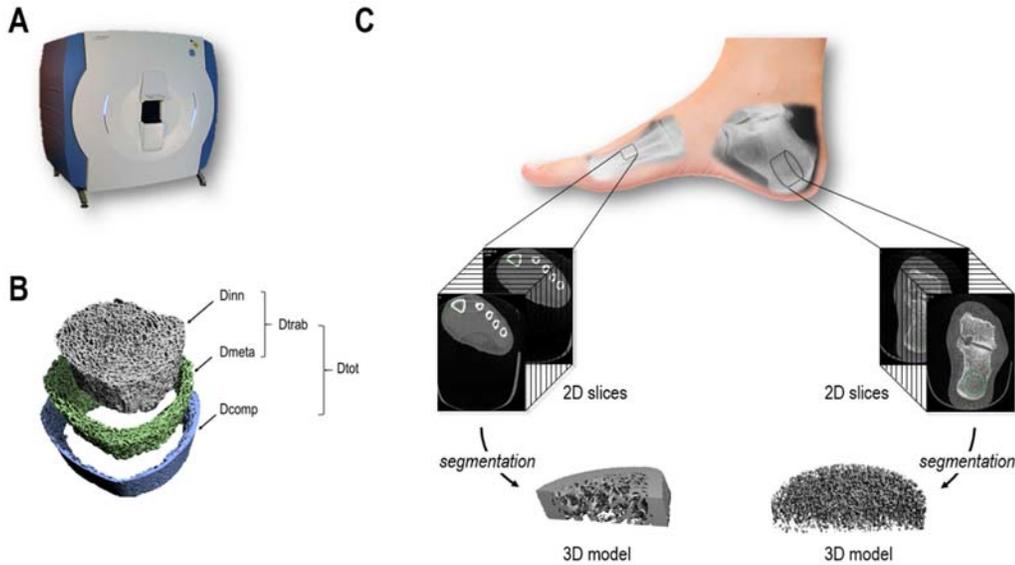


Figure 3-1 (A) High resolution peripheral quantitative computed tomography; (B) Example of Dtot (average bone density), Dtrab (trabecular bone density), Dinn (inner trabecular bone density), Dmeta (meta trabecular bone density), Dcomp (compact bone density); image adopted from Griffith & Genant (2008). (C) Region of interest for calcaneus and first metatarsus. Sequence of 2-dimensional slices are segmented to reconstruct a 3-dimensional model.

The following measurements were extracted from the images: average bone density (Dtot), trabecular bone density (Dtrab), meta trabecular bone density (Dmeta), inner trabecular bone density (Dinn), compact bone density (Dcomp), total volume (TV), bone volume (BV), bone volume with respect to total volume (BV/TV), trabecular number (Tb.N) as the inverse of the mean distance between the mid-axes of the trabeculae using 3D distance transformation (Laib & Rügsegger, 1999b). Derived trabecular thickness (Tb.Th) and separation (Tb.Sp) using plate-model assumptions (Laib & Rügsegger, 1999a). The StDev of Tb.1/N: Inhomogeneity of trabecular network (Tb.1/N.SD), and cortical thickness (Ct.Th). The anisotropy index (AI) was calculated as $1 - (\tau_1/\tau_3)$, where τ_1 , τ_2 , τ_3 are eigenvalues for the three eigenvectors representing the orientation in 3D space of the primary, secondary, and tertiary

material axes. As such, possible values for AI are confined between 0 (perfect isotropy) and 1 (perfect anisotropy) (Doube et al., 2010). Values of AI close to 0 can describe either a volume with numerous thin trabeculae that are randomly oriented or a volume that is completely filled with bone, both morphologies resulting in a lack of dominant orientations (Su, Wallace, & Nakatsukasa, 2013).

3.3.6 *Statistical analysis*

Mean and standard deviation were calculated for each muscle and bone dependent variables. For the muscle, independent t-tests were performed to assess for significant differences between RFS and FFS for muscle CSA, thickness, and toe flexor strength. For bone, a two-way ANOVA with within-factor *Bone* (two levels: calcaneus, metatarsus), and between-factors *Group* (two levels: RFS, FFS) was used to assess differences in each dependent variables of bone density and structural complexity. Level of significance was set at .05 in all statistical analyses. All statistics were performed using SPSS software (version 25, SPSS Inc., Chicago, IL, USA).

3.4 Results

3.4.1 Muscle

Contrary to what was expected, no statistically significant differences were found in CSA for ABH ($p = .261$), FDB ($p = .284$), FHB ($p = .451$), or QP ($p = .354$) between RFS and FFS (**Figure 3-2A**). Likewise, RFS and FFS had similar ($p = .193-.897$) muscle thickness, and similar ($p = .704-.926$) tendinous structure (plantar fascia and Achilles tendon) thickness (**Figure 3-2B**). There were no significant differences ($p = .974$) between groups for toe flexor force (**Figure 3-3**).

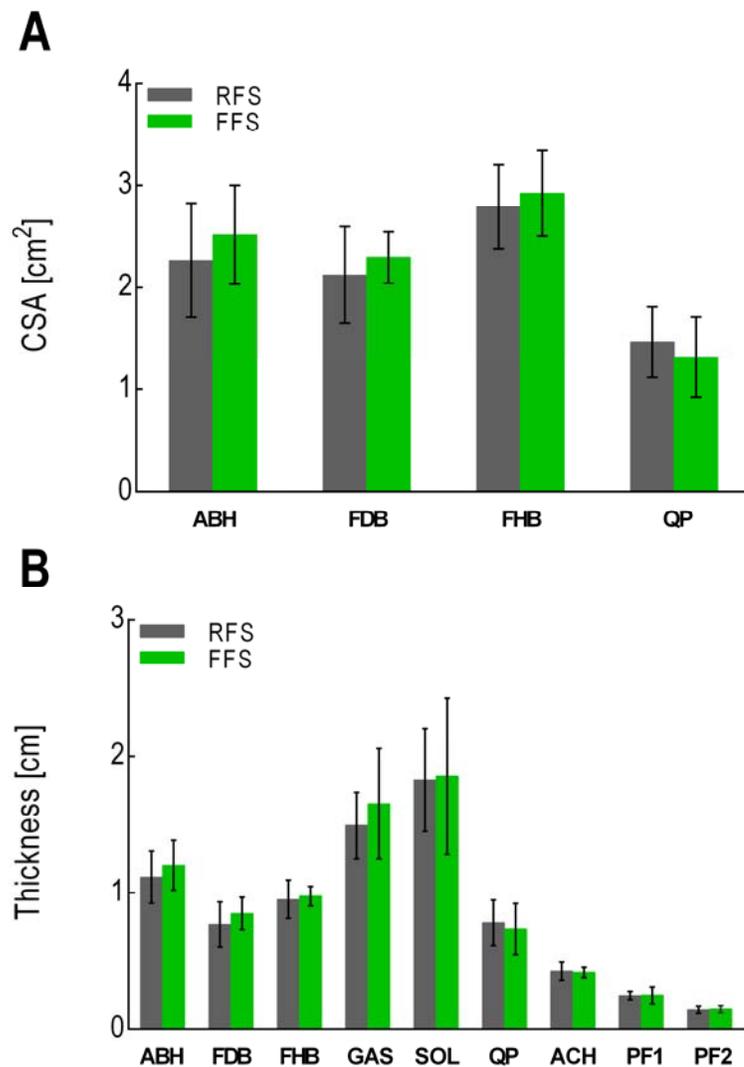


Figure 3-2 Cross-sectional area (**A**) and thickness (**B**) of the abductor hallucis (ABH), flexor digitorum brevis (FDB), flexor hallucis brevis (FHB), quadratus plantae (QP), gastrocnemius (GAS), soleus (SOL), Achilles tendon (ACH), plantar fascia calcaneal portion (PF1), and plantar fascia middle portion (PF2).

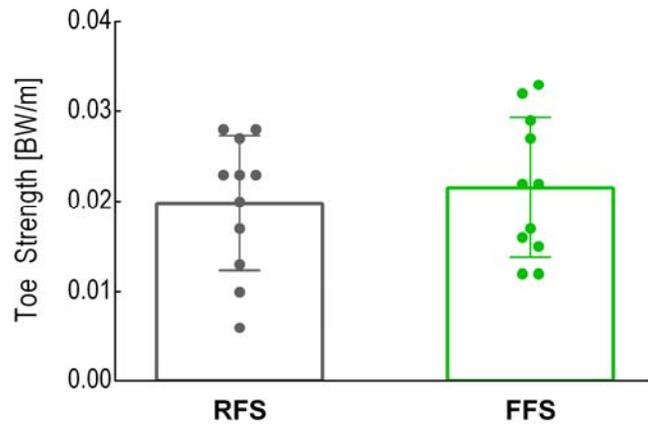
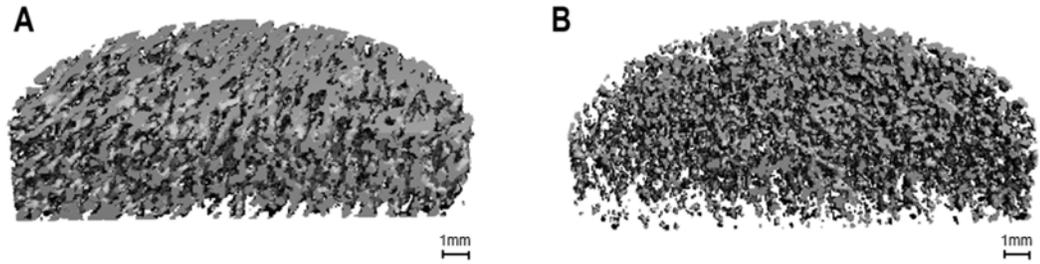


Figure 3-3 Mean and standard deviation of toe flexor strength (normalized to body weight). Comparison between rearfoot strikers (RFS) and forefoot strikers (FFS). Results from individuals are also reported.

3.4.2 Bone

Results from the bone scan of the calcaneus and first metatarsus are presented in **Figure 3-4** and **Figure 3-5** respectively. Main findings are a statistically lower (-67%, $p = .003$) trabecular area at mid shaft of the first metatarsal bone in RFS and a similar cortical area (-7%, $p = .3$) at the calcaneus. No main effect of Group ($F_{(1,8)} = 0.31$, $p = .692$) or interaction effects Group \times Bone ($F_{(1,8)} = 0.41$, $p = .845$), but main effect of Bone ($F_{(1,8)} = 14.20$, $p = .007$) showed that calcaneus has less number of trabeculae per normalized volume, but those trabeculae are thicker ($F_{(1,8)} = 66.71$, $p < .001$) and more spaced ($F_{(1,8)} = 7.57$, $p = .028$) compared to the 1st metatarsal.



Calcaneus		RFS		FFS	
<i>Density</i>		\bar{x}	<i>SD</i>	\bar{x}	<i>SD</i>
TV [mm ³]		1431.83	227.22	1445.75	114.34
BV [mm ³]		205.16	79.09	206.89	111.12
BV/TV [1]		0.15	0.09	0.14	0.07
<i>Structure</i>					
Tb.N [1/mm]		0.89	0.38	0.90	0.27
Tb.Th [mm]		0.17	0.03	0.15	0.03
Tb.Sp [mm]		1.11	0.50	1.05	0.41
AI [1]		0.09	0.05	0.09	0.05

Figure 3-4 Results for calcaneus. (A) Exemplar RFS (B) Exemplar FFS (C) Results for density measurements: TV (total volume), BV (bone volume), and BV/TV (bone volume with respect to total volume). For structure measurements: Tb.N (number of trabeculae), Tb.Th (thickness of trabeculae), Tb.Sp (space between trabeculae), and AI (anisotropy index).

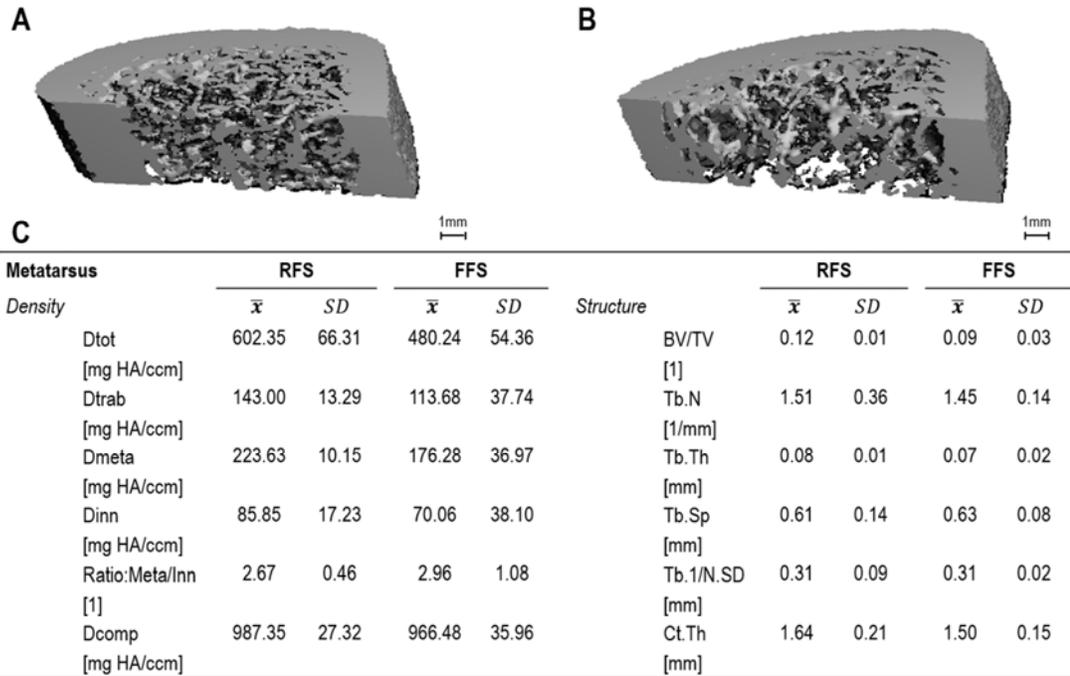


Figure 3-5 Results for first metatarsal (A) Exemplar RFS (B) Exemplar FFS (C) Results for density measurements: Dtot (average bone density), Dtrab (trabecular bone density), Dmeta (meta trabecular bone density), Dinn (inner trabecular bone density), and Dcomp (compact bone density). For structure measurements: BV/TV (trabecular bone volume with respect to tissue volume), Tb.N (number of trabeculae), Tb.Th (thickness of trabeculae), Tb.Sp (separation of trabeculae), Tb.1/N.SD (StDev of Tb.1/N: Inhomogeneity of trabecular network), and Ct.Th (cortical thickness).

3.5 Discussion

Very little is known about foot anatomical differences between RFS and FFS but based on substantial biomechanical differences while running (Daoud et al., 2012; Lieberman, 2012; Lieberman et al., 2015; Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, et al., 2010; Perl, Daoud, & Lieberman, 2012), it was reasonable to assume that foot muscles and bones may adapt to such diverse loading environments. However, our results suggest that neither the muscle size, nor the force they develop is affected by the habitual foot strike pattern. Only the first metatarsal bone presents greater trabecular volume in FFS compared to RFS.

It appears that long-distance running does not provide sufficient mechanical stress to enhance foot muscle size. The muscle CSA obtained in this study were similar to those obtained in control samples of previous studies (Angin, Crofts, Mickle, & Nester, 2014; Mickle et al., 2013). For example, Mickle et al. (2013) reported that muscle CSA in a sample of healthy and active (but not specifically runners), 5 males and 5 females (mean age 32.1 years) were $2.51 \pm 0.88 \text{ cm}^2$ for ABH, $2.15 \pm 0.54 \text{ cm}^2$ for FDB, $2.47 \pm 0.56 \text{ cm}^2$ for FHB, $1.75 \pm 0.58 \text{ cm}^2$ for QP. Similarly, the data from the present study reported $2.40 \pm 0.52 \text{ cm}^2$ for ABH, $2.22 \pm 0.38 \text{ cm}^2$ for FDB, $2.86 \pm 0.41 \text{ cm}^2$ for FHB, $1.39 \pm 0.37 \text{ cm}^2$ for QP. In contrast, our results for muscle thickness were relatively lower than those measured in sprinters (mean age 21.1 years) by Tanaka et al. (2018). Compared to sprinters, our sample of long distance runners have a -41% ABH thickness, -47% FDB thickness, -100% FHB thickness, and -29% GAS thickness. Although it is known that long-distance runners have thinner leg muscles than sprinters (Abe, Kumagai, & Brechue, 2000), there was no evidence of a similar adaptive response in foot muscles. Whether smaller muscle size results from training-specific adaptations due to running volume or due to footwear worn is still unclear (Garofolini & Taylor, 2019) (see Chapter 2). Certainly, longitudinal studies are necessary to understand the origin of foot muscle adaptations.

The measured toe flexor force in this study was similar between RFS and FFS. Given the above results, this was expected, as a clear relationship exists between foot muscles size and the force they can produce (Abe, Tayashiki, Nakatani, & Watanabe, 2016). In a previous study, habitual RFS running with a forefoot strike pattern have been found to have increased mechanical work performed by the intrinsic foot muscles (Kelly et al., 2018) therefore, one should expect those muscles to have increased size

and an increased ability to develop force in habitual FFS; however our results contradict this assumption. Because the activation of these muscles depends on loading requirements (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014), it is possible that habitual FFS have developed biomechanical features that reduce loading at the foot thus the need for active recruitment of foot muscles. This minimizes energy cost of running and it suggests that habitual FFS are able to rely on the passive structures of the foot with minimal intervention. This is speculation (and should be noted as such). It should also note that our group of FFS run habitually in traditional or less supportive shoes but not minimal shoes. The similarity with RFS may partially depend on shoe assistance. For instance, increased foot muscles size and stiffer longitudinal foot arch were found in minimally-shod populations compared to conventionally shod counterpart (Holowka, Wallace, & Lieberman, 2018).

The ability of bone to resist the external loads applied during running depends on the ability of the trabecular bone to transfer mechanical loads from the articular surfaces to the cortical bone (Nordin & Frankel, 2001; Oftadeh, Perez-Viloria, Villacamacho, Vaziri, & Nazarian, 2015). Although these results are based on a smaller sample of our participants, they endorse our assumption that foot strike loading has an effect on bone structure and density. The increased trabecular density (+67%) we found in the first metatarsal of FFS suggests that an habitual forefoot strike pattern may result in a more complex trabecular organisation, while an habitual rearfoot strike may lead to a simpler structural organisation. As trabecular bone is more metabolically active and responsive to stimuli than cortical bone (Jacobs, 2000), variation in its architecture may be evidence of adaptation to a different environment. The metatarsus of RFS are subjected to an environment with lower stress, where trabecular bone may be reabsorbed and transformed to cortical bone (Hart et al., 2017), while metatarsus of FFS are subjected to higher stresses throughout the stance phase, where the structure of the trabecular bone needs to be more organised (i.e. complex). RFS put the metatarsus under high stress (from both muscle contraction and gravitational force) from mid-stance through toe-off (propulsive phase). During this period the role of the foot is to provide a stable lever to propel the body forward. Thus, high stresses are applied to the metatarsus. On the other hand, metatarsus of FFS are also subjected to strains during landing when the intrinsic foot muscles work to modulate the effective stiffness of the foot (Riddick et al., 2019). The foot posture of FFS prior to landing

may be critical in attenuating the impact forces but also in stimulating trabecular adaptations in the metatarsus.

The fact that the calcaneal bone density was similar between RFS and FFS was expected. Bone formation and degradation are stimulated by mechanical stresses in the form of both muscular contraction and impact loading. The volume of interest we selected was appropriate to capture the effect of both elements. Similarities are explained by a higher impact load in RFS that stimulates the calcaneal bone to a similar extent as the pulling force produced by the plantarflexor muscles (through the Achilles tendon). Indeed, different regions of the calcaneus may adapt differently, therefore exploration of a larger bone volume, or even the calcaneus as a whole, may reveal a better insight into bone adaptation.

The main limitation of this aspect of the study is the number of subjects we were able to scan. Despite the advantages in using a HR-pQCT (Geusens et al., 2014), this technique has high associated costs. Here, we reported preliminary results and therefore our interpretation should be considered within the limitations of our study. Indeed, a larger sample size and a deeper analysis may attain clearer differences between groups.

3.6 Conclusion

In summary, we have showed that contrary to what was expected, RFS and FFS have similar foot muscle sizes and toe flexor force production. Interestingly, muscle size in our pooled sample was lower compared to other types of runners and somehow similar to active subjects not specifically involved in running. We demonstrated that long-distance runners incur foot muscle adaptations, but contrary to what was expected, foot muscle size is not increased. Both RFS and FFS have similar toe flexor strength but they differ in the organisation of trabecular bone in the first metatarsus probably in response to a different loading environment. Our findings advance our understanding of biomechanical differences between these two groups, and this knowledge can practically advise better training programs for those who want to transition from one foot strike pattern to the other. Overall, these findings suggest that proper neuro-mechanical functioning of the foot does not require strength training.

3.7 References

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4 ANKLE JOINT DYNAMIC STIFFNESS IN LONG-DISTANCE RUNNERS: EFFECT OF FOOT STRIKE AND SHOE FEATURES

4.1 Abstract

Foot strike mode and footwear features are known as factors that affect ankle joint kinematics and loading patterns, but how those factors are related to the dynamic properties of the ankle is less clear. In our study, two distinct samples of experienced long-distance runners: habitual rearfoot strikers (n=10), and habitual forefoot strikers (n=10), were analysed while running at constant speed on an instrumented treadmill in three footwear conditions. The minimalist index (MI) was used to characterise their shoes (low MI means strong shoe-foot interaction, high MI means minimum interaction). No instructions were given about foot strike pattern. The joint dynamic stiffness was analysed for three sub-phases of the moment-angle plot: early rising, late rising, and descending. A two-way repeated measures analysis of variance was used to analyse the effect of group and footwear. Habitual rearfoot strikers displayed a statistically ($p < 0.05$) higher ankle dynamic stiffness in all combinations of shoes and sub-phases except in early stance in low MI. In high MI shoes, both groups had the lowest dynamic stiffness values for early and late rising (initial contact through mid-stance), whilst the highest stiffness values were at late rising in high MI shoes for both rearfoot and forefoot strikers (0.21 ± 0.04 , 0.24 ± 0.06 [Nm/kg \cdot $^{\circ}$ 100], respectively). Rearfoot strikers in high MI shoes had the highest net work value (27.8 ± 8 [Nm/kg \cdot $^{\circ}$ 100]), with an increase of both work absorbed and produced; however, the work ratio (absorbed/produced) for rearfoot strikers in this condition was statistically lower (0.55 vs 0.59) than for forefoot strikers. This means that rearfoot strikers rely more on muscle energy production than on elastic energy storage. In conclusion the habitual landing pattern and the adaptation to footwear characteristics, which are conditioning the moment-angle loop, seem to reflect the neurophysiological ability of the subject to control the characteristics of the plantar flexor muscle-tendon unit. Habitual forefoot strikers may have access to a wider physiological range of the muscle torque and joint angle. This increased potential may allow forefoot strikers to adapt to different footwear by regulating ankle dynamic stiffness depending upon motor task.

4.2 Introduction

There is an ongoing debate on whether the foot strike pattern of long-distance runners plays a role in defining performance and injury risk in this population (Bramble & Lieberman, 2004; Davis, Rice, & Wearing, 2017; Hamill & Gruber, 2017). Experienced long-distance runners are able to change their foot strike pattern during a competition (Larson et al., 2011) or if they are asked to (Hamill et al., 2014). Their ability to adopt a different foot strike pattern has been often interpreted as a sign of adaptability. These concepts have been previously shown to not be equivalent (Garofolini, Taylor, McLaughlin, Vaughan, & Wittich, 2017a). In this thesis adaptability is defined as the complexity (or level of organisation) embodied by the human locomotor control system (see Chapter 1.3); it refers to the richness of motor behaviours that equally accomplish the task-goal. We expect experienced runners to have developed a level of adaptability that depends on their habitual foot strike. To test this hypothesis, in this chapter we evaluate the ability of runners with antithetical foot strike patterns (i.e. rearfoot strikers verses forefoot strikers) to adapt the dynamic stiffness of the ankle in response to different shoe substrates.

The concept of dynamic stiffness (Crenna & Frigo, 2011; Gabriel et al., 2008), defined quasi-stiffness by Latash and Zatsiorsky (1993), can be used to characterize the ankle behaviour during the stance phase of running (Stefanyshyn & Nigg, 1998). Here the ankle exhibits two distinctive states: a loading state in which the internal plantarflexor moment rises during dorsiflexion, and the joint stores energy; and an unloading state in which the plantarflexion moment decreases while the joint plantarflexes, and the joint returns energy. The level of stiffness (or its inverse, compliance) can express both: (i) anatomical adaptations that happen in the muscle-tendon units surrounding this joint, and (ii) neural adaptations that control the characteristics of these muscle-tendon units (Duchateau & Enoka, 2016; Feldman, 1980; Guissard & Duchateau, 2006). For instance, long-term adaptations in muscle and tendon architecture in the lower limb, such as shorter gastrocnemius medialis fascicles (Cronin & Finni, 2013), thicker Achilles tendon (Lichtwark, Cresswell, & Newsham-West, 2013), and stiffer foot arch (Lieberman, 2014), were found in habitual forefoot strikers, who usually land with a plantar-flexed ankle. Such adaptations could lead to a different load distribution in the muscle-tendon unit (Kubo et al., 2017), in which the role of the elastic components is increased, and the muscle fibers contract at

a slower rate, which is advantageous for maximal power output and efficiency (Lichtwark, Bougoulas, & Wilson, 2007). Together, anatomical and neural adaptations define the dimensionality of the system (hence the available degrees of freedom) that can be used to regulate the ankle dynamic stiffness in the most efficient way (Latash, 2012).

Ankle dynamic stiffness can be computed as the slope of the tangent to the moment-angle curve (Crenna & Frigo, 2011). Using similar approaches previous studies have investigated dynamic ankle stiffness during running (Günther & Blickhan, 2002; Jin & Hahn, 2018; Schache, Brown, & Pandy, 2015). During the stance phase of running the ankle plays a dominant role in generating energy for propulsion (Jin & Hahn, 2018; Schache et al., 2015), suggesting that the joint angle at landing (i.e. foot strike angle) is a compromise between metabolic and control effort minimisation (Günther & Blickhan, 2002). To our knowledge, Hamill et al. (2014) were the only researchers testing change in ankle joint stiffness in two groups of runners with distinct foot strike patterns. Participants were classified as either rearfoot or forefoot strikers based on the presence of an impact peak on the vertical ground reaction force and on the ankle angle at landing. Although using these criteria runners may have been misclassified (Garofolini, Taylor, McLaughlin, Vaughan, & Wittich, 2017b), according with the author, habitual forefoot strikers showed a more compliant ankle, and more negative work done when running with their preferred foot strike pattern (forefoot), however, no differences were found with habitual rearfoot strikers running with a forefoot strike pattern (non-preferred).

All the studies concerning running and ankle stiffness, simplify the loading phase of the moment-angle loop as represented by the average linear slope fitted from foot contact to peak moment (**Figure 4-1**, dashed line), which overlooks potentially meaningful details within the loading phase. For instance, at initial foot contact the ankle moment increases with no change in angle (vertical red arrow in **Figure 4-1**) this state represents the ankle joint response to external loading at initial impact. Thereafter, the ankle starts to dorsiflex while the ankle moment is still increasing (inclined red arrow in **Figure 4-1**).

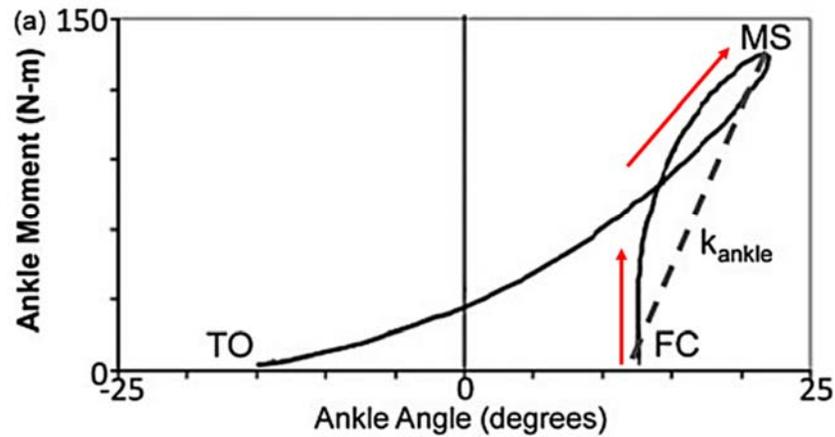


Figure 4-1 Example of moment-angle loop for the ankle joint. Adapted from Hamill, Gruber, & Derrick (2014)

This represents the loading of the passive structures of the muscle-tendon units. No studies have investigated the loading phase of the moment-angle dynamics in three task-relevant sub-phases, which we expect to yield a more sensitive insight of the differences between habitual rearfoot and forefoot strikers.

The aim of this study was to investigate if foot strike loading technique has an effect on the ankle moment-angle dynamics during the stance phase of running. We had three hypotheses. First, we expected FFS to have lower dynamic stiffness in all footwear conditions based on previous findings (Hamill et al., 2014). Second, we expect FFS to have a higher proportion of negative work relative to positive work because of their loading technique that allows them the ability to store and use potential energy in the foot-ankle anatomy. Third, because we expect that forefoot strikers will have a greater foot-ankle adaptability to external loading, we expect them to have a more invariant ankle stiffness throughout the stance phase. The latter will be expressed by stronger correlations in ankle stiffness between the three sub-phases of stance.

4.3 Methods

4.3.1 Participants

Forty male long-distance runners gave their personal consent to take part in this study. Participants were excluded if they had not been running for at least 5 years, with an average of at least 40 km/week, and had not been free of neurological, cardiovascular, or musculoskeletal problems within the previous six months. A number of 21 runners were found eligible. One subject was unable to complete the study protocol, which resulted in a tested sample of 20 subjects (age: 31.2 ± 6.9 yrs, height: 1.77 ± 0.07 cm, weight: 73.4 ± 7.9 kg). Participants were classified as rearfoot strikers (RFS, $n=10$) or forefoot strikers (FFS, $n=10$) based on their habitual mode of foot-ankle loading technique at ground contact. To classify their foot strike loading type, the participants were asked to run on an instrumented treadmill (AMTI Pty, Watertown, MA, USA) at their preferred speed, wearing their habitual running shoes. After a standardized 7-minutes of progressive warm-up and accommodation period, participants run for 3 minutes at their preferred running speed, which was identified from the protocol suggested by Jordan, Challis, and Newell (2007). Habitual foot strike mode was assessed on the basis of data collected on the last minute of running. A forefoot strike mode was based on the time spent performing an ankle plantarflexor moment within a short period at initial ground contact: defined between two events of foot contact and when first exceeding a vertical ground reaction force threshold of 1 body weight. Runners displaying an internal plantarflexor moment for at least 90% of this period were classified as forefoot strikers (FFS); conversely, those who displayed an internal dorsiflexor moment for at least 90% of the analysed period were classified as rearfoot strikers (RFS). This foot strike classification method was shown to perform best among other conventional methods (Garofolini et al., 2017b).

4.3.2 Experimental protocol

Tests were performed on an instrumented treadmill (Advanced Mechanical Technology Inc., Watertown, MA, USA) that collects ground reaction forces at a sampling rate of 1000 Hz. To minimize systematic force signal error associated with dynamic properties of instrumented treadmills, a wood frame was used to support the base and reduced the effect of low resonant frequencies Garofolini, Taylor, and Lepine (2018). Three-dimensional kinematics data of the lower extremities was recorded at a

sampling rate of 250 Hz from a 14-camera VICON B-10 system (Oxford Metrics Ltd, UK). Kinematic and ground reaction force data were synchronised using a VICON MX-Net control box and collected through Nexus 2.6 software (Vicon Motion Systems Ltd., Oxford, UK). A biomechanical model was reconstructed from 45 retroreflective markers placed on body segments.

After completing a standardized and progressive 7-minute warm-up, participants repeated a 5-minute running test three times, with a different shoe for each trial; the three shoe models were distinctly different by their minimalist indexes. The minimalist index is a classification that takes into account structure, flexibility, pronation support, and other footwear features, and ranges from 0% (maximum assistance) to 100% (least interaction with the foot) (Esculier, Dubois, Dionne, Leblond, & Roy, 2015). The shoes adopted in our experiments were classified at low MI (Mizuno® Wave Rider 21, MI= 18%), medium MI (Mizuno® Wave Sonic, MI= 56%), and high MI (Vibram® Five fingers, MI= 96%). Note: a low MI shoe is generally designed to provide maximum assistance for a runner that adopts a rearfoot loading pattern. The order of presentation was pseudo-random, that means that combinations were balanced within each group and equal between groups. Testing speed was fixed for all participants at 11 km/h.

4.3.3 Biomechanical Model

A set of retroreflective markers arranged in cluster setup were used to track 3D position of body segments, while landmark-derived virtual markers and movement-derived virtual markers were used to calibrate the position and orientation of the lower body skeletal system. Semi-rigid clusters of 4-5 markers were attached to lower-body segments so that the location of the cluster centroid was minimally affected by muscular contraction and related mass deformation. To minimize effects of skin movement artefact (Leardini, Chiari, Della Croce, & Cappozzo, 2005; Taylor et al., 2005), we secured the semi-rigid clusters over extra-long neoprene bands made of anti-migration material that wrapped and fastened on the thigh and shank segments. Individual trunk and pelvis retroreflective markers were placed over the 7th cervical vertebrae, sterno-clavicular notch, 10th thoracic vertebrae, posterior- and anterior-superior iliac spines. Virtual markers were used to identify medial and lateral epicondyles of the femur, medial and lateral malleoli. A custom version of the IOR

multi-segment foot model (Leardini et al., 2007) was adopted for the foot marker setup. Retroreflective markers were placed on calcanei, first metatarsal bases and heads, second metatarsal bases and heads, navicular bones and base and heads of the 5th metatarsals.

To fix the 9.5mm reflective markers on the foot we removed the internal screw from the markers, and replaced with a 6mm diameter x 1.5mm long Rare Earth Magnet fixed with superglue. After identifying the foot anatomical landmarks, we applied a similar magnet on the skin fixed with topical skin adhesive glue. Participants performed testing in socks and shoes. All shoes were modified with the circular holes cut at anatomical landmarks. Foot markers were attached to magnets that were pre-glued to the skin of the participants, ensuring repeatable marker location associated with reattachment process between footwear conditions (**Figure 4-2**).

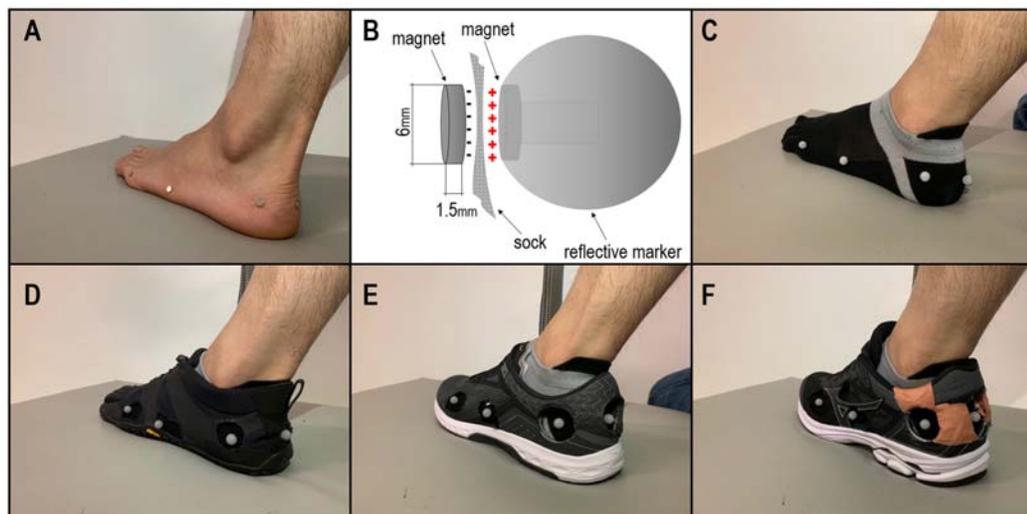


Figure 4-2 (A) Magnets glued to bony landmarks; (B) Schematic representation of magnets interaction; (C) markers placed over the sock maintaining the same position; (D-F) markers position in the three shoe conditions: Vibram® Five fingers (D), Mizuno® Wave Sonic (E), Mizuno® Wave Rider 21 (F).

Hip joint centre and knee joint axis of rotation were defined using functional movement trials according to Camomilla, Cereatti, Vannozzi, and Cappozzo (2006) and Schwartz and Rozumalski (2005). A six-degrees of freedom segment model was built for biomechanical analysis in Visual3D software (C-motion Inc., Rockville, USA). Standard methods were used to calibrate segment pose from marker setup and

reconstruct the subject biomechanical model in Visual3D. For joint rotations we used a right handed orthogonal coordinate systems where the z-axis represented the axial direction of the segment. The x-axis lied in the frontal plane perpendicular to the z-axis. The y-axis lied on the sagittal plane in the antero-posterior direction. In Visual3D, joint angles were calculated using an x-y-z Cardan-Euler sequence representing flexion/extension, abduction/adduction, and axial rotation of the thigh, shank, and foot (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2013). For the pelvis the Cardan sequence was reversed (z-y-x) as recommended by Baker (2001). Joint angles were normalized to the subject static reference position recorded as a ‘standing calibration trial’. For the scope of this thesis the segment movements of interest are those within the sagittal plane only, i.e. flexion/extension rotations.

The force signal recorded was assigned to relevant foot segment based on detection software in Visual3D. The estimated foot assigned to the force is based on the proximity between the location of the centre of mass of the foot and the transverse plane location of the centre of pressure on the force plate. Force signals were then used to compute joint moment (through inverse dynamic calculations) represented in the joint coordinate system (Schache & Baker, 2007).

4.3.4 Data analysis

Three-dimensional kinematics and kinetic data were analysed in Visual3D software (C-Motion, Inc, Rockville, MD, USA). A digital low-pass Butterworth filter (4th order, zero lag) was used to smooth raw kinematic and kinetic data with cut-off frequency of 15 and 35Hz, respectively. The ankle joint angle was calculated as the relative angle between the foot and the shank longitudinal axes, and normalised to the subject’s standing calibration posture. Joint moments were computed around flexion/extension axis using Newton-Euler inverse dynamics approach and normalized to body mass. Stance time was defined by gait events of initial and terminal foot contact (IC and TC) that were determined by a vertical ground reaction force threshold of 20 N. Stance time was normalised to 101 data points. The ankle (internal) moment was plotted as a function of the corresponding ankle angle (moment-angle plot) and the resultant curve was subdivided into three functionally relevant phases: early rising (ERP), late rising (LRP), and descending-phase (DP) according to methodology by

Crenna and Frigo (2011). [Note: these sub-phases equate to the impact (K1), loading (K2) and unloading (K3) phases described in Chapter 6.]

The slope of the angle-moment curve represents the level of joint stiffness at the ankle (K_{ankle}) in each functionally relevant phase. The area under the rising component and the descending component of the curve was integrated using a trapezoidal approximation. This gives the work absorbed (W_{abs}) and the work produced (W_{prod}) respectively. The net work (W_{net}) produced was computed as the difference between W_{prod} and W_{abs} . Finally, the work ratio ($W_{\text{ratio}} = W_{\text{abs}}/W_{\text{prod}}$) was computed as a measure of muscle efficiency (Holt & Askew, 2014).

4.3.5 Statistical analysis

An initial check for normal distribution (Shapiro-Wilk test) of the dependent variables, and homogeneity of variance (Levene's test) was performed. A three-way repeated measures ANOVA was used to test the effect of the between-factor *Group* (RFS, FFS) and within-factors *Shoe* (LOW, MED, and HIGH MI index), and *Slope* (ERP, LRP, and DP) on K_{ankle} . A two-way repeated measure ANOVA was used to test the effect of the between-factor *Group* (RFS, FFS) and within-factors *Shoe* for dependent variables W_{abs} , W_{prod} , W_{net} , and W_{ratio} . If ANOVA was significant, a post-hoc multiple comparison Tukey's test was used to determine where the differences were. Pearson Correlation coefficient (r) was calculated for all couples of dependent variables, while linear regression (r^2) was estimated between ERP, LRP, and DP. In case of non-normal distribution of data, the equivalent non-parametric tests (Kruskal-Wallis test, Dunn's multiple comparisons test, Spearman correlation) were used. All statistical analyses were performed using SPSS (version 25.0. Armonk, NY: IBM Corp.). Statistical significance was set at $p < .05$, with multiple pairwise comparisons corrected with Bonferroni adjustment method.

4.4 Results

An example of FFS (A) and RFS (B) ankle moment-angle relationship is reported in **Figure 4-3**. RFS show a distinct initial dorsiflexor angle, while FFS land with a more plantarflexed ankle. The moment as well exhibited a short dorsiflexion phase in RFS that was absent in FFS.

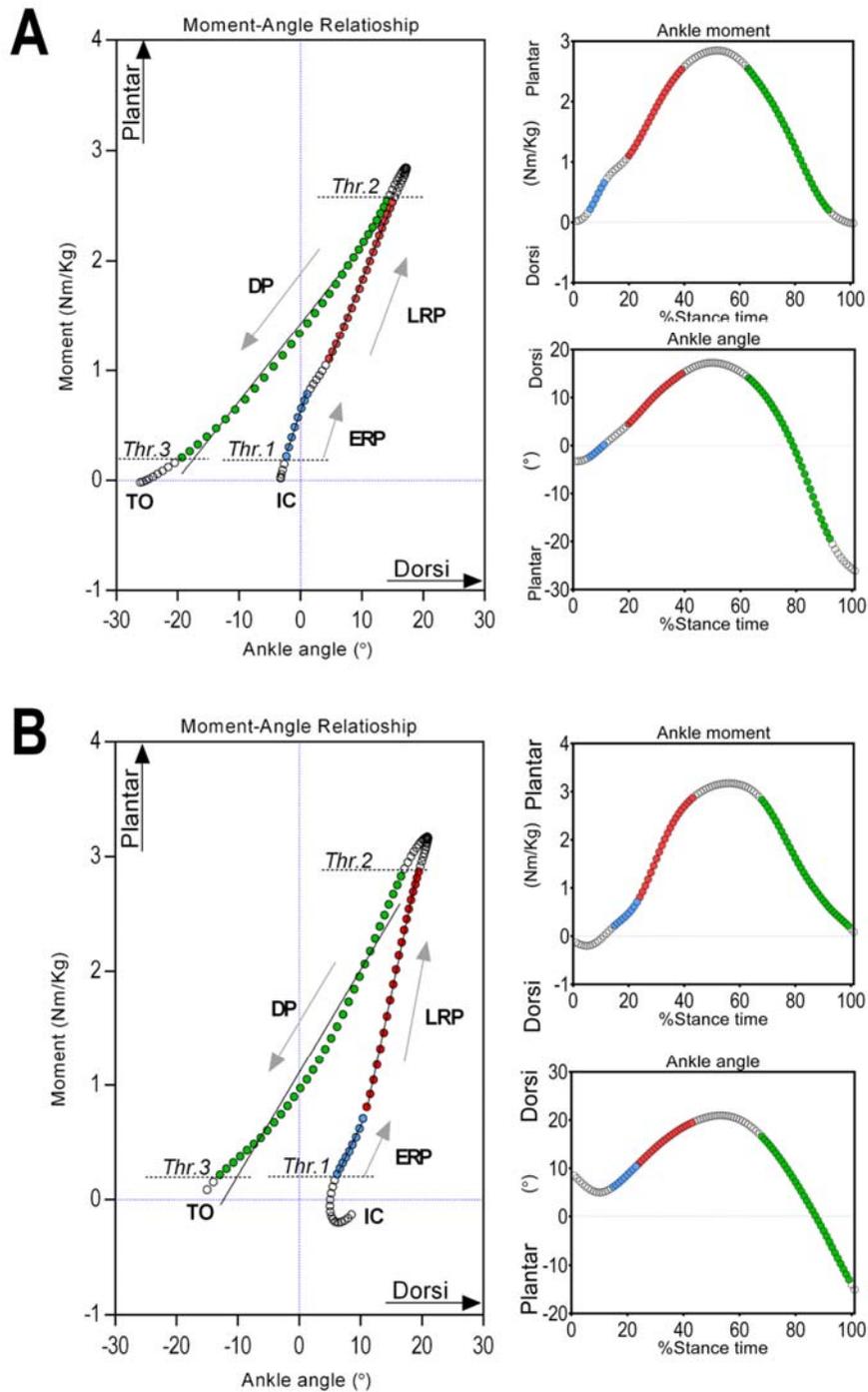


Figure 4-3 Example of ankle moment-angle relationship for a FFS subject (**top**) and a RFS subject (**bottom**) for the normalized stance phase from initial contact (IC) to toe-off (TO). The values for the quasi-stiffness is defined for the three phases of the moment-angle plot: early rising (ERP), late rising (LRP), and descending phase (DP). Thresholds are set to 0.2 ascending moment (Thr.1); 0.95 ascending/descending moments (Thr.2), and to 0.2 descending moment (Thr.3)

As shown in **Table 4-1**, no main effect of *Group* was found for K_{ankle} ($p = .164$) but main effect of *Shoe* ($p = .008$), and *Slope* ($p < .001$). Post-hoc analysis revealed K_{ankle} was 12% higher in med MI compared to high MI shoes ($p = .007$). **Table 4-2** and **Figure 4-4** shows mean and SD for K_{ankle} in the three sub-phases on stance and among the three shoe conditions. Significant differences were found among all sub-phases: ERP-LRP (.176±.01; .215±.01 Nm/kg·°·100; $p = .001$); ERP-DP (.176±.01; .091±.01 Nm/kg·°·100; $p < .001$); LRP-DP (.215±.01; .091±.01 Nm/kg·°·100; $p = .001$). Overall K_{ankle} was highest when wearing med MI shoes (although not different from low MI shoes; $p = .246$); K_{ankle} was highest during the loading phase (LRP) and lowest during the unloading phase (DP).

Table 4-1 Primary statistical results for differences between *Groups*, *Shoes*, and *Slopes* for mean ankle stiffness (K_{ankle}), work produced (W_{prod}), work absorbed (W_{abs}), work net (W_{net}), and work ratio (W_{ratio}). ANOVA results are given for main effects and interactions. Statistically significant findings are in bold.

Variable	Group	Shoe	Slope	Group x Shoe	Group x Slope	Shoe x Slope	Group x Shoe x Slope
K_{ankle}	$F_{(1,18)} = 2.11$ $p = .164$	$F_{(2,36)} = \mathbf{6.72}$ $p = \mathbf{.008}$	$F_{(2,36)} = \mathbf{144.34}$ $p < \mathbf{.001}$	$F_{(2,36)} = 0.719$ $p = .457$	$F_{(2,36)} = 1.15$ $p = .320$	$F_{(4,72)} = \mathbf{5.09}$ $p = \mathbf{.008}$	$F_{(4,72)} = 0.361$ $p = .732$
W_{prod}	$F_{(1,18)} = 2.92$ $p = .105$	$F_{(2,36)} = \mathbf{19.30}$ $p < \mathbf{.001}$		$F_{(2,36)} = 3.75$ $p = .051$			
W_{abs}	$F_{(1,18)} = 0.14$ $p = .716$	$F_{(2,36)} = \mathbf{13.29}$ $p = \mathbf{.001}$		$F_{(2,36)} = 2.81$ $p = .097$			
W_{net}	$F_{(1,18)} = \mathbf{9.71}$ $p = \mathbf{.006}$	$F_{(2,36)} = \mathbf{6.10}$ $p = \mathbf{.013}$		$F_{(2,36)} = 0.93$ $p = .376$			
W_{ratio}	$F_{(1,18)} = 4.29$ $p = .053$	$F_{(2,36)} = 0.53$ $p = .523$		$F_{(2,36)} = 0.49$ $p = .541$			

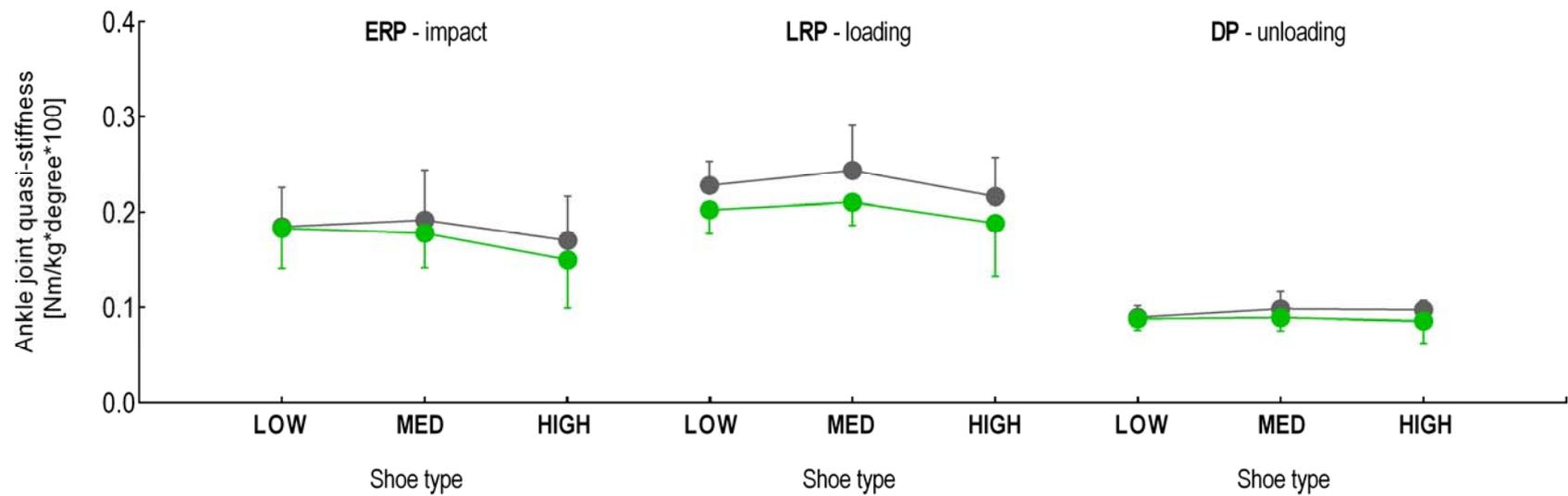


Figure 4-4 Mean and SD values for ankle joint dynamic stiffness of FFS and RFS for the three phases of stance, in the three shoe conditions. ERP early rising phase, LRP late rising phase, DP descending phase. Shoes conditions are termed as low MI (LOW), medium MI (MED), and high MI (HIGH).

Table 4-2 Mean and (SD) for *Groups, Shoes, and Slopes* for mean ankle stiffness, work [Nm/kg*degree*100] produced (W_{prod}), work absorbed (W_{abs}), work net (W_{net}), and work ratio (W_{ratio})

	LOW		MED		HIGH		POOLED	
	FFS	RFS	FFS	RFS	FFS	RFS	FFS	RFS
Slope ERP	0.183 (0.04)	0.185 (0.04)	0.178 (0.04)	0.192 (0.05)	0.15 (0.05)	0.171 (0.05)	0.171 (0.04)	0.182 (0.05)
Slope LRP	0.203 (0.03)	0.229 (0.03)	0.21 (0.02)	0.244 (0.05)	0.189 (0.06)	0.217 (0.04)	0.201 (0.04)	0.23 (0.04)
Slope DP	0.088 (0.01)	0.09 (0.01)	0.089 (0.01)	0.098 (0.02)	0.086 (0.02)	0.098 (0.01)	0.088 (0.02)	0.095 (0.01)
W_{prod}	46.8 (4.7)	48.1 (5.6)	49.2 (5.8)	52.7 (8)	52 (11.1)	61.3 (6.6)	49.3 (7.8)	54.1 (8.6)
W_{abs}	-28.8 (5.1)	-25.9 (3.6)	-29.2 (4.9)	-27.7 (3.5)	-31.6 (10.1)	-33.6 (3.8)	-29.9 (7)	-29.1 (4.8)
W_{net}	17.9 (4.1)	22.2 (4.9)	20 (4.2)	25 (6.4)	20.5 (4.4)	27.8 (5.5)	19.5 (4.2)	26.2 (5.7)
W_{ratio}	-0.6 (0.1)	-0.5 (0.1)	-0.6 (0.1)	-0.5 (0.1)	-0.6 (0.1)	-0.5 (0.1)	-0.6 (0.1)	-0.5 (0.1)

Runners in high MI shoes exhibit more compliant ankle during the impact phase (ERP) and loading phase (LRP); during the unloading phase (DP) low MI shoes allow the most compliant ankle. There was a *Shoe* × *Slope* interaction effect ($p = .008$; **Table 4-1**) for K_{ankle} (**Figure 4-4**, **Table 4-2**). Pairwise multiple comparisons shown that during the impact phase (ERP), K_{ankle} in high MI shoes was higher compared to both low MI and med MI shoes (+15%, $p = .013$; +16%, $p = .003$, respectively). During the loading phase (LRP) K_{ankle} was the highest in med MI shoes ($0.227 \pm 0.01 \text{ Nm/kg} \cdot \text{°} \cdot 100$) but only statistically different from high MI shoes (+12%, $p = .011$). During the unloading phase (DP), differences between shoes were only significant for low MI compared to med MI shoes (-6%, $p = .009$).

Figure 4-5 compares mean moment-angle loops for RFS and FFS. While curves are similar in low MI shoes, (**Figure 4-5**, top) the base (ankle range of motion) is shifted toward the left for FFS. This is also true for medium MI (**Figure 4-5**, middle), and high MI shoes (**Figure 4-5**, bottom). The insets in **Figure 4-5** show the linear regression between stiffness in the three sub-phases of stance. In low MI shoes, both groups present low regression values ($r^2 \leq 0.26$). In medium MI shoes, K_{ankle} of RFS during the loading phase (LRP) explained 49% of the K_{ankle} variance during the unloading phase (DP), while for FFS only 22% was explained. K_{ankle} of FFS in high MI shoes depends on the stiffness in the previous phase: that is, stiffness during the impact phase (ERP) explained 60% of the stiffness variance during the loading phase (LRP), and 65% of the stiffness variance during the unloading phase (DP); likewise, stiffness during the loading phase (LRP) explained 63% of the stiffness variance during the unloading phase (DP).

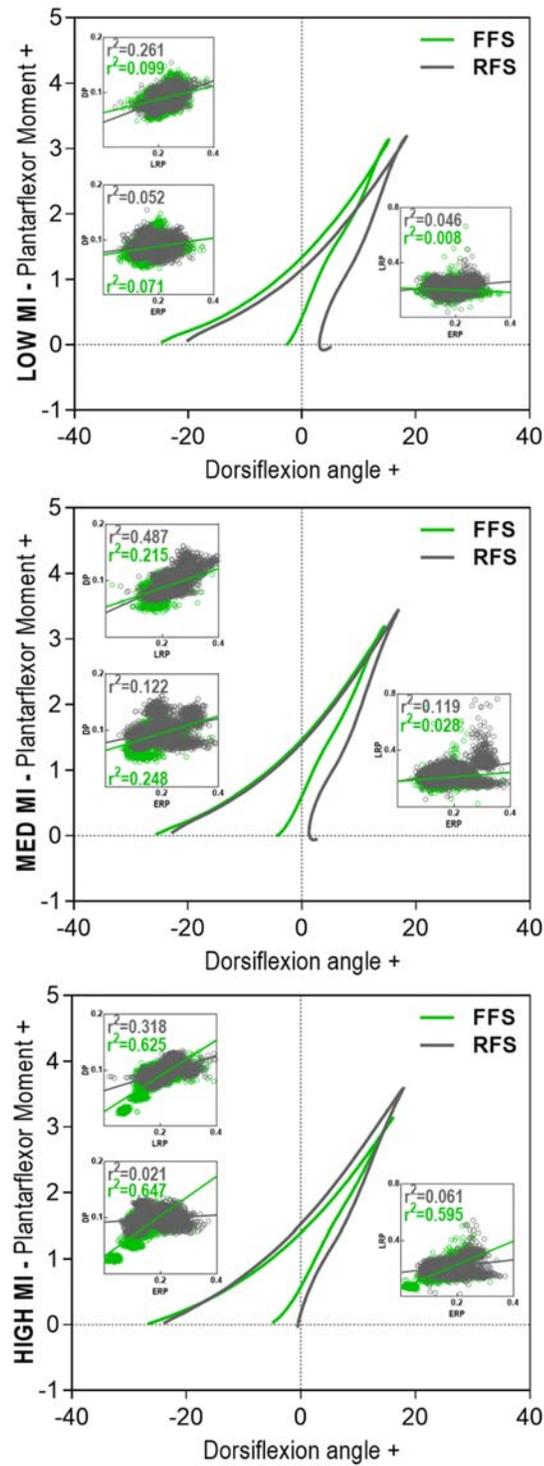


Figure 4-5 Ankle moment-angle plot. Group mean profiles comparison for low MI, medium MI and high MI shoes. Insets report linear regression lines between early rising phase (ERP), late rising phase (LRP), and descending phase (DP).

Table 4-1 also shows a main effect of *Shoes* for W_{abs} and W_{prod} ($p < .001$; $p = .001$) but no main effect of *Group* ($p = .105$; $p = .716$) or interaction effects for *Groups* \times *Shoes* were found ($p = .051$; $p = .097$). **Figure 4-6** shows W_{prod} by the ankle plantar flexors increases significantly from low MI to med MI shoes (7%, $p = .004$) and from med MI to high MI shoes (11%, $p = .017$); while W_{abs} by the ankle plantar flexors decreases as an inverse function of shoe MI index reaching highest values in high MI shoes ($-32.58 \pm 1.71 \text{ Nm/kg} \cdot \text{100}$). The latter was significantly lower than W_{abs} in low MI (-19%, $p = .002$) and med MI shoes (-14%, $p = .009$). RFS exhibited higher W_{net} compared to FFS (24.99 ± 1.25 versus 19.47 ± 1.25 ; $p = .006$); W_{net} increases with shoe MI index with runners in low MI shoes exhibiting statistically lower W_{net} (-12%; $p = .007$) compared to med MI shoes, and compared to high MI shoes (-20%; $p = .028$).

Rear foot strikers in high MI shoes had the highest W_{net} values ($27.8 \pm 8 \text{ [Nm/kg} \cdot \text{100]}$) explained by increased work absorbed (+28% from LOW, $p < .001$; +16% from MED, $p < .001$) and produced (+30% from LOW, $p < .001$; +21% from MED, $p < .001$) (**Figure 4-6**); however, the work ratio (absorbed/produced) for RFS was statistically lower than for FFS (0.55 vs 0.59). FFS increase positive work going from LOW to MED (+5%; $p < .001$) and from MED to HIGH (+6%; $p < .001$); while negative work was not statistically different from LOW (28.84 ± 5.8) and MED (29.21 ± 6.0 ; $p = .327$), but in HIGH, negative work was higher than both LOW (+9%; $p < .001$) and MED (+8%; $p < .001$); however, W_{net} in HIGH (20.4 ± 5.5) was similar ($p = .781$) to MED (20.2 ± 5.0) and LOW (18.8 ± 6).

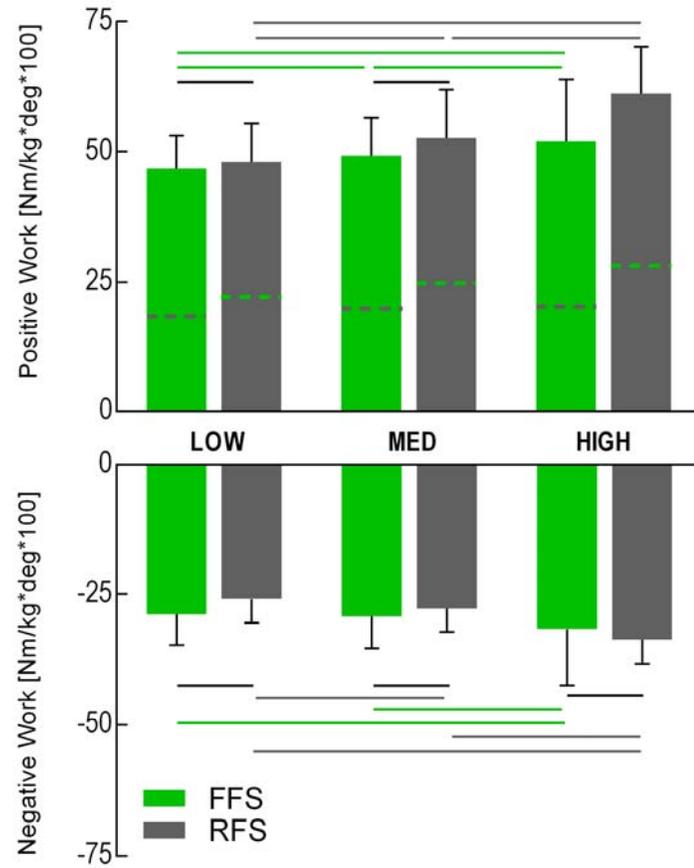


Figure 4-6 Mean and SD of ankle plantar flexors work for the three footwear conditions. Values are shown for positive and negative work for FFS and RFS. Dashed line indicates W_{net} , and solid lines signify a statistically significant ($p < .05$) difference.

Correlation between parameters of the moment-angle loop were computed and reported in **Table 4-3**. Overall, runners exhibiting high K_{ankle} during the loading phase (LRP), will have also high K_{ankle} during unloading phase (DP).

For FFS, the correlation between K_{ankle} in the impact phase (ERP) and in loading phase (LRP) increases with shoes' MI with the highest correlation ($r_s = 0.95$; $p < .01$) in high MI shoes. Similar trend is reported for correlations between K_{ankle} in impact phase (ERP) and in unloading (DP), and between K_{ankle} in loading phase (LRP) and in unloading (DP), with highest values in high MI condition ($r_s = 0.84$, $p < .01$; $r_s = 0.89$, $p < .01$, respectively). Values were only significant in high MI shoe conditions, this means that FFS in high MI shoes with high K_{ankle} during impact phase, will also have high K_{ankle} during the loading and unloading phases.

For RFS, correlations between K_{ankle} in impact phase (ERP) and in loading phase (LRP), and correlations between K_{ankle} in impact phase (ERP) and in unloading (DP) vary irrespectively to the shoe condition. The correlation between K_{ankle} in loading phase (LRP) and in unloading (DP) increases with shoes' MI with the highest correlation ($r_s = 0.92$; $p < .01$) in high MI shoes. This means, K_{ankle} during impact has less of an effect on the subsequent sub-phases in RFS; instead the loading phase plays a central role.

As for the correlation between energetic (work) measures, FFS exhibit high negative correlations values between W_{abs} and W_{prod} in all shoe conditions ($r_s \leq -0.69$) meaning that more work they absorb during loading, less work they need to produce during the unloading phase. RFS do not show such correlations, instead, they exhibit high positive correlations ($r_s \geq 0.60$) between W_{prod} and W_{net} meaning that the W_{net} increases as the W_{prod} increases.

Table 4-3 Correlations between moment-angle loop parameters (Spearman correlation coefficient rs). * represents statistically significant correlations ($p < .05$); ** represents statistically significant correlations ($p < .01$).

	FFS						RFS					
	Slope LRP	Slope DP	W _{abs}	W _{prod}	W _{net}	W _{ratio}	Slope LRP	Slope DP	W _{abs}	W _{prod}	W _{net}	W _{ratio}
LOW												
Slope ERP	-.16	.41	-.41	.04	-.46	-.39	.19	.20	-.27	.60	.31	-.01
Slope LRP		.10	.47	-.22	.26	.38		.58	-.03	.44	.36	.35
Slope DP			-.52	.09	-.65*	-.71*			-.24	.02	-.41	-.39
W _{abs}				-.69*	.37	.64*				-.50	.27	.60
W _{prod}					.36	.03					.60	.30
W _{net}						.93**						.88**
MED												
Slope ERP	.41	.67*	-.44	.32	-.09	-.22	.19	.42	-.60	.35	.29	-.02
Slope LRP		.76*	-.10	.13	-.29	-.21		.67*	-.01	-.07	-.02	.20
Slope DP			-.67*	.55	-.27	-.44			-.05	-.43	-.47	-.24
W _{abs}				-.82**	.08	.50				-.61	-.53	.16
W _{prod}					.39	-.08					.99**	.53
W _{net}						.86**						.61
HIGH												
Slope ERP	.95**	.84**	-.19	.30	.71*	.13	.09	-.02	.04	.21	.27	.31
Slope LRP		.89**	-.22	.30	.65*	.08		.92**	.26	-.08	.21	.43
Slope DP			-.53	.58	.58	.07			.14	-.08	.15	.25
W _{abs}				-.93**	-.01	.38				-.52	.19	.66*
W _{prod}					.26	-.16					.64*	.21
W _{net}						.61						.79**

4.5 Discussion

The purpose of this study was to explore the effect of foot strike modes and footwear features on the dynamic control of the ankle dynamics stiffness. There was no group main effect for ankle stiffness contrary to our first hypothesis that FFS would have had a lower ankle stiffness than RFS. Hamill et al., (2014) investigated stiffness during the phase of stance that corresponds most closely with the LRP region of our study. By examining a main effect of group within the LRP region (ignoring ERP and DP), we have also confirmed a statistically higher (+14%; $p = .005$) ankle stiffness in the RFS group. However, within the LRP, there was not a main effect of *Shoe* on ankle stiffness ($p = .163$). Previous studies found that changing shoe support alters the level of joint stiffness (Apps, Sterzing, O'Brien, & Lake, 2016; Sinclair, Atkins, & Taylor, 2016); where ankle dynamic stiffness increases as the shoe hardness decreased (Baltich, Maurer, & Nigg, 2015). While increasing stiffness may be functional in preventing excessive joint movement (Riemann, Myers, & Lephart, 2002), it has been identified as a possible risk of Achilles tendon injuries in runners (Lorimer & Hume, 2016).

The rearfoot strike loading technique generates more positive (produced) work by the ankle joint. This confirms our second hypothesis, and is consistent with previous studies that found ankle plantar flexor muscles to store more elastic energy (negative work) during the loading phase of fast running (i.e. forefoot strike) compared to positive work during unloading (Lai, Schache, Brown, & Pandy, 2016). The RFS group in our study exhibited 34% higher net work compared to FFS (**Table 4-2** and **Table 4-3**), which correlated strongly with the work produced (**Figure 4-5**); indicating that there was more muscle energy produced compared to elastic energy stored (Biewener & Roberts, 2000). Efficient running is achieved by efficiently storing and releasing elastic energy at each step ; our results are in line with previous literature that found FFS to store and return more elastic energy than RFS (Hasegawa, Yamauchi, & Kraemer, 2007; Lieberman et al., 2010; Perl, Daoud, & Lieberman, 2012). Despite this energetic advantage, FFS are consistently reported to be energetically inefficient (Gruber, Umberger, Braun, & Hamill, 2013; Ogueta-Alday, Rodríguez-Marroyo, & García-López, 2014). Therefore, it may be concluded that saving and releasing energy in the plantarflexor muscles may not significantly reduce the whole-body metabolic cost of running with a forefoot strike pattern (Gruber, Umberger, Miller, & Hamill, 2018).

The FFS group demonstrated a less variant ankle stiffness across the stance phase, especially for the high MI shoe condition fulfilling our third hypothesis. Furthermore, within the same shoe condition, the FFS group had strong correlations between ankle stiffness (K_{ankle}) during both impact and loading phases, with net work (W_{net}). By controlling ankle stiffness, the work around the ankle was modulated probably to achieve a functional redistribution of loading along the lower limb joints (Schache et al., 2015; Yen, Auyang, & Chang, 2009). Furthermore, **Figure 4-5** indicates that the K_{ankle} of FFS running in minimally supportive shoes is constant through the impact, loading and unloading sub-phases, suggesting that foot strike at landing is important in defying the ability to modulate ankle dynamic stiffness not only at impact, but also during the loading and unloading phases. Similar correlation has been found between the initial joint stiffness and maximal stiffness during the stance phase of hopping (Rapoport, Mizrahi, Kimmel, Verbitsky, & Isakov, 2003). One of the possible explanation for a constant ankle stiffness is that in that configuration (ankle plantarflexion with minimal support) the ankle-foot complex can express its spring-like function (Farris Dominic & Raiteri Brent, 2017; Kelly, Farris, Lichtwark, & Cresswell, 2018; Riddick, Farris, & Kelly, 2019); while increasing the support may introduce a level of instability that requires a trade-off between the task-goal of energy recycling and stable locomotion.

Shoes characteristics influenced the control of ankle dynamic stiffness. Both groups were able to reduce ankle dynamic stiffness during impact and loading phase when wearing high MI shoes (**Figure 4-4, Table 4-2**). However, both groups also increased the work produced and absorbed, so that the total net work done around the ankle during stance increased as a function of the shoe MI index (**Figure 4-6, Table 4-2**). Control and modulation of these loads need a certain level of adaptability of both the musculoskeletal and neuronal systems (Cronin, Carty, & Barrett, 2011). This may explain the high risk of certain injuries when changing from low to high MI shoes (Giuliani, Masini, Alitz, & Owens, 2011) or from RFS to FFS patterns (Daoud et al., 2012).

The main limitation of this study is that analysis was limited to the ankle joint. Indeed, adding analysis on the work done around knee and hip would have validated our assumption on leg-level force stabilization. However, inter-joint coordination and leg-level task stabilisation are the topics of the following chapters. Other limitations

are the assumed symmetry between dominant and non-dominant leg. The modulation of joint dynamic stiffness and the redistribution of joint work may vary if significant asymmetry exist (Exell, Irwin, Gittoes, & Kerwin, 2012).

4.6 Conclusion

In this study we investigated the effect of habitual rearfoot strike loading pattern, and the assistance of shoes, on ankle stiffness control. Our results suggest that RFS have reduced adaptability than FFS, but the constraint of this ability is dependent on the shoe worn. These findings reiterate the idea of this thesis that functional changes at joint level are important to define the redistribution of load along the lower-limb kinetic chain in order to solve leg-level force control (see Chapter 6 and 7). Shoes with a low MI may limit the ability to utilize the spring-like function of the ankle-foot complex, while shoes with high MI may promote the exploitation of the system redundancy. However, further studies are warranted to confirm the effect of shoes on ankle neuromuscular adaptations.

4.7 References

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5 THE PREFERRED LEG JOINTS COORDINATION PATH IN LONG-DISTANCE RUNNERS: EFFECT OF FOOT STRIKE AND SHOE FEATURES

5.1 Abstract

In this study we want to compare and contrast the joint coordination patterns of habitual forefoot and rearfoot strikers during steady-state running in different shoe types. One proposed method to describe coordination patterns is to implement the concept of the preferred movement path that represent the movement path runners naturally choose in response to their physical capacity and the external environment. We advanced from the current preferred movement path paradigm by addressing two of its main limitations: representativeness and quantification of deviations away from the preferred path. We conceptualized the “*preferred coordination path*” and use measures of trajectory consistency to quantify cycle-to-cycle variance as well as within trial variance in coordination pattern. Coordination variability is used to represent the richness of the system, thus its adaptability. In general, forefoot strikers tend to have greater coordination variability, and although shoe type did not have a clear effect on variability, in minimal supportive shoe rearfoot and forefoot runners had similar coordination patterns while in supportive shoes groups were the most different.

5.2 Introduction

While the locomotor system can express a variety of kinematic gait patterns via the lower limb, the many degrees of freedom available for intersegmental coordination appear to reduce into a few general modular properties or motor synergies (Ivanenko, Cappellini, Dominici, Poppele, & Lacquaniti, 2007; Lacquaniti, Ivanenko, & Zago, 2012). Recently, it was proposed that intersegmental covariance of running gait is attracted towards a preferred movement path that is unique to the participant and mostly invariant between gait cycles (Nigg, Baltich, Hoerzer, & Enders, 2015; Weir et al., 2018). The expression of the preferred movement path was quantified using kinematic gait trajectories, while the absolute divergence of these trajectories from their mean behaviour is considered a departure from the inherent preferred movement path (Nigg et al., 2017). Further, the preferred movement path is not sensitive to acute changes in footwear or surface conditions, but is a stable property inherent to the form and function of the neuro-musculoskeletal system. These ideas were based upon the finding that kinematic patterns of segment kinematics tracked using sub-cortical pins revealed consistent patterns insensitive to footwear and with non-systematic variations (Reinschmidt, Van Den Bogert, Lundberg, et al., 1997; Reinschmidt, Van Den Bogert, Nigg, Lundberg, & Murphy, 1997; Stacoff, Nigg, Reinschmidt, van den Bogert, & Lundberg, 2000; Stacoff, Reinschmidt, et al., 2000). The present study sought to investigate whether the preferred movement path of forefoot runners is sensitive to footwear and, therefore, adaptable.

One of the criticisms of the preferred movement path paradigm, is that it lacks clear integration with the inherited variability of human movement (Bernstein, 1967; Latash & Anson, 2006). In line with the uncontrolled manifold hypothesis (Scholz & Schönner, 1999) and the minimal intervention principle (Todorov & Jordan, 2002) introduced in Chapter 1.3, a certain amount of movement variability (functional to the task) is a sign of system complexity and may not require an active intervention from the nervous system (see Chapter 6 and 7). Therefore, the movement path should more accurately refer to similar trajectories that equally satisfy the motor task rather than be represented by the mean movement trajectory (Federolf, Doix, & Jochum, 2018). Recently, a change in gait mechanics has been found between high-volume and low-volume runners (Boyer, Silvernail, & Hamill, 2014) suggesting that training may change the preferred movement path. An expansion of movement variability around

the preferred movement path that does not alter the movement outcome during steady-state activities will be considered in this paper to represent a larger availability of redundant solutions for a given motor task, and hence a more adaptable system. We expect habitual forefoot strikers to exhibit a larger preferred movement path.

Previous studies examining the preferred movement path investigated individual joint angles (Nigg et al., 2017; Stacoff, Nigg, et al., 2000) without accounting for joint interdependency due to mechanical and neural constraints. That is, a change in angle in a single joint will influence a neighbouring joint angle, and thus alter their coupling (Federolf, Boyer, & Andriacchi, 2013). Changes in joint coupling may derive from a change in foot strike pattern (Pohl & Buckley, 2008) or more simply, from a change in shoe feature (DeLeo, Dierks, Ferber, & Davis, 2004). These changes could lead to an abrupt shift in stress to tissues not adapted for repetitive loading and arguably cause overuse injuries (DeLeo et al., 2004). Because of the frequency of these type of injuries, there has been an increased interest in interventions to modify individual running mechanics (Cheung & Davis, 2011; Crowell & Davis, 2011; Davis, Rice, & Wearing, 2017; Samaan, Rainbow, & Davis, 2014). Indeed, injured runners demonstrate altered shank-rearfoot (Rodrigues, TenBroek, & Hamill, 2013) and thigh-shank coordination (Hamill, van Emmerik, Heiderscheit, & Li, 1999). However, analysing joint angles individually does not represent how those joints work together to stabilize movement. In this chapter we will explore the multidimensional workspace on which changes in joint angles are functionally related.

Analysis of joint coupling requires accurate measurement of the trajectory shape (cyclograms) on an angle-angle plot rather than of individual joint kinematics. Conventional linear analyses are often used to capture running performance (Hall, Barton, Jones, & Morrissey, 2013; Moore, 2016; Williams & Cavanagh, 1987), but they lack the ability to provide an insight into the control system (Cavanagh & Grieve, 1973). On the other hand, cyclograms have the advantage of being described by geometric properties (Hershler & Milner, 1980) and give a more complete picture of coordinated movement of limb segments (Bartlett, 2007). It may be expected that because running with a rearfoot or forefoot strike pattern requires distinct temporospatial and kinematic adaptations (Lieberman et al., 2010), these two running styles may also display different lower-limb joint coupling. The inter-joint

coordination between ankle, knee, and hip will give shape to the *'preferred coordination path'*.

The purpose of the current study was to assess the variability in inter-joint coordination among habitual rearfoot strikers and habitual forefoot strikers and compare and contrast coordination variability between different shoe types. We hypothesised forefoot strikers to have developed, through experience, a more complex system. If this is true, they should exhibit lower indices of cycle-to-cycle consistency and higher variability compared to rearfoot strikers. Because of the different habitual foot strike pattern between groups, we hypothesised runners to have different preferred coordination paths, and for these coordination differences to be more evident at the ankle-knee coupling. In addition, we hypothesised runners in minimal supportive shoes to present the highest joint coupling variability, and the most supportive shoes to have an 'equalisation' effect (reduce differences) between groups.

5.3 Methods

Participants' characteristics and testing protocol are the same as per previous Chapters. Refer to Chapter 4.3 for details.

5.3.1 Data Analysis

Kinematic raw data were exported to Visual 3D (C-motion) and low-pass filtered using a Butterworth filter (4th order, zero lag) with a cut-off frequency of 15 Hz. Hip, knee, and ankle joint angles from the last 400 gait cycles of each condition (group-footwear) were cut into individual cycles (foot contact (FC) to following FC) and time-normalized to 500 samples using linear interpolation. Data were then exported to Matlab (The MathWorks Inc., Massachusetts, US) for further analysis.

The intra-limb coordination was analysed by means of hip-ankle, hip-knee, and knee-ankle cyclograms. The cycle-to-cycle consistency of the cyclograms for each participant was quantified using the angular component of the coefficient of correspondence (ACC) (Field-Fote & Tepavac, 2002), a vectorisation technique that indicates the overall variability of the joint-joint relationship for all cycles. The change in angle frame-by-frame is used to build a vector (l) with both direction and magnitude, joining frame n to frame $n+1$, so that:

$$l_{n,n+1} = \sqrt{(x_{n,n+1})^2 + (y_{n,n+1})^2} \quad (1)$$

where $x_{n,n+1}$ and $y_{n,n+1}$ represent the change in angle for the x joint and the y joint from the n frame to the subsequent ($n+1$). Vectors among consecutive cycles are compared to derive the degree of dispersion of the joint-joint values about the mean over multiple cycles for that frame ($a_{n,n+1}$) calculated as:

$$a_{n,n+1} = \sqrt{(\cos \bar{\theta}_{n,n+1})^2 + (\sin \bar{\theta}_{n,n+1})^2} \quad (2)$$

where the mean of cosine ($\cos\theta$) and sine ($\sin\theta$) are derived from the $l_{n,n+1}$ vector using simple trigonometry. The average dispersion (\bar{a}) of all cycles is then computed as:

$$\bar{a} = a_{1,2} + a_{2,3} + a_{3,4} \dots + a_{n-1,n}/n \quad (3)$$

where n is the number of cycles and \bar{a} is the angular component of the ACC. The larger the ACC value (between 0 and 1), the less variable (less randomly distributed, more consistent) is the joint-joint relationship. ACC values were then averaged across group and condition for further analysis.

The intra-subject cycle variability was calculated computing the average sum of squared distances (SSD) using the approach presented by Awai and Curt (2014). After translation of the cyclogram centroids to the origin and normalisation of the angle signals to the interval [0 1], we computed the cumulative ellipse area with half axes (a and b) corresponding to the between-subject standard deviation of every two joint coupled angles (i.e. hip-ankle, hip-knee, and knee-ankle) for 20 equal bins of time-normalized cyclograms:

$$Var_n = \sum_{i=1}^{20} \Pi * a_{n,i} b_{n,i} \quad (4)$$

where n represents the subject number, and i is the bin number. The sum of variance was calculated as the cumulated elliptic area for the 20 bins. The within-group SSD was then obtained comparing the mean group cyclograms in each joint couple-footwear combination as:

$$SSD_{j,k} = \sqrt{\sum_i (\alpha_{j,i} - \alpha_{k,i})^2 + (\beta_{j,i} - \beta_{k,i})^2} \quad (4)$$

where j and k are consecutive cyclograms, and α and β are the transformed and scaled joint angles at sample point i . The preferred coordination path was obtained by projecting the normalised ankle, knee, and hip joint angles on a 3-dimensional space. To further analyse differences in joint coordination patterns between the two groups, we applied a variation of a previously presented method (Giese & Poggio, 2000; Ilg, Rorig, Thier, & Giese, 2007) for modelling the space-time characteristics of multi-joint movements. Spatial correspondence between two trajectories was defined by a set of linear displacements (vectors) that map the first trajectory onto the second (**Figure 5-1**).

The magnitude of the vectors was then used as a measure of spatial variance between the two groups. Variance was then computed for single and multidimensional spaces and plotted as a function of the normalized gait cycle. All analysis were carried out using custom scripts in Matlab (Math Works Inc., USA).

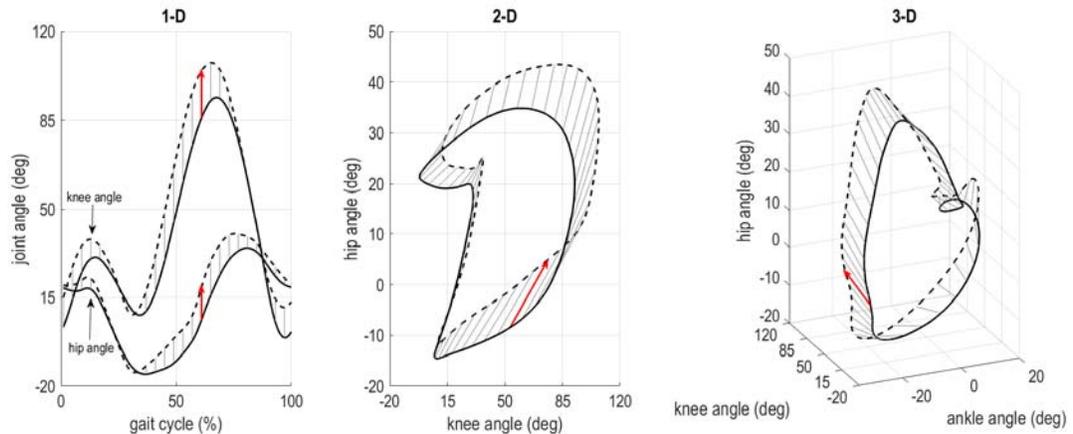


Figure 5-1 Analysis of the spatial variability in one-dimensional (1-D), and multidimensional spaces (2-D, 3-D).

5.3.2 Statistical analysis

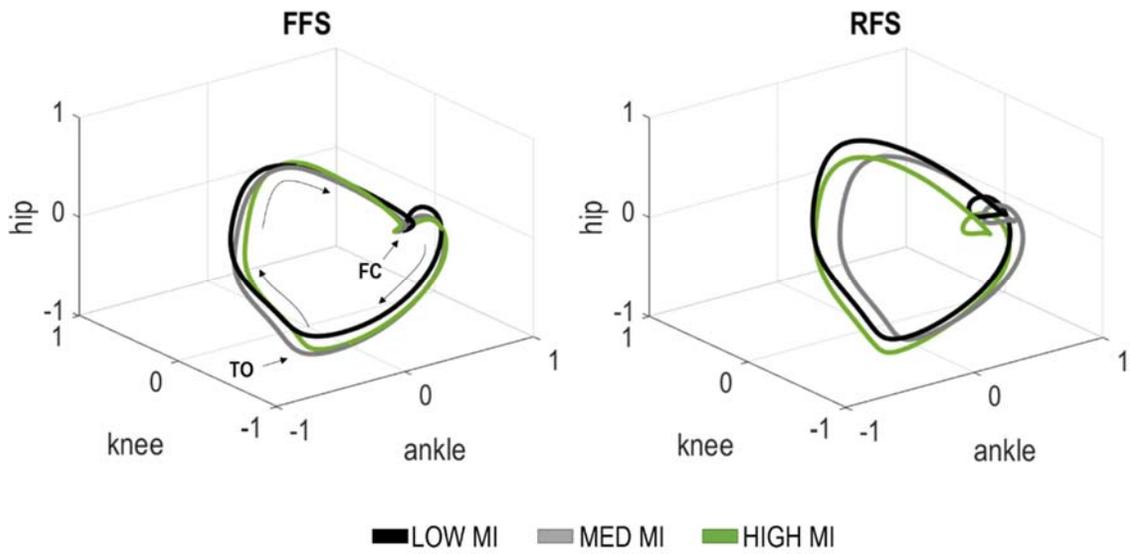
Mean, standard and deviation (SD) were computed for each *Group x Shoe x Phase* condition. To test the hypothesis that different coordination patterns of the lower leg joint angles exists between habitual forefoot strikers and rearfoot strikers, and to evaluate the influence of footwear characteristics, a mixed design 3-factor (shoe x phase x group) repeated-measures ANOVA was used to examine the interaction and main effects of within-subject factors of *Shoe* (3 levels: low MI, medium MI, high MI) and *Joint couple* (3 levels: hip-ankle, hip-knee, knee-ankle), and between-subject factor of foot loading *Group* (2 levels: forefoot, rearfoot) on the three dependent variables of variance: ACC, SSD, and sum of variance. Significance was set at 0.05 for all tests. Planned contrasts examined specific levels of an interaction effect between group, joint and shoe. Tukey post-hoc analysis was used to test multiple pairwise comparisons. All statistics were performed using SPSS software (version 25, SPSS Inc., Chicago, IL, USA).

5.4 Results

Figure 5-2A shows exemplar preferred coordination path for FFS and RFS in each footwear condition; **Figure 5-2B** compares groups within each footwear condition. There was no main effect *Group* ($p = .989$), or *Shoe* ($p = 0.667$) but a main effect of *Joint coupling* ($p < .001$, **Table 5-1**) for ACC values. Indicating that cycle-to-cycle consistency was dependent on the joint couple. Post hoc analysis shows that runners have the most uniform cyclogram shapes at knee-hip level (**Table 5-2**), while ankle-hip coordination showed lower consistency than ankle-knee and knee-hip cyclograms ($p < .001$).

There was a *Group x Shoe x Joint coupling* interaction effect ($p = .019$, **Table 5-2**), indicating the ankle-hip couple to be the least consistent in all footwear conditions for both groups ($p < .05$). Post-hoc tests revealed that for RFS ankle-hip coupling had the lowest consistency in all footwear condition ($p \leq .004$); while for FFS, ankle-hip coupling was only less consistent than ankle-knee in all footwear conditions ($p \leq .030$).

A



B

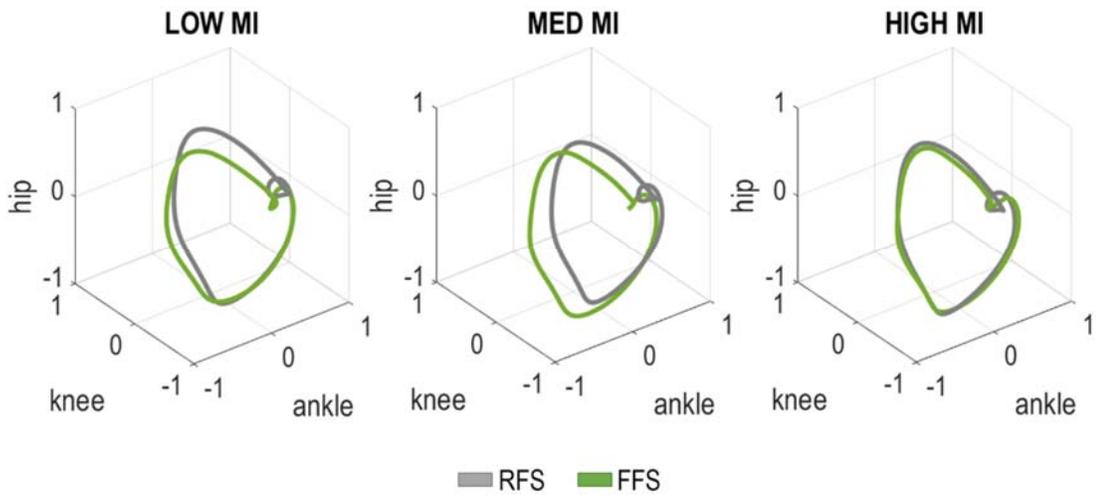


Figure 5-2 (A) Three dimensional plot of the mean preferred coordination path for FFS and RFS. Comparison is made between the three footwear conditions: low MI, med MI, and high MI. FC = foot contact; TO = toe off. **(B)** Comparison of mean group within each footwear condition.

Table 5-1 Main effects for group, shoe type, and joint coupling, and interaction effects for the coefficient of correspondence (ACC), mean sum of variance and the square root of the sum of squared distances (SSD). For SSD, main effect for shoe comparison instead of shoe is reported. Statistically significant results ($p < .05$) are reported in bold.

Variable	Group	Shoe	Joint Coupling	Group x Shoe	Group x Joint C.	Shoe x Joint C.	Group x Shoe x Joint C.
ACC	$F_{(1,38)} = 0.00$ $p = .989$	$F_{(2,76)} = 0.41$ $p = .667$	$F_{(2,76)} = \mathbf{34.94}$ $p < .001$	$F_{(2,76)} = 0.51$ $p = .562$	$F_{(2,76)} = 0.54$ $p = .529$	$F_{(4,152)} = 2.08$ $p = .138$	$F_{(4,152)} = \mathbf{4.35}$ $p = .019$
Sum of Variance	$F_{(1,38)} = 0.54$ $p = .474$	$F_{(2,76)} = 1.10$ $p = .347$	$F_{(2,76)} = \mathbf{31.34}$ $p < .001$	$F_{(2,76)} = 1.48$ $p = .247$	$F_{(2,76)} = 1.04$ $p = .336$	$F_{(4,152)} = 0.72$ $p = .479$	$F_{(4,152)} = 0.23$ $p = .767$
	Group	Shoe Comp	Joint Coupling	Group x Shoe C.	Group x Joint C.	Shoe C. x Joint C.	Group x Shoe x Joint C.
SSD	$F_{(1,38)} = 0.01$ $p = .942$	$F_{(2,76)} = \mathbf{4.57}$ $p = .018$	$F_{(2,76)} = \mathbf{7.92}$ $p = .002$	$F_{(2,76)} = \mathbf{4.37}$ $p = .022$	$F_{(2,76)} = 0.29$ $p = .751$	$F_{(4,152)} = \mathbf{3.95}$ $p = .006$	$F_{(4,152)} = \mathbf{4.54}$ $p = .003$

Congruent with the ACC results, there was a main effect of *Joint coupling* ($p < .001$, **Table 5-1**) for the cumulative variability (Sum of variance) along the 20 equal time bins. The knee-hip cyclograms exhibited the lowest variance (**Table 5-2**), while the cumulative variability in ankle-knee was three-time larger ($p < .001$) than ankle-hip, and eight-times larger ($p < 0.001$) than knee-hip sum of variance. The larger sum of variance at the ankle-knee coupling is indicative of changing behaviour within a trial. Table 2 shows that, although not significant, FFS tend to have a more variable ankle-knee coupling in all footwear conditions, but similar combinations of ankle-hip and knee-hip coupling, compared to RFS.

Table 5-2 Mean \pm standard deviation for the coefficient of correspondence (ACC), and sum of variance [mm^2]. Group comparison for the three joint couples: ankle-knee (AK), ankle-hip (AH), and knee-hip (KH), in each footwear condition.

Group comparison	ACC		Sum of Variance	
	RFS	FFS	RFS	FFS
AK				
Low MI	0.98 \pm 0.00	0.98 \pm 0.01	5142.2 \pm 3332.9	7008.5 \pm 4942.2
Med MI	0.98 \pm 0.00	0.98 \pm 0.01	6604.7 \pm 4905.7	7593.4 \pm 4344.8
High MI	0.98 \pm 0.00	0.98 \pm 0.01	5383.0 \pm 4762.1	8197.5 \pm 4414.0
AH				
Low MI	0.96 \pm 0.02	0.96 \pm 0.02	1520.6 \pm 1151.6	1232.5 \pm 1576.6
Med MI	0.96 \pm 0.01	0.97 \pm 0.02	1512.1 \pm 1334.7	1022.2 \pm 980.5
High MI	0.96 \pm 0.01	0.96 \pm 0.02	1722.1 \pm 1640.2	2121.4 \pm 2924.7
KH				
Low MI	0.98 \pm 0.00	0.98 \pm 0.01	840.5 \pm 938.7	569.8 \pm 569.3
Med MI	0.98 \pm 0.01	0.98 \pm 0.01	852.1 \pm 942.2	646.2 \pm 573.4
High MI	0.98 \pm 0.01	0.98 \pm 0.01	654.3 \pm 754.7	643.5 \pm 689.5

The SSD values reported in **Table 5-3** represent the amount of shape difference after uniform scaling and translation of the centroid. Table 1 shows that for SSD there was a main effect of *Joint coupling comparison* ($p = .022$) and *Shoe* ($p = .018$), but no main effect of *Group* ($p = .942$). Although reported, statistical effects have no low relevance for SSD values as they are based on differences between combinations of shoe and joint coupling, therefore their interpretation is meaningless.

Table 5-3 Mean±SD squared root of the sum of squared distances (SSD) group comparison.

Group comparison	SSD			
	Low MI		Med MI	
	RFS	FFS	RFS	FFS
Ankle-Hip				
High MI	143±118	204±153	275±151	128±147
Low MI			145±117	229±197
Ankle-Knee				
High MI	251±178	279±122	157±118	147±101
Low MI			163±81	199±121
Knee-Hip				
High MI	145±78	92±27	81±52	83±44
Low MI			94±51	69±46

As expected, differences in cyclograms shapes (SSD) were the highest between low MI and high MI shoes (**Table 5-2**), and lowest for the knee-hip coupling in accordance with the ACC results. This indicates that even after normalisation of cyclograms, knee-hip coupling has the highest consistency and the lowest amount of shape difference between shoe conditions; while shoes effects the joint phase.

Another quantitative characterisation of the differences in coordination patterns between RFS and FFS can be obtained with the spatial variance analysis of joint angles on one dimensional (**Figure 5-3A**) and multidimensional spaces (**Figure**

5-3B). RFS showed a more dorsiflexed ankle at both foot contact and toe-off, and a more extended hip and knee joint throughout the stance phase. Variance due to joint couples is reported in **Figure 5-3B**. Differences between groups at FC can be attributed to a larger difference in knee-hip and ankle-knee coordination, while at toe-off (TO) knee-hip couple are similar between groups, and total variance at this point is due to differences in ankle-hip and ankle-knee coupling.

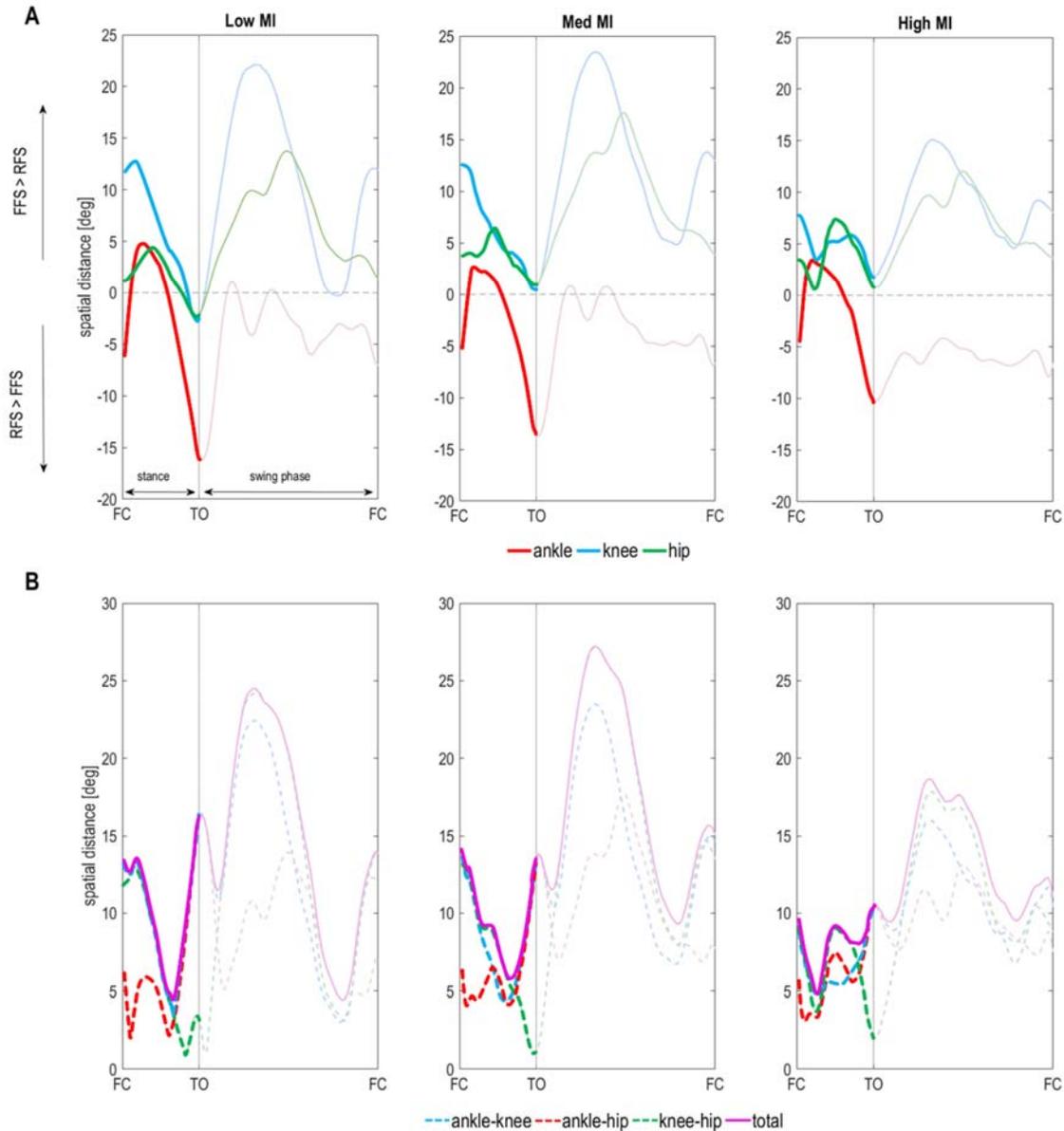


Figure 5-3 Spatial variance quantification expressed as a function of the stance phase (foot contact – FC to toe off – TO). Results for the one-dimensional analysis (**A**) and for the multidimensional analysis (**B**) are reported. Comparisons are made among the three footwear conditions.

Shoes had an effect on spatial variance between groups. Peak ankle angle difference was at TO and decreased from 16° in low MI shoes to 14° in med MI shoes, to reach the lowest values in high MI shoes (10°). Knee peak difference was at FC and similar in low and med MI shoes (12°, 13°. respectively) but lower in high MI shoes (8°). Hip peak difference was at mid-stance and increased slightly from low MI to med MI shoes (from 4° to 6°) and stays the same from med MI to high MI (from 6° to 7°). The contribution of coupled joints to the total (3D) variance also depended on the shoe's minimal index. In high MI shoes, groups are more similar in their coordination patterns, while in both low and med MI shoes groups differ more, in particular at FC and TO. At FC, the high variance in ankle-knee and knee-hip coordination decreases in high MI shoes, while the ankle-hip couple remain similar (~ 6°) for all shoe conditions. During the stance phase, there is a drop in total variance (more evident in low MI shoes) that is followed by a rise up to TO. Peak total variance at TO decreases as an inverse function of shoe MI, so that the highest difference is in low MI and the highest difference is shown in high MI shoes.

5.5 Discussion

In this study, we used treadmill steady-state running to explore coordination variability within lower limb joint couplings - described as the preferred coordination path. As the motor task is stable, we expected the preferred coordination path to represent self-organisation of the system, hence its entropy. We used three measure of variability to characterize the preferred coordination path: ACC to indicate the cycle-to-cycle consistency; the sum of variance to indicate the richness of the joint coupling along the preferred coordination path with higher values representing higher redundancy of the system; and, the SSD to indicate the normalized shape mean differences between groups and within conditions (i.e. shoe type).

The preferred coordination path is a step forward from the movement path (Nigg et al., 2015). It considers all three lower limb joints simultaneously and quantifies variability around the mean trajectory as an expression of system complexity. Figure 2 displays mean preferred coordination path for both rearfoot and forefoot strikes. During the stance phase, the coordination path is constrained by the external forces acting on the body, and from muscle activity controlling the distribution of stiffness among the joints to enable energy transfer in the limb (Zajac, Neptune, & Kautz, 2002). During swing, the mechanical constraints inherited in the system define the path.

We hypothesised differences in the preferred coordination path to be more evident in high MI shoes. We found the opposite to be true (**Figure 5-2B**). RFS and FFS have similar coordination paths in high MI shoes while in low MI shoes FFS have a more plantarflexed ankle at foot contact (**Figure 5-2B**, Supplementary A) compared to RFS which alters the coupling with both knee and hip (Supplementary B). Individual joint kinematics would have led to the conclusion that shoes do not affect the preferred movement path. The preferred coordination path leads to the same conclusion but it gives a more in-depth understanding of joint coordination. By changing the ankle angle, leg segments can still be similarly oriented, but to maintain stable locomotion, the inter-joint coordination needs to adapt. This inevitably changes the distribution of joint loadings and thus joint angles (Yen & Chang, 2010).

RFS display greater consistency in the preferred coordination path among shoe types (**Table 5-2**) but this may result in less flexibility. These results are in line with recent studies investigating the effect of different shoes on the preferred movement

path in habitual rearfoot strikers (Weir et al., 2018). To our knowledge, our study is the first to investigate adaptation in forefoot strikers.

Forefoot strikers tend to have greater coordination variability. FFS tend to have lower cycle-to-cycle consistency (ACC values), and larger sum of variance (**Table 5-1**) compared to RFS, thus partially fulfilling our first hypothesis that FFS have a larger movement solution space. In addition, FFS tend to use more combinations of ankle-knee coupling in all footwear conditions. Such richness of coordinative variability has been proposed to be indicative of a more flexible system (Hamill, Palmer, & Van Emmerik, 2012). These results are in accordance to the higher adaptability of ankle stiffness in the FFS group described in the previous Chapter (Chapter 4). The end point kinematics is mainly achieved by controlling ankle joint stiffness (Yen & Chang, 2010) and thus the relative rotation of segments. Covariance among limb segments can be reduced to two principal components that stabilize leg length and leg orientation (Ivanenko et al., 2007). Similarly here, the coordination between joint angles can be assumed to stabilize the leg length and orientation, and hence, the body centre of mass position. By adapting the ankle angle, FFS define the range of possible movement solutions along the other joints, so that either by compensation or collaboration, inter-joint coupling produces stable performance.

Coordination variability is effected by shoe type. Our findings do not reveal an effect of shoes on any index of cycle-to-cycle variability. However, from analysis of the spatial differences between the group mean cyclograms (**Figure 5-3**) one can appreciate the effect of the shoe features in ankle-knee-hip coordination. Assuming that each footwear condition required a unique movement plane and therefore unique joint coupling, the strategies used by the two groups were the least different in high MI shoes. This is consistent with the similar preferred coordination paths displayed in **Figure 5-2B**.

Reduced differences may be caused by the absence of cushioning materials underneath the heel or the medial aspect of the shoe in high MI shoes. In this condition – high MI shoes - RFS may be able to ‘mimic’ the coordinative patterns of FFS, by adopting a more plantarflexed ankle (McCallion, Donne, Fleming, & Blanksby, 2014; Squadrone, Rodano, Hamill, & Preatoni, 2015). However, as indicated by the lower sum of variance (**Table 5-2**), the amount of variability the RFS have available in this condition may still not be enough to acquire an adaptable pattern. In low MI shoes, the

ankle at TO was more dorsiflexed for RFS than FFS, and it changes as an inverse function of shoe minimal index; the change in ankle joint affected the coupling of this joint (ankle) with the other joints (knee and hip).

The difference in joint coupling during swing may represent a neuro-mechanical adaptation (Cavagna, 2006). The knee-hip coupling was the most consistent and also the most similar between groups (**Table 5-2**). We expected such a coupling to be the least sensitive to change, or to be the most difficult to change, based on previous studies that also found knee-hip coupling to be more in phase than knee-ankle in sprinters performing at maximal speed (Gittoes & Wilson, 2010).

Indeed, using a treadmill to test our hypothesis may have limited variability to some extent (Dingwell, Cusumano, Cavanagh, & Sternad, 2001), but the treadmill allowed us to analyse continuous gait cycles and avoid subjective selection of cycles and analysis of a rather low number of steps. Another possible limitation is the absence of kinetic data that may have helped confirm some of our hypotheses.

Moreover, most of the results did not reach statistical significance when testing for differences between groups. This can be partly explained by a small sample size and the individual adaptations that each runner involved in the study may have developed through their own running experience. Nevertheless, both visual inspection of the preferred coordination path and quantification of spatial variability are relevant tools that qualitatively and quantitatively describe differences between these two groups of runners.

We presented a rather simple methodology to calculate differences between cyclograms. The vectorisation technique we used is based on basic trigonometry and easily applicable. Other methods such as continuous relative phase (Hamill et al., 1999) could also be used to describe joint coordination, but it implies the transformation of the data, calculation of phase angles, and calculation of the continuous relative phase. Although this technique has indisputable clinical relevance (Lamb & Stöckl, 2014), it did not serve the scope of our research. Lastly, we did not extend computation of variability indexes on the 3-dimensional coordination path. The interest was to investigate joint coupling at first; the application of the ACC, Sum of variance, and SSD on the 3D coordination path will strengthen the qualitative results presented here.

5.6 Conclusion

The ability of runners to coordinate lower-limb segments and joints represents aspects of gait that complement the information on running adaptability reported in Chapter 4, and provides additional insights into the underlying mechanics explaining stable performance. The preferred coordination path is inherently stable among subjects; however, FFS exhibited a greater ability to change gait behaviour to accommodate environmental conditions. Habitual FFS may have developed, through their running experience, a coordinative pattern that is more variable in essence, and is equipped to better respond to different shoe conditions.

5.7 References

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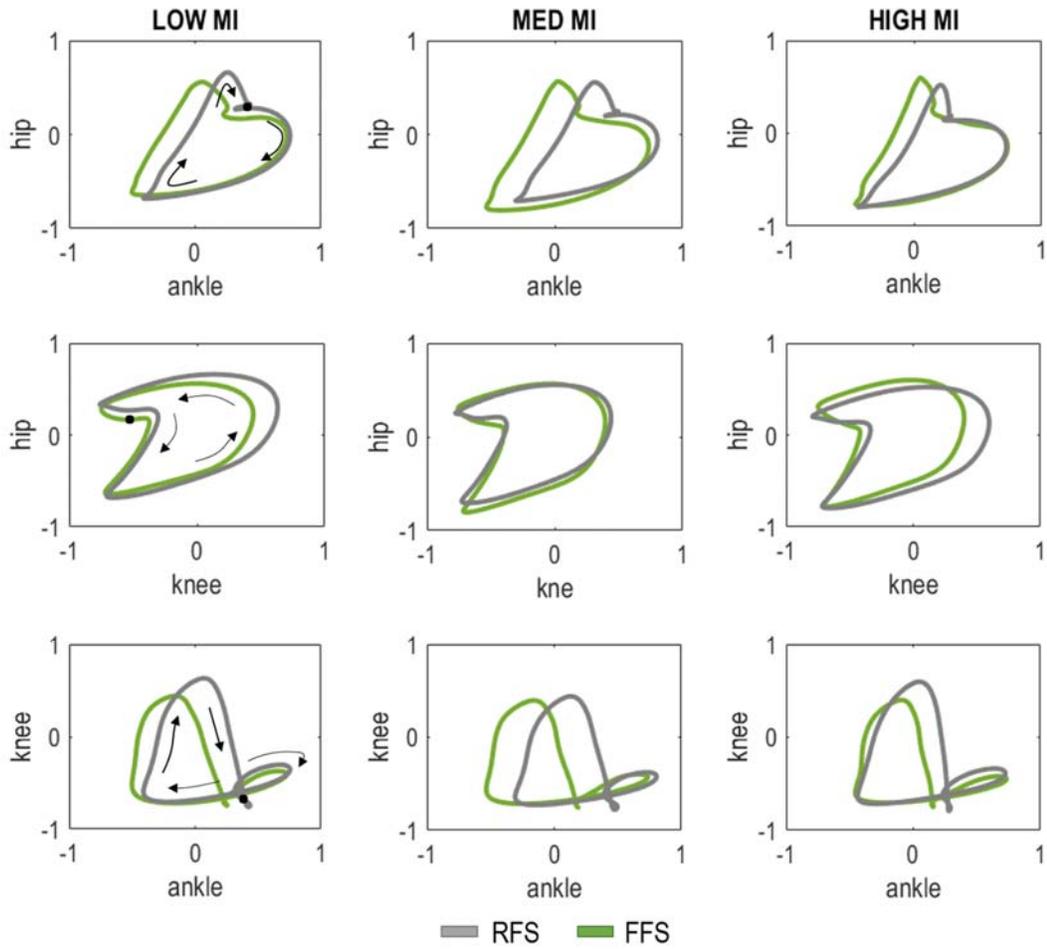
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5.8 Supplementary A

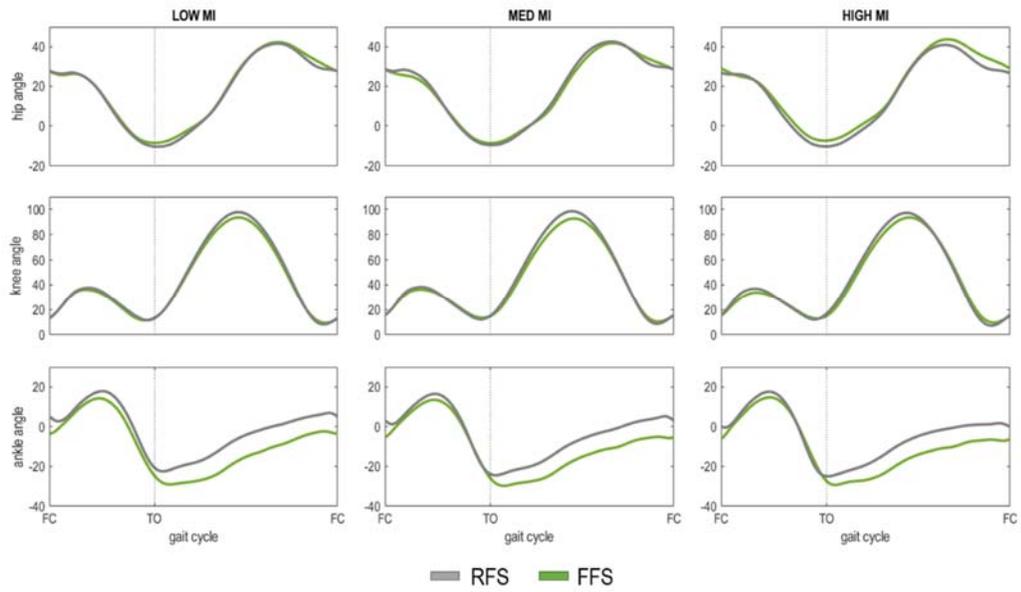
Group mean of two-dimensional cyclograms: hip-ankle (**top**), hip-knee (**middle**), and knee-ankle (**bottom**). Comparison are showed for each footwear condition.

• indicates foot contact.



5.9 Supplementary B

Joint angles for hip, knee, and ankle.



6 LEG STIFFNESS CONTROL IN LONG-DISTANCE RUNNERS: EFFECT OF FOOT STRIKE AND SHOE FEATURES

6.1 Abstract

Be able to adjust leg stiffness in response to different conditions is vital for the health and performance of runners. However, the ability to control leg stiffness may be influenced by the habitual loading pattern of runners and by the support provided by the shoes they wear. In this chapter we explore the modulation of leg stiffness through the loading and unloading phase of running using analysis of persistence in long time series. Differences and similarities between rearfoot and forefoot striker runners are interpreted within the two theoretical framework of optimal feedback control and dynamic system theory. First, by running correlations between level of leg stiffness control and leg stiffness variance, we found that regulation of leg stiffness is task-dependent: the high-level controller is responsible for leg stiffness control during the loading phase, while the low level controller is responsible for leg stiffness control at impact and during unloading phase of running. At group level, we found that rearfoot strikers have restricted neuro-locomotor entropy that is relevant to leg stiffness control; and contrary to what expected, we found regulation of leg stiffness control to be independent from shoe support.

6.2 Introduction

For humans that engage in regular and long periods of running, the factors that affect the control of leg stiffness are most relevant (Almeida, Davis, & Lopes, 2015; LeBlanc & Ferkranus, 2018; Valenzuela, Lynn, Mikelson, Noffal, & Judelson, 2015). Experimental data and theoretical models from human and animal studies indicate that steady-state running is optimal when the combined costs of energy, posture instability and injury risk are minimised; and critically, leg stiffness appears as the essential biomechanical parameter that mediates these goals (Daley, Voloshina, & Biewener, 2009; Seyfarth, Geyer, Günther, & Blickhan, 2002; Shen & Seipel, 2015b, 2018). Common locomotor control theory suggests that a runner's control policy requires the attribute of leg stiffness to be adaptive in order for it to shift between its competing priorities (Birn-Jeffery et al., 2014; Shen & Seipel, 2015b, 2018). For example, adaptable landing patterns during the loading phase of stance can mitigate the effect of external forces that threaten to perturb the body into unsafe and destabilising biomechanical states (Latash, Scholz, & Schoner, 2007). Furthermore, loading patterns that are controlled by an adaptable neuro-locomotor system might enable a more energy efficient solution during the subsequent unloading period (Kuo, 2002; Ruina, Bertram, & Srinivasan, 2005).

Indeed, shoe and foot posture are well researched topics in human running biomechanics, and this is not surprising because they are two critical factors that influence the legs' force-length dynamics during both loading and unloading phases of stance (Addison & Lieberman, 2015; Bishop, Fiolkowski, Conrad, Brunt, & Horodyski, 2006; Divert, Baur, Mornieux, Mayer, & Belli, 2005; Krogt et al., 2009). Clinical studies of human running suggest that too much stiffness may be associated with skeletal injuries, while too little stiffness may be associated with muscle-tendon injuries (Granata, Padua, & Wilson, 2002; Williams, McClay Davis, Scholz, Hamill, & Buchanan, 2003). Theoretical studies suggest there is an ideal range of leg stiffness that allows a runner to optimize the priorities of energy and stability (Shen & Seipel, 2015b, 2018). Meeting this leg stiffness range might be simplified by shoe-assisted rearfoot loading. Also, shoe can assist with minimising the energy cost of limb unloading during the propulsive phase of stance (Oh & Park, 2017). While these benefits of shoe are appealing, there is actually very limited information about the

long-term effect on the neuro-locomotor control system that arises from frequent intensive periods of shoe-assisted rearfoot loading patterns.

The essential properties of the embodied neuro-musculoskeletal system that influence the leg force-length dynamics during loading and unloading phases of running are often expressed using a variant of the spring loaded inverted pendulum model (SLIP). The model uses a spring-damping function to express the leg length-force behaviour during loading, and a spring-actuation function to express leg biomechanics during unloading (**Figure 6-1**). Leg stiffness relates to the force-length ratio of the curve and there are different methods for its calculation (Blum, Lipfert, & Seyfarth, 2009). When running at preferred speed, the peak of the ground reaction force signal generally occurs between 40-45% of the stance period (Cavanagh & LaFortune, 1980; Frederick & Hagy, 1986), and prior to peak leg compression (Cavagna, 2006; Cavagna, Legramandi, & Peyre-Tartaruga, 2008). This underscores the asymmetric force-time profile of running. The force-length graph is equivalent to an examination of the collective stress-strain property of the leg system components, such as elasticity, hysteresis and energy loss. In a non-actuated passive leg system, all the stored potential energy created during the loading phase is completely returned to the system during the unloading phase; the system is considered elastic and the curve is linear and symmetric. In situations where the passive leg system loading-unloading profile is asymmetric but the initial and final length is equivalent, the stored potential energy is lost; i.e. hysteresis. Experimental data shows that the loading-unloading force-length profile across stance phase of a human shod runner is asymmetric and irregular after the load exceeds body weight (Cavagna, 2006; Farley & Morgenroth, 1999); whilst experimental data from animals (unshod) show profiles that are more symmetric and regular (Birn-Jeffery et al., 2014). Simulations using various SLIP models confirm that a combination of factors affect the storage and recovery of energy during loading-unloading, including inter-joint coordination, timing of muscle actuation, foot and limb posture at initial contact, shoe and surface material (Kram, 2000; Kram & Taylor, 1990). Of these types of studies, there are few that have directly investigated leg length-force dynamics and the differences between rearfoot and forefoot loading patterns (Miller & Hamill, 2015; Viale, Dalleau, Freychat, Lacour, & Belli, 1998). In their study, Miller and Hamill (2015) used a more advanced method (musculo-skeletal modelling) to investigate which cost functions were minimized by

which foot strike pattern (i.e. rearfoot versus forefoot). From the 44 different cost functions tested they found RFS were optimal in minimizing metabolic cost, while FFS were optimal in minimizing lower limb loading at the cost of ankle loading.

Empirical data shows that forefoot strikers have higher leg stiffness compared to rearfoot strikers (Laughton, Davis, & Hamill, 2003), but this can provide a misleading message due to two reasons. First, the collective biomechanical degrees of freedom that govern leg length changes due to additional foot and ankle compliance is naturally higher in a forefoot loading technique (Nigg, 2010). Second, the commonly adopted method for defining leg stiffness – as the ratio between peak force and change in leg length – overlooks the time-course of the force profile as loading evolves up to peak force. For instance, a high rate of ground reaction force loading is likely to be associated with high stiffness (assuming corresponding change in leg length remains fixed), and this will get missed with effective leg stiffness calculation. Indeed, studies that compare rearfoot and forefoot landing techniques report higher force-time loading rates for the rearfoot technique (Boyer, Rooney, & Derrick, 2014; Hamill & Gruber, 2017; Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, et al., 2010). Studies rarely report instantaneous leg stiffness during early loading period (Oliver & Smith, 2010), but biomechanical theory suggests that it is more likely that rearfoot landing technique would demonstrate a remarkable increase in instantaneous leg stiffness during initial impact period compared to forefoot landing technique. During initial impact phase the force-time and force-length dynamics shows a dependence on landing technique, with changes to force frequency content (Gruber, Edwards, Hamill, Derrick, & Boyer, 2017) and changes to leg effective mass (Clark, Ryan, & Weyand, 2017; Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, et al., 2010). Therefore, it is plausible that there is a sequence of two task-relevant sub-phases with different goals (and cost policies), which occur during the time-course of the loading phase. In following the optimal feedback control theory framework (see Chapter 1.1.3), the locomotor controller is likely to adopt a cost policy that shifts priorities as the loading period evolves. The policy is likely to reward states that meet stability and safety during initial impact phase, and as loading evolves towards peak force the policy shifts the reward on energy economy states (Shen & Seipel, 2018). No studies that we are aware of, have confirmed the nature of a dual-goal policy during the loading period when

running. However, if such a policy exists, then conventional methods that calculate the effective leg stiffness will not be sensitive.

Advanced biomechanical modelling studies of jumping have demonstrated the role of passively generating potential energy in the properties of muscle-tendon units during loading phase result in minimal energy cost from muscle actuation during unloading (Bobbert, Yeadon, & Nigg, 1992; Wade, Lichtwark, & Farris, 2018). A similar experiment design has not yet examined the comparative effect between footwear-assisted rearfoot loading (RFS) and minimal-assisted forefoot loading (FFS) on the biomechanical behaviour of the system during unloading phase. We have contributing evidence to this story of FFS runners transferring energy stored from the loading phase and recovering it for unloading. In Chapter 4 we observed that RFS produce relatively higher positive ankle work compared to negative work across the stance phase. Moreover, we observed in the FFS group that ankle stiffness during loading sub-phase explains 63% of ankle stiffness variance during the unloading sub-phase when wearing minimal supportive shoes; and this did not occur for any RFS conditions. Such evidence can suggest that shoe-assisted rearfoot loading would be associated with less elastic loading of the ankle-foot muscle-tendon units and this will have flow-on consequences with motor command strategy and energy efficiency during unloading.

Evidence shows that leg stiffness is a control parameter of the locomotor control system and therefore any change to the system should be directly expressed by the behaviour of leg stiffness control (Shen & Seipel, 2015a, 2018). Chapter 1 illustrated how the neuro-locomotor control system can be effectively modelled from a combination of two theories: dynamical systems theory and optimal feedback control theory (Chapter 1.1.3). The system is supervised by an active high-level controller that adheres to the principle of minimum intervention (Dingwell, John, & Cusumano, 2010; Latash, Gorniak, & Zatsiorsky, 2008; Todorov & Jordan, 2002), preferring control to be managed at a low-level by a complex self-organised system with biomechanical trajectories attracted to passively stable states (Goswami, Espiau, & Keramane, 1996, 1997). This model demonstrates good accuracy with experimental data, and therefore it allows a framework for interpreting influential factors of locomotor control. The property of the model belonging to complexity and dynamical systems theory has relevance for the question in this chapter: *how does shoe-assisted*

rearfoot loading influence the adaptability of the neuro-locomotor control system. A high degree of system complexity (rich dimensionality of system resources) is important to the high-level controller that prefers minimal regulation of control, and quantifying control regulation can infer the state of complexity in the system. The concept of entropy was introduced in Chapter 1, and entropy regression is a property of a system losing its potential for adaptable solutions. In this chapter we aim to investigate whether there is evidence of system entropy (and loss of potential for adaptability) in long distance runners habituated with a shoe-assisted rearfoot loading pattern. By selecting leg stiffness as the parameter of interest, we can expect a more sensitive appraisal of the systems resources and how they are governed to effect a goal-oriented outcome. Furthermore, we can expect that this goal-relevant parameter has consistent weighting of priority between participants.

Among many tools that quantify system complexity, one approach has successfully demonstrated an ability to detect the level of effort by the central nervous system to regulate locomotor control by examining persistence (i.e. a scale of self-similar structure) in the time series of a known control parameter or performance variable of gait (Bohnsack-McLagan, Cusumano, & Dingwell, 2016; Cusumano & Dingwell, 2013; Dingwell & Cusumano, 2015). Gait parameters that demonstrate persistent correlations of their time-series signal are considered to be an expression of a complex self-organised system (Hausdorff et al., 1997; Scafetta, Marchi, & West, 2009; Warlop, Detrembleur, Stoquart, Lejeune, & Jeanjean, 2018), while random correlations and anti-persistent structure suggests higher level active intervention (Dingwell, Bohnsack-McLagan, & Cusumano, 2018; Dingwell et al., 2010). It has been shown that signal complexity is reduced in locomotor systems affected by disease and ageing (Hausdorff et al., 1997), and from fatigue and injury (Meardon, Hamill, & Derrick, 2011). Essentially, these biologically affected locomotor systems also demonstrate a loss of persistence; but in contrast to control regulation effects on persistence, the biological effects are indicators of a more permanent regression of system entropy and an indicator of an inherently less complex and adaptable system. Two investigations by Dingwell et al. (2018) and Dingwell et al. (2010) validated their experimental data and theory – that persistence is an indicator of central nervous system intervention to correct goal-relevant deviations of gait parameters – with a simulation model of locomotor control that adheres to the minimum intervention

principle when supervising a dynamical system. In this chapter we adopt this signal analysis tool and general control regulation theory – but without the model validation – and employ it to assess empirical data of stride-to-stride leg stiffness time-series.

A system with an expanded level of entropy (higher complexity) will express persistence in time-series and its processes will functionally interact within and between spatio-temporal scales (van Emmerik, Ducharme, Amado, & Hamill, 2016). Such adaptive system will have a larger set of abundant solutions to satisfy the goals (length-force dynamics) of the control system (Costa, Peng, Goldberger, & Hausdorff, 2003). There is more likelihood that the high entropy system will self-regulate divergent trajectories to a stable state through its inherent allometric control processes (West, 2010); which suggests that an optimal leg length-force state can emerge as a goal-relevant solution from a low-level control process. Therefore, in a high entropy system, there will be less need for intervention on divergent trajectories, and such parameters represented as a time series will show relatively high statistical persistence (approximating $1/f$ -type noise). In essence, the allometric control processes of a high entropy system is highly adaptive. Nevertheless, certain goal-relevant locomotor variables that regularly deviate from a target biomechanical state will be controlled according to optimal feedback control theory and regular higher-level central nervous system intervention will express low statistical (anti-) persistence in the time series (Dingwell & Cusumano, 2010).

The purpose of this chapter is to evaluate if habitual loading technique has an effect on the level of control of a parameter that is directly relevant to the task goals of running. The premise is that a reduction in statistical persistence when the task is known to be under minimal control regulation is an indicator of a reduction in adaptability to perform this task. We evaluate the level of leg stiffness control regulation in two groups of long distance runners, distinguished by their habituation to shoe-assisted rearfoot loading and minimal-assisted forefoot loading. We also investigate the acute effect of shoe structure on their control system.

There are three general hypotheses of this study. First, control regulation is dependent on the phase of the task: the two control systems are not equally responsible for statistical persistence in leg stiffness time series throughout the stance phase. We expect that central nervous system control regulation occurs when leg stiffness persistence correlates with a change in leg stiffness performance (variance). We expect

central nervous system control regulation of leg stiffness to be highest in the loading phase (safety goal) and lowest in the unloading phase (economy goal). Second, the habituation of footwear-assisted rearfoot loading technique and long distance running will reduce neuro-locomotor adaptability to perform the task of regulating leg stiffness during loading. Third, the level of structural assistance provided by the shoe will affect leg stiffness control differently for runners habituated to a minimal-assisted forefoot loading technique compared to runners habituated to cushioned-shoe-assisted rearfoot loading technique. For RFS, high-assistance footwear will require less control regulation of leg stiffness compared to the unfamiliar minimal assistance footwear. In contrast, FFS with minimal-assistance footwear will require less control regulation of leg stiffness compared to the unfamiliar high-assistance footwear.

6.3 Methods

Participants' characteristics and testing protocol are the same as per previous Chapters. Refer to Chapter 4.3 for details.

6.3.1 Data Analysis

Raw kinematic and kinetic data was exported from Nexus 2.6 (VICON) to Visual 3D (C-motion Pty, USA) for processing and parameterisation. The kinematic and kinetic signals were low-pass filtered using a Butterworth filter (4th order, zero lag) with a cut-off frequency of 15 Hz and 35 Hz respectively. Gait events were defined using the vertical component of the ground reaction force - an ascending and descending threshold of 20N identified foot contact (FC), and toe-off (TO) respectively. Within this time period, four other events were created from the body-weight normalised ground reaction force signal exceeded 0.2, 1.0 body weight (BW), when it reached a maximum, and when it fell below 0.2 BW. These events were used to sub-divide the body stance phase of running into three task-relevant phases: 0.2-1 BW, impact (K1); 1-max, loading (K2); and max to 0.2 BW unloading (K3). These phases display a unique leg stiffness profile: while K1 and K3 are almost linear, K2 may lose linearity depending on foot strike (**Figure 6-1**). Each participant's lower limb was modelled as a planar spring-mass system (Blickhan, 1989) from which leg stiffness, k_{leg} , was calculated as $\Delta F/\Delta L$, where ΔF is the change in ground reaction resultant force, while ΔL represents the change in leg length (normalized to the recorded leg length in the standing position) and equal to the change in length of the 3D distance vector starting

at the pelvis centre of mass and ending at the centre of pressure (Liew, Morris, Masters, & Netto, 2017). Leg stiffness was then computed for each of the three phases: K1, K2, and K3. The leg stiffness time series for each condition was exported to Matlab (The MathWorks Inc., Massachusetts, US) for processing statistics of control.

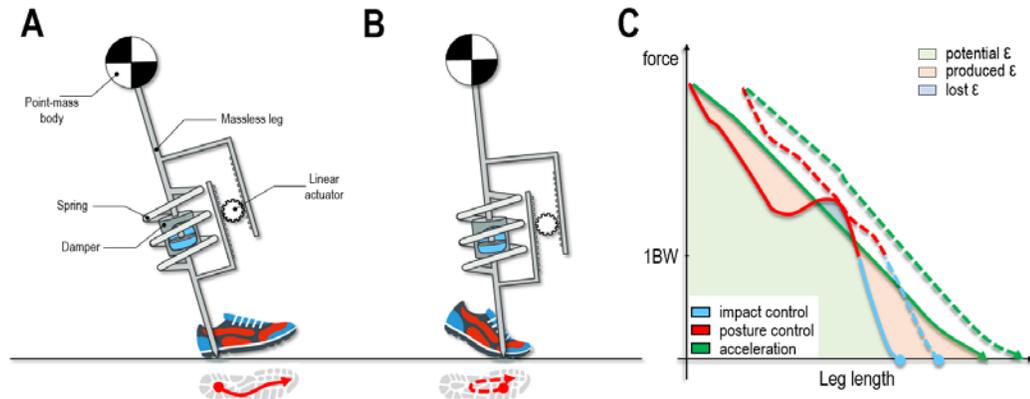


Figure 6-1 (A) Schematic virtual leg-spring model used to simulate running with a rearfoot strike pattern, and (B) with forefoot striker pattern (Adapted from Birn-Jeffery et al., 2014). Centre of pressure trajectory beneath the shoe is also displayed. (C) Comparison of rearfoot loading (solid line) and forefoot loading (broken line) landing types and their ground reaction force changes as a function of leg length. Curves are divided into three task-relevant sub-phases: impact control, loading, unloading. The slope and area features of the graph represent leg stiffness and energy respectively. Leg stiffness is largest during the first sub-phase. The area under the curves represent the potential energy, produced energy, and lost energy during the stance phase.

6.3.2 Detrended Fluctuation Analysis

Detrended Fluctuation Analysis (DFA) was a method originally designed to measure the scaling index (known as α) of long range correlations and fractal-like (self-similar) structure of time-series signals arising in parameters representing complex systems (Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995). When there are no assumptions about the underlying origins of the long-range correlations, the DFA method is more conservatively used to quantify statistical persistence of a time series (Dingwell & Cusumano, 2010). Empirical data and simulation models of the locomotor control system demonstrate that either cognitive stresses (control regulation) or reduction in system complexity (i.e. specialisation or low-dimensionality) can cause a breakdown

in statistical persistence by presenting random-like fluctuations in the signal (Goldberger et al., 2002; Scafetta et al., 2009; Yogev et al., 2005). Statistical persistence is present when α values are between 0.6 and 1.0; while a break-down of persistent structure occurs when α values approach 0.5 (Dingwell & Cusumano, 2010; Peng et al., 1995). Under the model of hierarchical locomotor control, the minimum intervention principle and dynamical systems theory; α values are interpreted as the product of both control regulation and system complexity. Using this interpretation, high α values (≈ 1.0) will be due to loose control regulation and a high-dimensional complex system (Dingwell et al., 2010). In this case, a trend of small deviations are free to persist in future gait cycles. In contrast, low α values (≈ 0.5) represent tight control regulation or a system that has reduced complexity and interacting components have become low dimensional. In this case, small deviations do not persist between consecutive gait cycles (Dingwell et al., 2010).

Statistical persistence was computed in Matlab (The MathWorks Inc., Massachusetts, US) from a customised program (Taylor, 2012) adapted from conventional detrended fluctuation analysis method (Bashan, Bartsch, Kantelhardt, & Havlin, 2008; Hausdorff et al., 1995). The general procedure for calculating the scaling exponent, $DFA\alpha$, followed these five general steps: 1) obtain a random-walk time-series profile (Y_n) by integrating the original time series (x_n) by partial summation; 2) divide the integrated time series (Y_n) into non-overlapping equal sized windows (time scales) of $w = \{9, 17, 33, 65, 129\}$; 3) detrend the integrated random-walk profile (Y_n) within each window segment, w , by peicewise fitting a linear trend to each window by a least squares fitting function and concatenating the residuals to form a new detrended time series \tilde{Y}_n ; 4) compute the average fluctuation variance F , within each window scale w , of the detrended time series \tilde{Y}_n ; 5) plot the average fluctuations F , per window size w , on a log-log graph and determine the linear relationship using a least-squares linear fitting function. The DFA scaling exponent α is the slope determined from step 5.

6.3.3 Statistical analysis

Mean, standard deviation (SD), and coefficient of variation (CV) were computed for each *Group* \times *Shoe* \times *Phase* condition. Because the biomechanical attributes and functional roles between left and right limbs can often be asymmetric, we considered

dominant and non-dominant limbs of the participants as separate cases (i.e. $n_{FF} = 20$, $n_{RF} = 20$). A mixed design 3-factor (*Group* \times *Shoe* \times *Phase*) repeated-measures ANOVA was used to examine the interaction and main effects of within-subject factors of *Shoe* (3 levels: low MI, medium MI, high MI) and task-dependent *Phase* (3 levels: K1, K2 and K3 – these acronyms relate to leg stiffness during impact, loading, unloading sub-phases), and between-subject factor of foot loading type *Group* (2 levels: forefoot, rearfoot) on the four dependent variables of leg stiffness (mean) and leg stiffness control (SD, CV, DFA α). Significance was set at 0.05 for all tests. Planned contrasts examined specific levels of an interaction effect between *Group*, *Phase* and *Shoe*. Tukey post-hoc analysis was used to test multiple pairwise comparisons. All statistics were performed using SPSS software (version 25, SPSS Inc., Chicago, IL, USA).

6.4 Results

6.4.1 Reconciling control system responsibility for causes of low DFA α

Prior to addressing results for hypotheses 2 and 3, it is important to acknowledge the source of control that underlies the DFA α values. The interpretation of results related to DFA α require an understanding based on a two-system control hierarchy model consistent with an optimal feedback control theory framework (Todorov & Jordan, 2002). The high-level control system sets the control policy and optimises leg stiffness performance by adhering to the minimum intervention principle. The low-level control system is represented by dynamical systems theory (Kelso & Schöner, 1988), and concepts of self-organisation and allometric control govern coordination of the embodied elemental components of the system. Under this model, both control systems can independently effect a reduction in statistical persistence; reflecting a constraint of entropy at the low level, or increased control regulation from high level.

This is a critical issue that is overlooked in nearly all studies that employ DFA to gain insight into the human locomotor system, which is why many of these studies lack precision when interpreting how an experimental treatment causes changes to the DFA scaling index (e.g. Fuller et al. (2016); Meardon et al. (2011)). With the exception of Dingwell, Salinas, and Cusumano (2017), no studies of gait control have attempted to reconcile DFA α results within a two-level hierarchical control system. While this study was not specifically designed to reconcile these dual-contributions of control

regulation and system complexity on DFA α , we can make some plausible deductions based on the relationship between DFA α and CV, and take advantage of the repeat-test design of this study. The first premise is that if the high level controller causes the DFA α to reduce by top-down intervention, then there should exist a sensitive change to the task performance; otherwise, the high-level controller would not choose to intervene. The second premise, is that within an embodied system there will be no changes to the complexity of the system provided that the conditions of the task are consistent. This is possible when a minimal-assisted forefoot loading runner (FFS) performs a running trial in moderately assisted shoe (med-MI) and then repeats the condition but in minimal assisted shoe (high MI). Likewise, when a shoe-assisted rearfoot loading runner (RFS) performs a running trial in high assistance shoe (low MI) and then repeats the condition in moderate assisted shoe (med-MI). By comparing the differences in DFA α and CV within subject and within limb, we analyse how change in control process (DFA α) correlates with change in performance outcome (CV). By combining the results of both FFS and RFS within each task-dependent phase of stance (K1, K2 and K3), we find correlations of $r = 0.2, -0.6, 0.1$ respectively. By separating the data into RFS and FFS, the results are consistent: $r = 0.2, -0.7, 0.1$ for RFS, and $r = 0.2, -0.6, 0.4$ for FFS. The results indicate that both groups adopt the same strategy of control: the high-level controller is responsible for leg stiffness control at K2, while the low level controller is responsible for leg stiffness control at K1 and K3.

6.4.2 *Effect of Group and Phase*

There was a main effect for *Group*, where FFS have a higher ($p = .027$) DFA α compared to RFS. Phase had a significant main effect on DFA α ($p < .001$, **Table 6-1**), indicating that on average, DFA α was dependent on the phase of the stance task. Post hoc tests reveal that DFA α is higher ($p < .001$) at K3 compared to K2.

There was a trend for a *Group* \times *Phase* interaction effect on the DFA α ($p = .113$, **Table 6-1**, **Figure 6-2**), indicating a potential difference between groups in the way they regulate the control of stiffness between phases. Planned contrasts compared the groups between phases K1 and K2 ($p = .017$) and between K2 and K3 ($p = .067$). These contrasts showed that there is a difference between groups in transition of control behaviour from impact phase (K1) to loading phase (K2). Moreover, while

both groups reduce tight control of leg stiffness during transition from loading to unloading (i.e. from K2 to K3), the FFS group made a relatively higher change to DFA α compared to RFS (**Figure 6-2**). For direct within-group pairwise comparisons between K1 and K2, the FFS group had a higher ($p = .044$) DFA α at the impact phase (K1). For direct within-group pairwise comparisons between K2 and K3 the FFS group reveal a higher ($p < .001$) DFA α at K3.

For the dependent variables CV and mean leg stiffness the significant main effect of *Phase* ($p < .05$; **Table 6-1**, Figure 3) was not unexpected. Pairwise comparisons show that leg stiffness is stronger ($p < .001$) and more inconsistent ($p < .001$) at K1 compared to K2; while comparing between K2 and K3, leg stiffness at K2 is stronger ($p < .001$) and more inconsistent ($p < .001$) compared to K3. However, while both groups display a similar mean ($p > .05$) and CV ($p > .05$) of leg stiffness during K2, their behaviour at K1 and K3 is different. Hence, there was a significant interaction effect of '*Phase \times Group*' on mean and CV of leg stiffness ($p < .05$; **Table 6-1**, **Figure 6-3**). For direct within-group pairwise comparisons between K1 and K3, both groups had a stronger ($p < .001$) mean leg stiffness and a larger ($p < .001$) CV at the impact phase (K1). For between-group comparisons, RFS exhibited a stronger ($p = .034$) mean leg stiffness, while FFS exhibiting larger ($p = .023$) CV.

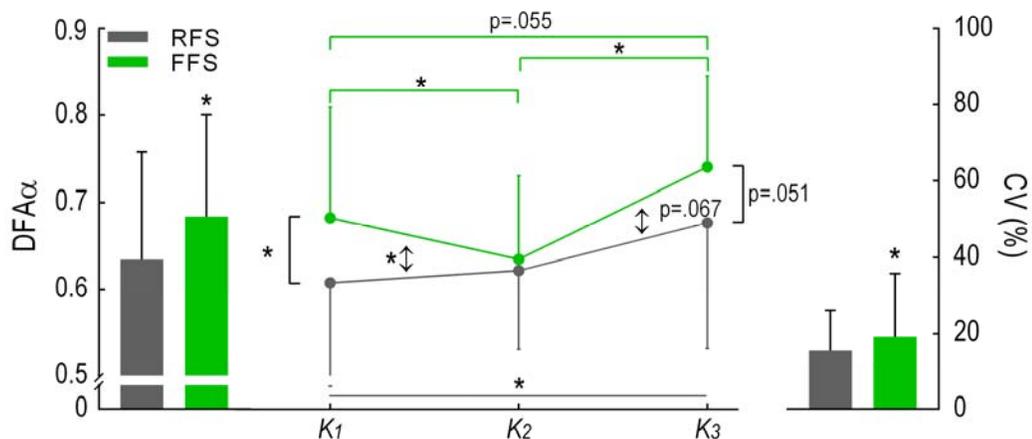


Figure 6-2 Group mean and SD of DFA α values averaged across shoe types for each group, and over the three task-relevant sub-phases of the stance phase. Bar graphs show between-group (FFS vs RFS) differences for average DFA α and average CV across sub-phases and shoe type. * represents significance level $p < .05$; for group \times phase interaction effects, and pairwise comparisons for between group and between phase.

Table 6-1 Primary statistical results for differences between Groups, Shoes, and Phase for mean leg stiffness, standard deviation (SD), coefficient of variation (CV), and mean DFA α values. ANOVA results are given for main effects and interactions. Statistically significant findings are in bold.

Variable	Group	Shoe	Phase	Group \times Shoe	Group \times Phase	Shoe \times Phase	Group \times Shoe \times Phase
Mean	$F_{(1,38)} = \mathbf{5.35}$ $p = \mathbf{.026}$	$F_{(2,76)} = \mathbf{17.55}$ $p < \mathbf{.001}$	$F_{(2,76)} = \mathbf{176.94}$ $p < \mathbf{.001}$	$F_{(2,76)} = \mathbf{10.83}$ $p < \mathbf{.001}$	$F_{(2,76)} = \mathbf{4.42}$ $p = \mathbf{.041}$	$F_{(4,152)} = \mathbf{19.33}$ $p < \mathbf{.001}$	$F_{(4,152)} = \mathbf{11.94}$ $p < \mathbf{.001}$
SD	$F_{(1,38)} = 0.82$ $p = .372$	$F_{(2,76)} = \mathbf{11.81}$ $p = \mathbf{.001}$	$F_{(2,76)} = \mathbf{69.07}$ $p < \mathbf{.001}$	$F_{(2,76)} = 2.06$ $p = .156$	$F_{(2,76)} = 0.81$ $p = .374$	$F_{(4,152)} = \mathbf{10.54}$ $p = \mathbf{.001}$	$F_{(4,152)} = 1.07$ $p = .319$
CV	$F_{(1,38)} = \mathbf{5.31}$ $p = \mathbf{.027}$	$F_{(2,76)} = \mathbf{12.97}$ $p < \mathbf{.001}$	$F_{(2,76)} = \mathbf{90.06}$ $p < \mathbf{.001}$	$F_{(2,76)} = \mathbf{8.03}$ $p = \mathbf{.003}$	$F_{(2,76)} = \mathbf{4.79}$ $p = \mathbf{.031}$	$F_{(4,152)} = \mathbf{5.66}$ $p = \mathbf{.007}$	$F_{(4,152)} = 0.54$ $p = .561$
DFA α	$F_{(1,38)} = \mathbf{5.31}$ $p = \mathbf{.027}$	$F_{(2,76)} = 1.42$ $p = .250$	$F_{(2,76)} = \mathbf{14.69}$ $p < \mathbf{.001}$	$F_{(2,76)} = 0.06$ $p = .942$	$F_{(2,76)} = 2.25$ $p = .113$	$F_{(4,152)} = 0.35$ $p = .846$	$F_{(4,152)} = 1.60$ $p = .178$

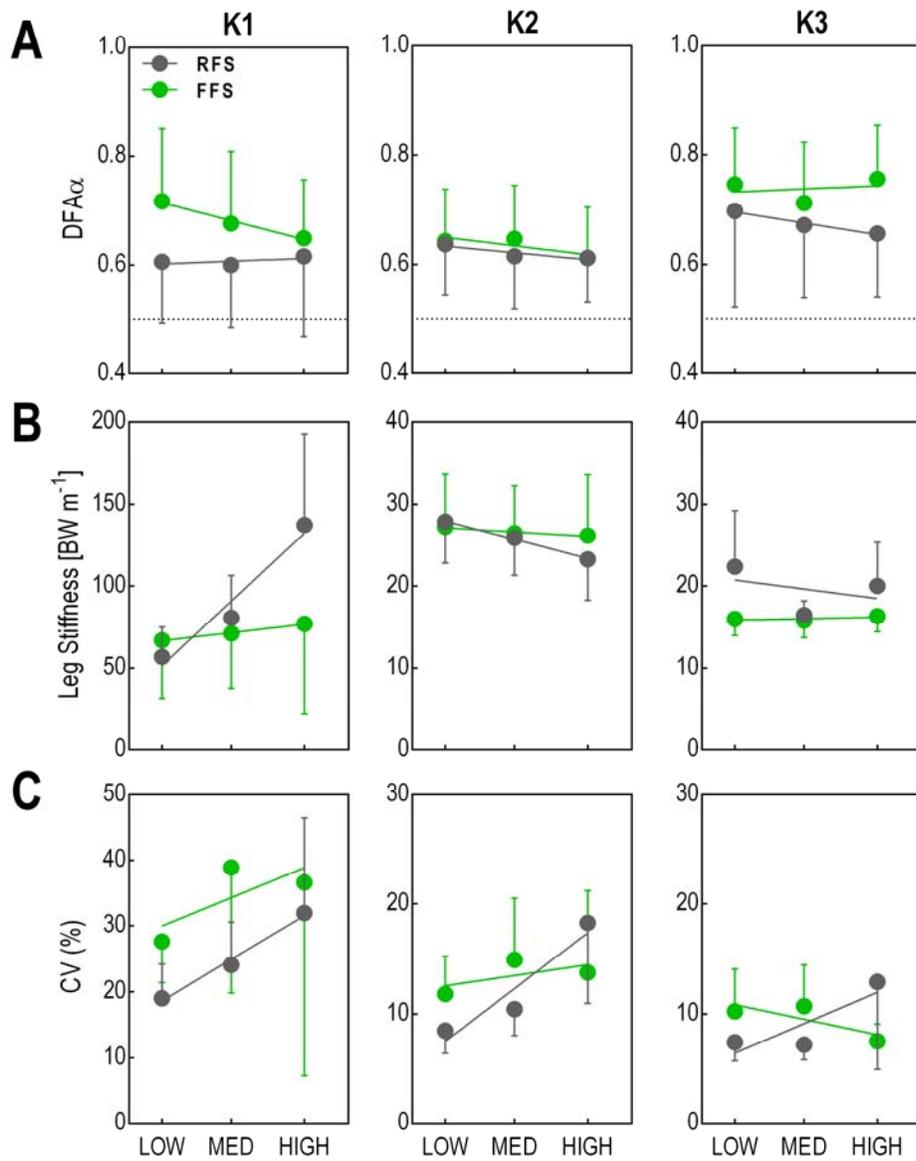


Figure 6-3 Group mean and SD represented for each task-relevant phase (K1-K3) and shoe type (LOW, MED, HIGH) for dependent variables: (A) DFA α of leg stiffness, (B) mean leg stiffness, and (C) CV of leg stiffness.

Note: magnified scale in (B) and (C) for sub-phases K2 and K3.

6.4.3 Effect of Shoe

For the dependant variable DFA α : shoe did not have a significant effect on the interaction between *Group* \times *Phase* ($p = .178$; **Table 6-1, Figure 6-3**); there was no main effect for Shoe ($p = .250$), nor interaction effects for Shoe \times Group ($p = .942$) or Shoe \times Phase ($p = .846$). Therefore, the interaction effect of *Shoe* did not change the *Group* \times *Phase* behaviour identified in Hypothesis 1.

For both dependent variables CV and mean leg stiffness: there was a significant main effect of *Shoe* ($p < .05$; **Table 6-1, Figure 6-3**); and significant interaction effects of *Shoe* \times *Phase* and *Shoe* \times *Group* ($p < .05$; **Table 6-1, Figure 6-3**). Pairwise comparison of mean leg stiffness revealed RFS have stronger (+44%, $p = .001$, **Table 6-2**) mean leg stiffness with high MI shoes during K1, and stronger (+19%, $p = .006$, **Table 6-2**) leg stiffness during K3. Also, RFS have stronger (+29%, $p < .001$) mean leg stiffness when running in low MI in phase K3. As shown in **Figure 6-3**, the habitual rearfoot loading group (RFS) increase stiffness as the minimal index of the shoe increases (LOW-MED $p = .083$; MED-HIGH $p < .001$). In contrast, the habitual forefoot loading group (FFS) produced mean leg stiffness that did not change significantly with shoe ($p > .05$). Pairwise comparisons of CV revealed FFS have higher (+42%, $p < .001$, Table 2) CV in low MI shoes, and higher (+55%, $p < .001$, **Table 6-2**) CV in med MI shoes compared to RFS. In all phases, CV values tend to increase from low MI to high MI shoes. Differences were not significant only for med MI compared to high MI in phase K1 ($p = .882$), and among all shoe types in phase K3 ($p \geq .479$).

Table 6-2 Group mean and (SD) for leg stiffness mean, SD, CV and DFA α values in the three functional phases of impact (K1), loading (K2), and unloading (K3). Comparisons are made among the three shoe type (LOW, MED, HIGH) and pooled data.

	LOW		MED		HIGH		POOLED	
	FFS	RFS	FFS	RFS	FFS	RFS	FFS	RFS
K1								
mean	66.95 (35.34)	56.78 (18.4)	71.15 (33.7)	80.56 (25.82)	76.96 (55.30)	137.43 (55.26)	71.69 (42.07)	91.59 (49.72)
SD	19.86 (15.89)	10.86 (4.33)	30.5 (22.59)	19.29 (7.84)	37.04 (54.52)	41.6 (17.97)	29.13 (35.41)	23.91 (17.34)
CV	27.6 (6.21)	19.04 (5.13)	38.92 (19.11)	24.06 (6.48)	36.63 (29.33)	31.97 (14.51)	34.38 (20.77)	25.02 (10.89)
DFA α	0.72 (0.13)	0.61 (0.11)	0.68 (0.13)	0.60 (0.11)	0.65 (0.11)	0.62 (0.15)	0.68 (0.13)	0.61 (0.12)
K2								
mean	27.23 (6.45)	27.86 (5.04)	26.46 (5.81)	25.89 (4.55)	26.17 (7.43)	23.29 (5.04)	26.62 (6.50)	25.68 (5.16)
SD	3.17 (1.11)	2.36 (0.74)	3.92 (2.02)	2.69 (0.76)	3.33 (1.62)	4.13 (1.72)	3.47 (1.63)	3.06 (1.38)
CV	11.8 (3.38)	8.47 (2.05)	14.9 (5.60)	10.41 (2.37)	13.74 (7.46)	18.26 (7.31)	13.48 (5.78)	12.38 (6.21)
DFA α	0.64 (0.09)	0.64 (0.09)	0.65 (0.10)	0.61 (0.10)	0.61 (0.09)	0.61 (0.08)	0.63 (0.09)	0.62 (0.09)
K3								
mean	16 (2.00)	22.37 (6.84)	15.78 (2.05)	16.45 (1.73)	16.32 (1.87)	20.03 (5.33)	16.03 (1.95)	19.62 (5.58)
SD	1.64 (0.65)	1.7 (0.75)	1.69 (0.65)	1.18 (0.24)	1.23 (0.29)	2.68 (1.83)	1.52 (0.59)	1.86 (1.29)
CV	10.22 (3.82)	7.42 (1.68)	10.69 (3.72)	7.2 (1.35)	7.53 (1.55)	12.88 (7.93)	9.48 (3.45)	9.17 (5.36)
DFA α	0.75 (0.10)	0.7 (0.18)	0.71 (0.11)	0.67 (0.13)	0.76 (0.10)	0.66 (0.12)	0.74 (0.10)	0.68 (0.14)

6.5 Discussion

To gain appropriate insight into neuro-locomotor system complexity and its control regulation of a goal-relevant parameter of running, the quantification of leg stiffness variability parameters was measured at three goal-relevant subtasks of the stance phase: impact (stability goal, K1), loading (safety goal, K2), and unloading (economy goal, K3). The first general hypothesis of this study was that statistical persistence can detect system entropy, which is relevant to the control and adaptability of leg stiffness during running. The second hypothesis was that control regulation of leg stiffness will be different between groups and it will be task-dependent. The third general hypothesis of this study was that control regulation of leg stiffness will be affected by the level of shoe assistance and this effect would be different between groups.

6.5.1 The DFA-CV results support the first hypothesis that control regulation of leg stiffness involves the interaction of two control systems and this varies with the time-course of stance.

The scaling exponent during the loading phase (K2) suggests the expression of intervention regularity, where higher-level central nervous system control is most responsible. Therefore, during this period it is not appropriate to infer that persistence measured by the DFA scaling exponent is a representation of system complexity and allometric control. In contrast, because changes to statistical persistence are not associated with outcome performance during the impact and unloading phases, there is a reliance on allometric control during these sub-phases, and hence the persistence measured within these periods provides for a more exclusive expression of system entropy and adaptable self-regulation. The implication of this result on reconciling control responsibility of leg stiffness allows for more precise interpretation hereafter.

6.5.2 The DFA-CV results support the second hypothesis that control regulation of leg stiffness is phase and group dependent.

6.5.2.1 Footwear-assisted rearfoot loading technique for leg stiffness control is associated with reduced system adaptability.

Long distance runners that adopt a habitual shoe-assisted rearfoot loading technique generally exhibit lower persistence (**Figure 6-2, Table 6-1 and 2**), lower variability

of performance at early loading phase (K1) and unloading phase (K3), compared to subject-matched minimal-assisted forefoot loading runners. Why do we expect larger CV with large DFA α values for leg stiffness control by the FFS group? Based on a general theory that large entropy would require sufficiently higher level of neural control costs to achieve performance precision (Manohar et al., 2015; Tassa, Erez, & Todorov, 2011); however, a neuro-locomotor system with large entropy (high dimensional workspace) is a pre-requisite for embedding within it an abundant reserve of multiple stable limit cycle attractors that are somewhat imprecise (due to expanded entropy) but require limited control cost (Seyfarth et al., 2002; Tassa et al., 2011). This idea of locomotor control is what the results of this study appear to be reflecting (albeit without verification by a test simulation model). High regularity (low CV) and low statistical persistence may be indicative of low entropic locomotor control system that is less adaptable (Costa, Goldberger, & Peng, 2002), and consistent with ageing (Hausdorff et al., 1997), pathology (Gruber et al., 2011; Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998; Manor et al., 2010), running fatigue (Meardon et al., 2011), and possibly running speed (Fuller et al., 2016). The results suggest that a habitual shoe-assisted rearfoot loading technique has an embodied system that has become less adaptable for controlling leg stiffness.

6.5.2.2 The control of leg stiffness during loading phase (K2) is tightly regulated by a higher-level control system.

Statistical persistence of leg stiffness changes during the time-course of stance phase; as loading transitions from impact towards peak loading, the control of leg stiffness transitions from dependence on system entropy towards increased intervention from central nervous system control. This phenomenon appears consistent for both groups. After we established which control level is operating in the different task-relevant phases, these results suggest that repetitive shoe-assisted rearfoot loading from long-distance running may enhance specialisation of biomechanical patterns (low CV) but at the expense of neuro-locomotor adaptability (low DFA α). Adaptability relates to stability (persistence) and flexibility (variability) of performance (Li, Haddad, & Hamill, 2005) and they are equally essential to execute skilled movements. For instance, expert athletes display regular movement patterns that are not fixed into rigidly stable solutions, but they can functionally adapt in response to environmental constraints (Davids, Bennett, & Newell, 2006; Glazier & Davids, 2009). In contrast,

novices tend to exhibit similar behaviour in all situations (Chow, Davids, Button, & Koh, 2007) suggesting less adaptability.

Control of leg stiffness during impact phase K1 depends on feedforward adaptations of leg posture pre landing. Impact phase is a too short time period for the control system to process any afferent sensory feedback; with a neural short-latency of 20-50 ms post perturbation feedback responses may arise only after the external limb loading reaches 100% body weight (end of K1) (Cavagna et al., 2008). Motor control in very fast action (as impact phase is) depends on the accurate prediction of the state estimate (Crevecoeur & Scott, 2014) based on feedforward strategies. Previous studies have shown a more plantarflexed ankle before touchdown when running across unpredictable terrains (Müller, Ernst, & Blickhan, 2012; Müller, Häufle, & Blickhan, 2015). This feedforward guided strategy in preparation to landing can be regarded as the exploration of the system mechanics (Blickhan et al., 2006) under low level control (Haeufle, Günther, Wunner, & Schmitt, 2014). Although visual feedback will modulate the feedforward strategies (Müller et al., 2015), we could not discern between the two. However, we assume the constraint of our experiment (indoor steady-state treadmill running) would have provide the same visual feedback information to all participant, thus equalising its effects. Therefore the higher level of DFA α that FFS have during the impact phase (**Figure 6-2, Table 6-2**) shows that their system has a larger flexibility resulting in an adaptable body configuration at landing that better deal with impact forces.

During general loading phase (K2), leg stiffness is tightly regulated by both groups. There are at least three reasons why leg stiffness control is tightly regulated during this period. First, a mismatch between the predicted and required leg stiffness has to be adjusted by active control of feedback information. Short and long-latency (>50 ms) sensory feedback responses are now available (Pruszynski & Scott, 2012); these signals arrive to the nervous system with noise (Faisal, Selen, & Wolpert, 2008) that needs to be processed presumably by allocation of neural resources (Faisal et al., 2008) or higher firing rate (Manohar et al., 2015) which implies higher control. Second, after irrelevant information are attenuated and relevant signals properly represented, this information needs to be translated in precise co-contraction of agonist and antagonist muscles, in order to safely load the passive elements of the skeletal muscle tendon units – controlled leg effector stiffness. Reduction of motor errors

entails isolating the motor system from competing affordances that requires higher control (Manohar et al., 2015). Third, muscle loading requires eccentric contractions which pose an increased risk of injuries. Previous studies have shown poorer movement accuracy during eccentric compared to concentric contraction (Fang, Siemionow, Sahgal, Xiong, & Yue, 2004; Yao et al., 2014); this has been linked to a different underlying control mechanism of motor neuron excitability at corticospinal (Duclay, Pasquet, Martin, & Duchateau, 2011) as well as cerebral level (Yao et al., 2016). Taken together, the system in K2 is dealing with filtering noisy sensory feedbacks, producing precise movement coordination, and avoiding possible injurious muscle contractions; thus the benefit of intervening to optimize performance outbalance the cost of intervention (Manohar et al., 2015).

During the unloading phase (K3) leg stiffness does not requires higher control. According to the principle of minimal intervention (Todorov & Jordan, 2002) higher intervention will compromise the objective (goal state) of this phase: energy minimisation. Unloading of the leg is achieved by transferring contact force stored as elastic energy in passive elements during the loading phase into upward momentum of the body (Wade et al., 2018). Imposing a control over leg stiffness will mean losing the stored energy, while requiring an increased energy production. Adaptability in this phase is important for performance (Ueno et al., 2018a; Ueno et al., 2018b) and it relates to the tendon elastic strain energy in the ankle plantar flexor (Lai, Schache, Brown, & Pandy, 2016). The more energy is stored in the elastic components, the less energy (positive work) will be required for forward progression (see Chapter 4).

The differences in control strategy between K2 and K3 can be exemplified with a simple illustration: imagine compressing a spring between two fingers; the phase that requires active control and is more precarious (risk of losing the grip on the spring) is the loading phase. If we are able to fully load the spring, then the unloading phase requires less control and indeed the goal is to minimize the resistance applied to the spring during this phase, to maximise the energy potential.

6.5.3 *The DFA-CV results do not support the third hypothesis that control regulation of leg stiffness is shoe-dependent.*

Regulation of leg stiffness control is independent from shoe support.

Our third hypothesis, that runners would require less control regulation when running in familiar shoes compared to unfamiliar shoes, was not supported. However, during phase K1 there was a tendency for system entropy of FFS to reduce as shoe offered less support (**Figure 6-3A, Table 6-2**); this was unexpected because less supportive shoes (high MI index) were expected to allow more freedom (more degrees of freedom available) for the system to express its complexity (Lawrence, Gottwald, Khan, & Kramer, 2012; Newell & Vaillancourt, 2001). Nevertheless, compared to RFS that kept a constant level of DFA α , FFS adapted the control of leg stiffness as they changed shoe. A constraint in entropy was necessary in less supportive shoes (high MI) to maintain an invariant level of leg stiffness (**Figure 6-3C, Table 6-2**). It is possible that FFS had to constrain their entropy to find functional solutions passing from fully supportive to non-supportive shoes; this hypothesis is partially supported by a decrease in CV values changing from medium supportive shoes to non-supportive shoes. Support also comes from studies investigating the change in system dimensionality as a function of task constraint (Araújo, Davids, Bennett, Button, & Chapman, 2004; McGregor, Busa, Skufca, Yaggie, & Bollt, 2009; Newell, Broderick, Deutsch, & Slifkin, 2003). These studies found that dimensionality is subjected to change depending on the task constraints. In contrast, as the shoe support reduces, the system entropy of RFS remained similar (**Figure 6-3B, Table 6-2**). A possible explanation for RFS unchanged entropy is that RFS are unable to correctly estimate the consequences of landing (inaccuracy in the state estimate), or they are unable to use a functional feedforward strategy (see section 4.2). Although RFS attempted to change their kinematical configuration at landing passing from low MI to high MI shoes (see Chapter 4, and 5), they were unable to find functional solutions to maintain a constant level of stiffness. Similar increase in leg stiffness was found in habitual rearfoot strikers passing from wearing supportive shoes to minimal-supportive shoes (Lussiana, Hébert-Losier, & Mourot, 2015). This results suggest that shoes support may have an effect on the control of leg stiffness, and FFS running in minimal supportive shoes may constrain the entropy of the system in order to keep a constant level of stiffness during landing.

In summary, in this study we reconciled DFA α with the control hierarchy (Figure 6-4) founding that regulation of leg stiffness is task-dependent. This is new information that can help in interpreting past results and formulating new hypothesis for future studies (see Chapter 10.3).

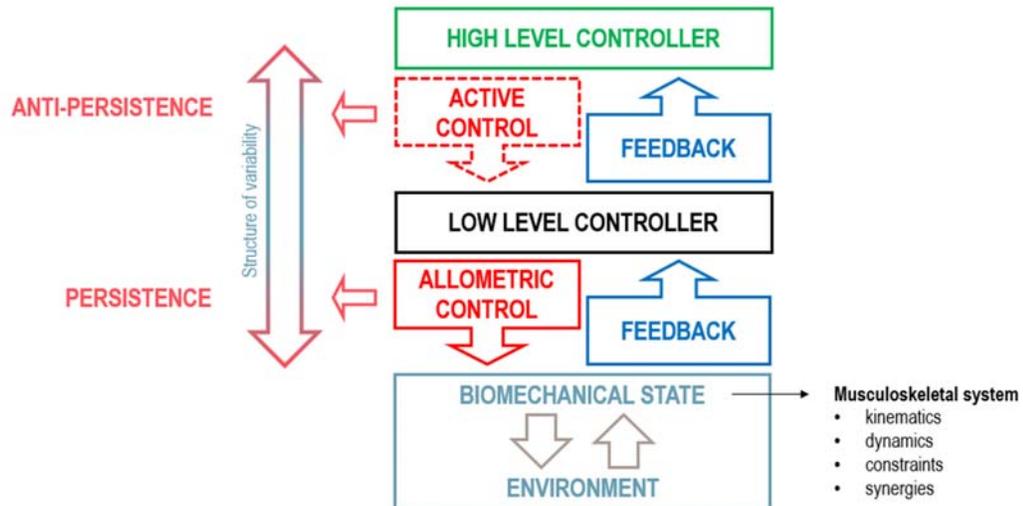


Figure 6-4 Conceptual control diagram. Active intervention from the high level controller will cause the DFA α to increase toward anti-persistence if the cost policy is not meet (i.e. too high, too low leg stiffness). If cost policy is meet, despite high movement variability, the high level controller will not intervene but rather leave the low level controller to exercise its allometric control over the biomechanical state. This will make the DFA α to decrease toward persistence.

Moreover, rearfoot strikers have restricted neuro-locomotor entropy that is relevant to leg stiffness control; and contrary to what was expected, we found regulation of leg stiffness control to be independent from shoe support. This is critical information for performance and injury prevention. Runners may want to consider introducing more variability in their daily training in order to challenge their system so that it can expand. One possible change may be to gradually increase the time spent running with a forefoot strike pattern. This is a well-known prescription, however, the neurophysiological advantages of adopting a forefoot strike pattern were not defined so far. We provide convincing evidence that habitual forefoot strikers develop a more adaptable system; and adaptable system by definition may be better at dealing with

external perturbations than more rigid systems, preventing overload of anatomical structures by optimal organising of motor redundancy.

6.5.4 *Study Limitations*

This study has several limitations. First, we considered that shoe classified by a minimalist index 'MI' provides equivalent loading and unloading control assistance for both RFS and FFS runners. It is possible that assistance can change between loading and unloading. Furthermore, a low MI shoe could be assistive for a RFS runner but unassistive for a FFS runner. The different effects of shoe on group could have prevented the identification of optimal shoe-type for optimal loading-unloading control. Second, we interpret DFA α results as representing the dual-effect of system complexity (high-dimensional degrees of freedom) and higher-level control regulation (Chapter 1.3). We based our interpretation on the link between DFA α and CV (see Section 6.1), and the premise that when their relative change is correlated there must be top-down control intervention. This theory will need evidence from appropriate model simulations that support empirical data. Third, only by examining statistical persistence in the covariant and redundant variables of force and leg length, will we ascertain a clear insight into how leg stiffness is being controlled (see Chapter 10.3). Fourth, we used the term entropy to describe system complexity (and therefore its adaptability) but we did not measure system entropy directly. For example, multiscale entropy (MSE) has been applied to time-series to evaluate complexity across multiple scales in standing and walking (Costa et al., 2002; Costa, Goldberger, & Peng, 2005; Costa et al., 2003; Lipsitz & Goldberger, 1992; van Emmerik et al., 2016) and while DFA has shown an association with entropy (Costa et al., 2002), further work is required to prove that DFA of leg stiffness during loading phase is an accurate representation of system adaptability. Part of the solution to this issue will require quantifying system entropy from a different experiment design to that of this study, one that produces a larger data set but minimises causes that can lead to fluctuating control regulation (e.g. distraction, fatigue). Last, we have to acknowledge an appropriate but limited sample size and a gender restriction that limits generalisation of the results (see Chapter 1.4). The strict inclusion criteria was necessary to ensure the sample of selected runners was an appropriate representation of the population they were intended to represent and their demographics were equivalent between groups

(i.e. body mass, average running load per week). Of the original 40 willing participants, 50% were excluded after familiarisation tests. In addition, we selected only male participants because it reduced confounding gender-relevant factors that are associated with interpreting bone density and structure (Riggs et al., 2004); i.e. results from Chapter 3.

6.6 Conclusion

In this study we used a theoretical framework of neuro-locomotor control and DFA to investigate the hierarchical control systems that govern leg stiffness during loading and unloading phases of running. We found that both FFS and RFS runners generally abide by the same degree of control regulation. However, we reason that the embodied complexity of the shoe-assisted rearfoot loading pattern has evolved differently and will influence adaptable motor-command patterns during the unloading phase of stance.

6.7 References

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7 LIMB EFFECTOR CONTROL DURING THE LANDING PHASE OF RUNNING: EFFECT OF FOOT STRIKE AND SHOE FEATURES

7.1 Abstract

The task goal during landing is to safely control the external forces that can destabilize the body. Those forces are controlled deploying the abundant degrees of freedom in our system to stabilize leg length and leg orientation. Differences may exist in how these variables are controlled at landing depending on the habitual foot strike pattern of runners and the shoes they wear. In this study we investigate how the nervous system manages the abundant degrees of freedom in segment angles in order to stabilize the performance variables leg length and orientation. We utilised uncontrolled manifold theory and method to quantify goal-relevant and goal-irrelevant deviations from a set of consistent gait cycles taken from a larger set of trials produced from five minutes of treadmill running. To investigate the effect of foot strike and footwear, we compared the running pattern of habitual forefoot strikers and habitual rearfoot strikers in three shoe conditions. First we established that variance at segment level is structured to stabilise leg length and orientation during landing. Second, we found runners adopt a similar control policy, where deviations that are goal-relevant (i.e. they influence performance) are corrected, and deviations that are goal-irrelevant (i.e. they do not influence performance) are allowed. Pre landing, the goal-relevant deviations in leg length are minimized and deviations in leg orientation allowed. This helps the system in achieving critical tasks such as stability, energy, and injury prevention. The rapid shift in control structure prior to landing, indicates that control of the kinematic state of the leg at impact is most reliant on feedforward prediction rather than fast feedback from proprioception senses. During initial impact phase (called K1 in Chapter 6), there is a decrease to goal-relevant deviations relative to goal-irrelevant deviations, thereby ensuring a stronger synergy that produces a more consistent leg length during this period. Between groups we found habitual forefoot strikers tend to have higher level of goal-irrelevant deviations (high system entropy) but differences are affected by shoe conditions.

7.2 Introduction

Running consists of repetitive jump-land sequences performed successfully at a rate of about 1500 cycles per mile (930 per km) (Hoeger, Bond, Ransdell, Shimon, & Merugu, 2008), which for a long distance runner that regularly completes 40km per week will be approximately 37,000 impacts that load the limb with 2.5 times their body weight. Although landing seems an apparently easy and common task, many runners become injured as a result of excessive tissue stress from an accumulation of these repetitive loading events (Daoud et al., 2012; Messier et al., 2018).

The posture of the leg can be represented by a kinematic vector spanning the joints (i.e. leg effector), where the vector components can define the effective leg length and orientation (Auyang, Yen, & Chang, 2009); by organising the multiple degrees of freedom at the three main joints (hip, knee, and ankle) the problem of global kinematics motor redundancy of the leg can be reduced to the control of the overall leg length and leg orientation (Ivanenko, d'Avella, Poppele, & Lacquaniti, 2008). Therefore, leg posture and orientation can be used as goal-level variables to test the effect of different landing styles and type of footwear. For instance, the loading phase of running can be successfully achieved with the foot approaching the ground from a range of different foot posture and loading patterns (Lieberman et al., 2010). One study investigated long-distance runners of similar demographics and found that habitual shod runners have reduced locomotor variations during loading compared to habitual barefoot runners (Lieberman et al., 2015). However, it is unknown how the kinematic elements of the leg organise so that any noise or deviation in their covariant coordination has limited effect on leg length and orientation around the impact and loading phase of running; and whether this is affected by landing style or foot loading strategy and type of footwear assistance.

There have been several different approaches to investigate how the complex neuro-locomotor system achieves control over the many redundant degrees-of-freedom of elemental variables (EV) for achieving a goal-relevant performance variable (PV). For example, principle component analysis (Ivanenko, Cappellini, Dominici, Poppele, & Lacquaniti, 2007), and covariation by randomization (Müller & Sternad, 2003), have been used among others. The principal component analysis method reduces the redundancy in elemental variables (i.e. three joints) by plotting joints angles on each other (phase relationship) thus defining a plane with two principal

components (i.e. performance variables) that explain the majority of the variance (Ivanenko et al., 2007). The covariation method compares variability at the goal-level between empirical and de-correlated surrogate data. By looking at correlation between elemental variables, the structure of the variance that is not caused by the correlation is not detected (Schöner & Scholz, 2007). The most popular method that seek to discover the structure of variance has been the uncontrolled manifold theory (UCM), which was first proposed by (Scholz & Schöner, 1999). It has since been applied in a range of gait-related tasks, such as walking (Cusumano, John, & Dingwell, 2008; Monaco, Tropea, Rinaldi, & Micera, 2018; Papi, Rowe, & Pomeroy, 2015), and hopping (Auyang et al., 2009), but it has not been applied to running. The UCM has an advantage over other methods because it maps the covariance of elemental variables to the performance variable within the same (geometric/physical) space and units as the performance variable (Schöner & Scholz, 2007).

The UCM hypothesis shares the same theory as the minimum intervention principle – MIP (Todorov, 2004); the variability about the manifold space (UCM) represents a two-level control hierarchy scheme. **Figure 7-1** illustrates the low-level controller of the UCM, defined by covariance of elemental variables (EV1 and EV2, e.g. segment angles) that vary freely within the manifold space of a performance variable (V_{UCM} , i.e. goal-relevant task such as leg length), while consistent with theory of MIP, the high-level controller intervenes only when cooperating element variables deviate orthogonal to the manifold space (V_{ORTH}). The ratio of the variance formed by the set of parallel deviations and orthogonal deviations to this manifold space is defined by motor control theorists as the effect of a motor synergy (Latash, Scholz, & Schöner, 2007; Todorov, Li, & Pan, 2005). By observing how the control system partitions kinematic variance to stabilize leg length and orientation, we can gain an insight into the ability of the locomotor system to deal with a critical phase of running such as landing.

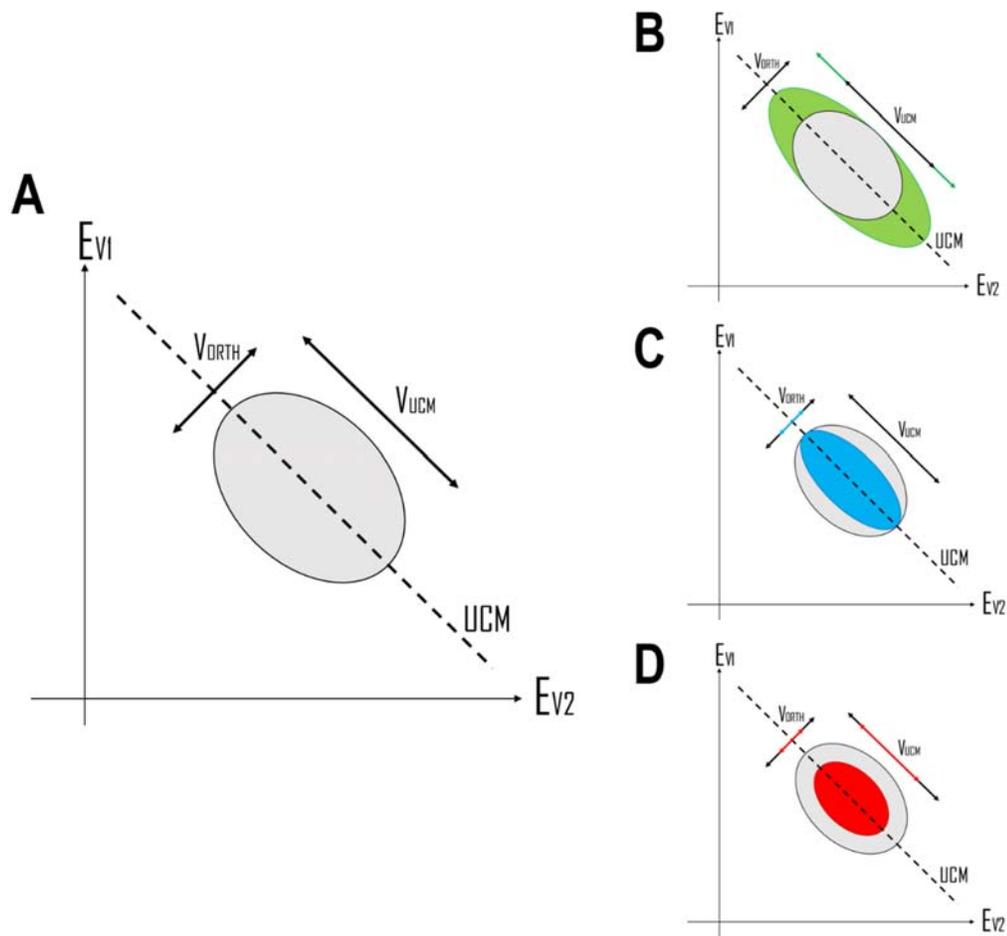


Figure 7-1 (A) Multi-dimensional manifold represented in 2D space, showing two elemental variables (EV1-2) and one performance variable (UCM, projected as a line). (B) Expanded V_{UCM} , (C) constricted V_{ORTH} , (D) constricted both V_{UCM} and V_{ORTH} .

The ability to control the forces generating during landing is critical (Selgrade & Chang, 2015; Yen & Chang, 2010), and optimal leg stiffness appears to be a goal state for the locomotor system (Birn-Jeffery et al., 2014; Shen & Seipel, 2018) see also Chapter 6. The change in the state of leg length and orientation will directly contribute to leg stiffness and hence the external force applied to the body (i.e. leg dynamics) (Arampatzis, Schade, Walsh, & Brüggemann, 2001; Hobara et al., 2010). The locomotor system objective during the loading phase of running is to safely absorb and harness the kinetic and potential energy of the body, while maintaining balance (Daley

& Biewener, 2006; Seyfarth, Geyer, Günther, & Blickhan, 2002). The previous chapter described running biomechanics using the concept of the leg behaving like a spring-loaded inverted pendulum with actuation and feedback control (see section 6.1): the leg adopts a certain stiffness to attain goal-relevant properties of stability, safety and energy efficiency during the loading and unloading phases of ground contact (Shen & Seipel, 2018). Leg stiffness and load stress from the external ground reaction force is influenced by the rate of change in leg posture (Ivanenko et al., 2008); moreover, the foot-ankle posture at landing influences leg stiffness (Yen, Auyang, & Chang, 2009). To enact the fine-control task of spring-like action of the leg, the central nervous system coordinates high-dimensional elements of an embodied neuro-muscular-tendon-skeletal workspace into cohesive low-dimensional synergies that provide primitive control of joint torques and segment angles to minimise effect of perturbations on the goal of precise alteration in the state of leg orientation and length.

7.2.1 Can the measure of GID(par) be used as an indicator of motor abundance and system flexibility?

Selgrade and Chang (2015) demonstrated that setting a target/goal peak force for a hopping task resulted in a reduced GRD(orth), while the GID(par) remained unchanged. This indicates that control of the orthogonal goal-relevant deviations was restricted due to higher-level CNS control without affecting the parallel goal-irrelevant deviations (lower-level). It appears from this result, that the parallel variance might be a true representation of the motor abundance (Latash & Anson, 2006; Yang & Scholz, 2005). When the task was designed to adapt to a new target Force, there was a change to GID(par), but the GRD(orth) remained consistent with baseline behaviour. This result suggests that motor abundance [GID(par)] can be expanded with only a minor increase to GRD(orth). Therefore, it appears that GID(par) is a fixed entity unless adaptation to a new condition is required. If the adaptation task is challenging, it can be expected that GRD(par) is relatively larger to the familiar task. Whether this control of goal-relevant variance behaviour is evident in runners that change their footwear type is unknown. Yen and Chang (2010) used a UCM analysis to demonstrate that the loading phase was consistent with the minimum intervention principle; and the beginning of the stance phase of a hop (loading) had more GID(par) relative to GRD(orth). These authors were able to separate covariation and individual joint

variation, and by quantifying the sensitivity of each joint they found that the ankle is always the most important joint defining vertical force variance. Robert, Bennett, Russell, Zirker, and Abel (2009) used DFA and UCM to demonstrate that GRD(orth) was being controlled by CNS process; but it wasn't known whether direct CNS control would also restrict the measure of motor abundance (GIDpar). Approaching the ground, we expect the ratio of the variance structure to change in order to optimise performance (Liu & Todorov, 2007); however, a constraint in a joint may result in decreased availability of degrees of freedom, resulting in a constraint of the GID(par); whether constraints at elemental variables do modify the structure of the variance is still unknown.

The primary purpose of this study was to determine if components of the leg effector (length and orientation) are stabilized during impact phase of running, and to examine whether this stability is affected by foot loading type of the runner or by footwear type. In other terms, we want to know how lower limb segment angles covary in goal-relevant space of leg length and orientation (i.e. vertical and horizontal dimension of the leg effector) while subjects inherently control their leg force-length dynamics when running in different shoes. There were three hypotheses. First, we expected that both performance variables leg length and leg orientation will be stabilized during impact phase with evidence from proportionately more variance of goal-irrelevant deviations relative to goal-relevant deviations. Because vertical dimension is most relevant to leg force-length dynamics, we also expect a stronger ratio of variance for stabilising leg length performance compared to leg orientation.

Second, because of an expected constraint on the kinematic degrees of freedom during impact, we expected that a habitual rearfoot loading technique will have less variance of goal-irrelevant deviations and less variance of goal-relevant deviations compared to forefoot loading technique. We had no expectation of how a proportionate structure of variance between goal-relevant and goal-irrelevant deviations would be different between the groups because we expected a general decrease in both components for the rearfoot loading technique. Third, we expected that increased footwear assistance would reduce the amount of goal-relevant deviations more than goal-irrelevant deviations for the rearfoot loading group, because the shoe is supposed to augment leg force-length dynamics for this group. Overall, we hypothesize that habitual forefoot loading technique will have relatively higher variance in goal-

irrelevant deviations in both vertical and fore-aft performance tasks, for all footwear conditions, and this between-group difference is expected to be greatest for minimal assistive shoes.

7.3 Methods

Participants' characteristics and testing protocol are the same as per previous Chapters. Refer to Chapter 4, section 3 for details.

7.3.1 *Data processing and analysis*

Joint position was recorded from 21 retro-reflective markers (14 or 9 mm diameter) attached to pelvis, thigh, shank and feet. To describe body locomotion, the body was represented as a planar system of 7 rigid segments (pelvis, thigh, shank, and feet) with six degrees of freedom (**Figure 7-2**). Raw data was exported to Visual 3D (C-motion) and filtered using a low-pass Butterworth filter (4th order, zero lag) with a cut-off frequency of 15 Hz.

Gait events were defined using the vertical component of the ground reaction force: an ascending and descending threshold of 20N identified foot contact (FC), and foot off respectively. Two other events were created 40 ms before foot contact (FC-10), and 40 ms after foot contact (FC+10) defining the pre landing phase (PRE) and post landing phase (POST). The latter, Post landing phase, can be referred to the impact phase (K1, ≈ 50 ms) investigated in the Chapter 6. Data were then exported into Matlab (The MathWorks Inc., Massachusetts, US) to evaluate the structure of variances within the UCM framework.

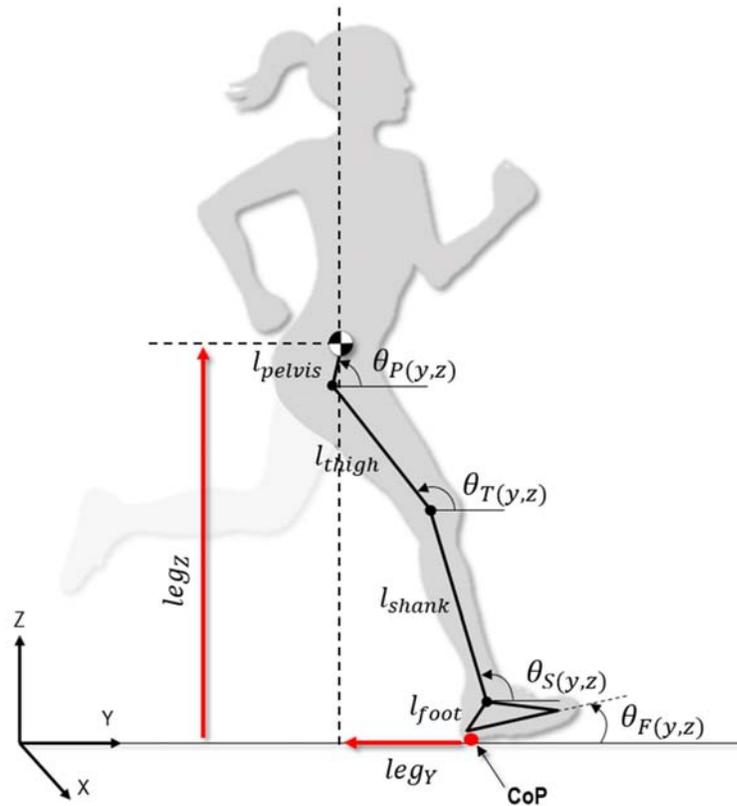


Figure 7-2 Geometric model used to estimate performance variables and joint angles.

7.3.2 Uncontrolled Manifold formulation

To understand how the locomotor control is addressing the problem of ubiquitous variance in the redundant kinematic leg effector system (segment angles) are being coordinated and controlled during the leg landing phase of running, to achieve consistency in the performance variables of the 2D leg effector end-point (i.e. the leg orientation and length, L_Y , L_Z respectively). The set of covariant solutions of these elemental variables (i.e. foot, shank, thigh, and pelvis angles, θ_F , θ_S , θ_T , θ_P respectively) is defined as a low-dimensional synergy that work together in order to assist the controller by stabilizing (or destabilize) the performance variable. We consider the kinematic leg effector vector (spanning the kinematic segment chain) is coordinated

and controlled by a two-level motor control hierarchy (Diedrichsen, Shadmehr, & Ivry, 2010; Scholz & Schöner, 1999).

The UCM analysis was computed at each time slice of the landing phase period. Each time slice corresponded to a time period of 4ms. The UCM parameters were calculated using a customised Matlab program that was based on the conventional UCM method (Scholz & Schöner, 1999). The UCM analysis method can be described in four general steps below:

7.3.2.1 Step 1: Define the Geometric Model

The kinematic leg effector is defined by a geometric function that maps the segment angles with the 2D effector end-point:

$$(L_Y, L_Z) = f(\theta_F, \theta_S, \theta_T, \theta_P) \quad (1)$$

The leg effector was defined by a vector spanning between a fixed point at the pelvis segment centre of mass and the location of the centre of pressure beneath the foot. From equation 1 the 2D position of the leg effector is defined by a specific geometric model that directly maps the end-effector in the same space as the elemental variables. The performance variables of L_Y and L_Z are associated geometrically with the elemental variable details, segment angles and segment lengths:

$$\begin{cases} L_Y = l_{ft} \cdot \cos(\theta_F) + l_{sh} \cdot \cos \theta_S + l_{th} \cdot \cos \theta_T + l_{pv} \cdot \cos \theta_P \\ L_Z = l_{ft} \cdot \sin(\theta_F) + l_{sh} \cdot \sin \theta_S + l_{th} \cdot \sin \theta_T + l_{pv} \cdot \sin \theta_P \end{cases} \quad (2)$$

where l_{ft} , l_{sh} , l_{th} , and l_{pv} are the lengths of the foot, shank, thigh, and pelvis respectively; while θ_F , θ_S , θ_T , and θ_P are the segment angles (with respect to the horizontal axes).

7.3.2.2 Step 2: Linear Approximation of the UCM

A deviation matrix (DV) from the mean joint configuration at each i th time instant was computed for each j th stride:

$$DV(i, j) = \begin{bmatrix} \theta_F(i, j) - \bar{\theta}_F(i) \\ \theta_S(i, j) - \bar{\theta}_S(i) \\ \theta_T(i, j) - \bar{\theta}_T(i) \\ \theta_P(i, j) - \bar{\theta}_P(i) \end{bmatrix} \quad (3)$$

The Jacobian matrix (J) relating partial changes in elemental variables (i.e. $\theta_F, \theta_S, \theta_T, \theta_P$) to partial changes in the performance variables (i.e. L_Y, L_Z), was computed around the mean joint configuration (i.e. $\bar{\theta}_F, \bar{\theta}_S, \bar{\theta}_T, \bar{\theta}_P$) across the set of strides (trials) for each time slice of the period ($i=21$).

7.3.2.3 Step 3: Projecting the joint configuration onto the UCM

The next step was to compute the null space of the Jacobian matrix ($N(J)$). The null space is the linear subspace of all the segment angle combinations that result in no change to the end-effector position. Linearization of the UCM is necessary in order to compute variance (linear concept) from a nonlinear geometric model (the UCM) (Latash, Scholz, & Schoner, 2007). The null space spanned by the basis vectors ε_{n-d} has a dimension equal to the difference between the number of elemental variables ($n=4$) and the number of performance variables ($d=2$).

$$N(J) = \begin{bmatrix} \varepsilon_{11} & \varepsilon_{12} \\ \varepsilon_{21} & \varepsilon_{22} \\ \varepsilon_{31} & \varepsilon_{32} \\ \varepsilon_{41} & \varepsilon_{42} \end{bmatrix} \quad (4)$$

The deviation matrix then decomposed into components parallel (DV_{\parallel}) and perpendicular (DV_{\perp}) to the null space:

$$DV_{\parallel}(i, j) = \sum_{k=1}^{n-d} (N(J)_k^T \cdot DV(i, j)) \cdot N(J)_k \quad (5)$$

$$DV_{\perp}(i, j) = DV(i, j) - DV_{\parallel}(i, j) \quad (6)$$

7.3.2.4 Step 4: Computing the variance of V_{UCM} and V_{ORTH}

The variance of these projections were estimated and normalized per degree of freedom of each subspace as:

$$\sigma_{\parallel}^2(i) = \frac{\sum_{j=1}^N DV_{\parallel}^2(i, j)}{(n - d)N} \quad (7)$$

$$\sigma_{\perp}^2(i) = \frac{\sum_{j=1}^N DV_{\perp}^2(i, j)}{dN} \quad (8)$$

Variance of goal-irrelevant deviations are parallel to the UCM are indicated as V_{UCM} ($\sigma_{\parallel}^2(i)$), and goal-relevant deviations are orthogonal to the UCM are indicated as V_{ORTH} ($\sigma_{\perp}^2(i)$). The variances were computed at each time instance and compared across conditions. The ratio of variability V_{RATIO} was the third UCM parameter and was computed in a form suggested by Papi, Rowe, and Pomeroy (2015):

$$Ratio = \left(\frac{2\sigma_{\parallel}^2}{\sigma_{\parallel}^2 + \sigma_{\perp}^2} \right) - 1 \quad (9)$$

this formulation expresses the ratio in a range from -1 and +1 with 0 as midpoint avoiding the symmetrical and statistical problem related to the original formulation $\sigma_{\parallel}^2/\sigma_{\perp}^2$ (Scholz & Schöner, 1999). This ratio reflects the need for intervention in order to control the performance variable during landing. If the ratio is greater than 0, the effector system has a coordination strategy that produces a stable goal variable and is indicative of motor redundancy. On the contrary, ratios less than 0 will indicate that variations in coordination will have a larger effect on the performance variable.

Although the UCM analysis does not require temporal order of trials, it does presume that between performance trials the effector of elemental variables will address the same task-goal from the same initial conditions. In this case the task-goal was a kinematic orientation of the leg segment angles in the sagittal plane and the initial condition was the effective leg length and foot orientation at ground contact. The task goal is presumed to be inherent to the locomotor control system, which is the change the leg length and orientation. This kinematic variable is under the fine control of leg stiffness adjustments from joint torques, which ultimate enable the appropriate

load applied to the muscle-tendon units to meet the determinant goals of running. Therefore, we restricted the number of trials for UCM analysis to meet the criteria of consistent initial condition (foot angle at ground contact) and consistent response to meet the performance goal (change in leg length). From the set of 375 stride cycles, we twice sub-divided into groups rank-ordered according to the above criteria. First, we rank-ordered the cycles by magnitude of sagittal plane foot angle at the ground contact event (FC) and then sub-divided this ranked set into equal tercile groups: low-tercile (below-average initial conditions), mid-tercile (average initial conditions), and upper-tercile (above average initial conditions). Within each group, the cycles were rank-ordered according to the second criterion: change in leg length from FC-10 (i.e. 40milliseconds prior to FC event) to FC, where the groups were again partitioned into tercile subgroups: small-change (low-tercile), average-change (mid-tercile), and large-change (upper-tercile). After removing 5% of extreme cases in each sub-group, this process of rank-ordering trials created nine subgroups of 40 cycles. We performed UCM analysis on the middle set of data, which was the average-change in leg length and average foot angle at impact. This was expected to be most representative of the task goal and consistent conditions for each subject.

7.4 Results

V_{RATIO} group mean and standard error is plotted across the time course of the landing phase – which includes pre-landing (PRE) and landing (POST) phases – for the performance task of vertical leg length and fore-aft orientation in all footwear conditions (**Figure 7-3**). In all footwear conditions the ratio for the vertical (Z) component increases almost linearly during the PRE phase, and from foot contact (FC) it then plateaus during POST phase. For the Y component, V_{RATIO} maintains a relatively uniform constant value throughout the entire landing phase. V_{RATIO} values for both the Z and Y performance variables were statistically different from zero for most of the landing phase for all shoe conditions.

FFS displayed higher V_{RATIO} values in low MI shoes at the beginning of the PRE phase (**Figure 7-3**), and in high MI shoes at the end of the POST phase on the Y component. On the Z component, FFS had an earlier peak in med MI shoes than RFS.

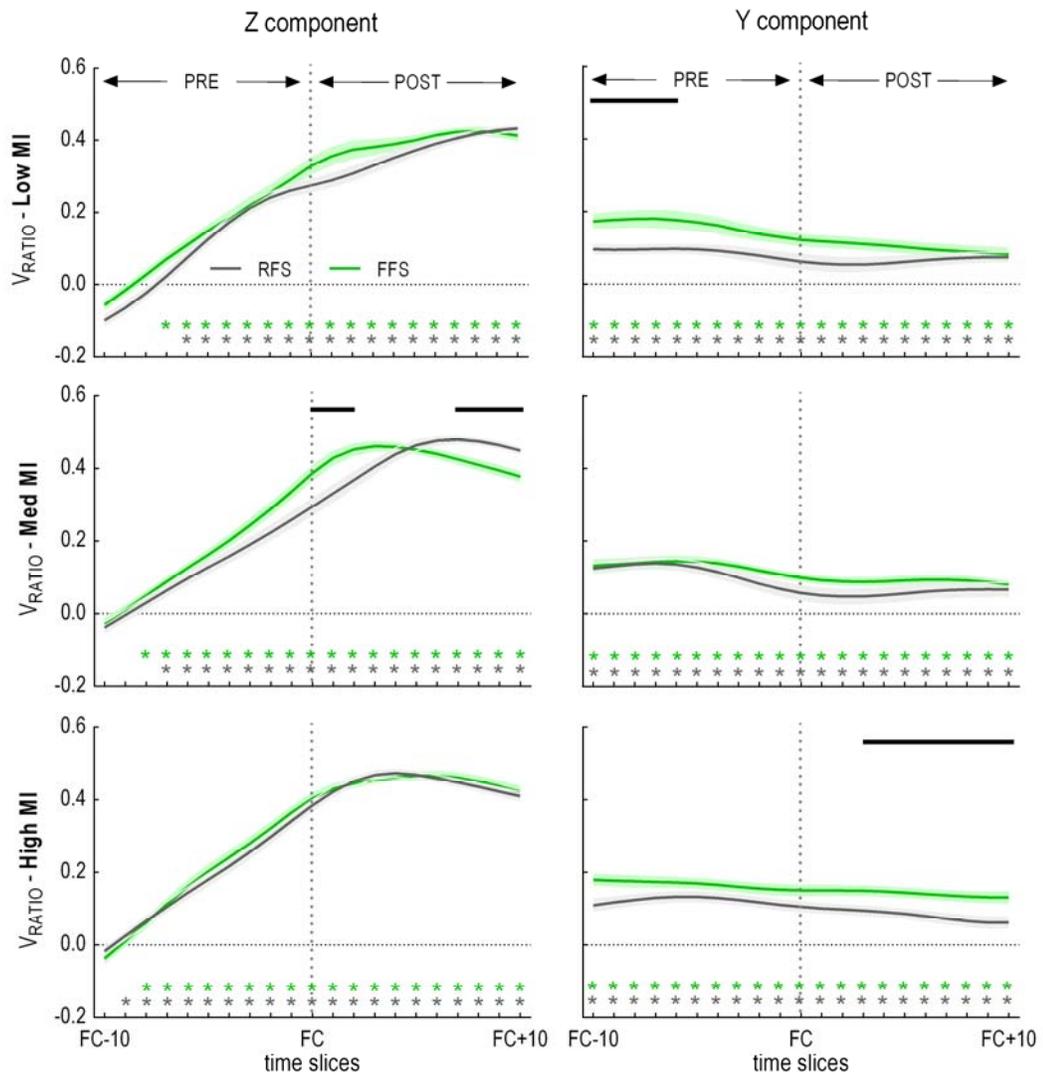


Figure 7-3 Mean±SE ratio values for RFS and FFS groups. Time has been divided in two phases: PRE from 10 frames before foot contact (FC-10) to foot contact (FC); and POST from FC to 10 frames after foot contact (FC+10). Solid lines indicate a statistically significant difference between groups ($p < .05$). * indicates statistically significant difference from zero ($V_{UCM} > V_{ORTH}$). Note: frames correspond to absolute time (mmsec); 1 frame = 4mmsec. FC+10 is ~ 15% of stance.

Figure 7-4 shows the time course of V_{UCM} and V_{ORTH} along the landing phase. RFS and FFS groups show similar behaviours, although FFS tend to have higher values for V_{UCM} in low MI shoes in both Z and Y components. Overall, V_{ORTH} decreases in the PRE phase and remains constant after FC showing a statistical difference between PRE and POST phase for the both Z and Y components (**Table 7-1**). For both groups, the elbow in the curve happens earlier in low MI shoes than in med or high MI shoes.

Table 7-1 Primary statistical results for differences between Groups, Shoes, and Phase for variance parallel to the UCM (V_{UCM}), variance orthogonal (V_{ORTH}), and ratio (V_{RATIO}) for the vertical component (Z) and horizontal component (Y). ANOVA results are given for main effects and interactions. Statistically significant findings are in bold.

Variable	Group	Shoe	Phase	Group x Shoe	Group x Phase	Shoe x Phase	Group x Shoe x Phase
Z comp							
V_{UCM}	$F_{(1,18)} = 0.29$ $p = .598$	$F_{(2,36)} = 0.21$ $p = .707$	$F_{(1,18)} = 6.57$ $p = .020$	$F_{(2,36)} = 0.66$ $p = .460$	$F_{(1,18)} = 0.29$ $p = .599$	$F_{(2,36)} = 2.33$ $p = .112$	$F_{(2,36)} = 1.32$ $p = .279$
V_{ORTH}	$F_{(1,18)} = 0.10$ $p = .753$	$F_{(2,36)} = 6.57$ $p = .004$	$F_{(1,18)} = 179.96$ $p < .001$	$F_{(2,36)} = 1.25$ $p = .299$	$F_{(1,18)} = 0.05$ $p = .824$	$F_{(2,36)} = 1.35$ $p = .272$	$F_{(2,36)} = 1.25$ $p = .300$
V_{RATIO}	$F_{(1,18)} = 0.36$ $p = .555$	$F_{(2,36)} = 6.63$ $p = .004$	$F_{(1,18)} = 355.46$ $p < .001$	$F_{(2,36)} = 0.23$ $p = .796$	$F_{(1,18)} = 0.34$ $p = .569$	$F_{(2,36)} = 0.96$ $p = .394$	$F_{(2,36)} = 0.63$ $p = .542$
Y comp							
V_{UCM}	$F_{(1,18)} = 0.34$ $p = .568$	$F_{(2,36)} = 0.47$ $p = .538$	$F_{(1,18)} = 23.96$ $p < .001$	$F_{(2,36)} = 0.90$ $p = .415$	$F_{(1,18)} = 0.21$ $p = .654$	$F_{(2,36)} = 1.88$ $p = .167$	$F_{(2,36)} = 1.15$ $p = .330$
V_{ORTH}	$F_{(1,18)} = 0.10$ $p = .921$	$F_{(2,36)} = 0.53$ $p = .596$	$F_{(1,18)} = 11.17$ $p = .004$	$F_{(2,36)} = 0.22$ $p = .806$	$F_{(1,18)} = 0.36$ $p = .554$	$F_{(2,36)} = 1.07$ $p = .353$	$F_{(2,36)} = 0.42$ $p = .659$
V_{RATIO}	$F_{(1,18)} = 2.95$ $p = .103$	$F_{(2,36)} = 0.62$ $p = .543$	$F_{(1,18)} = 6.77$ $p = .018$	$F_{(2,36)} = 0.18$ $p = .840$	$F_{(1,18)} = 0.01$ $p = .936$	$F_{(2,36)} = 0.39$ $p = .678$	$F_{(2,36)} = 1.23$ $p = .305$

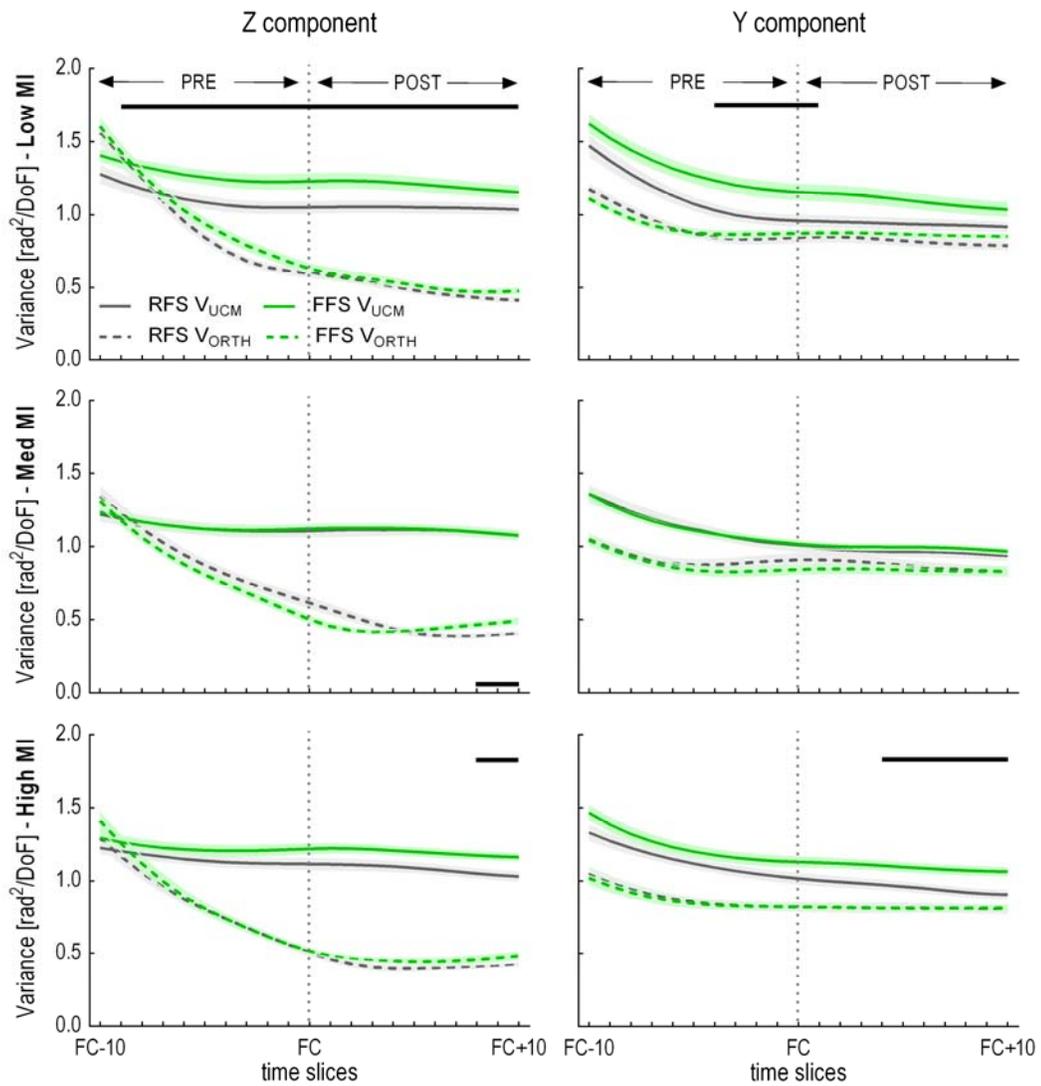


Figure 7-4 Mean \pm SE of Variance components parallel (solid lines) and orthogonal (dashed lines) to the linearized UCM. Note: frames correspond to absolute time (mmsec); 1 frame = 4mmsec. FC+10 is \sim 15% of stance. Solid lines indicate a statistically significant difference between groups ($p < .05$).

Table 7-2 Mean \pm standard deviation for variance parallel (V_{UCM}), orthogonal (V_{ORTH}), and ratio (V_{RATIO}) across the three footwear conditions for the vertical (Z) component and horizontal (Y) component.

Z component	PRE		POST	
	RFS	FFS	RFS	FFS
Low MI				
V_{UCM}	1.12 \pm 0.31	1.28 \pm 0.30	1.03 \pm 0.30	1.09 \pm 0.44
V_{ORTH}	0.98 \pm 0.25	1.05 \pm 0.26	0.49 \pm 0.14	0.52 \pm 0.13
V_{RATIO}	0.09 \pm 0.10	0.12 \pm 0.11	0.37 \pm 0.08	0.40 \pm 0.08
Med MI				
V_{UCM}	1.14 \pm 0.29	1.15 \pm 0.19	1.08 \pm 0.30	1.05 \pm 0.28
V_{ORTH}	0.95 \pm 0.26	0.88 \pm 0.15	0.44 \pm 0.12	0.45 \pm 0.10
V_{RATIO}	0.11 \pm 0.11	0.15 \pm 0.08	0.44 \pm 0.09	0.43 \pm 0.06
High MI				
V_{UCM}	1.16 \pm 0.23	1.23 \pm 0.26	1.05 \pm 0.30	1.12 \pm 0.30
V_{ORTH}	0.87 \pm 0.24	0.91 \pm 0.24	0.41 \pm 0.10	0.46 \pm 0.12
V_{RATIO}	0.16 \pm 0.10	0.17 \pm 0.10	0.45 \pm 0.06	0.45 \pm 0.07

Y component	PRE		POST	
	RFS	FFS	RFS	FFS
Low MI				
V_{UCM}	1.15 \pm 0.33	1.33 \pm 0.36	0.90 \pm 0.33	0.97 \pm 0.48
V_{ORTH}	0.94 \pm 0.17	0.93 \pm 0.16	0.82 \pm 0.21	0.87 \pm 0.10
V_{RATIO}	0.09 \pm 0.09	0.16 \pm 0.14	0.06 \pm 0.11	0.10 \pm 0.11
Med MI				
V_{UCM}	1.16 \pm 0.29	1.15 \pm 0.17	0.93 \pm 0.30	0.90 \pm 0.32
V_{ORTH}	0.92 \pm 0.24	0.89 \pm 0.21	0.87 \pm 0.25	0.84 \pm 0.21
V_{RATIO}	0.12 \pm 0.10	0.13 \pm 0.09	0.06 \pm 0.14	0.09 \pm 0.07
High MI				
V_{UCM}	1.15 \pm 0.25	1.25 \pm 0.27	0.91 \pm 0.33	0.99 \pm 0.36
V_{ORTH}	0.90 \pm 0.20	0.88 \pm 0.20	0.81 \pm 0.20	0.82 \pm 0.12
V_{RATIO}	0.12 \pm 0.09	0.17 \pm 0.09	0.08 \pm 0.09	0.14 \pm 0.09

7.4.1 Variance parallel to the UCM, V_{UCM}

There was a main effect of Phase ($p = .020$; $p < .001$, **Table 7-1**) for the vertical (Z) and horizontal (Y) performance variable, respectively. Indicating that on average V_{UCM} was dependent on the phase of the landing task. Post-hoc analysis reveal that V_{UCM} pre landing is higher compared to post landing for both performance variables (**Figure 7-3, Table 7-2**).

7.4.2 Variance orthogonal to the UCM, V_{ORTH}

Similar results were found when testing the differences in variance orthogonal to the UCM (**Table 7-1**). There was a main effect of Phase ($p < .001$; $p = .004$) for the vertical (Z) and horizontal (Y) performance variable, respectively. Shoe had a significant main effect on V_{ORTH} ($p = .004$) for the vertical component only. Post-hoc tests reveal that V_{ORTH} is higher in low MI shoes compared to med MI and high MI ($p = .021$; $p = .028$, respectively). Indicating that more supportive shoes may induce V_{ORTH} to increase (**Figure 7-4, Table 7-2**).

7.4.3 Ratio of variances perpendicular and orthogonal to the UCM, V_{RATIO}

There was a main effect of Phase ($p < .001$, $p = .018$, **Table 7-1**) for the vertical (Z) and horizontal (Y) performance variable, respectively. Shoe had a significant main effect on V_{RATIO} ($p = .004$) for the vertical component only. Post-hoc tests reveal that V_{RATIO} is higher ($p = .013$) in high MI shoes compared to low MI shoes. Indicating that more supportive shoes may induce V_{RATIO} to decrease (**Figure 7-3, Table 7-2**).

7.5 Discussion

The purpose of this study was to determine if redundant segment angles of the leg effector are stabilized to control leg length and orientation during impact phase of running, and to examine whether this stability is affected by foot loading type of the runner or by footwear type. Specifically, we examined how the elemental variables of limb segment angles co-vary in a common space with the goal task of the performance variable leg length and orientation. This is the first study known to us to examine these features of running.

7.5.1 Redundancy is exploited for leg length and orientation stabilisation

The variance was structured according to our expectations. First, the V_{RATIO} was significantly greater than zero (**Figure 7-3, Table 7-1**), indicating that a kinematic synergy stabilises the performance variables across the landing phase. Second, V_{ORTH} was rapidly reduced as the impact phase approached, demonstrating that control over limb length and orientation is relevant to the goal of the locomotor control system. The relatively high V_{ORTH} values found during early period of pre-landing confirm the minimal intervention principle (Todorov & Jordan, 2002). In contrast, V_{UCM} remained relatively constant throughout the landing period for both performance variables leg length and leg orientation ($V_{UCM} > V_{ORTH}$). The hypothesis that there would be a more stable leg length relative to leg orientation was also confirmed ($V_{RATIO-Z} > V_{RATIO-Y}$). The hypothesis that variance is structured to provide increased stability of leg length and orientation as the impact phase approached was also confirmed ($V_{RATIO-POST} > V_{RATIO-PRE}$). The idea of a strong synergy being responsible for stability of the kinematic leg effector during late swing and early stance of running has also been demonstrated by (Blum, Lipfert, Rummel, & Seyfarth, 2010; Blum et al., 2014; Daley & Usherwood, 2010; Ivanenko et al., 2007).

Our results support the idea that stabilisation of performance variables (leg length and orientation) are indicators of a hierarchical control system and subjected to higher-level cost policy that is task-relevant. The observation that the redundant elements (segment angles) of the two performance variables were managed differently reveals the priority of the controller. Both performance variables demonstrated relatively similar V_{UCM} , however, they differed in the way the V_{ORTH} was reduced (**Figure 7-3, Table 7-2**). The leg length (Z component) was stabilized rapidly (V_{RATIO}) prior to foot

contact due to the rapid decrease in V_{ORTH} . This rapid reduction in V_{ORTH} is unlikely to come from low-level control processes, but rather it is most plausible that there is intervention from higher-level central nervous system control. The tools to appropriately reconcile the responsible source of this rapid change in V_{ORTH} would require extended analysis that combines surrogate data sets where the segment angle correlations are randomised to reduce their non-trivial covariance structure and then perform the UCM method (Scholz & Schöner, 2014). Nevertheless, we can reasonably conclude that a consistent leg length is a goal for the landing task of running. In contrast, the leg orientation (Y component) was relatively less stable (lower V_{RATIO}) during the same period, due to a modest reduction in V_{ORTH} . However, the V_{RATIO} of leg orientation is significantly greater than 0 indicating there is a significant non-trivial structuring of the covariance. Control of the redundant combinations of leg segment angles that lead to consistent leg orientation (i.e. fore-aft dimension) suggests attention afforded by the high-level controller to stabilise this task. Leg posture at landing determines stance goals, such as stability (Seyfarth, Geyer, & Herr, 2003), and leg loading (Vejdani, Blum, Daley, & Hurst, 2013) (see also Chapter 6); while both leg length and orientation are important, the system cannot simultaneously optimize both, resulting in a trade-off between two simultaneous performance goals: keeping a stable body trajectory while minimizing leg loading (Karszen, Haberland, Wisse, & Kim, 2011). Reducing the GRD in leg length pre landing allows control of impact forces (Cusumano & Cesari, 2006) (see also Chapter 4) and disturbance rejection (Blum et al., 2014). By minimally intervening on leg orientation the system allows this performance variable to be more flexible. Destabilisation before a quick change of state ensures adaptability to external perturbations by attenuating synergies that would otherwise interfere with the change in the performance variable (Klous, Mikulic, & Latash, 2011). Therefore, while stabilisation of the leg length is needed for energy and steady state locomotion, destabilisation of the leg orientation is also important for injury avoidance.

7.5.2 *Effect of foot strike on GID and GRD*

The hypothesis that habitual rearfoot loading technique will have restricted variance, evident by a reduction in the variance in both goal-relevant and goal-irrelevant deviations was partially supported under certain conditions (**Figure 7-4, Table 7-2**). There was however, no evidence to support a contrary hypothesis. The effect of group on the restriction of V_{UCM} occurred for both leg length and orientation, and was mostly evident within the low MI shoe (high load assistance), but also apparent within the high MI shoe (low load assistance) (**Figure 7-3, Table 7-2**). The restriction of V_{ORTH} by the rearfoot loading technique was not conclusive. The restriction of V_{UCM} by the rearfoot loading technique group represents a reduction in redundant motor solutions that can equally produce an equivalent leg length (i.e. equifinality). In contrast, a forefoot loading technique is associated with a more adaptable repertoire of covariant segment angle combinations that are available to their system, which lead to consistent performance of establishing a desired leg length during landing. This expansion of goal irrelevant deviations (GID) is related to the concept of an abundant repertoire of flexible solutions available for passive low-level controller (allometric controller). Having such a diverse system could be important if there is a large dependence on passive control at the beginning of the landing phase (Krogt et al., 2009; Moritz & Farley, 2005). The RFS group have likely developed a technique that has restricted their available degrees of freedom and the potential to find flexible motor solutions. As the rearfoot strike pattern is likely caused by the cushioning provided by the footwear (Gruber, Silvernail, Brueggemann, Rohr, & Hamill, 2013), the implication of our findings may be extended to the risk of providing new generation of runners with heel raised running shoes.

Many studies report on the habitual forefoot strike pattern in barefoot populations (Hatala et al., 2013; Larson, 2014; Pontzer et al., 2014) and the benefit that may be associated with it; for instance increased foot muscle size (Cheung, Sze, Chen, & Davis, 2016), higher foot arch (Miller, Whitcome, Lieberman, Norton, & Dyer, 2014), and increased sensory feedback (Shinohara & Gribble, 2013). Those adaptations are more likely what RFS are not developing. By supporting the foot and make it “comfortable”, shoes may have been desensitizing the system from its elements. The necessity for the system to be flexible may have been thus compromised. Although injuries among forefoot and rearfoot runners are equal in

numbers but different in location (Warr et al., 2015), no studies considered a reduction in adaptability as a possible chronic injury related with running. We are the first in approaching changes in adaptability from a dynamic system theory perspective; we did so by analysing the two main mechanical variables that can possibly influence neuromuscular adaptability: foot strike, and footwear. This study may be lay the foundations for a change in perspective on how to approach running-related injuries.

7.5.3 *Effect of shoes on GID and GRD*

The hypothesis that increased footwear assistance would reduce the amount of goal-relevant deviations more than goal-irrelevant deviations in the group of long distance runners with rearfoot loading technique was not supported. For the task of stabilising leg length and leg orientation, the runners with a rearfoot loading technique demonstrated similar profile to the forefoot loading runners.

Because a main effect of footwear has not been found for all deviation components, it can be assume that shoes may have a minimal effect on the organisation of redundant degrees of freedom. This can be explained by the ability of the body's system to adapt quickly to external constraints (Ferris, Liang, & Farley, 1999; Marigold & Patla, 2005). Rearfoot strikers have changed their foot strike landing with a more plantarflexed ankle in less supportive shoes, it is therefore difficult to discern the effect of foot strike pattern from shoe. Forcing runners to adopt a specific foot strike can give access to the effect of shoes. However, by giving runners the task goal of keeping a certain leg orientation, the stability of that variable will increase by a reduction in performance variance (Dingwell, John, & Cusumano, 2010). Further investigation will be needed to clearly define if shoe support has an effect on the organisation of variability along leg segments during landing.

The UCM approach (and DFA analysis Chapter 6) gave another insight into the effect of rearfoot loading technique on the adaptability of the locomotor control system. With the UCM, we explore explicitly the state of the low-level control system and observe the effect of regular long distance running. If the level of locomotor abundance is being restricted by footwear influences, then studies like this are important for footwear companies to identify new design solutions that minimise this effect: can a shoe be designed that can assist with sustaining locomotor abundance that allows runners lifelong benefits towards healthy mobility. The development of

barefoot-like shoes was born on an ideological effort to bring back evolutionary principles from which we are separating through the use of assistive technology. Although barefoot running is anachronistic for many earth inhabitants, a shoe designed to enhance not only physical performance but it acts to augment neuromuscular adaptability and health. Such a pursuit of technology can assist the reduction of injuries and ageing populations; and be another reason to claim that running is an essential exercise for lifelong health.

A limitation of this study is related to the biomechanical model, as we assumed the CoM of the pelvis to be a suitable surrogate of the body CoM. The exclusion of the trunk and arm from the model may affect the position of the body centre of mass during landing. In addition, we did not test the sensitivity of the system to individual joints. While we assumed that elemental variables equally influenced the performance variable, it may be that the system becomes more sensitive to certain elemental variables in unfamiliar conditions (Yen & Chang, 2010). Further, we did not verify whether the structure of variance is a true representation of covariation, this can be obtained performing correlation between surrogate datasets and actual data (Scholz & Schöner, 2014). Lastly, we inferred adaptability but we did not measure it directly. A true measure of adaptability is when a system is able to effect a desired outcome from an undesirable starting point, or initial state. In future, we aim to investigate running trials that have a different initial condition of leg length at impact, but they produce an equivalent response in limb length change. The question is whether there is a group effect on the achievement of a desired peak force.

7.6 Conclusion

Leg length and orientation is an important parameter to control during landing period. The results from our study suggest that habitual RFS may have restricted their level of adaptability (lower V_{UCM}) compared to FFS, however, FFS and RFS are equally able to reduce task-relevant variance in order to stabilize the performance variables leg length and leg orientation. Runners strongly stabilize the leg length (vertical component), and during this process they have less stability of leg orientation (horizontal component); which, is useful if they need to change this performance parameter more rapidly in response to external perturbations. Further studies however are needed to consolidate these findings. The indirect implications of our results expand from the results on leg stiffness control (Chapter 6), here we also found that running with a rearfoot strike pattern leads to partial neural control degeneracy, hence decrease entropy. Although speculative, there may be more advantage to pursue a forefoot strike pattern for long-distance runners that goes beyond tangible, anatomical changes. While running is often quantified by how the movement is performed (see Chapter 5), in this study we offer a new insight into how the movement is controlled and its variance organised.

7.7 References

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8 REPEATABILITY AND ACCURACY OF A FOOT MUSCLE STRENGTH DYNAMOMETER

This chapter is an amended version of the manuscript: Garofolini, A., Taylor, S., McLaughlin, P., Stokes, R., Kusel, M. & Mickle, KJ. Repeatability and accuracy of a foot muscle strength dynamometer. *Journal of Medical Engineering and Physics*, 2019. Published version in appendix B.

8.1 Abstract

Toe flexor strength is a pivotal biomechanical contributor for effecting balance and gait. However, there are limited reports that evaluate measurement accuracy and repeatability of this important attribute. Dynamometers are designed to measure force which can be used to derive joint torque if the perpendicular distance to the joint axis is known. However, an accurate and reliable measurement method to assess the ability of the toe flexor muscles to produce torque, is lacking. Here we describe a new device and method, designed to quantify the toe flexor torque developed at the metatarsal phalangeal joint. We evaluate measurement bias and the ability of the instrument to consistently measure what it is supposed to measure (Interclass Correlation Coefficient). Results suggest that our device is an accurate tool for measuring angle and torque with a small (0.10° and 0.07 Nm, respectively) bias. When tested for reliability and repeatability in measuring toe flexor torque ($n = 10$), our device showed high interclass correlation ($ICC=0.99$), small bias (-1.13 Nm) and small repeatability coefficient ($CR = 3.9$). We suggest mean bias and CR to be reported for future measurement methods and our protocol used as standard approach to measure maximal toe flexor torque.

8.2 Introduction

Adequate foot muscle strength is imperative for efficient performance of sport and activities of daily living (Landers, Hunter, Wetzstein, Bamman, & Weinsier, 2001). When we stand, foot muscles provide the basis for upright balance, but during locomotion the foot has a dual function: it forms a rigid lever at foot-strike and push-off, and a shock-absorber during mid-support (McKeon, Hertel, Bramble, & Davis, 2014). This is accomplished through the deformation of the arch, which is controlled and supported by small intrinsic (foot) and large extrinsic (leg) muscles. Although critical to locomotion, our ability to measure and evaluate foot muscle strength accurately is rather limited (Miller, Whitcome, Lieberman, Norton, & Dyer, 2014; Soysa, Hiller, Refshauge, & Burns, 2012).

Dynamometers are suggested to directly measure muscle force. They all rely on the assumption that (i) the external moment of force measured around the device axis represents the moment of the force produced by the muscles, and (ii) the force that produces such moment is equal to the muscle force. For semantical precision, hereon we will refer to torque – external moment of force – when referring to what a dynamometer is measuring.

Previous toe dynamometers described in the literature have had technical limitations: some rely on the tester providing resistance (Spink, Fotoohabadi, & Menz, 2010), while others allow gripping of the toes and, therefore have a greater contribution from the extrinsic toe flexors (Uritani, Fukumoto, & Matsumoto, 2012). An alternative is a fixed dynamometer whereby participants press their toes against a fixed sensor plate (i.e. force sensors) (Mickle, Munro, Lord, Menz, & Steele, 2009; Senda et al., 1999). In this way, Endo, Ashton-Miller, and Alexander (2002) used the signal from a force plate to quantify toe flexor torque around the metatarsophalangeal joint (MPJ); however, the movement was not isolated: the contribution of the moment generated among the other (bigger) joints was not accounted for. Goldmann and Brüggemann (2012) introduced a system of Velcro® straps to fix the forefoot, midfoot, and rearfoot to the dynamometer while keeping the body into a standardized position. Although giving repeatable measurements, their device was not tested for accuracy and reliability. Based on the device built by Goldmann and Brüggemann (2012), we developed a custom-made toe dynamometer addressing the technical limitations of previous studies while ensuring accurate measurements of torque produced by toe

flexor muscles. The purpose of the present study was: 1) to assess the accuracy between the known measures for angle and torque measured by the novel dynamometer device; and 2) to assess the device re-test repeatability of maximal isometric contractions of toe flexor muscles.

8.3 Methods

In this study, we quantified the moment of force generated by toe flexor muscles around the axis of the dynamometer during maximal isometric contraction. Our design addressed two important issues when assessing toe muscle strength: angular orientation of the metatarsal heads and foot size.

8.3.1 Hardware and software

The device is an improved version of a previously proposed machine (Goldmann & Brüggemann, 2012) to which we added flexibility, and adaptability. It has been designed to allow measurements to be taken in either a seated or standing position. For operation in the seated position, a knee-thigh clamping mechanism is included, with both vertical and longitudinal adjustment features (**Figure 8-1a**). The device can be set in a locked angular position to monitor a subject's ability to apply static torque, or can be set to allow free angular range of motion with adjustable mechanical limits. The height of the transverse axis of the MPJ is a function of foot size; therefore, we secured the plate on three adjustable screws with fixed rulers such that the plate position can be recorded and readjusted according to the participant's foot size. The angular orientation of the metatarsal heads also needed to be taken into consideration (Raychoudhury, Hu, & Ren, 2014; Smith, Lake, Lees, & Worsfold, 2012). We designed a plate with a matrix of holes to which locking pins and straps can be tethered for strapping the subject's foot into different orientations. A requirement to provide the capacity to impose and resist up to 50 Nm of torque has been met with the use of dumbbell weights loaded on to a carrier (**Figure 8-1b**), and a pulley arrangement (**Figure 8-1c**).

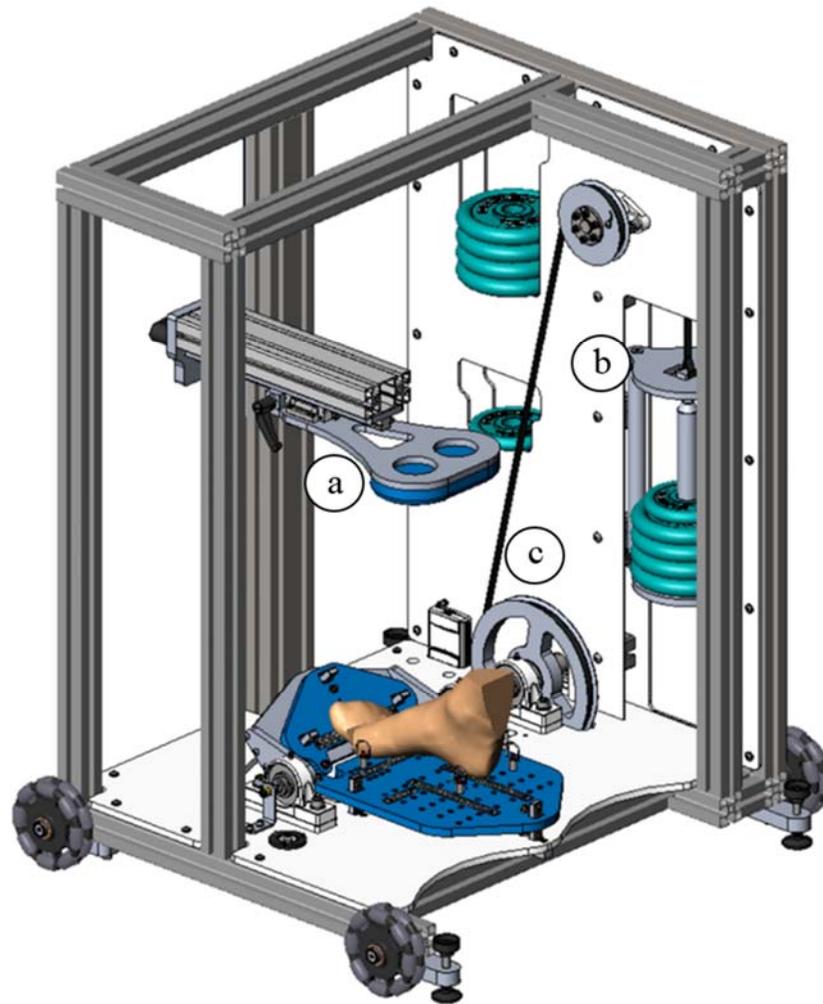


Figure 8-1 Overview of the toe flexors strength device: **a** knee-thigh clamping mechanism, **b** carrier, and **c** pulley arrangement

The tension $[t_p]$ in the primary strap is the weight of the mass load. The tension in the secondary strap $[t_s]$ is equivalent to the tension in the primary strap multiplied by the ratio of the primary $[r_p]$ and secondary $[r_s]$ pulley radii. The torque $[T]$ imposed on the phalanges shaft is the product of the secondary strap tension and the driven pulley radius $[r_d]$. The effective radius of each pulley is the sum of the radius of the pulley surface and half the thickness of the tension strap. The primary pulley effective radius was 0.100 m, the secondary pulley radius was 0.049 m, and the driven pulley radius r_d was 0.100m; therefore:

$$T[Nm] = m[Kg] * g * (rp/rs) * rd \quad (8-1)$$

$$T = m * 9.81 * (.100/.049) * .100$$

$$T = m * 2.002$$

The phalanges rotation shaft carries an absolute angle rotary encoder (**Figure 8-2a**) on its end, which produces an analogue output voltage signal. The shaft assembly also includes a torsion strain cylinder element (**Figure 8-2b**), which is connected to the assembly in such a way as to ensure that the link transmits torque without being exposed to any bending, tensile or compressive loads. The main foot and phalanges resting surface plates are designed and built to provide a large range of height adjustment so that any subject's proximal phalanges centre of rotation can be aligned with the device's rotation shaft. This allows simulation of a tilted MPJ mediolateral axis of rotation, through adjustment of jacking screws accordingly on both the main foot and phalanges tooling plates. The tarsal resting surface plate includes a matrix of holes to which locking pins and straps can be tethered for strapping the subject's foot into position. Both the main foot and phalanges resting plates include millimetre linear scales for foot positioning reference (**Figure 8-2c**).

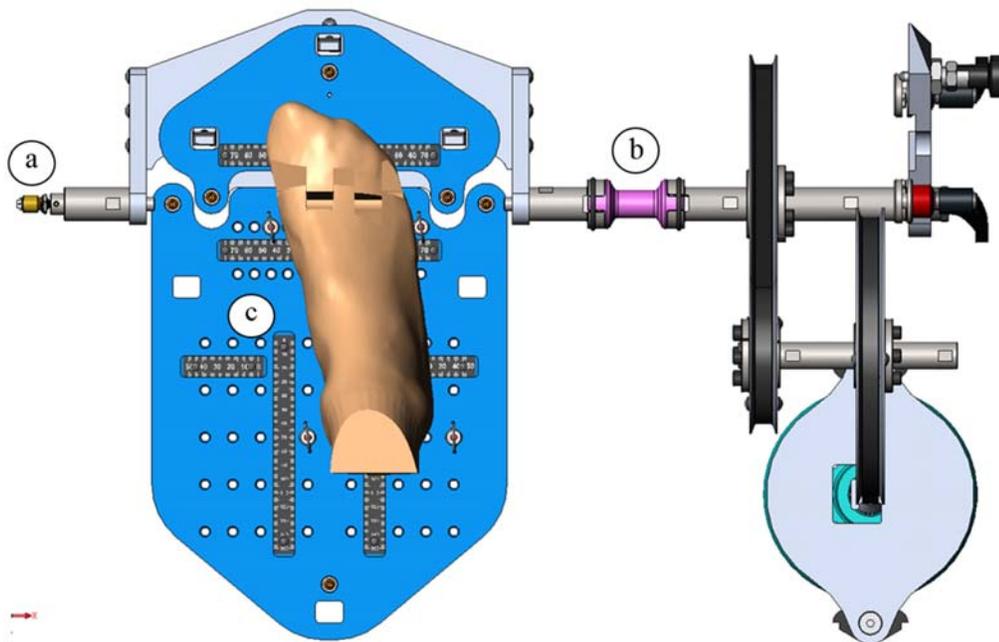


Figure 8-2 Schematic of the main foot and phalanges plates. **a** rotary encoder, **b** torsion strain cylinder, and **c** millimetre linear scales

The electronic instrumentation comprises two transducers, their associated signal conditioning circuitry, and a custom Labview data acquisition system running on a laptop PC and employing an NI-6009 14-bit USB DAQ module to sample the 2 analogue quantities. An absolute angle encoder (US Digital MA3 with analogue output) is directly coupled to the shaft end of the toe plate and thus directly monitors the -20 to +50 degrees' angular range of the toe plate. This transducer has a resolution of 10 bits which equates to 0.33 degrees measurement resolution.

A torque transducer and its associated amplifier monitors the torque applied by the toes to the toe plate. It covers a torque range of 0-50 Nm. The transducer was constructed in-house using a Micro-Measurements CEA-06-250US-350 full bridge strain gauge bonded to a custom designed hollow shaft and rated for 50 Nm full load. The associated strain gauge amplifier has a gain of 500 to provide an output voltage of approx. 4V at 50 Nm. Custom Labview code (National Instruments) samples the above 2 analogue channels at 100 Hz and applies the appropriate scaling factors and offsets to produce actual torque and angle values which are displayed in real-time (**Figure 8-3a,b**).

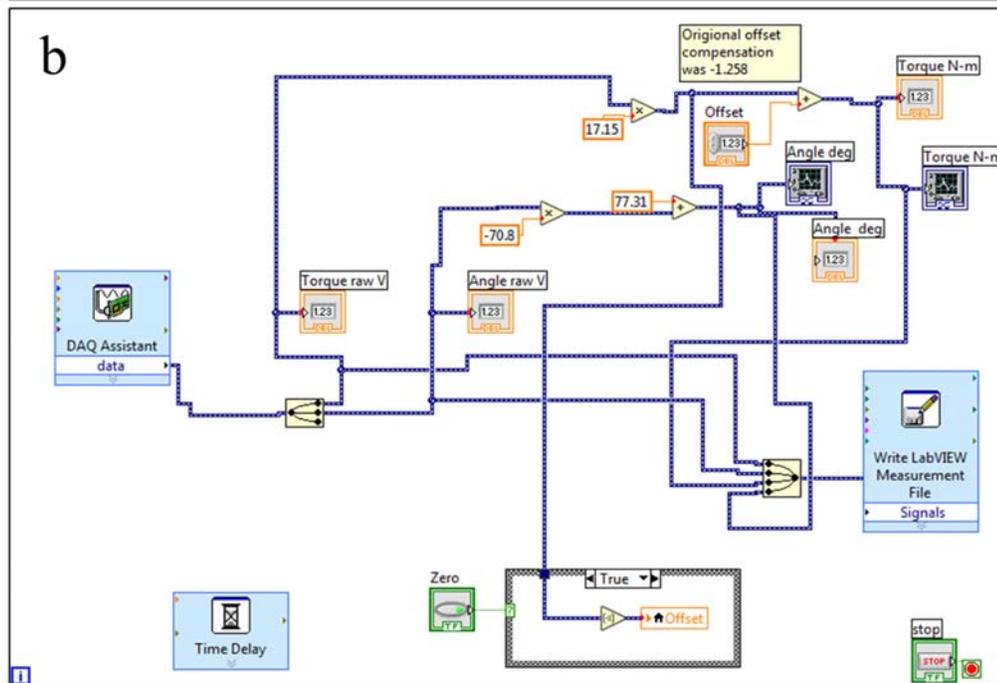
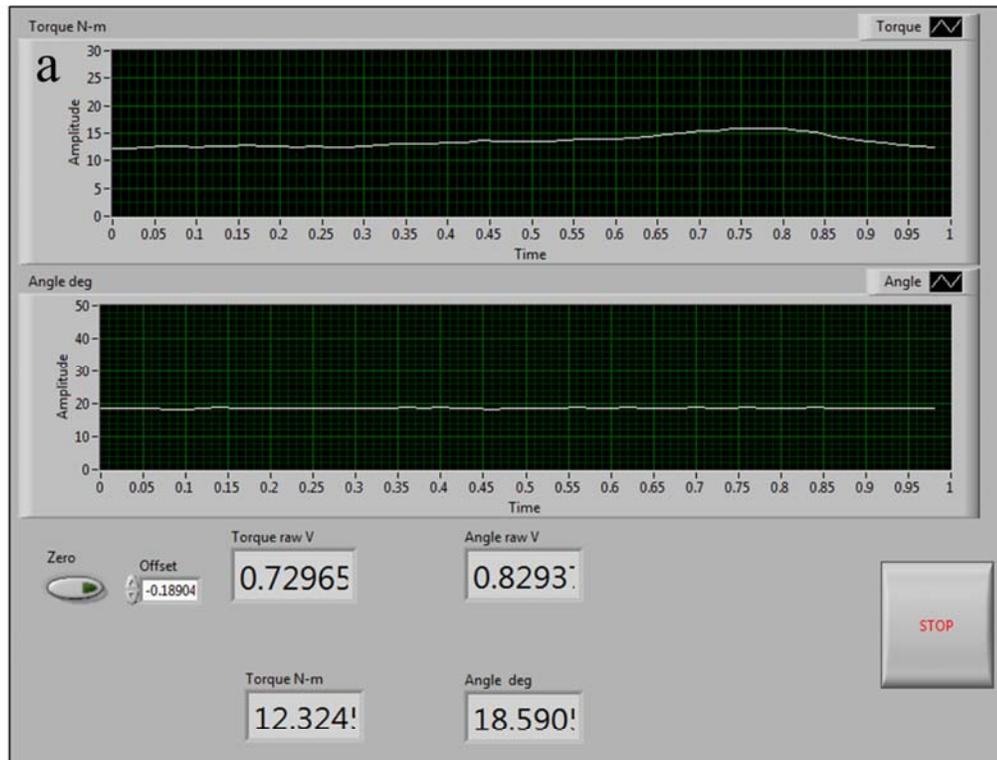


Figure 8-3 Labview software interface (a) and block diagram (b)

8.3.2 Accuracy

Accuracy is intended here as the description of the systematic error (statistical bias) and random error (statistical variability) associated with a measurement (Menditto, Patriarca, & Magnusson, 2007). In this study, limits of agreement (LoA) and mean bias were used as a measure of accuracy (Bland & Altman, 1986).

8.3.2.1 Angle

The predicted angle was compared to the software readings for that angle (i.e. plate fixed at 50° and record the angle). All angles from 50° dorsiflexion to 20° plantarflexion (in 10° increments) were tested. Results are reported in Table 1. For each angle, we computed the mean of 500-recorded values (10 sec).

8.3.2.2 Torque

Starting with zero weight, the weight of the carrier was added; then additional 2.5 kg calibrated weights were added. For each load, a 10 sec period was allocated before adding the next weight. The expected torque was compared to the software readings for that weight. The frontal plate was kept in a neutral position and weights were added perpendicularly to it.

8.3.2.3 Statistical analysis

For each angle, 500 values were averaged and the standard deviation calculated. The same computational process was performed for the torque. The Bland-Altman plot (Bland & Altman, 1986) was used to visually inspect the differences between the computed theoretical values and the measured values (of both torque and angle); and how the differences might change in proportion to the magnitude of the measure. Limits of agreement (Bland & Altman, 2003) were used to assess differences between two types of evaluation methods: 1) device accuracy from concurrent tests, and 2) device repeatability from the same re-test conditions. The LoA provides an estimate that 95% of measured observations can be expected to lie within limits of agreement defined by the mean bias and coefficient of repeatability. Specifically, $LoA_{\text{between}} = \text{Mean difference}_{\text{between}} \pm CR_{\text{between}}$. For the accuracy test, the mean difference was defined by

$$\frac{\sum_i^{500}(x_e - x_m)}{500} \pm 95\% CI \quad (2)$$

where x_e is the expected value and x_m is the measured value. The coefficient of repeatability ($\pm CR_{\text{between}}$) is computed by $CR_{\text{between}} = 1.96 \times SD_{\text{between}}$, where SD_{between} is the standard deviation of the between method differences ($x_e - x_m$).

8.3.3 Repeatability and Reliability

A study was conducted to establish the repeatability and reliability of the dynamometer in measuring the joint torques produced by the toe flexor muscles. Ten participants (7 men and 3 women, mean height 1.75 ± 0.1 m; mean weight 74.9 ± 15.5 ; mean BMI 24.3 ± 3.2) gave their informed consent to undergo a familiarisation and two testing sessions conducted on different (non-sequential) days.

Each participant reported to the laboratory at the same time of the day. The protocol consisted of a pre warm-up period of 1 min where the participants repeatedly performed toe flexion/extension movements with no resistance applied followed by submaximal isometric contractions with incremental exertion up to maximal contraction. After a 3-minute rest, three 5 second-maximal contractions were performed. Protocol design was such that learning effect was minimized, different ability to contract foot muscles accounted for, and maximal muscle pre-activation achieved.

Participants sat on a chair with their knee and ankle fixed at 90 degrees. Metatarsal-phalangeal joints (MPJs) were fixed at 30 degrees of dorsiflexion as recommended for optimal torque production (Goldmann, Sanno, Willwacher, Heinrich, & Brüggemann, 2013) and secured to the bottom plate through a means of Velcro® straps. The head of the metatarsals (1-5) were in line with the transverse axis of the device. Raw data were filtered using a 101-point (2 sec) moving average. The highest torque value among the trials (1-3) was used for analysis.

8.3.3.1 Statistical analysis

For repeatability, mean and standard deviation of the differences between the two sessions were used to calculate the limits of agreement using the Bland-Altman plot as described previously. The coefficient of repeatability and mean bias were also

computed. For reliability, a two way mixed single measures (absolute agreement) was used to calculate Interclass Correlation Coefficients (ICC; 3,1). All statistics were run in SPSS (Version 24, SPSS Inc., Chicago, IL). The level of significance was set to $\alpha=0.01$.

8.4 Results

8.4.1 Accuracy

Results from the accuracy study are showed in **Table 8-1** (and Supplementary Figure 1). For angle, the largest difference between expected and measured values (0.23°) was at 10 degrees dorsiflexion, while the lowest error (0.03°) was recorded at 0 and 20 degrees plantarflexion. Overall, the absolute mean difference was 0.12° and the absolute percentage difference was 0.81%. For torque, the highest difference between expected and measured values (0.34 Nm) was recorded at the highest load (42.93 Nm), while the highest percentage difference (2.9%) was recorded at 7.93 Nm expected torque. Overall, the absolute average difference was 0.16 Nm with an absolute percentage difference of 0.85%. Mean bias of measurement for torque was -0.07 Nm with a CR of 0.39 Nm. For the angle, the mean bias was 0.10° with a CR of 0.21° (**Supplementary Figure 1**).

Table 8-1 Validity results for the angle and torque measurements. Difference (Diff) between predicted values and measured are reported; Absolute Average Difference (Abs Avg Diff) is also reported as raw and percentage. Typical error and Coefficient of variation (Coeff of var) are reported as raw and percentage respectively.

Angle ($^\circ$)					
Predicted	Measured mean \pm SD	Diff (%)	Abs Avg Diff (%)	Typical error	Coeff of var
50	49.78 \pm 0.16	-0.22 (-0.44)	0.12 (0.81)	0.08	0.6%
40	40.06 \pm 0.17	0.06 (0.15)			
30	29.91 \pm 0.17	-0.09 (-0.30)			
20	19.83 \pm 0.16	-0.17 (-0.85)			
10	9.77 \pm 0.17	0.23 (2.30)			
0	0.03 \pm 0.16	0.03 (-)			
-10	-10.15 \pm 0.16	-0.15 (-1.50)			
-20	-20.03 \pm 0.17	-0.03 (-0.15)			

Torque (Nm)					
Predicted	Measured mean \pm SD	Diff (%)	Abs Avg Diff (%)	Typical error	Coeff of var
0	0.01 \pm 0.07	-0.01 (-)	0.16 (0.85)	0.14	0.9%
2.93	2.93 \pm 0.06	0.00 (0)			
7.93	7.70 \pm 0.07	-0.23 (-2.90)			
12.93	12.76 \pm 0.07	-0.17 (-1.31)			
17.93	17.89 \pm 0.07	-0.04 (-0.22)			
22.93	22.98 \pm 0.07	0.05 (0.22)			
27.93	28.06 \pm 0.06	0.13 (0.47)			
32.93	33.24 \pm 0.07	0.31 (0.94)			
37.93	38.25 \pm 0.06	0.32 (0.84)			
42.93	43.27 \pm 0.07	0.34 (0.79)			

8.4.2 Repeatability and reliability

Results from the repeatability test are reported in **Table 8-2** (and **Supplementary Figure 8-1**). The two testing sessions were not significantly different ($t(9) = -2.11$, $p = 0.64$) with a mean bias of -1.13 ± 3.9 Nm.

The average measures interclass correlation coefficient was excellent ($ICC = 0.99$); with 95% of the samples having confidence intervals (CI) between 0.95 and 1.00 which shows high reliability. The within-observation variance was also found to be low (3.96 [Nm]²) with a between-observation variance of 92.28 [Nm]².

Table 8-2 Mean (\pm SD) torque produced by toe flexor muscles (in a 30° of dorsiflexion at the MPJ joint) for session one (test) and two (retest). Results reported for Interclass Correlation Coefficient (ICC), within-observation and between-observation variance [Nm]², mean bias, and coefficient of repeatability (\pm CR).

	test	retest	ICC	within	between	mean bias
	mean \pm SD	mean \pm SD	(95%CI)	variance	variance	(\pm CR)
Torque (Nm)	18.75 \pm 9.2	19.88 \pm 10.5	0.99 (0.95-1.00)	3.96	92.28	-1.13(\pm 3.9)

8.5 Discussion

In this study, we tested the accuracy, repeatability and reliability of a method to test toe flexor strength. Results suggest that our bespoke dynamometer is an accurate tool for measuring angular position and torque: mean bias for torque measurements (-0.07 Nm) and for angular position measurements (0.1°) were less than a unit; the CR for torque (0.39) and for angle (0.21) were also small. Therefore, our device is not only accurate, but it has a small instrument error (noise in the measuring device).

When tested for between-session repeatability and reliability in measuring toe flexor strength, our device showed low bias (-1.13 ± 3.9) confirming its repeatability, and high interclass correlation coefficient (ICC=0.99) confirming its reliability. Although torque measurements in the second session were generally higher than in the first, the not significant ($p = 0.41$) difference (+1.13 Nm or +6%), gives confidence on the accuracy of the number of sessions (one familiarisation and two tests) and the warm-up protocol defined, to minimizing any learning effect.

It has been reported that measurement of torque is affected by many technical factors, such as the applied methodology (Mickle, Nester, Crofts, & Steele, 2013), and joint orientation (Goldmann & Brüggemann, 2012). Here we propose an accurate and reliable standardized methodology – with an improved design – compared with previous devices (Goldmann & Brüggemann, 2012; Miyazaki & Yamamoto, 1993). The first metatarsal bone has a higher (from ground level) effective centre of rotation than the smaller toe bones, therefore the effective axis of all phalanges working together is tilted relative to the ground plane. We included an additional degree of freedom to account for the mediolateral slope of the effective rotational axis of the phalanges.

Our study is the first to propose an estimate of instrument repeatability (Limits of Agreement) when performing toe flexor strength tests by dynamometer. The importance in reporting the degree of measurement accuracy is well-documented (Denegar & Ball, 1993; Hopkins, 2000; Smith & Hopkins, 2011). Poor accuracy reduces the ability to monitor changes over time - both in clinical and experimental contexts; studies not reporting the amount of bias inherent in the measurement may over- or under-estimate the true moment of force produced, therefore their results need to be interpreted with caution.

Our device also has the potential to be used as a training tool, instead of just for evaluation. Strengthening of the foot muscles is commonly achieved with toe-flexion exercises such as towel crunches or marble pickups (Chung, Lee, & Lee, 2016; Feger & Hertel, 2014), short-foot exercises that involve drawing the heads of the metatarsals toward the calcaneus without curling the toes (Lynn, Padilla, & Tsang, 2012), or exercises performed using exercise bands with progressive resistance (Mickle, Caputi, Potter, & Steele, 2016). However, in those exercises the extrinsic foot muscles are activated to some extent, the resistance applied is difficult to quantify exactly, and the efficiency of the training is dependent on the position held by the performer. Our device could potentially be a more effective method to reinforce foot muscles and it could simplify the training plan by setting a constant individualized position, and by setting specific resistive progression while minimizing the contribution of extrinsic foot muscles.

Although the device was accurate in measuring torque and angle, and showed a small measurement bias, it is not possible to confidently assume that the device is able to isolate toe muscles and measure only their strength. The set-up of the machine was such that muscles not crossing the MPJ should have had a small (if any) effect on torque production around that joint, however, this is not certain. It is also acknowledged that during a maximal isometric contraction the extrinsic muscles help in stabilizing the adjacent foot joints therefore, they may have an indirect role in force production. In future, concurrent use of motion capture system, electromyography, and/or foot plantar pressure devices with dynamometers will better define if any secondary movements (i.e. imperceptible heel raising) play a role in the development of torque around the MPJ.

8.6 Conclusion

This study evaluated the performance of a bespoke dynamometer, which had been designed to measure maximal toe flexor strength. The results indicate that the device is accurate when measuring torque and flexion angle, and repeatable and reliable when measuring maximal joint torque developed by toe flexor muscles. In future studies, the ability of the device to reliably discriminate between different groups of people (i.e. different gender or sport) should be tested in a larger sample.

8.7 References

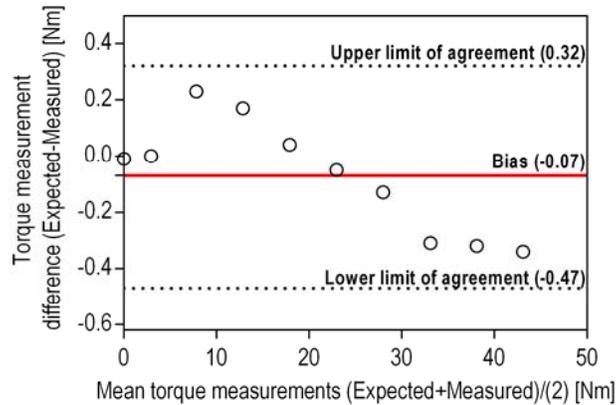
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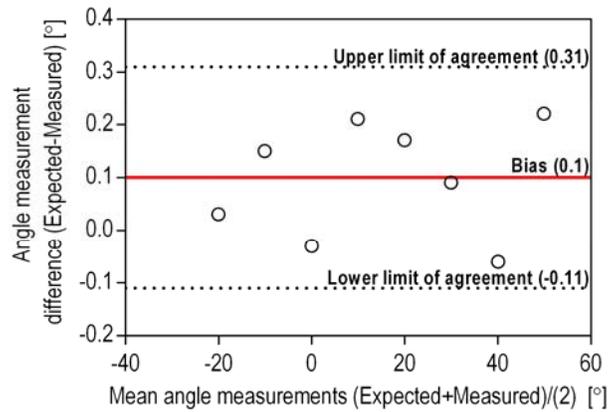
8.8 Supplementary Figure 1

Bland-Altman plots for torque (A), angle (B), and toe strength test (C).

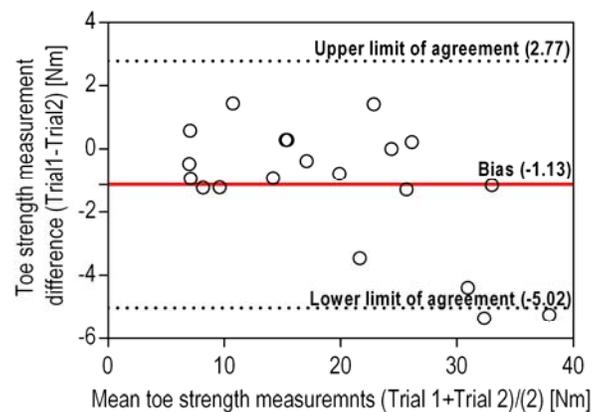
A



B



C



9 EVALUATING DYNAMIC ERROR OF AN INSTRUMENTED TREADMILL AND THE EFFECT ON MEASURED KINETIC GAIT PARAMETERS: IMPLICATIONS AND POSSIBLE SOLUTIONS

This chapter is an amended version of the manuscript: Garofolini, A., Taylor, S., & Lepine, J. (2019). Evaluating dynamic error of a treadmill and the effect on measured kinetic gait parameters: Implications and possible solutions. *Journal of Biomechanics*, 82, 156-163. Published version in Appendix C.

9.1 Abstract

The dynamic properties of instrumented treadmills influence the force measurement of the embedded force platform. We investigated these properties using a frequency response function, which evaluates the ratio between the measured and applied forces in the frequency domain. For comparison, the procedure was also performed on the gold-standard ground-embedded force platform. A predictive model of the systematic error of both types of force platform was then developed and tested against different input signals that represent three types of running patterns. Results show that the treadmill structure distorts the measured force signal. We then modified this structure with a simple stiffening frame in an attempt to reduce measurement error. Consequently, the overall absolute error was reduced (-22%), and the error in force-derived metrics was also sufficiently reduced: -68% for average loading rate error and -80% for impact peak error. Our procedure shows how to measure, predict, and reduce systematic dynamic error associated with treadmill-installed force platforms. We suggest this procedure should be implemented to appraise data quality, and frequency response function values should be included in research reports.

9.2 Introduction

Force platforms are an essential measurement device in many biomechanical studies, from which kinetic parameters are derived to evaluate gait. As an adjunct to the common ground-installed force platform sensor (G_{FS}), the treadmill-installed force platform sensor (T_{FS}) is becoming popular in gait research laboratories (Dierick, Penta, Renault, & Detrembleur, 2004; Riley et al., 2008; Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007). Given that kinetic parameters depend on accurate force signal measurements (Pàmies-Vilà, Font-Llagunes, Cuadrado, & Alonso, 2012; Silva & Ambrósio, 2004), data quality and research integrity relies upon the known degree of measurement error associated with these force-instrumented treadmills. The precision of a force measurement device is dependent upon the inherent natural frequency of its structure. Depending on the mass and stiffness of a treadmill structure, and on the force sensor size (Dierick et al., 2004), treadmill dynamic behavior may generate mechanical vibrations and mode shapes at specific frequencies (natural frequencies) that could approach the frequency content of applied forces from human gait and create artefacts in the measurements. While the ground-installed force platforms have natural frequencies much higher than the frequency content of the exerted force (Antonsson & Mann, 1985), the natural frequencies of the treadmill installed platforms have been reported to be as low as 16 Hz in some cases (Draper, 2000) that is within the frequency content of normal gait (reported as 35-50 Hz (Antonsson & Mann, 1985; Blackmore, Willy, & Creaby, 2016)), affecting the accuracy of the measured force by the strain gauges (force sensors) (Willems & Gosseye, 2013). Nowadays, there is a rise in research that uses parameters derived by treadmill-installed force platforms data for training and retraining (rehabilitative) interventions, in both sport (Crowell & Davis, 2011) and clinical settings (Van den Noort, Steenbrink, Roeles, & Harlaar, 2015), as well as for development of new technologies (Mooney & Herr, 2016). Although accurate measurement of force data is paramount, it is not common practice to include an independent report on the frequency response and the expected measurement error of the forces.

The error inherent within force measurement is best detected and evaluated from frequency domain analysis (Gruber, Boyer, Derrick, & Hamill, 2014; Gruber, Davis, & Hamill, 2011). Therefore, this study will evaluate the Ground Reaction Force signal (GRF) in the frequency domain and describe its harmonic contents, as per

White, Agouris, & Fletcher (2005). The inherent error in the GRF created by the natural frequency of the treadmill is not a random noise that may disappear by taking the average or integration of measured signals across gait cycles. Instead, this error is systematic; it has the same effect on each measurement episode. Bias created by the natural frequency is not related to the magnitude of signal noise that can be overcome by smoothing process that produces a best-fit line (De Bièvre, 2009), but it is related to the degree of difference between the measured and smoothed signal and the true signal (Menditto, Patriarca, & Magnusson, 2007). Therefore, bias is an essential feature to consider when comparing measurements obtained across different force platform systems.

At the author's best knowledge, only one study included the issue of natural frequency testing on instrumented treadmills (Sloot, Houdijk, & Harlaar, 2015). They presented a new approach to test the performance of treadmills, assessing the accuracy of forces and center of pressure, including assessment of the natural frequency. However, they did not explore the effect of low natural frequencies on force signals, nor propose any solution to improve treadmill performance. Our study continues upon this theme by outlining a standardized method to evaluate natural frequencies and their effect on measurement bias. The three aims of this study were: i) to evaluate measurement bias (systematic error) of an instrumented treadmill using a test for frequency-dependent behavior of a force platform; ii) to develop and evaluate a model that is designed to predict measurement bias of the force platform frequency response; and iii) to reduce measurement bias of an instrumented treadmill.

9.3 Methods

The aims were addressed in three stages. Stage 1 assessed the dynamic behavior of the instrumented treadmill using Frequency Response Function (FRF) (Rao & Yap, 2011). This was achieved by evaluating the signal frequency ratio between two interacting force measurement devices. We used a hammer installed force sensor (H_{FS}) to apply an impact force to a treadmill-installed force platform sensor (T_{FS}), and to a ground-installed force platform sensor (G_{FS}). Stage 2 evaluated a model that was developed to predict the dynamic behavior of the treadmill (refer to (Rao & Yap, 2011) for more details on the mathematical procedure used to develop the model). Stage 3 assessed a solution to improve the dynamic behavior of T_{FS} by altering the support structure of

the treadmill. We then assessed the dynamic behaviour of the new TW_{FS} using the predictive model.

9.3.1 Stage 1

9.3.1.1 Analysis of treadmill frequency response

The Fourier transform represents any signal - such as the force signal - as a sum of periodic waveforms (e.g. sine functions). Each waveform is characterized by a frequency (ω), an amplitude (A) and a phase (ϕ). This allows investigation of how the signal's amplitude and phase vary for any given frequency. The systematic error of the force platforms (T_{FS} or G_{FS}) can be represented in the frequency domain using a FRF. The FRF is a frequency dependent modulation system that alters the frequency properties of the input signal (**Figure 9-1**). For example, the amplitude (A_i) and phase (ϕ_i) of the input signal pass through the modulation function, where the signal is transformed into an output signal with new amplitude (A_o) and phase (ϕ_o).

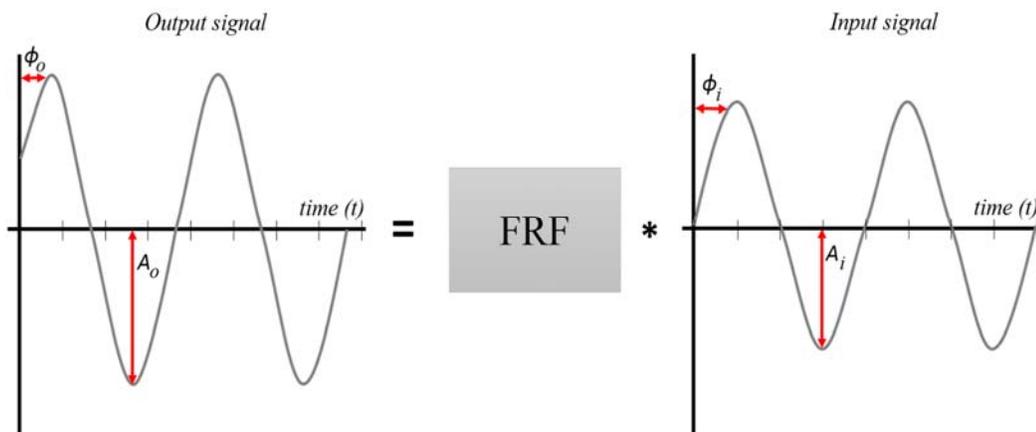


Figure 9-1 Response of a linear time-invariant system to a sinusoidal input (right). The steady state output (left) depends on the characteristics of the system (FRF).

The computed FRF can predict how the output signal of T_{FS} (or G_{FS}) diverges from the input signal by comparing the amplitude (A_i) and phase (ϕ_i) of the H_{FS} (input), with the amplitude (A_o) and the phase (ϕ_o) of the output signal (T_{FS} or G_{FS}) at each frequency. The output signal is described at each frequency by equation 9-1:

$$(A_i(j\omega)\angle\phi_i(j\omega))(A_{FRF}(j\omega)\angle\phi_{FRF}(j\omega)) = A_o(j\omega)\angle\phi_o(j\omega) \quad (9-1)$$

where ω is $2\pi f$, and f is frequency in Hz. The input signal ($A_i\angle\phi_i$) is multiplied by the modulation system ($A_{FRF}\angle\phi_{FRF}$). This can be rewritten in terms of the modulation system as:

$$A_{FRF}(j\omega)\angle\phi_{FRF}(j\omega) = \frac{A_o(j\omega)\angle\phi_o(j\omega)}{A_i(j\omega)\angle\phi_i(j\omega)} \quad (9-2)$$

Now, it is possible to look at how the system (FRF) reacts for each frequency of the input signal using the following transfer function estimator:

$$FRF(\omega) = \frac{FP(\omega)}{H(\omega)} \quad (9-3)$$

where $FP(\omega)$ is the Fourier transform of the force platform signal and $H(\omega)$ is the Fourier transform of the hammer signal. The change in amplitude and phase caused by the modulation system can then be represented as:

$$A_{FRF}(\omega) = |FRF(\omega)| \quad (9-4)$$

$$\phi_{FRF}(\omega) = \angle FRF(\omega) \quad (9-4i)$$

where A_{FRF} defines how the system affects the amplitude of the input signal (in absolute terms) for any given frequency, and ϕ_{FRF} defines how the system affects the phase of the input signal for any given frequency.

9.3.1.2 Measurement

The H_{FS} was composed of a high precision force sensor (PCB Piezotronics, 218A) fixed on the head of a modified hammer, so-called impact hammer. The G_{FS} were embedded into a ground-installed force platform (BP600900TT, AMTI, USA). The T_{FS} were embedded into a treadmill-installed force platform (DBCEEWI, AMTI, USA). The impact hammer has been calibrated using a known mass and accelerometer (Waltham & Kotlicki, 2009) and connected to a 2 channel charge amplifier (Rion, UV-16). The devices were synchronized using Nexus data acquisition system (Oxford Metrics Ltd, Oxford, UK) at a sample frequency of 2000 Hz. The H_{FS} has a flat response up to 1000 Hz, therefore it provides an accurate measure of the force applied to the platforms. The ratio between the output from platform force sensors and the H_{FS} shows how the measurement is affected by the dynamic behavior of the system. When the response is 1 N/N, it means that the force measured by both instruments perfectly match.

Using the hammer we generated a set of 20 vertical impacts at five locations on each platform (four corners and the platform center). The average magnitude of the impacts was 100.2 ± 39.7 N, which is the linear range of the force platform (0-8800 N) meaning that the measured FRF is valid for any force below 8800 N. The FRF linearity was validated with a coherence function which was above 0.90 between 5-200 Hz (Randall, 2008). Data were exported to Matlab (Math Works Inc., USA) for FRF analysis, averaging the 20 impacts to achieve adequate coherence function between 0 and 100 Hz. In order to evaluate the dynamic behavior of the treadmill, the FRF was computed from the force signals of force platforms and hammer using the so-called H1 estimator (Rocklin, Crowley, & Vold, 1985), which reduces the effect of the measurement noise in the force platforms signal, therefore:

$$\text{FRF}(\omega) = \frac{P_{FPH}}{P_{HH}} \quad (9-5)$$

where P_{FPH} is the cross-spectrum between the force platform and the hammer signals, and P_{HH} is the auto-spectrum of the H_{FS} signal (Randall, 2008). Amplitude and phase were then evaluated to investigate the occurrence of the first mode of vibration (i.e. natural frequency).

9.3.2 Stage 2

9.3.2.1 Predictive Model

The FRF of the measurement devices (e.g. force platform on the treadmill) represents, in the frequency domain, how a force measurement is distorted at every frequency by the dynamic behavior of the measurement device (e.g. natural frequency of the structure). An ideal measurement device would have a flat FRF throughout its frequency range which means that there would be no amplification nor delay between the real input (e.g. applied force) and reading (e.g. measured force).

Effect of the amplification and delay on the measurement can be assessed in the time domain using a predictive model. To do so, the first step was to transform the FRF into the time domain using the inverse Fast Fourier transform (Randall, 2008). The transformed FRF is known as the Impulse Response Function (IRF). The reading on the measurement device, $y_{FP}(t)$, in response to a certain input, $x(t)$, can be predicted by convolving the IFR with x :

$$y_{FP}(t) = \text{IRF}(t) * x(t) \triangleq \int \text{IRF}(\tau)x(t - \tau)d\tau \quad (9-6)$$

where τ is a time lag integration variable.

The accuracy of the treadmill and ground-installed force-platforms measurements can be assessed by comparing the predicted response of both measurement devices for different inputs. We selected three archetypal signals that represent the vertical component of typical ground reaction force vectors (VGRF) generated by humans when running (data collected in a previous experiment). These archetypes had distinct impact transients associated with low, medium, and high loading (**Figure 9-2**).

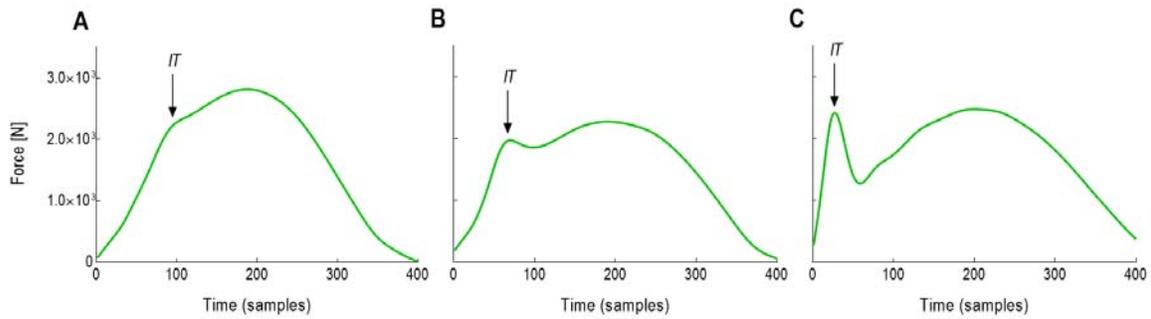


Figure 9-2 GRF archetypal signals with different impact transient properties. The intensity of the loading is low (**A**), moderate (**B**) and high (**C**); IT indicates the Impact Transient.

9.3.3 Stage 3

9.3.3.1 Application and evaluation of a stiffening frame

The treadmill-installed force platforms are supported by a framework structure of steel beams (**Figure 9-3**). The rectangular shape of the treadmill frame lays upon four feet posted at the corners. To stiffen the long axis of the frame and increase the natural frequency, we positioned two wooden support bearers under each long side of the treadmill frame (**Figure 9-3**). To evaluate the bias of the new system, TWFS response was modelled and tested using the three archetypal signals as input. Bias is reported as root mean squared error (RMSE). The natural frequency didn't shift between tests and the coherence function was close to one, which suggests that the supports behave linearly throughout all the tests.

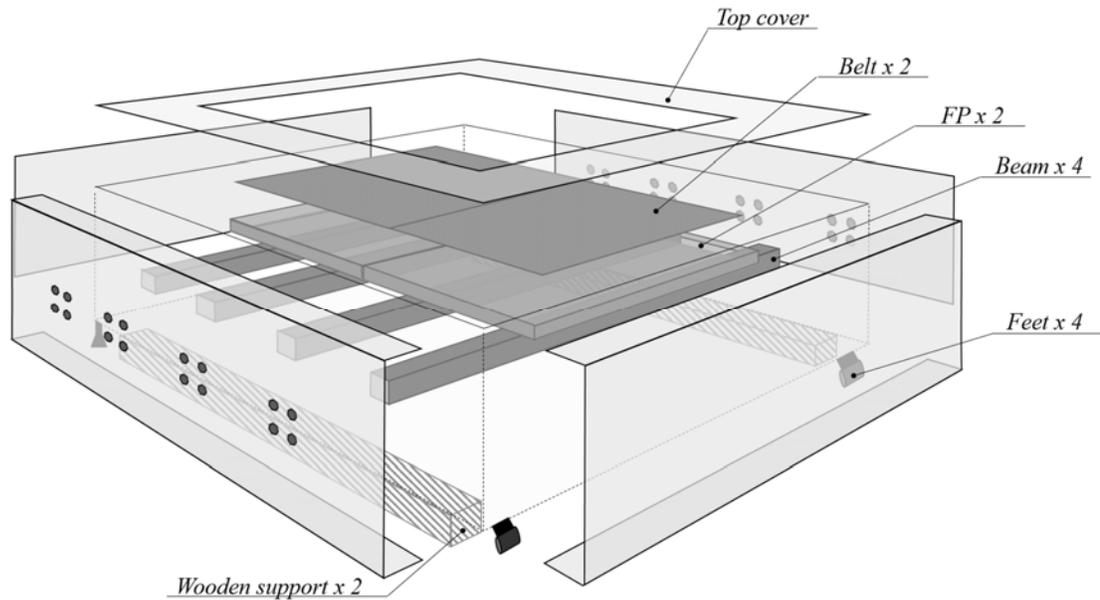
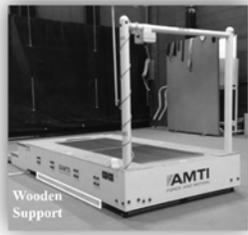


Figure 9-3 Structural components of the instrumented treadmill. Wooden supports were added underneath the lateral sides of the treadmill frame to improve overall stiffness of the device. Treadmill was resting on the wooden supports instead of on the four legs during the experiment.

9.4 Results

9.4.1 Treadmill frequency response

Figure 9-4 presents the amplitude (a) and phase shift (b) features of the FRFs produced from the hammer test on the three measurement systems: G_{FS} , T_{FS} , and TW_{FS} .

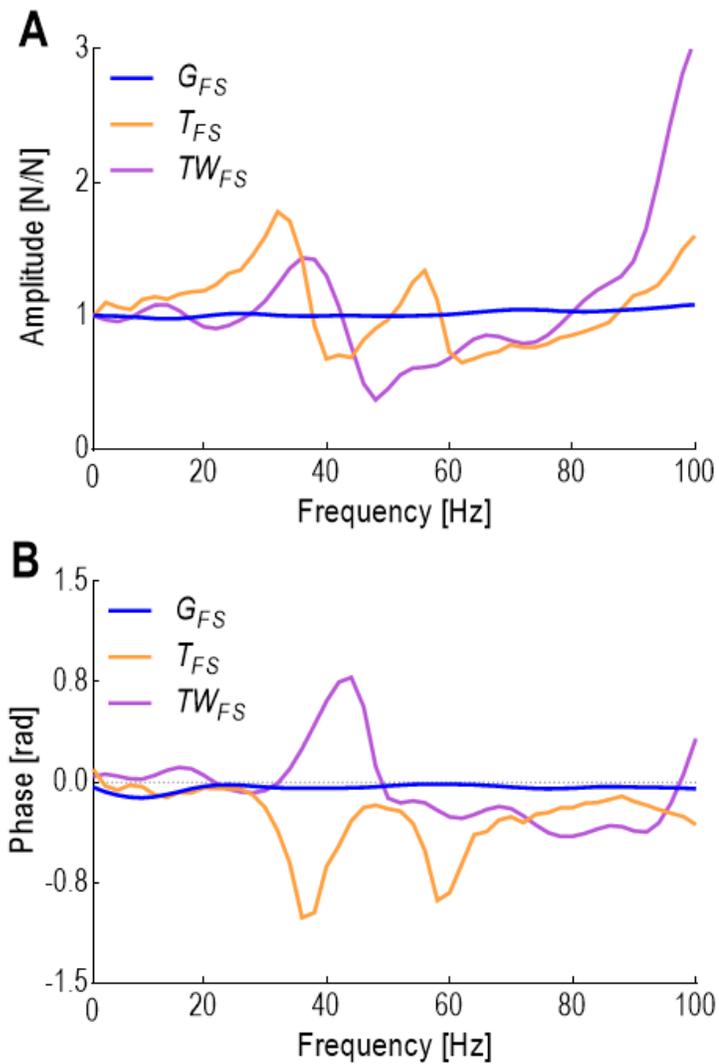


Figure 9-4 Frequency Response Function test displayed in the Amplitude (**A**) and Phase (**B**) domain. FRF outcomes of the three hammer tests are over-ground sensor (GFS, blue), treadmill sensor (TFS, orange), and treadmill with wood sensor (TWFS, purple).

For the amplitude, a FRF < 1 implies there is an underestimation of the signal at that frequency, whereas a FRF > 1 implies that there is an overestimation at that frequency. For instance, at 30 Hz the ratio between the applied force and the measured one is 1.6, which means the measured force at 30 Hz is 37% greater than what it is in reality (i.e. the force applied by the hammer). At 32 Hz there is a 10% increase with respect to 30 Hz. Thus, between 32 ms and 33 ms of the loading phase, the measured signal will show a 10% increase in the first peak force that does not exist in reality. At 40 Hz (ratio 0.68) the measurement by the T_{FS} will underestimate the force by 47%.

The T_{FS} FRF presents two peaks at 32 Hz and 55 Hz; whereas the G_{FS} shows the relatively flat response that is expected from a gold-standard force measurement device (**Figure 9-4a**). After applying wooden bearers to the treadmill, the first natural frequency shifted from 32 to 36 Hz. For the phase, T_{FS} shows two main shifts at the two natural frequencies (32 and 55 Hz) and TW_{FS} has also a phase shift in correspondence of its first natural frequency (36 Hz). In contrast, the G_{FS} shows no phase shift among the analyzed frequencies.

9.4.2 *Effect of improved treadmill stiffness*

Table 9-1 lists the level of agreement between the three archetypal signals and the model-predicted VGRF signals derived from the FRF. The degree of overlap between the measured and archetypal signals for the three different types of impact intensity and force sensor type is shown in **Figure 9-5**. The measurement error of the G_{FS} increases as loading intensity increases while, the lowest error for the T_{FS} was at Medium load (52.5 N) and the highest value was at High loading (127.8 N), representing a 243% relative increase. TW_{FS} follows a similar trend to T_{FS} . The largest difference between T_{FS} and TW_{FS} was in High loading condition with a reduction in RMSE of 48%. Overall the TW_{FS} displays less error (-22%) compared to the T_{FS} . The modified frame reduced the error in the variables related to the impact transient, such as average loading rate (ALR) and impact peak. The TW_{FS} exhibits an error 3-times lower in the ALR (a reduction of 68 percentage points), and an error 5-times lower in the impact peak (a reduction of 80 percentage points; see **Table 9-1**).

Table 9-1 Root mean squared error (RMSE) is reported as a measure of bias. The error of over-ground force platform sensor (G_{FS}), treadmill-installed force platform sensor (T_{FS}), and adapted treadmill (TW_{FS}) are reported for low loading (Low), medium loading (Med) and high loading profiles (High). The average (AVG) is also reported. RMSE is reported as raw values [N], percentage of peak force, and percentage of mean force. Average loading rate (ALR) and Impact peak are reported as percentage change from the archetypal VGRF signals. ALR was computed between 20-90% of impact peak.

		Loading pattern			
		Low	Med	High	AVG
RMSE [N]					
	G_{FS}	3.9	7.0	8.4	6.4
	T_{FS}	56.7	52.5	127.8	79.0
	TW_{FS}	68.4	54.9	60.7	61.3
RMSE % Peak Force					
	G_{FS}	0.1	0.3	0.3	0.3
	T_{FS}	2.0	2.3	5.2	3.2
	TW_{FS}	2.4	2.4	2.4	2.4
RMSE % Mean Force					
	G_{FS}	0.2	0.5	0.5	0.4
	T_{FS}	3.5	3.5	7.2	4.7
	TW_{FS}	4.2	3.6	3.4	3.7
ALR ($\Delta\%$)					
	G_{FS}	-2.0	-3.8	-1.3	2.4
	T_{FS}	1.8	12.3	3.7	5.9
	TW_{FS}	-1.5	3.4	0.8	1.9
IMPACT PEAK ($\Delta\%$)					
	G_{FS}	-0.4	0.0	0.4	0.3
	T_{FS}	4.1	4.8	9.2	6
	TW_{FS}	1.1	1.3	1.1	1.2

Figure 9-5 (a-c) shows the three archetypal signals (a – low; b – medium; c – high) compared against the predicted force reading for the G_{FS} , T_{FS} and TW_{FS} . **Figure 9-5** (d-f) represents the raw error for each condition. Main error for the T_{FS} is in the first half of stance at high loading with an evident oscillatory behavior that decays over

time. TW_{FS} consistently overestimates the force measurement in early stance and underestimates it from mid stance forward. G_{FS} almost perfectly measures force applied in any loading condition.

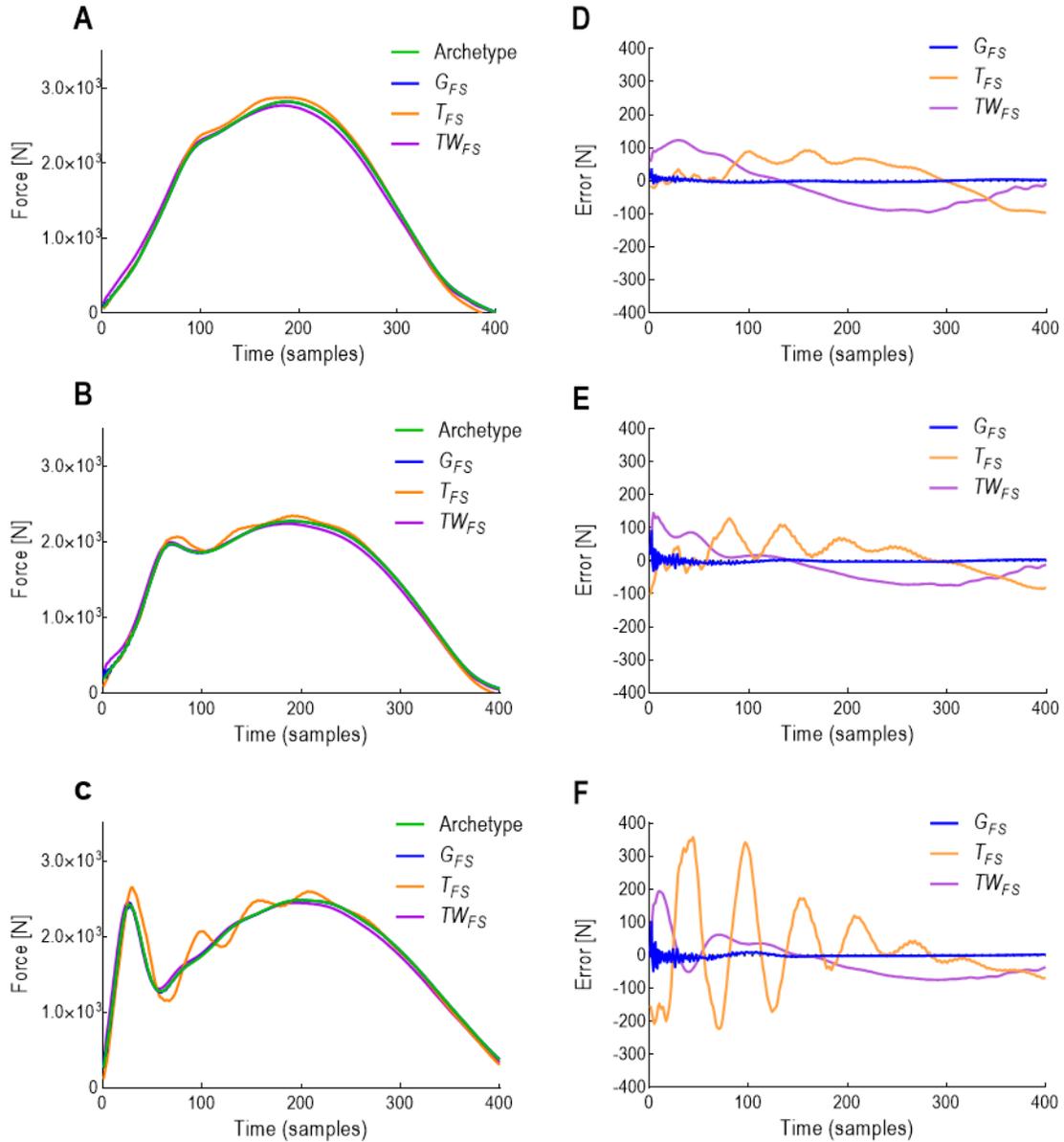


Figure 9-5 Archetypal VGRF signals from over-ground running with low loading (A), medium loading (B), and high loading (C). Archetypal VGRF signal (green) is compared against over-ground model-prediction (G_{FS} blue), treadmill model-prediction (T_{FS} orange), and new treadmill configuration (with wood bearers) model-prediction (TW_{FS} purple). Error for each model is reported for low loading (D), medium loading (E), and high loading (F).

9.5 Discussion

The general aim of this study was to evaluate the force measurement bias from a typical T_{FS} by comparing it against a ‘gold standard’ G_{FS} . The force reading from the G_{FS} is precise across a range of analyzed frequencies (1-100 Hz), whilst the signal from the T_{FS} has some measurement bias. Any applied force to the T_{FS} that is above 10 Hz will either over- or under-estimate the true magnitude of the applied force and this measurement error will depend on the frequency content of the applied force.

The measurement error of the treadmill followed a different trend compared to the ground-installed force platform. While the G_{FS} showed a consistent increase with the loading intensity, the T_{FS} was inconsistent between these three archetypal signals. This is explained by the number and position of the treadmill’s natural frequencies. The G_{FS} has a very high first natural frequency (> 500 Hz), while the treadmill has two natural frequencies at approximately 32 and 55 Hz. Therefore, as the frequency content of the applied force increases with increased loading intensity, it is adjacent to the first natural frequency at Low loading, it sits between the two natural frequencies at Medium loading and it is adjacent to the second natural frequency at High loading. As the application of wood support bearers does not eliminate the natural frequencies, the trend is similar for the TW_{FS} .

The first natural frequency of the treadmill was identified at 59 Hz prior to shipping. This suggests that the measured first natural frequency (32 Hz) was either not identified by the manufacturer, or the testing conditions were different. For instance, the soft elastic floor covering the ground (Mondo®) in our laboratory creates a compliant substrate of the treadmill-floor interface, which may have changed modes in the frequency bandwidth of interest. To further investigate the reasons for these discrepancies, a full modal analysis of the treadmill including several degree of freedom must be performed in different laboratory environments (e.g. floor structure, and mounting conditions). This type of systematic study would highlight how the dynamic behaviors of the system depend on its boundary conditions and establish general guideline for instrumented-treadmill installation.

The position where the measurements are made could also affect the number of natural frequencies appearing in the frequency response function. If the excitation or the measurement has been made on a ‘node’ of a mode shape, the natural frequency of this mode doesn’t appear on the FRF. As the tests presented in this paper were

conducted at the point where the runner most commonly hits the platforms (i.e. its center), we ensured that all the relevant natural frequencies were measured. After modelling the FRF for the G_{FS} , T_{FS} and the adapted TW_{FS} , we then compared their output force measurement with archetypal signals. While the G_{FS} seems to be more consistent in measurement error between loading intensities, the T_{FS} behaves differently depending on the type of VGRF profiles (**Figure 9-5**): it may be the case that the frequency content of the input signal is actually increasing as the loading profile of the VGRF increases. VGRF with high loading profile has a frequency content close to a resonance frequency of the treadmill, therefore the measured force signal is amplified. Instead, when the VGRF curve becomes smoother the frequency content changes - reduce - moving away from a resonance frequency; as a result, the signal is minimally amplified due to the structural damping.

Due to the low natural frequencies of the treadmill, the T_{FS} VGRF profile degenerates, leading to errors in measures of gait particulars associated with the impact transient (**Table 9-1**). For instance, the recorded signals by the T_{FS} show that there can be errors in impact transient parameters of up to 12%. Accurate measurement of impact transient parameters is important for clinical evaluation of running performance and risk of injuries (Davis, Milner, & Hamill, 2004; Milner, Ferber, Pollard, Hamill, & Davis, 2006). Moreover, results from running retraining studies (Crowell, Milner, Hamill, & Davis, 2010) aiming to reduce the impact transient may be affected by the dynamic behavior of the instrumented treadmill. The measurement bias could be either systematic or random - because it is dependent upon frequency; hence if a person applies different load intensities the observed error could vary (under/over) between foot contacts within a trial. Therefore, pre-post intervention differences may be partially contributed by the bias associated with the dynamic (vibratory) behavior of the treadmill. For many future studies using instrumented treadmills, researchers could evaluate the confidence they have in their data by using the FRF and IRF method. Indeed this is performed by manufacturers prior to shipping, however, this evaluation also needs to be conducted in the lab setting.

It is worth noticing that measurement errors – related to the dynamic behavior of the treadmill – will pass undetected when error evaluation techniques are employed with conventional static calibrations (Gill & O'Connor, 1997; Hsieh, Lu, Chen, Chang, & Hung, 2011). The results from the dynamic validation method performed in this

study demonstrates the effect that a T_{FS} can have on the data quality within a biomechanics lab, and raises the necessity to include such an evaluation procedure as regular practice prior to the reporting of data. The evaluation of the modified TW_{FS} is indicative of why a T_{FS} should be tested in its specific environment and condition. The application of supports underneath the body of the treadmill showed an overall improvement of the ratio between input (hammer) and output (force platform), reducing the measurement error of the VGRF. Although the natural frequency has been increased slightly (from 32 Hz to 36 Hz), the reduction of the error is remarkable. For instance, at 30 Hz the ratio decreased from 1.60 to 1.15, reducing the 37% artificial increase in force recording to just 13%. When comparing the amount of measurement bias (RMSE) and the change in loading variables across the different loading conditions, the modified TW_{FS} shows a smaller average error (**Table 9-1**). Although a benchmark of an acceptable error limit will vary according to derived parameters, we can consider a level of error equivalent to that of the ground embedded force platform as the gold standard benchmark. Achieving this will require improvement in two areas: (i) mathematical models of the frequency response, and (ii) engineering a stiffening frame comparable to a ground embedded force platform. A mathematical model will minimize the effect of systematic error; while an improved frame structure will increase resonance frequency and provide a more reliable measurement of high frequency forces.

Indeed, the effect of systematic artifact will have a greater impact on certain users and their analyses, while others might find these levels acceptable. For example, the ground reaction force orientation may be sufficiently altered to affect joint kinetic parameters, particularly the hip joint moments (where a combination of both kinematic and kinetic errors would exist). In another context, the appeal of using instrumented treadmills is that they accommodate analyses that require long continuous data sets. However, analyses that quantify time-series behavior of gait parameters (e.g. (Dingwell, John, & Cusumano, 2010; Hausdorff et al., 1996) should be cautious when considering similar analyses on gait parameters measured from instrumented treadmills, particularly impact transient.

An alternative method to avoid sensor natural frequency related error is to use a digital low-pass filter. Commonly, in running studies, force signals are low-pass filtered with a cut-off frequency of 50 Hz (Baggaley, Willy, & Meardon, 2017; Cheung

& Rainbow, 2014; Kulmala, Avela, Pasanen, & Parkkari, 2013) with some using 100 Hz (Hobara, Sato, Sakaguchi, Sato, & Nakazawa, 2012). As the frequency content of the force signal recorded during running can reach frequencies up to 50 Hz (Blackmore et al., 2016; Shorten & Mientjes, 2011), any cut-off frequency lower than 50 Hz will necessarily delete part of the true signal. In our case, as the first natural frequency started affecting the signal at 10 Hz, a lower cut-off frequency (i.e. 6 Hz) would be needed to remove the amplification effect caused by the treadmill dynamic behavior, however, it will also smooth every sharp change in the signal (i.e. rising portion of the GRFv). Therefore, when applying a low-pass filter to the force signal, the user should appreciate the effect of three influential factors: (1) the natural frequency of the treadmill; (2) the typical frequency content of the force signal being recorded (i.e. influence of different types of impact); and (3) the type of bias that the treadmill's dynamic behavior has on the force signal. In this study we showed how to address those issues with a rather simple test. Results will give confidence not only on the validity of the force signal, but also on the adequacy of low-pass filter cut-off frequency.

The main limitation of this study is the generalizability of our results. As the laboratory environment affects the natural frequency, the error found and solution proposed is only applicable to our treadmill. However, with this study we highlight the need of ensuring appropriate system quality check and report of measurement associated error which should be a priority for any biomechanical laboratory. Although our method was able to raise the natural frequency of the treadmill, it improved force reading accuracy without suppressing the bias. However, the procedure presented highlights that an evaluation of T_{FS} measurements performed in the frequency domain provide sensitive characteristics of the force signal that can expose any presence of systematic error – this form of measurement error would otherwise be undetected through time domain procedures. Such an evaluation should always be performed *in situ*, that is, in the specific environment and condition in which the treadmill is used, and results should accompany any reported data for quality assurance.

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10 CONCLUSIONS

10.1 Summary of results

This thesis has presented a comprehensive series of studies on adaptability in long-distance runners. As introduced in Section 1.3, system entropy can define the expansion, preservation, or regression of the workspace dimensionality. In the case of habitual RFS runners, the neuro-locomotor workspace was expected to reduce in dimensionality; therefore, it was expected that adaptation to structural, functional, and control properties of the system would be observed.

This relates to five hypotheses:

10.1.1 Rearfoot strikers have reduced foot bone density and simpler structural organisation

In chapter 3 it was reported that RFS have similar bone density at the calcaneus but reduced trabecular area at the metatarsus to FFS, indicating the direction of the external force may be important in shaping bone structure. The result of this different foot posture at landing, is for the FFS to stress the metatarsal bone. When landing on the forefoot this force has a more variable direction, which requires a rearrangement of the trabecular bone; therefore, our results suggest that the external force magnitude may not be the only important factor as previously assumed in bone formation (Kersting & Bruggemann, 1999).

10.1.2 Rearfoot strikers have reduced foot muscle size, tendon thickness, and foot strength

In Chapter 3 it was reported that the repeated loading from a foot strike type does not affect the anthropometry of foot muscles and tendons. The fact that RFS and FFS had similar muscle size and force produced, can likely be explained by the high running volume experienced by both groups of runners. These findings are consistent with the conclusions reported from the systematic review presented in Chapter 2.

10.1.3 Rearfoot strikers have reduced ankle stiffness and joint coupling variability

Compared to RFS, the FFS runners have increased ankle stiffness adaptability, but the exploitation of such ability is dependent on the shoe worn. For both groups of runners, shoes with a low MI index reduce adaptable behaviour while shoes with high MI index enhance adaptability (Chapter 4). Similarly, RFS have a less variable coordinative pattern (Chapter 5); adding further support to the claim of lost adaptability in this group of runners.

10.1.4 Rearfoot strikers have reduced control of leg length-force dynamics during stance

Through analysis of statistical persistence along time-series of leg stiffness (Chapter 6) this thesis was able to describe the relationship between a high-level goal (leg stiffness) that needs to be optimised, and the real-time sensorimotor control of muscle stiffness. This control is achieved by a fine balance between exploitation of passive structures and active control. Passive control has the advantage of being energetically cheaper and highly complex. That is, the system uses a wide variety of elements to achieve similar functions. Alternately, active control, constrains the movement to a reduced set of elements. In general, FFS have a more complex and adaptable system compared to RFS, therefore requiring less tight (high level) control regulation of leg stiffness.

10.1.5 Rearfoot strikers have reduced kinematic synergies of leg length and orientation during impact

Runners stabilize the vertical position of the body centre of mass allowing the horizontal position to change more rapidly in response to external perturbations (Chapter 7). However, FFS exhibited a wider distribution of functional kinematic synergies along the uncontrolled manifold. This result confirms the hypothesis that habitual forefoot strikers have developed a higher level of adaptability, while RFS have a less entropic system.

In addition, Chapter 8 and 9 report the measurement error (accuracy) associated with the two primary instruments used to evaluate function and control properties of the groups. A new device that provides reliable and accurate measurement of toe flexor

strength was developed. In chapter 9 the measurement, prediction, and reduction in systematic dynamic error associated with treadmill-installed force platforms was reported.

10.2 Executive summary

As a whole, this thesis provides new knowledge, toward a better understanding of the complexity of the neuro-musculoskeletal system and its adaptability to external forces during long-distance running. Our bodies are highly adaptable to changing external conditions, but we also reproduce highly consistent movements so that control intervention is minimized. Adaptable systems are therefore those able to find multiple stable solutions that optimize performance. As runners accumulate experience with an antithetical foot strike pattern, not only do biological tissue properties change, but a plastic remodelling of neural network connectivity may occur so that coordination habits are affected.

From the results presented in this thesis, it is not possible to say that one foot strike pattern is ‘better’ than another. However, evidence is provided that runners with antithetical foot strike patterns adapted differently, and these differences are not confined to movement kinematics and kinetics but also to how the control system organizes movement. Rearfoot strikers exhibited a shrinking workspace of neural and biological elements reducing their system complexity, thus they require more active intervention to absorb, disperse, and recycle energy during the stance phase of running. On the other hand, forefoot strikers rely on self-organizing optimality of key variables deployed to disperse the mechanical stresses inherent to running.

Footwear also effects adaptation. Rearfoot strikers in this study’s sample relied more on the shoe substrate to absorb external forces. However, the habit of using such a strategy made them (RFS) reluctant to adopt any other kinematic solution to a different substrate – evidence of reduced complexity. Therefore, while rearfoot strikers may be ‘safe’ while running in controlled external conditions, their system is not expanding through their experiences, and thus, by comparison, they are gravitating around deterministic types of motor behaviours, that represent a restricted entropic system, and reduced adaptability.

10.3 Potential queries for future work

Results from this thesis indicate multiple opportunities for future research. In this last section a series of questions is posed that may be answered in the short-term using the data already reported here, and in the long-term if the line of thinking here is taken further.

10.3.1 Does the difference in bone architecture between RFS and FFS result in a different stress distribution along the metatarsus?

The information gathered from the bone scans can be used for computational approaches such as micro-finite element analysis (μ FEA) of bone (Pistoia et al., 2002). This method simulates in vivo conditions using complex geometric models with defined material properties (i.e. stress-strain relationship). However, any material has specific properties, and in the case of the bone, the trabecular and cortical bone differs substantially in density and morphological composition. One of the limitations in current FE analysis is the assumption that bone is homogeneous and isotropic. With HR-pQCT this thesis indicated different morphological compositions of these components. Utilising this information in μ FEA may allow a more detailed analysis of stress distribution and resistance to external forces.

10.3.2 Does the flight phase of running reveal adaptive strategies?

Functionally, this thesis analysed the behaviour and control of the landing leg during the support phase (Chapter 4, 6, and 7). It would be interesting to extend the analytical methods used in Chapter 6 and 7 to the flight phase. There is mounting evidence that proper load of passive elastic tissues during flight phase may reduce the energy expended during running (Simpson et al., 2018). Although the position of the body centre of mass during the flight phase cannot be changed, the position and posture of the legs are determined based on previous take-off conditions and expected future landing conditions. Therefore, analysis of the flight phase may further explain the adaptive strategies utilized in response to previous interactions with the ground and in preparation for landing. Indeed, computing stiffness of leg segments when they are not in contact with the ground would require calculation of segment's inertial properties and their velocity.

10.3.3 Is there a compensatory control between dominant and non-dominant limbs?

It would be interesting to analyse differences in behaviour and control between the dominant and non-dominant legs. Asymmetry in performance is often related to an increased risk of injury (Brumitt, Heiderscheit, Manske, Niemuth, & Rauh, 2013), and a wide body of literature has quantified asymmetry using indexes (Carpes, Mota, & Faria, 2010). However, those indexes have not been used to examine mechanisms of neuromuscular control. A normalized symmetry index (Gouwanda & Senanayake, 2011) can be applied to results from detrended fluctuation analysis (Chapter 6) and the uncontrolled manifold analysis (Chapter 7). This will link asymmetry to motor performance and neurological control.

10.3.4 Can DFA be used to distinguish between the two hierarchical levels of control?

Based on interpretation of the statistical persistence in time-series data utilised in this thesis (Dingwell & Cusumano, 2010), it is expected that analysis of the statistical persistence in the elemental variables (i.e. force and leg length) used to compute leg stiffness would demonstrate if those variables are free to vary (high statistical persistence) while leg stiffness time-series will hold low persistence (more active control).

To do so, an experiment would need to be set up to allow leg stiffness to be computed in real-time and visually displayed so that participants are given a clear task goal of keeping leg stiffness within certain limits. Based on the knowledge that when healthy humans walk in time with a metronome their stride times become less strongly correlated (Hausdorff et al., 1996; Terrier, Turner, & Schutz, 2005), it is expected that participants would hold low persistence on leg stiffness (more active intervention), while force and/or leg length may be free to vary.

10.3.5 Can control of leg stiffness be trained?

If it is assumed that control of leg stiffness depends on the sensitivity the system has towards this variable, in order to teach how to control leg stiffness it is necessary to make the value of leg stiffness manifest (evident) to the participant in real-time (i.e. visual feedback). In this way, the sensory information related to a value of leg stiffness can be recognised and internalized. Real-time feedback computing the resultant GRF and leg length change is needed. Utilising the force analog output of an instrumented

treadmill and the 3-dimensional position of the body centre of mass, change in resultant force and change in leg length can be computed. Using a microcontroller-based signal processor and an analog-to-digital converter, it is possible to generate an analog (or digital) of leg stiffness in real-time. In line with the principle of self-organisation of motor action (Schoner & Kelso, 1988), the task of controlling leg stiffness should be the goal of the participants – perhaps using thresholds under which the stiffness values should be kept – but no instructions on how to achieve this should be provided.

10.3.6 Can the model used for motor control be linked to physiological processes?

This thesis utilised a framework based on two theories of motor control: optimal feedback control theory (Todorov & Jordan, 2002), and dynamic system theory (Kelso & Schöner, 1988). Although Todorov and Jordan (2002) described the mathematical model of optimal control, presenting data acquired from both experimental and simulation experiments, the model did not effectively show how this is implemented physiologically, or how stochastic processes such as growth or learning affects these models. Similarly, the holistic view of the dynamic system theory, although useful in describing system self-organisation, does not address the physiological basis of such organisation. Therefore, it worth pondering what it would take to break down the optimal feedback control processes in biologically plausible processes. Only one theory, the threshold control theory (Feldman & Levin, 2009) brings forward a biologically relevant basis to explain motor control. In this view, the only parameter (or control variable) the nervous system is able to change is the activation threshold of alpha-motor neurons. By shifting this threshold, muscles will generate an adequate force to pass from an equilibrium point to another, or better, from one stable posture to another. This model has been shown to fit experimental data in single (Abdusamatov, Adamovich, & Feldman, 1987), and in two-joint movements (Flanagan, Ostry, & Feldman, 1993).

All of these theories are not mutually exclusive, instead, they look at motor control from a different perspective; one (Todorov & Jordan, 2002) from a computational point of view, one (Kelso & Schöner, 1988) from an ecological point of view, the last one (Feldman & Levin, 2009) from a physiologically sound

perspective. In future, integration of these perspectives will give rise to more precise model of motor control.

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APPENDIX A Published manuscript chapter 2

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The effect of running on foot muscles and bones: A systematic review



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ABSTRACT

Despite the widespread evidence of running as a health-preserving exercise, little is known concerning its effect on the foot musculature and bones. While running may influence anatomical foot adaptation, it remains unclear to what extent these adaptations occur. The aim of this paper is to provide a systematic review of the studies that investigated the effects of running and the adaptations that occur in foot muscles and bones. The search was performed following the PRISMA guidelines. Relevant keywords were used for the search through PubMed/MEDLINE, Scopus and SPORTDiscus. The methodological quality of intervention studies was assessed using the Downs and Black checklist. For cross-sectional studies, the Newcastle-Ottawa scale was used. Sixteen studies were found meeting the inclusion criteria. In general, the included studies were deemed to be of moderate methodological quality. Although results of relevant literature are limited and somewhat contradictory, the outcome suggests that running may increase foot muscle volume, muscle cross-sectional area and bone density, but this seems to depend on training volume and experience. Future studies conducted in this area should aim for a standard way of reporting foot muscle/bone characteristics. Also, herein, suggestions for future research are provided.

1. Introduction

Running is an important form of exercise because it is inexpensive, accessible, and it provides many health benefits (Lee et al., 2017); however, many of these benefits can only occur through repetitive loading of anatomical structures, and the effect of overload will lead to musculoskeletal injury and non-participation (Nohren, Davis, & Hamill, 2007; Pepper, Akuthota, & McCarty, 2006). Bones and muscles are adaptive tissues that develop in structure and function in response to mechanical load and metabolic demands, which is a demonstration of activity-dependant plasticity (Kiely & Collins, 2016). However, tissue can also be maladaptive. While repetitive load may cause a positive hypertrophic response in bone (Chen, Beaupré, & Carter, 2010) and muscles (Seynnes, de Boer, & Narici, 2007); the converse occurs with a reduction (or removal) of load – due to immobilization, physical inactivity, or microgravity exposure – resulting in tissue decay through the process of bone resorption (Holick, 2000; Kiratli, Smith, Nauenberg, Kallfelz, & Perkash, 2000) and muscle atrophy (Powers, Kavazis, & DeRuisseau, 2005). Runners can modulate the nature of the stresses experienced by bone and muscle by altering limb kinematics at impact (Li, Zhang, Gu, & Ren, 2017), or by selecting compliance variations in terrain surface and footwear substrates (Firminger, Fung, Loundagin, & Edwards, 2017); this is because both approaches will effect a change in the direction and magnitude of the external and internal forces applied to the lower limbs. In accordance with activity-dependent plasticity principle, there will exist certain kinematic-substrate combinations that lead to optimal

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adaptation of foot structure and function and help mitigate injury risk for runners, whereas other combinations will amplify risk. To adequately understand the pathological effect of maladaptive foot structure and function on running injury, a prerequisite step is to first understand the effect of repetitive running load on changes to foot anatomy. The motivation for this review is that this mechanistic effect remains largely unknown due to limited research exploration (Lee et al., 2017).

Repetitive stress injuries are very common among runners, especially stress fractures of the foot (van Gent et al., 2007). Around 55% of these fractures occur in the metatarsals – mostly second and third (Petzer & Wright, 2006); the calcaneus, talus, navicular and sesamoid account for 6% (Groshar et al., 1997; Pelletier-Galarneau, Martineau, Gaudreault, & Pham, 2015). Long distance runners tend to be afflicted by metatarsal stress fractures more than other athletes (Brukner, Bradshaw, Khan, White, & Crossley, 1996). This high injury rate might be related to training distance (van Gent et al., 2007), training volume (Hreljac, 2004), and runners' biomechanical adaptations (Davis, Rice, & Wearing, 2017). During running, human locomotor system broadens the distribution of stress that arises from impact forces (Hart et al., 2017) by active modulation of muscle activity (Olin & Gutierrez, 2013) and hence joint torques and rotational energy (Lieberman et al., 2010). Because the foot is the most proximal aspect of the lower limb to the external ground forces, the effect of the stresses will be larger than elsewhere in the lower limb (Lieberman et al., 2010; Lieberman, 2012a, 2012b); furthermore, the foot may happen to have the most sensitive anatomy of the lower limb to exhibit activity-dependent plasticity (McKeon, Hertel, Bramble, & Davis, 2014).

Previous studies have shown an increased incidence in bone stress in runners who were transitioning from 'cushioned' footwear to minimal shoes (Johnson, Myrer, Mitchell, Hunter, & Ridge, 2016). The authors found that those who transitioned without negative effects to minimal shoes developed larger adductor hallucis muscles, while those who developed bone stress had smaller foot muscles. Popp et al. (2017) investigated the association between tibial cortical bone density and stress fractures in runners, finding substantially weaker bones in the stress fracture group at the mid-shaft of the tibia. Results from the previous studies (although based on acute interventions) suggest that stronger foot muscles and bones may be protective, while weak feet may be more likely to be injured. However, the long-term effect of the loads generated in the foot bones and muscles during running remains unknown. This knowledge could be used to study the contribution of mechanical load to foot musculoskeletal development and health maintenance, which is essential information for devising methods of injury prevention and treatment.

Measuring bone and muscle adaptations is difficult *in vivo*. Even if bone strength can be approximated by dual-energy X-ray absorptiometry (DXA) (Cummings, Bates, & Black, 2002) and computed tomography techniques (Norton & Gamble, 2001), the problem remains that bone mineral density (BMD) is not the only determinant of bone strength. Innovative 3D analysis of high-resolution images can now provide an insight into bone microstructure and architecture; this technique has shown to be less dependent on bone density than DXA (Geusens et al., 2014), outperforming ultrasound and previous X-ray scanning techniques in terms of image resolution (up to 82 μm) and level of radiation exposure ($< 3 \mu\text{S}$) (Cheung et al., 2013). Muscles have been imaged by techniques other than conventional radiography, such as magnetic resonance imaging (MRI), and ultrasound scanning. Compared to the former, ultrasound imaging (US) is widely available and rather inexpensive, allowing valid measure of muscle size through real-time high-resolution imaging (Mickle, Nester, Crofts, & Steele, 2013).

The load-related changes (adaptations) in foot muscle and bone may influence more variable running form and biomechanical solutions (Lieberman et al., 2015), resulting in minimisation of an accumulation of repeat stresses, however, solid evidence on the effect of running on the anatomical foot structure is needed to perorate this claim. Several original papers (Bobbert, Yeadon, & Nigg, 1992; Bramble & Lieberman, 2004; Bus, 2003; Davis et al., 2017; Gruber, Davis, & Hamill, 2011; Hasegawa, Yamauchi, & Kraemew, 2007; Hunter, Marshall, & McNair, 2005; Kasmer, Wren, & Hoffman, 2014; Lieberman et al., 2010, 2015; Lieberman, 2012a, 2012b, 2014; Nigg, 2010; Nigg, De Boer, & Fisher, 1995; Shu et al., 2015; Stefanyshyn & Nigg, 1997), as well as systematic reviews (Almeida, Davis, & Lopes, 2015; Hall, Barton, Jones, & Morrissey, 2013; Hollander, Heidt, Van Der Zwaard, Braumann, & Zech, 2017; Perkins, Hanney, & Rothschild, 2014; Schubert, Kempf, & Heiderscheit, 2014) analysed kinematics and kinetics of runners, with only some (Hollander et al., 2017; Shu et al., 2015) reporting findings on the long-term effect of running on foot morphology. The review by Hollander et al. (2017) concluded that habitual barefoot runners have wider feet and a reduced hallux angle than individuals that habitually wear shoes. However, most of the studies included in their review did not control for likely confounding variables such as body weight or running experience. Indeed, any structural change has also to be related to running volume and the amount of time spent resting between runs. Moreover, although they reported changes in foot morphology, the review by Hollander et al. (2017) focused on the differences between barefoot and shod populations, and they did not address adaptations to intrinsic foot muscle or bone. Therefore, the aim of the present paper is to review the evidence regarding the effect of running on foot musculoskeletal adaptations.

2. Methods

2.1. Search strategy

A systematic search of the literature was conducted in accordance with the PRISMA guidelines (Moher, Liberati, Tetzlaff, Altman, & Group, 2009). PubMed/MEDLINE, Scopus, and SPORTDiscus databases were used to search for relevant literature from the inception of indexing up to the 1st November 2018. Combinations of the following keywords were used as search: running AND ("foot muscle" OR "foot muscles" OR "bone density" OR "bone strength" OR "bone composition" OR "muscle cross sectional area" OR "muscle volume" OR "foot morphology" OR "foot muscle morphology" OR "muscle strength" OR "foot strength"). Secondary searches were performed by checking the reference list of included articles as suggested by Greenhalgh and Peacock (2005). Forward citation tracking of the included studies was performed in Google Scholar.

2.2. Eligibility criteria

Studies were considered eligible if they met the following inclusion criteria: (1) published in English language; (2) published in a peer-reviewed journal; (3) included human participants; (4) used a randomized controlled trial (RCT), a case-control, a prospective cohort, or a cross-sectional study design; (5) measured foot muscle characteristics and/or foot bone characteristics; (6) at least one of the included groups was comprised of active runners. Exclusion criteria were studies reporting on groups or individuals with pre-existing medical conditions, such as metabolic diseases or foot anatomical deformation.

2.3. Coding of studies

The following information was extracted from the included studies: (i) sample size; (ii) groups description; (iii) main findings related to muscle/bone characteristics; and (iv) methods used to measure muscle/bone characteristics.

2.4. Methodological quality

Methodological quality of the included intervention studies was assessed using the validated Downs and Black scale (Downs & Black, 1998). For assessing cross-sectional studies, the modified Newcastle-Ottawa Scale was used (Wells et al., 1999). For the Downs and Black scale, studies scoring from 0 to 8 points were considered as being of poor methodological quality, studies scoring from 9 to 17 points were considered as being of moderate quality, and studies that scored 18 to 27 points were considered as being of high methodological quality. The maximum score on the Newcastle-Ottawa scale is 10 points. Based on the total score on the Newcastle-Ottawa Scale the studies were defined as either low quality (score ≤ 3 points), moderate quality (4–7 points), or high quality (score > 7 points). The datasets analysed during the current study are available from the corresponding author on reasonable request.

3. Results

3.1. Search results

The initial search resulted with 5487 search results. After the removal of duplicates, 3677 papers were screened, and excluded based on title, abstract, or in some cases, based on the full-text. In total, 41 full-text papers were read. Thirteen studies met the inclusion criteria (Best, Holt, Troy, & Hamill, 2017; Chen, Sze, Davis, & Cheung, 2016; Escamilla-Martinez et al., 2016; Fredericson et al., 2007; Fuller et al., 2018; Harber, Webber, Sutton, & MacDougall, 1991; Johnson, Myrer, Mitchell, Hunter, & Ridge, 2015; Kersting & Bruggemann, 1999; Laabes, Vanderjagt, Obadofin, Sendeht, & Glew, 2008; Lara et al., 2016; Miller, Whitcome, Lieberman, Norton, & Dyer, 2014; Senda et al., 1999; Zhang, Delabastita, Lissens, De Beenhouwer, & Vanwanseele, 2018). After screening the reference lists of the included studies, three additional studies were included (Drysdale, Collins, Walters, Bird, & Hinkley, 2007; Williams, Wagner, Wasnich, & Heilbrun, 1984). Forward citation tracking of the included studies did not result in the inclusion of additional studies. Thus, the total number of included studies was 16. Fig. 1 reports the flow diagram of the search process.

3.2. Study characteristics

Ten studies used a cross-sectional design (Best et al., 2017; Drysdale et al., 2007; Escamilla-Martinez et al., 2016; Fredericson et al., 2007; Harber et al., 1991; Kemmler et al., 2006; Laabes et al., 2008; Lara et al., 2016; Senda et al., 1999; Zhang et al., 2018) with a sample size ranged from 11 to 401 (median = 45). Four studies (Chen et al., 2016; Fuller et al., 2018; Johnson et al., 2015; Miller et al., 2014) used a RCT design, with sample sizes of $n = 20$, $n = 19$, $n = 18$ and $n = 33$, respectively, one study (Kersting & Bruggemann, 1999) used a 20-week long non-randomized intervention ($n = 8$), and one study (Williams et al., 1984) used a 9 months controlled before-and-after study design ($n = 7$). Two of the RCT studies (Johnson et al., 2015; Miller et al., 2014) were short in duration (10 and 12 weeks, respectively) while the study by Chen et al. (2016) had a 6-month transitioning program.

3.3. Sample characteristics

Overall, 624 males and 347 females (mean = 39M and 22F; median = 20M and 4F) were tested. Eight studies did not include female subjects while two did not include males. Runners ranged on average from 20 to 50 years old (mean = 32) and their body weight ranged from 46 to 78 kg (mean = 68) (Fig. 2A). Habitual training volume was quantified as km/week by ten studies (Best et al., 2017; Chen et al., 2016; Fuller et al., 2018; Johnson et al., 2015; Kemmler et al., 2006; Kersting & Bruggemann, 1999; Laabes et al., 2008; Lara et al., 2016; Miller et al., 2014; Zhang et al., 2018) and was on average 40 km/week (ranged from 25 to 69); whilst two studies (Kemmler et al., 2006; Laabes et al., 2008) reported training volume as kcal/kg/day (mean = 27 ± 12) and min/week (mean = 555 ± 129) respectively, making those studies incomparable with others (Fig. 2B).

Only three studies (Fredericson et al., 2007; Kemmler et al., 2006; Senda et al., 1999) included elite long distance runners, whose definition was not given by Fredericson et al. (2007); while Senda et al. (1999) defined 'elite level' using personal best time for the 3000 m run (mean 9 min and 19 s) and Kaup index (14.8–21.9). Kemmler et al. (2006) defined elite runners as those having a running history of at least 5 years and a running volume of 75 km/week and a time of less than 1.15 h for a half-marathon (or $< 32:30$ min for 10,000 m). The other studies involved 'recreational runners' whose definition was also inconsistent. For instance, Miller et al. (2014)

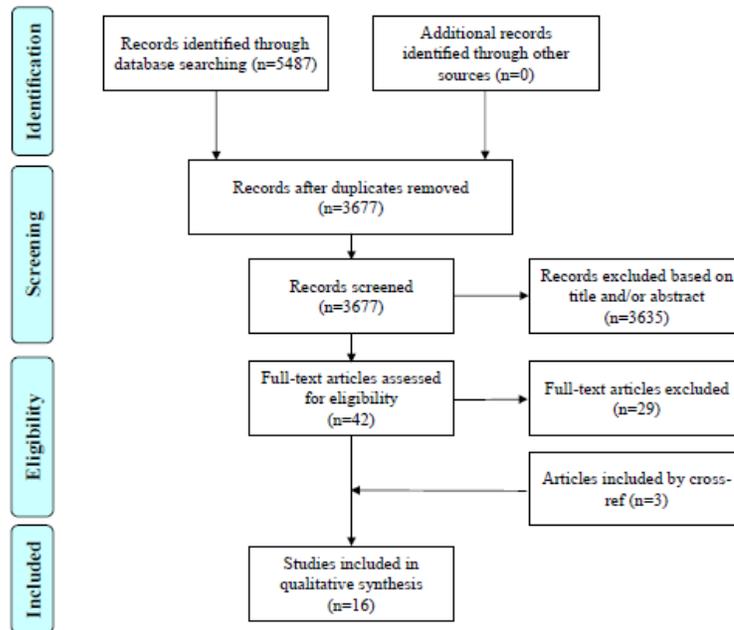


Fig. 1. Flow chart of the search strategy.

defined recreational as those who run an average of 30 miles per week (48.3 km) for a minimum of 12 months. Similarly, for Johnson et al. (2015) recreational was defined as an individual who runs an average of 24–48 km/week for the 6 months prior to the start of the study. However, Escamilla-Martinez et al. (2016) defined recreational runners as those who had been distance running as amateurs for at least five years and training at least three times per week with minimum per session duration of one hour.

3.4. Measuring techniques characteristics

Methods used to measure foot muscle or bone characteristics also varied between the studies. Ultrasound-transmission velocity and broadband ultrasound attenuation were the main methods used to quantify bone density. Other techniques reported were photon absorptiometry, compton scattering technique, and peripheral instantaneous X-ray imaging. Only one study, (Best et al., 2017) used high resolution peripheral computed tomography to analyse trabecula characteristics of the calcaneus. For muscle measures, ultrasound and magnetic resonance imaging were most commonly used along with a custom toe dynamometer. Table 1 summarize the details of studies included in the analysis.

3.5. Methodological quality

Quality scores for the Downs and Black scale and the modified Newcastle-Ottawa Scale are reported in Table 2. The RCTs (Chen et al., 2016; Fuller et al., 2018; Johnson et al., 2015; Miller et al., 2014) had a score ≥ 18 points and were classified as being of high methodological quality. The non-randomized studies (Kersting & Bruggemann, 1999; Williams et al., 1984) scored 10 points and were classified as being of moderate methodological quality (Table 2A). Eight of the ten cross-sectional studies (Best et al., 2017; Escamilla-Martinez et al., 2016; Fredericson et al., 2007; Harber et al., 1991; Kemmler et al., 2006; Laabes et al., 2008; Lara et al., 2016; Senda et al., 1999) scored between 4 and 7 points on the Newcastle-Ottawa Scale, and, therefore, they were all classified as being of moderate quality (Table 2B). Only the Drysdale et al. (2007) and Zhang et al. (2018) studies were classified as of high quality (8 points).

4. Discussion

This systematic review summarises findings related to the effect of running on foot muscle and bone characteristics from 16 studies. The current body of evidence on this topic is limited, which highlights the need for future studies. In the next sections, we discuss the most significant findings and provide recommendations for future research in this area. Fig. 3 depicts the main findings of

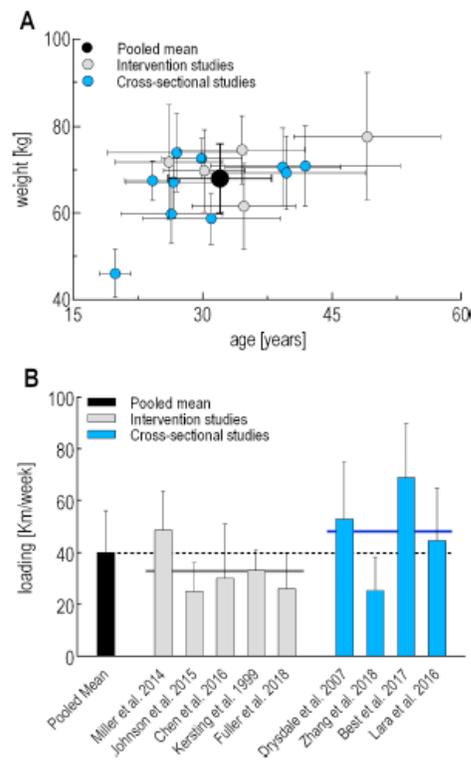


Fig. 2. (A) Sample age by weight distribution for all studies but Zhang et al. (2018) who did not report weight but body mass index; (B) training loading for studies reporting load as km per week. Solid lines represent the mean of the group. Dotted line is the grand mean.

this review and what is still unknown.

4.1. Effect on muscles

Very limited evidence exists indicating that running is associated with increased foot muscle size. Chen et al. (2016) found a muscle growth (+8.8%, $p = 0.01$) in intrinsic foot muscles (measured as a whole) after a 6-month transitioning program to minimal shoes. However, a muscle-strengthening program was also part of the intervention, which may partially explain the change in muscle volume. The control group running in traditional shoes showed no change in foot muscle volume after the program.

Short training intervention may be more effective in increasing muscle size. Johnson et al. (2015) reported a significant increase (+10.6%, $p = 0.01$) in abductor hallucis cross-sectional area after 10 weeks of training in minimal running shoes compared with the change (pre-post) in the control group (+1.8%) who were using traditional running shoes; however, no significant differences were found among all the other intrinsic muscles that were examined. Similarly, after a 12 weeks transitioning period, a +24.7% increase was found in the abductor digiti minimi muscle volume ($p = 0.009$) and a +18.0% increase in the abductor digiti minimi muscle cross-sectional area ($p = 0.007$) of recreational runners (Miller et al., 2014). For the other tested muscles no significant differences were found, and furthermore, no statistically significant differences were found between pre-and post-training in the control group running in traditional shoes.

Based on the limited evidence available, there is an indication that intrinsic muscle strength and muscle size may increase with running but this is dependent on type of footwear and the associated biomechanical changes (Davis et al., 2017; Lieberman, 2012a, 2012b). A stronger foot may better control loading redistribution at each step (McKeon et al., 2014) while reduced strength may limit the ability to control inter-joint movements resulting in increased soft tissue strain; therefore, greater foot strength may be a beneficial adaptation in response to the repetitive loading imposed on the foot during running, which may contribute to a decreased incidence of injuries (McKeon & Fourchet, 2015). When controlling for the shoe worn, loading seems to have less of an effect in stimulating muscle growth: while comparing 4 type of running shoes (neutral, motion control, minimalistic, and neutral with insoles), Zhang et al. (2018) found that among all intrinsic foot muscles selected, only abductor hallucis showed a significant difference

Table 1
Characteristics of the included studies.

Study	Total n	Design	Grouping	Age (y) – BW (kg)	Footwear	Training volume	Intervention duration	Muscle measures	Method	Findings
Senda et al. (1999)	49	Cross-sectional	12 top level marathon runners – 37 healthy control	19.9 ± 1.8 y – 46.1 ± 5.5 kg	//	//	//	Total toe flexors power, abductor power of 1st and 5th	TD	Running (in conventional running shoes) decreases total flexor power
Miller et al. (2014)	33	Randomized control study	Control (recreational runners; n = 16) – recreational runners + intervention (n = 17)	30.2 ± 4.7 y – 69.8 ± 9.5 kg	TBS	48.7 ± 15 km/week	12-Week training regime	MV and CSA of the FDB, abductor digiti minimi (ADM), and ABDH	MRI	Running in minimal shoes (with 4 mm offset or less) strengthen the foot
Johnson et al. (2015)	37	Randomized control study	Sex-blocked randomization. 19 control (recreational runners) – 18 recreational runners + intervention	26.1 ± 6.2 y – 71.8 ± 13.3 kg	TBS	25 ± 11 km/week	10-Week transition period	ABDH CSA (cm ²), FDB CSA (cm ²), FHB thickness (cm), EDB thickness (cm)	MRI, USI	Significant 10.6% increase in abductor hadtule cross-sectional area in the Vibram FiveFingers™ group compared with the control group (p = 0.01)
Chen et al. (2016)	38	Randomized, single-blinded control study	Control (training program in TBS; n = 18) – intervention (training program in MIS + transition exercises + transitioning tips; n = 20)	34.8 ± 6 y – 61.6 ± 9.9 kg	TBS (heel-toe drop > 5 mm)	30.4 ± 21.3 km/week	6-Month transition period	IFM volume	MRI	MIS group had significantly larger foot (p = 0.01, Cohen's d = 0.62) muscles after transition. The forefoot mainly contributed to foot muscle growth (continued on next page)

Table 1 (continued)

Muscle										
Study	Total sbj	Design	Grouping	Age (y) – BW (kg)	Footwear Foot-strike	Training volume	Intervention duration	Muscle measures	Method	Findings
Zhang et al. (2018)	38	Cross-sectional	Neutral shoes (n = 11); motion control shoes (n = 10); minimalist shoe (n = 7); insole (n = 10)	26.3 ± 6.9 y – 22 ± 2.1 BMI	Mixed shoe molds	25.4 ± 13 km/week	//	ADDH CSA (mm ²) and FDB CSA (mm ²) and thickness (mm)-FDB thickness (mm)	US	Runners in minimal shoes had the thickest abductor hallucis
Bone										
Study	Total sbj	Design	Grouping	Age (y) – BW (kg)	Footwear Foot-strike	Training volume	Intervention duration	Bone measures	Method	Findings
Williams et al. (1984)	30	Controlled before-and-after study	Consistent runners (n = 7); inconsistent runners (n = 13); control (n = 10)	49.1 ± 8.5 y – 77.7 ± 14.6 kg	//	//	9Months	Calcaneal bone mineral content	PA	Calcaneal bone mineral content is dependent on training volume. Post intervention, subject training more than 16 km per month has significantly higher bone mineral content than control
Harder et al. (1991)	42	Cross-sectional	Group A (eumenorrheic nonovovative females) n = 14 subjects who reported 9 or more months per year and who exercised fewer than 3 times per week but did not participate in any formal exercise; Group B (eumenorrheic athletes) n = 17 runners who reported 9 or more months per year and who trained 7–12 times per week; Group C (amenorrheic athletes) n = 11 runners who reported no menses in the last 12 months and who trained 7–12 times per week	26.4 ± 5.9 y – 59.8 ± 6.9 kg	//	//	//	Calcaneal density	CST	Amenorrhea in athletes is not associated with any reduction in heel bone density. However, bone turnover rate is significantly greater in athletes

(continued on next page)

Table 1 (continued)

Study	Total sbj	Design	Grouping	Age (y) – BW (kg)	Footwear Foot-strike	Training volume	In intervention duration	Muscle measures	Method	Findings
Kersting and Bruggemann (1999)	26	Non-randomized intervention	3 groups, running shoes of similar construction but different midsole hardness: 45° (n = 9), 53° (n = 9) and 61° (n = 8)	34.6 ± 7.2 y – 74.7 ± 7.9 kg	RFS	33.8 ± 8.2 km/week	20-Week training regime	Calcaneal density	SOS, MRI	No relationship between midsole hardness and external or in-shoe impacts. Bone parameters showed specific differences for all groups which are pronounced in runners with intermediate impacts. Runners displayed significantly higher SOS and BUA than control.
Remmler et al. (2006)	31	Cross-sectional	Endurance trained male runners (n = 20), BMI-matched control (n = 11), aged 20–35 years	26.6 ± 5.2 y – 67.2 ± 6.7 kg	//	555 ± 129 m h/week	//	Calcaneal density	SOS, BUA	The rate of decline of BMD appeared to be reduced significantly in marathon runners compared with the normative group. Running is associated with higher BMD at directly loaded sites (the calcaneus) but not at relatively unloaded sites (the spine).
Drysdale et al. (2007)	401	Cross-sectional	Marathon runners (n = 401; 217 M, 184F), control group from previous studies (n = 601; 267 M, 334F).	41.9 ± 11 y – 70.9 ± 9.3 kg	//	53.8 ± 22.3 km/week	//	Calcaneal density	BUA	The rate of decline of BMD appeared to be reduced significantly in marathon runners compared with the normative group. Running is associated with higher BMD at directly loaded sites (the calcaneus) but not at relatively unloaded sites (the spine).
Frederickson et al. (2007)	45	Cross-sectional	Elite male soccer players (n = 15), elite male long-distance runners (n = 15) and sedentary male controls (n = 15) aged 20–30 years	24.2 ± 3.2 y – 67.5 ± 4.6 kg	//	//	//	Total and regional bone mineral density	DXA	Repetitive skeletal loading at the heel has the potential to improve bone density in black male athletes. The magnitude of increase may be higher in medium impact sports such as soccer and running compared with low or non-impact sports. Distance running seems to have a negative effect on calcaneal bone mass density during the course of a 700-km training season.
Laibson et al. (2008)	102	Cross-sectional	Football (n = 68), running (n = 15), handball (n = 7), taekwondo (n = 6), cycling (n = 2), judo (n = 1), badminton (n = 1) and high jump (n = 1)	31 ± 8 y – 58.7 ± 6 kg	//	27 ± 12 kcal/kg/d (runners only)	//	Calcaneal bone stiffness index	BUA	Repetitive skeletal loading at the heel has the potential to improve bone density in black male athletes. The magnitude of increase may be higher in medium impact sports such as soccer and running compared with low or non-impact sports. Distance running seems to have a negative effect on calcaneal bone mass density during the course of a 700-km training season.
Ecamilla et al. (2016)	95	Cross-sectional	Amateur runners (n = 33); control (n = 62)	39.3 ± 6.7 y – 70.7 ± 9.1 kg	RFS	//	//	Calcaneal density	BUA	Repetitive skeletal loading at the heel has the potential to improve bone density in black male athletes. The magnitude of increase may be higher in medium impact sports such as soccer and running compared with low or non-impact sports. Distance running seems to have a negative effect on calcaneal bone mass density during the course of a 700-km training season.

(continued on next page)

Table 1 (continued)

Study	Total subj	Design	Grouping	Age (y) – BW (kg)	Footwear Foot-strike	Training volume	Intervention duration	Muscle measures	Method	Findings
Butler et al. (2018)	39	Randomized control study	Minimalist shoes (n = 19); conventional shoes (n = 20)	27 ± 8 y – 74 ± 9.1 kg	TRS and MRS	26 ± 14 km/week	20 w/week training regime	Calcaneal and metatarsal (1st to 5th) mineral density (g cm ³)	DXA	Minimalist shoes did not affect bone mineral density after 20 weeks follow-up
Best et al. (2017)	18	Cross-sectional	FFS (n = 6); RFS (n = 6); control (n = 6)	29.9 ± 4.6 y – 72.7 ± 4.6 kg	TRS and MRS	68.8 ± 20.9 km/week	//	Calcaneal volumetric density; trabecular thickness, number, distance between; DA	HRpQCT	Trabecular thickness and mineral density were greatest in forefoot runners with strong effect sizes (< 0.80). Trabecular thickness was positively correlated with weekly running distance (r ² = 0.417, p < 0.05) and years running (r ² = 0.359, p < 0.05). Individuals with the greatest summative loading stimulus had, after body mass adjustment, the thickest trabeculae
Lara et al. (2016)	278	Cross-sectional	Long-distance runners (n = 122); short distance runners (n = 81); control (n = 75)	39.7 ± 9.2 y – 69.3 ± 8.5 kg	//	44.7 ± 20 km/week	//	Calcaneal bone stiffness	BUA, SOS	Long distance runners and short distance runners presented higher values than sedentary counterparts in SOS (p < 0.05), and calcaneus stiffness (p < 0.05). However, there were no significant differences between longer distance and shorter distance runners

MV muscle volume, CSA cross-sectional area, FDB flexor digitorum brevis, ADM abductor digiti minimi, ABDH abductor hallucis, FHB flexor hallucis brevis, EDB extensor digitorum brevis, TD toe dynamometer, PA photon absorptiometry, CST Compton scattering technique, MRI magnetic resonance imaging, USI ultrasound imaging, TRS traditional running shoes, RFS rear foot strike, FFS fore foot strike, MRS minimalist running shoes, IEM intrinsic foot muscles, SOS speed of sound, BUA broadband ultrasound attenuation, DXA dual-energy x-ray absorptiometry, HRpQCT high resolution peripheral computed tomography, DA degree of anisotropy.

Table 2
Methodological quality evaluation using (A) the Downs and Black methodological quality assessment, and (B) the adapted Newcastle-Ottawa Scale.

Study	Scale items																											Total	
	A – Non cross-sectional													B – Cross-sectional															
	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24	25	26	27		
Williams et al. (1984)	1	1	0	1	0	1	1	0	0	1	0	0	0 ^a	0	0	1	0 ^a	1	0	0	0 ^a	1	0	0	0	1	0	0	10
Keršing and Bruggemann (1999)	1	1	1	1	1	1	1	0	0	1	0 ^a	0 ^a	0 ^a	0 ^a	0 ^a	0	0 ^a	1	0 ^a	1	0 ^a	0	0	0	0 ^a	0 ^a	0	0	10
Miller et al. (2014)	1	1	1	1	1	1	1	1	0	1	1	0	0 ^a	0	1	0	0 ^a	1	1	1	1	1	1	0 ^a	0	1	0	18	
Johnson et al. (2015)	1	1	1	1	1	1	1	1	1	1	1	0	0	0	1	1	1	1	1	0	1	1	1	0 ^a	1	1	1	22	
Chen et al. (2016)	1	1	1	1	1	1	1	0	1	1	1	0	0 ^a	1	1	0	0 ^a	1	1	1	1	1	1	1	0	1	0	20	
Fuller et al. (2018)	1	1	1	1	1	1	1	1	1	1	1	0 ^a	0 ^a	0	0 ^a	1	1	1	1	1	1	1	1	0 ^a	1	1	1	21	
B – Cross-sectional																													
Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24	25	26	27	Total	
Harber et al. (1991)	0	0	0	0	0	0	0	0	0	0	0	2	1	1	1	1	1	1	1	1	2	1	1	1	1	1	1	6	
Serdin et al. (1999)	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	4	
Kemmler et al. (2006)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	7	
Drysdale et al. (2007)	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	8	
Frederickson et al. (2007)	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	6	
Laubes et al. (2008)	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	5	
Escamilla-Martinez et al. (2016)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	6	
Lara et al. (2016)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	6	
Zhang et al. (2018)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	8	
Best et al. (2017)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	2	2	1	1	1	1	1	7	

Items 1–10 are related to reporting, items 11–13 are related to external validity, items 14–25 are related to internal validity, item 27 is related to statistical power.
1 criteria met, 0 criteria not met.
^a Item was unable to be determined, scored 0.

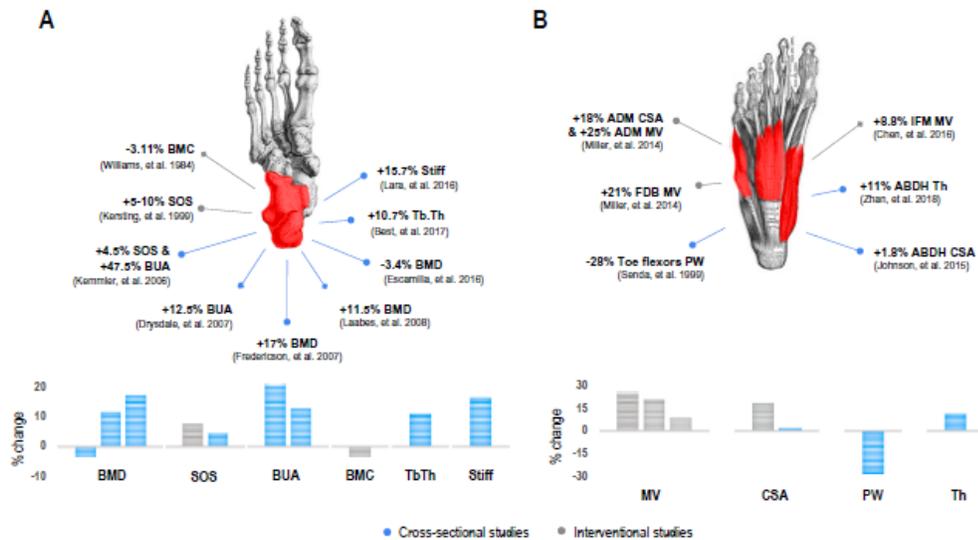


Fig. 3. Results summary of the effect of running on foot bones (A) and foot muscles (B). BMC bone mineral content; SOS speed of sound; BUA broadband ultrasound attenuation; BMD bone mineral density; Tb.Th trabecular thickness; Stiff bone stiffness. CSA cross-sectional area; MV muscle volume; Th thickness; PW power; ADM abductor digiti minimi; FDB flexor digitorum brevis; Abd Hal abductor hallucis; IFM intrinsic foot muscles.

between groups. Runners using minimalistic shoes had the thickest abductor hallucis. More cushioning and restrictive design of traditional shoes may neutralize the action of the intrinsic foot muscles making runners relying more on extrinsic foot muscles for loading redistribution (Murley, Landorf, Menz, & Bird, 2009). Muscle imbalance could explain the lower (-28%) global foot power recorded in marathoners compared against a control group (Senda et al., 1999). Long-term, muscle imbalance may cause foot deformity (Kwon, Tuttle, Johnson, & Mueller, 2009) and increase risk of injury (Nigg et al., 2017; Page, Frank, & Lardner, 2010).

4.2. Effect on bones

A number of studies (Pocock, Eisman, Yeates, Sambrook, & Eberl, 1986; Strobe et al., 2015; Whitfield, Kohrt, Gabriel, Rahbar, & Kohl, 2015) suggest that increased physical activity can result in an increase in bone mineral density (BMD) in common skeletal loading sites. In long-distance runners the calcaneus showed greater ($+17\%$, $p = 0.002$) BMD compared with sedentary controls (Fredericson et al., 2007), greater ($+3.1\%$) mineral content in 'consistent' (> 16 km/month) runners compared with a control group ($p < 0.05$) (Williams et al., 1984), and greater ($+12\%$) stiffness compared to sedentary counterparts (Lara et al., 2016). Greater ($+11.5\%$) calcaneus BMD was also reported in male runners (sprinters, middle distance and marathoners) when compared with athletes from low or no-impact disciplines; running was a significant ($p < 0.001$) determinant of BMD and independent of age and body weight (Laabes et al., 2008).

The repetitive high forces generated during running should theoretically increase foot bone density (Hart et al., 2017); Kersting and Bruggemann (1999) speculated that impact forces are constantly, and directly, regulating calcaneal bone adaptations. For example, Kemmler et al. (2006) compared high volume runners (> 75 km/week) with BMI-matched controls (≤ 2 h exercise/week) and reported that runners display a significantly higher calcaneal density. Similarly, in a large cross-sectional study involving marathon runners ($n = 401$; 217 men and 184 women) the rate of decline of BMD appeared to be reduced significantly in marathon runners compared with a normative group (Drysdale et al., 2007).

Overall, runners have higher calcaneus BMD than sedentary population; however, due to their continued practice the accelerated bone turnover (Harber et al., 1991) would inevitably decrease bone mass (Hetland, Haarbo, & Christiansen, 1993). For instance, Escamilla-Martinez et al. (2016) reported distance running to have a negative effect on calcaneal BMD during a 700-km training season in amateur runners ($n = 33$); similarly, Fuller et al. (2018) found no differences ($p \geq 0.319$) at the 20-week follow-up of a minimalist training intervention. Regular high volume of running may therefore decrease foot bone strength, increasing the risk of osteopenia and/or stress fracture.

4.3. Research limitations

The main limitations of the included studies are (i) the inconsistency on the dependent variable chosen as a proxy for foot muscles

strength, (ii) primarily only one site (the calcaneus) was chosen to investigate foot bone characteristics, (iii) the inconsistency on the methodology used to measure muscles and bone properties, and (iv) the incomplete information regarding the footwear, pattern of foot strike (heel vs. fore foot), physical activity background (training volume) of participants of the studies.

Experimental devices have been designed to measure foot muscles strength (Goldmann & Brüggemann, 2012; Senda et al., 1999); however, no device is able to distinguish between intrinsic and extrinsic muscles. Moreover, other biomechanical factors such as the moment arms of intrinsic foot muscles and muscle-tendon length may also influence the capacity of these muscles to generate force. An accurate measure of intrinsic foot muscles may provide valuable insight into their ability to produce force; however, such a technology still needs to be developed.

Although the calcaneus is considered an important peripheral site for osteoporosis assessment (Frost, Blake, & Fogelman, 2000; Glüer et al., 2004), prediction of the risk of hip fracture (Ross et al., 2000), and often used as a representation of skeletal status (Baroncelli, 2008; Langton & Langton, 2000), foot accounts for 26 bones with a unique shape that varies the magnitude and direction of the load they are subjected to. The choice of the calcaneus as an indicator of bone characteristics is questionable as this bone seems to be less affected by stress fractures than others. For example, the evidence indicates that sites of high risk stress fractures include the tarsal navicular, base of the fifth metatarsal, talus, base of the second metatarsal, sesamoids, and medial malleolus (Boden & Osbahr, 2000). While low-risk fractures in the foot and ankle include the calcaneus, and the second through fifth metatarsals (Boden, Osbahr, & Jimenez, 2001).

Moreover, bone density is only a proxy of bone strength that also depends on bone geometry, bone quality (metabolism and collagen cross-linking), cortical and trabecular morphology (Ammann & Rizzoli, 2003; Saito et al., 2010; Seeman, 2008). Only one study (Best et al., 2017) investigated trabecular characteristics using high resolution peripheral quantitative computed tomography – HR-pQCT; they found trabecular thickness to be positively correlated to weekly running distance ($r^2 = 0.417$, $p < 0.05$) and experience ($r^2 = 0.339$, $p < 0.05$). Clearly, more study of other foot bones and their specifics, other than density, may unveil new perspective on the effect of running on foot bones. Furthermore, bone density is not only influenced by mechanical external stresses (i.e. physical activity level), but also by age, diet, hormonal characteristics and genotype (Herbert et al., 2018), these internal physiological mechanisms together are suggested to explain around 50–85% of bone density; it is therefore important for future studies to consider those possible confounding variables when seeking to explain the effect of exercise (i.e. running) on bone density.

Finally, no standard protocols to investigate foot muscles and bones characteristics have been developed that would allow comparison between studies. These limitations could be addressed in future. Besides the comparison of runners and nonrunners, it would be interesting to compare foot anatomical characteristics in individuals with similar running experiences (i.e. weekly mileage and years of running) but different footwear choices. Despite the generalized perception that running is good for health, there are still questions that need to be answered: what is the impact of running on foot health? Do the shoes worn affect the potential benefits associated with running?

5. Conclusion

The present review systematically appraises the current level of knowledge on the effect of running on foot anatomical structures. Due to the moderate-quality and small sample size (and possible low statistical power) of the majority of the included studies, caution must be used when attempting to generalize their results to the wider population. The limited body of evidence suggests that running may increase foot muscles size and calcaneal BMD, but this seems to depend on training volume, running experience, and footwear.

The lack of details on the shoes worn by participants involved does not allow any inference on the contribution of footwear (and the associated biomechanical changes) on foot anatomical adaptations. It is evident that the role of footwear in ‘modelling’ the foot has not received enough attention and further experimental investigations are warranted. Future research should therefore, more closely, examine the links between running and foot musculoskeletal adaptations.

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Author contribution

The authors (AG and ST) conducted the search and coding process independently. AG performed methodological quality assessment, while ST checked the accuracy of the data. All authors were involved in drafting and reviewing the manuscript.

Additional information

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Technical note

Repeatability and accuracy of a foot muscle strength dynamometer

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ABSTRACT

Toe flexor strength is a pivotal biomechanical contributor for effecting balance and gait. However, there are limited reports that evaluate measurement accuracy and repeatability of this important attribute. Dynamometers are designed to measure force which can be used to derive joint torque if the perpendicular distance to the joint axis is known. However, an accurate and reliable measurement method to assess the ability of the toe flexor muscles to produce torque, is lacking. Here we describe a new device and method, designed to quantify the toe flexor torque developed at the metatarsal phalangeal joint. We evaluate measurement bias and the ability of the instrument to consistently measure what it is supposed to measure (Intraclass Correlation Coefficient). Results suggest that our device is an accurate tool for measuring angle and torque with a small (0.10° and 0.07 Nm, respectively) bias. When tested for reliability and repeatability in measuring toe flexor torque ($n = 10$), our device showed high interclass correlation (ICC = 0.99), small bias (-1.13 Nm) and small repeatability coefficient (CR = 3.9). We suggest mean bias and CR to be reported for future measurement methods and our protocol used as standard approach to measure maximal toe flexor torque.

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1. Introduction

Adequate foot muscle strength is imperative for efficient performance of sport and activities of daily living [1]. When we stand, foot muscles provide the basis for upright balance, but during locomotion the foot has a dual function: it forms a rigid lever at foot-strike and push-off, and a shock-absorber during mid-support [2]. This is accomplished through the deformation of the arch, which is controlled and supported by small intrinsic (foot) and large extrinsic (leg) muscles. Although critical to locomotion, our ability to measure and evaluate foot muscle strength accurately is rather limited [3,4].

Dynamometers are suggested to directly measure muscle force. They all rely on the assumption that (i) the external moment of force measured around the device axis represents the moment of the force produced by the muscles, and (ii) the force that produces

such moment is equal to the muscle force. For semantical precision, hereon we will refer to torque – external moment of force – when referring to what a dynamometer is measuring.

Previous toe dynamometers described in the literature have had technical limitations: some rely on the tester providing resistance [5], while others allow gripping of the toes and, therefore have a greater contribution from the extrinsic toe flexors [6]. An alternative is a fixed dynamometer whereby participants press their toes against a fixed sensor plate (i.e. force sensors) [7,8]. In this way, Endo, Ashton-Miller [9] used the signal from a force plate to quantify toe flexor torque around the metatarsophalangeal joint (MPJ); however, the movement was not isolated; the contribution of the moment generated among the other (bigger) joints was not accounted for. Goldmann and Brüggemann [10] introduced a system of Velcro® straps to fix the forefoot, midfoot, and rearfoot to the dynamometer while keeping the body into a standardized position. Although giving repeatable measurements, their device was not tested for accuracy and reliability. Based on the device built by Goldmann and Brüggemann [10], we developed a custom-made toe dynamometer addressing the technical limitations of previous studies while ensuring accurate measurements of torque produced by toe flexor muscles. The purpose of the present study was: (1) to assess the accuracy between the known measures for angle and

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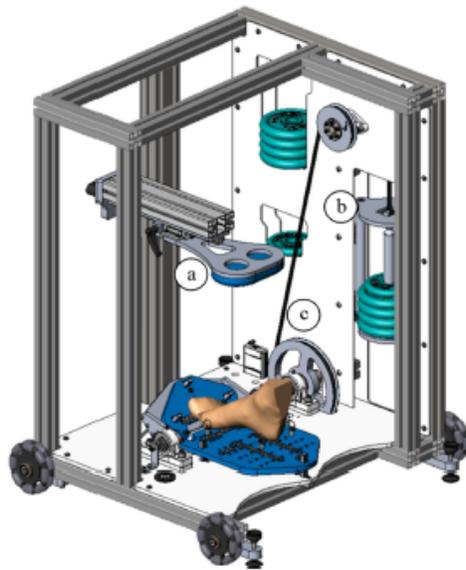


Fig. 1. Overview of the toe flexors strength device: a knee-thigh clamping mechanism, b carrier, and c pulley arrangement.

torque measured by the novel dynamometer device; and (2) to assess the device re-test repeatability of maximal isometric contractions of toe flexor muscles.

2. Methods

In this study, we quantified the moment of force generated by toe flexor muscles around the axis of the dynamometer during maximal isometric contraction. Our design addressed two important issues when assessing toe muscle strength: angular orientation of the metatarsal heads and foot size.

2.1. Hardware and software

The device is an improved version of a previously proposed machine [10] to which we added flexibility, and adaptability. It has been designed to allow measurements to be taken in either a seated or standing position. For operation in the seated position, a knee-thigh clamping mechanism is included, with both vertical and longitudinal adjustment features (Fig. 1(a)). The device can be set in a locked angular position to monitor a subject's ability to apply static torque, or can be set to allow free angular range of motion with adjustable mechanical limits. The height of the transverse axis of the MPJ is a function of foot size; therefore, we secured the plate on three adjustable screws with fixed rulers such that the plate position can be recorded and readjusted according to the participant's foot size. The angular orientation of the metatarsal heads also needed to be taken into consideration [11,12]. We designed a plate with a matrix of holes to which locking pins and straps can be tethered for strapping the subject's foot into different orientations. A requirement to provide the capacity

to impose and resist up to 50 Nm of torque has been met with the use of dumbbell weights loaded on to a carrier (Fig. 1(b)), and a pulley arrangement (Fig. 1(c)).

The tension (t_p) in the primary strap is the weight of the mass load. The tension in the secondary strap (t_s) is equivalent to the tension in the primary strap multiplied by the ratio of the primary (r_p) and secondary (r_s) pulley radii. The torque (T) imposed on the phalanges shaft is the product of the secondary strap tension and the driven pulley radius (r_d). The effective radius of each pulley is the sum of the radius of the pulley surface and half the thickness of the tension strap. The primary pulley effective radius was 0.100 m, the secondary pulley radius was 0.049 m, and the driven pulley radius r_d was 0.100 m; therefore:

$$\begin{aligned} T[\text{Nm}] &= m[\text{Kg}] * g * (r_p/r_s) * r_d \\ T &= m * 9.81 * (.100/.049) * .100 \\ T &= m * 2.002 \end{aligned} \quad (1)$$

The phalanges rotation shaft carries an absolute angle rotary encoder (Fig. 2(a)) on its end, which produces an analogue output voltage signal. The shaft assembly also includes a torsion strain cylinder element (Fig. 2(b)), which is connected to the assembly in such a way as to ensure that the link transmits torque without being exposed to any bending, tensile or compressive loads. The main foot and phalanges resting surface plates are designed and built to provide a large range of height adjustment so that any subject's proximal phalanges centre of rotation can be aligned with the device's rotation shaft. This allows simulation of a tilted MPJ mediolateral axis of rotation, through adjustment of jacking screws accordingly on both the main foot and phalanges tooling plates. The tarsal resting surface plate includes a matrix of holes to which locking pins and straps can be tethered for strapping the subject's foot into position. Both the main foot and phalanges resting plates include millimetre linear scales for foot positioning reference (Fig. 2(c)).

The electronic instrumentation comprises two transducers, their associated signal conditioning circuitry, and a custom Labview data acquisition system running on a laptop PC and employing an NI-6009 14-bit USB DAQ module to sample the 2 analogue quantities. An absolute angle encoder (US Digital MA3 with analogue output) is directly coupled to the shaft end of the toe plate and thus directly monitors the -20 to $+50^\circ$ angular range of the toe plate. This transducer has a resolution of 10 bits which equates to 0.33° measurement resolution.

A torque transducer and its associated amplifier monitors the torque applied by the toes to the toe plate. It covers a torque range of 0–50 Nm. The transducer was constructed in-house using a Micro-Measurements CEA-06-250US-350 full bridge strain gauge bonded to a custom designed hollow shaft and rated for 50 Nm full load. The associated strain gauge amplifier has a gain of 500 to provide an output voltage of approx. 4V at 50 Nm. Custom Labview code (National Instruments) samples the above 2 analogue channels at 100 Hz and applies the appropriate scaling factors and offsets to produce actual torque and angle values which are displayed in real-time (Fig. 3(a) and (b)).

2.2. Accuracy

Accuracy is intended here as the description of the systematic error (statistical bias) and random error (statistical variability) associated with a measurement [13]. In this study, limits of agreement (LoA) and mean bias were used as a measure of accuracy [14].

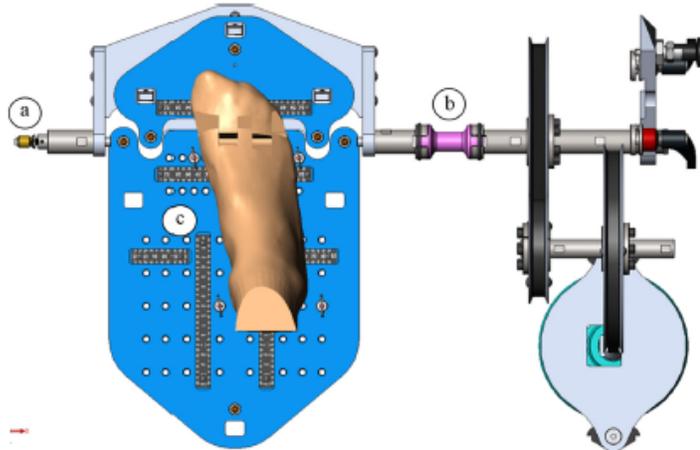


Fig. 2. Schematic of the main foot and phalanges plates. a rotary encoder, b torsion strain cylinder, and c millimetre linear scales.

Table 1

Accuracy results for the angle and torque measurements. Difference (Diff) between expected values and measured are reported; Absolute Average Difference (Abs Avg Diff) is also reported as raw and percentage. Mean Bias and limits of agreement (LoA) are also reported.

Angle (°)						
Expected	Measured mean \pm SD	Diff (%)	Abs Avg Diff (%)	Mean bias	LoA lower	upper
50	49.78 \pm 0.16	-0.22 (-0.44)	0.12 (0.81)	0.10	-0.11	0.31
40	40.06 \pm 0.17	0.06 (0.15)				
30	29.91 \pm 0.17	-0.09 (-0.30)				
20	19.83 \pm 0.16	-0.17 (-0.85)				
10	9.77 \pm 0.17	0.23 (2.30)				
0	0.03 \pm 0.16	0.03 (-)				
-10	-10.15 \pm 0.16	-0.15 (-1.50)				
-20	-20.03 \pm 0.17	-0.03 (-0.15)				
Torque (Nm)						
Expected	Measured mean \pm SD	Diff (%)	Abs Avg Diff (%)	Mean bias	LoA lower	upper
0	0.01 \pm 0.07	-0.01 (-)	0.16 (0.85)	-0.07	-0.47	0.32
2.93	2.93 \pm 0.06	0.00 (0)				
7.93	7.70 \pm 0.07	-0.23 (-2.90)				
12.93	12.76 \pm 0.07	-0.17 (-1.31)				
17.93	17.89 \pm 0.07	-0.04 (-0.22)				
22.93	22.98 \pm 0.07	0.05 (0.22)				
27.93	28.06 \pm 0.06	0.13 (0.47)				
32.93	33.24 \pm 0.07	0.31 (0.94)				
37.93	38.25 \pm 0.06	0.32 (0.84)				
42.93	43.27 \pm 0.07	0.34 (0.79)				

2.3. Angle

The predicted angle was compared to the software readings for that angle (i.e. plate fixed at 50° and record the angle). All angles from 50° dorsiflexion to 20° plantarflexion (in 10° increments) were tested. Results are reported in Table 1. For each angle, we computed the mean of 500-recorded values (10s).

2.4. Torque

Starting with zero weight, the weight of the carrier was added; then additional 2.5 kg calibrated weights were added. For each load, a 10s period was allocated before adding the next weight.

The expected torque was compared to the software readings for that weight. The frontal plate was kept in a neutral position and weights were added perpendicularly to it.

2.5. Statistical analysis

For each angle, 500 values were averaged and the standard deviation calculated. The same computational process was performed for the torque. The Bland-Altman plot [14] was used to visually inspect the differences between the computed theoretical values and the measured values (of both torque and angle); and how the differences might change in proportion to the magnitude of the measure. Limits of agreement [15] were used to assess differences

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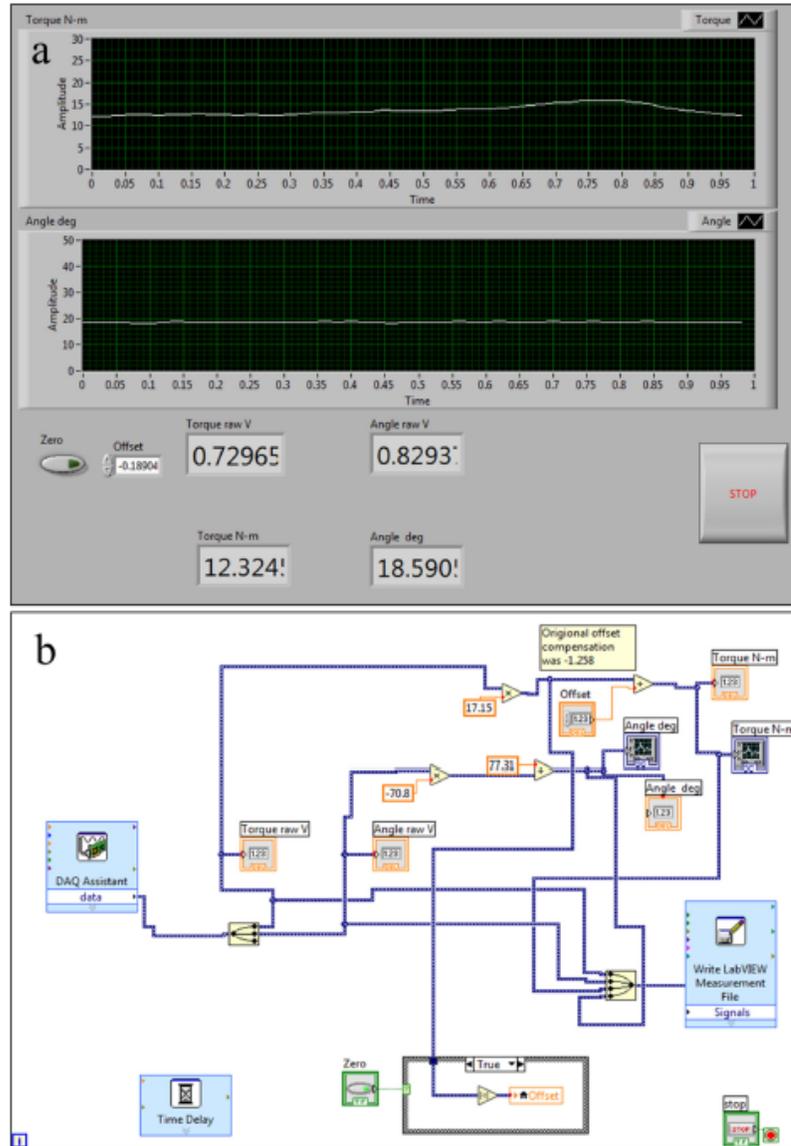


Fig. 3. Labview software interface (a) and block diagram (b).

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Table 2

Mean (\pm SD) torque produced by toe flexor muscles (in a 30° of dorsiflexion at the MPJ joint) for session one (test) and two (retest). Results reported for Interclass Correlation Coefficient (ICC), within-observation and between-observation variance (Nm^2), mean bias, and coefficient of repeatability (\pm CR).

	Test Mean \pm SD	Retest Mean \pm SD	ICC (95%CI)	Within Variance	Between Variance	Mean bias (\pm CR)
Torque (Nm)	18.75 \pm 9.2	19.88 \pm 10.5	0.99 (0.95-1.00)	3.96	92.28	-1.13(\pm 3.9)

between two types of evaluation methods: (1) device accuracy from concurrent tests, and (2) device repeatability from the same re-test conditions. The LoA provides an estimate that 95% of measured observations can be expected to lie within limits of agreement defined by the mean bias and coefficient of repeatability. Specifically, $\text{LoA}_{\text{between}} = \text{Mean difference}_{\text{between}} \pm \text{CR}_{\text{between}}$. For the accuracy test, the mean difference was defined by

$$\frac{\sum_{i=1}^{500} (x_e - x_m)}{500} \pm 95\% \text{ CI} \quad (2)$$

where x_e is the expected value and x_m is the measured value. The coefficient of repeatability ($\pm \text{CR}_{\text{between}}$) is computed by $\text{CR}_{\text{between}} = 1.96 \times \text{SD}_{\text{between}}$, where $\text{SD}_{\text{between}}$ is the standard deviation of the between method differences ($x_e - x_m$).

2.6. Repeatability and reliability

A study was conducted to establish the repeatability and reliability of the dynamometer in measuring the joint torques produced by the toe flexor muscles. Ten participants (7 men and 3 women, mean height 1.75 \pm 0.1 m; mean weight 74.9 \pm 15.5; mean BMI 24.3 \pm 3.2) gave their informed consent to undergo a familiarization and two testing sessions conducted on different (non-sequential) days.

Each participant reported to the laboratory at the same time of the day. The protocol consisted of a pre warm-up period of 1 min where the participants repeatedly performed toe flexion/extension movements with no resistance applied followed by submaximal isometric contractions with incremental exertion up to maximal contraction. After a 3-minute rest, three 5 s-maximal contractions were performed. Protocol design was such that learning effect was minimized, different ability to contract foot muscles accounted for, and maximal muscle pre-activation achieved.

Participants sat on a chair with their knee and ankle fixed at 90°. Metatarsal-phalangeal joints (MPJs) were fixed at 30° of dorsiflexion as recommended for optimal torque production [16] and secured to the bottom plate through a means of Velcro® straps. The head of the metatarsals (1–5) were in line with the transverse axis of the device. Raw data were filtered using a 101-point (2 s) moving average. The highest torque value among the trials (1–3) was used for analysis.

2.7. Statistical analysis

For repeatability, mean and standard deviation of the differences between the two sessions were used to calculate the limits of agreement using the Bland-Altman plot as described previously. The coefficient of repeatability and mean bias were also computed. For reliability, a two way mixed single measures (absolute agreement) was used to calculate Interclass Correlation Coefficients (ICC; 3,1). All statistics were run in SPSS (Version 24, SPSS Inc., Chicago, IL). The level of significance was set to $\alpha = 0.01$.

3. Results

3.1. Accuracy

Results from the accuracy study are showed in Table 1 (and Appendix A). For angle, the largest difference between expected and measured values (0.23°) was at 10° dorsiflexion, while the lowest error (0.03°) was recorded at 0 and 20° plantarflexion. Overall, the absolute mean difference was 0.12° and the absolute percentage difference was 0.81%. For torque, the highest difference between expected and measured values (0.34 Nm) was recorded at the highest load (42.93 Nm), while the highest percentage difference (2.9%) was recorded at 7.93 Nm expected torque. Overall, the absolute average difference was 0.16 Nm with an absolute percentage difference of 0.85%. Mean bias of measurement for torque was -0.07 Nm with a CR of 0.39 Nm. For the angle, the mean bias was 0.10° with a CR of 0.21° (Appendix A).

3.2. Repeatability and reliability

Results from the repeatability test are reported in Table 2 (and Appendix A). The two testing sessions were not significantly different ($t(9) = -2.11$, $p = 0.64$) with a mean bias of -1.13 ± 3.9 Nm.

The average measures interclass correlation coefficient was excellent (ICC=0.99); with 95% of the samples having confidence intervals (CI) between 0.95 and 1.00 which shows high reliability. The within-observation variance was also found to be low (3.96 (Nm^2)) with a between-observation variance of 92.28 (Nm^2).

4. Discussion

In this study, we tested the accuracy, repeatability and reliability of a method to test toe flexor strength. Results suggest that our bespoke dynamometer is an accurate tool for measuring angular position and torque: mean bias for torque measurements (-0.07 Nm) and for angular position measurements (0.1°) were less than a unit; the CR for torque (0.39) and for angle (0.21) were also small. Therefore, our device is not only accurate, but it has a small instrument error (noise in the measuring device).

When tested for between-session repeatability and reliability in measuring toe flexor strength, our device showed low bias (-1.13 ± 3.9) confirming its repeatability, and high interclass correlation coefficient (ICC=0.99) confirming its reliability. Although torque measurements in the second session were generally higher than in the first, the not significant ($p = 0.41$) difference ($+1.13$ Nm or +6%), gives confidence on the accuracy of the number of sessions (one familiarization and two tests) and the warm-up protocol defined, to minimizing any learning effect.

It has been reported that measurement of torque is affected by many technical factors, such as the applied methodology [17], and joint orientation [10]. Here we propose an accurate and reliable standardized methodology - with an improved design - compared with previous devices [10,18]. The first metatarsal bone has a higher (from ground level) effective centre of rotation than

the smaller toe bones, therefore the effective axis of all phalanges working together is tilted relative to the ground plane. We included an additional degree of freedom to account for the mediolateral slope of the effective rotational axis of the phalanges.

Our study is the first to propose an estimate of instrument repeatability (Limits of Agreement) when performing toe flexor strength tests by dynamometer. The importance in reporting the degree of measurement accuracy is well-documented [19–21]. Poor accuracy reduces the ability to monitor changes over time – both in clinical and experimental contexts; studies not reporting the amount of bias inherent in the measurement may over- or under-estimate the true moment of force produced, therefore their results need to be interpreted with caution.

Our device also has the potential to be used as a training tool, instead of just for evaluation. Strengthening of the foot muscles is commonly achieved with toe-flexion exercises such as towel crunches or marble pickups [22,23], short-foot exercises that involve drawing the heads of the metatarsals toward the calcaneus without curling the toes [24], or exercises performed using exercise bands with progressive resistance [25]. However, in those exercises the extrinsic foot muscles are activated to some extent, the resistance applied is difficult to quantify exactly, and the efficiency of the training is dependent on the position held by the performer. Our device could potentially be a more effective method to reinforce foot muscles and it could simplify the training plan by setting a constant individualized position, and by setting specific resistive progression while minimizing the contribution of extrinsic foot muscles.

Although the device was accurate in measuring torque and angle, and showed a small measurement bias, it is not possible to confidently assume that the device is able to isolate toe muscles and measure only their strength. The set-up of the machine was such that muscles not crossing the MPJ should have had a small (if any) effect on torque production around that joint, however, this is not certain. It is also acknowledged that during a maximal isometric contraction the extrinsic muscles help in stabilizing the adjacent foot joints therefore, they may have an indirect role in force production. In future, concurrent use of motion capture system, electromyography, and/or foot plantar pressure devices with dynamometers will better define if any secondary movements (i.e. imperceptible heel raising) play a role in the development of torque around the MPJ.

5. Conclusion

This study evaluated the performance of a bespoke dynamometer, which had been designed to measure maximal toe flexor strength. The results indicate that the device is accurate when measuring torque and flexion angle, and repeatable and reliable when measuring maximal joint torque developed by toe flexor muscles. In future studies, the ability of the device to reliably discriminate between different groups of people (i.e. different gender or sport) should be tested in a larger sample.

Acknowledgment

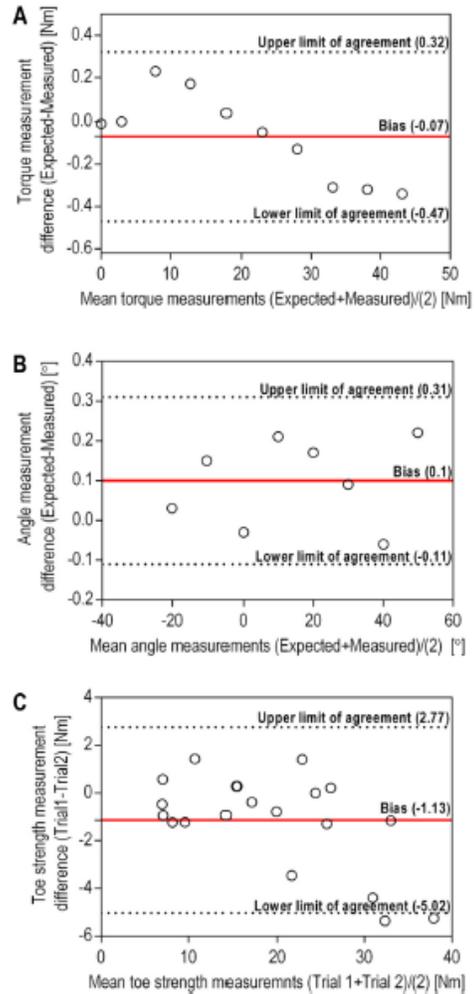
The University's Human Research Ethics Committee (ref 24315) approved this study. The authors report no conflicts of interest.

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Appendix A. Bland-Altman plots for torque (A), angle (B), and toe strength test (C)



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Evaluating dynamic error of a treadmill and the effect on measured kinetic gait parameters: Implications and possible solutions



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ABSTRACT

The dynamic properties of instrumented treadmills influence the force measurement of the embedded force platform. We investigated these properties using a frequency response function, which evaluates the ratio between the measured and applied forces in the frequency domain. For comparison, the procedure was also performed on the gold-standard ground-embedded force platform. A predictive model of the systematic error of both types of force platform was then developed and tested against different input signals that represent three types of running patterns. Results show that the treadmill structure distorts the measured force signal. We then modified this structure with a simple stiffening frame in an attempt to reduce measurement error. Consequently, the overall absolute error was reduced (–22%), and the error in force-derived metrics was also sufficiently reduced: –68% for average loading rate error and –80% for impact peak error. Our procedure shows how to measure, predict, and reduce systematic dynamic error associated with treadmill-installed force platforms. We suggest this procedure should be implemented to appraise data quality, and frequency response function values should be included in research reports.

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1. Introduction

Force platforms are an essential measurement device in many biomechanical studies, from which kinetic parameters are derived to evaluate gait. As an adjunct to the common ground-installed force platform sensor (G_{RS}), the treadmill-installed force platform sensor (T_{RS}) is becoming popular in gait research laboratories (Dierick et al., 2004; Riley et al., 2008, 2007). Given that kinetic parameters depend on accurate force signal measurements (Pàmies-Vilà et al., 2012; Silva and Ambrósio, 2004), data quality and research integrity relies upon the known degree of measurement error associated with these force-instrumented treadmills. The precision of a force measurement device is dependent upon the inherent natural frequency of its structure. Depending on the mass and stiffness of a treadmill structure, and on the force sensor size (Dierick et al., 2004), treadmill dynamic behavior may generate mechanical vibrations and mode shapes at specific frequencies (natural frequencies) that could approach the frequency content of applied forces from human gait and create artefacts in the measurements. While the ground-installed force platforms have natural frequencies much higher than the frequency content of the

exerted force (Antonsson and Mann, 1985), the natural frequencies of the treadmill installed platforms have been reported to be as low as 16 Hz in some cases (Draper, 2000) that is within the frequency content of normal gait (reported as 35–50 Hz (Antonsson and Mann, 1985; Blackmore et al., 2016)), affecting the accuracy of the measured force by the strain gauges (force sensors) (Willems and Gosseye, 2013). Nowadays, there is a rise in research that uses parameters derived by treadmill-installed force platforms data for training and retraining (rehabilitative) interventions, in both sport (Crowell and Davis, 2011) and clinical settings (Van den Noort et al., 2015), as well as for development of new technologies (Mooney and Herr, 2016). Although accurate measurement of force data is paramount, it is not common practice to include an independent report on the frequency response and the expected measurement error of the forces.

The error inherent within force measurement is best detected and evaluated from frequency domain analysis (Gruber et al., 2014, 2011). Therefore, this study will evaluate the Ground Reaction Force signal (GRF) in the frequency domain and describe its harmonic contents, as per (White et al., 2005). The inherent error in the GRF created by the natural frequency of the treadmill is not a random noise that may disappear by taking the average or integration of measured signals across gait cycles. Instead, this error is systematic; it has the same effect on each measurement

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episode. Bias created by the natural frequency is not related to the magnitude of signal noise that can be overcome by smoothing process that produces a best-fit line (De Bièvre, 2009) but it is related to the degree of difference between the measured and smoothed signal and the true signal (Menditto et al., 2007). Therefore, bias is an essential feature to consider when comparing measurements obtained across different force platform systems.

At the authors best knowledge, only one study included the issue of natural frequency testing on instrumented treadmills (Sloot et al., 2015). They presented a new approach to test the performance of treadmills, assessing the accuracy of forces and center of pressure, including assessment of the natural frequency. However, they did not explore the effect of low natural frequencies on force signals, nor propose any solution to improve treadmill performance. Our study continues upon this theme by outlining a standardized method to evaluate natural frequencies and their effect on measurement bias. The three aims of this study were: (i) to evaluate measurement bias (systematic error) of an instrumented treadmill using a test for frequency-dependent behavior of a force platform; (ii) to develop and evaluate a model that is designed to predict measurement bias of the force platform frequency response; and (iii) to reduce measurement bias of an instrumented treadmill.

2. Methods

The aims were addressed in three stages. Stage 1 assessed the dynamic behavior of the instrumented treadmill using Frequency Response Function (FRF) (Rao and Yap, 2011). This was achieved by evaluating the signal frequency ratio between two interacting force measurement devices. We used a hammer installed force sensor (H_{FS}) to apply an impact force to a treadmill-installed force platform sensor (T_{FS}), and to a ground-installed force platform sensor (G_{FS}). Stage 2 evaluated a model that was developed to predict the dynamic behavior of the treadmill (refer to (Rao and Yap, 2011) for more details on the mathematical procedure used to develop the model). Stage 3 assessed a solution to improve the dynamic behavior of T_{FS} by altering the support structure of the treadmill. We then assessed the dynamic behavior of the new T_{FS} using the predictive model.

2.1. Stage 1

2.1.1. Analysis of treadmill frequency response

The Fourier transform represents any signal – such as the force signal – as a sum of periodic waveforms (e.g. sine functions). Each waveform is characterized by a frequency (ω), an amplitude (A)

and a phase (ϕ). This allows investigation of how the signal's amplitude and phase vary for any given frequency. The systematic error of the force platforms (T_{FS} or G_{FS}) can be represented in the frequency domain using a FRF. The FRF is a frequency dependent modulation system that alters the frequency properties of the input signal (Fig. 1). For example, the amplitude (A_i) and phase (ϕ_i) of the input signal pass through the modulation function, where the signal is transformed into an output signal with new amplitude (A_o) and phase (ϕ_o).

The computed FRF can predict how the output signal of T_{FS} (or G_{FS}) diverges from the input signal by comparing the amplitude (A_i) and phase (ϕ_i) of the H_{FS} (input), with the amplitude (A_o) and the phase (ϕ_o) of the output signal (T_{FS} or G_{FS}) at each frequency. The output signal is described at each frequency by Eq. (1):

$$(A_i(j\omega) \angle \phi_i(j\omega))(A_{FRF}(j\omega) \angle \phi_{FRF}(j\omega)) = A_o(j\omega) \angle \phi_o(j\omega) \tag{1}$$

where ω is $2\pi f$, and f is frequency in Hz. The input signal ($A_i \angle \phi_i$) is multiplied by the modulation system ($A_{FRF} \angle \phi_{FRF}$). This can be rewritten in terms of the modulation system as:

$$A_{FRF}(j\omega) \angle \phi_{FRF}(j\omega) = \frac{A_o(j\omega) \angle \phi_o(j\omega)}{A_i(j\omega) \angle \phi_i(j\omega)} \tag{2}$$

Now, it is possible to look at how the system (FRF) reacts for each frequency of the input signal using the following transfer function estimator:

$$FRF(\omega) = \frac{FP(\omega)}{H(\omega)} \tag{3}$$

where $FP(\omega)$ is the Fourier transform of the force platform signal and $H(\omega)$ is the Fourier transform of the hammer signal. The change in amplitude and phase caused by the modulation system can then be represented as:

$$A_{FRF}(\omega) = |FRF(\omega)| \tag{4}$$

$$\phi_{FRF}(\omega) = \angle FRF(\omega) \tag{4i}$$

where A_{FRF} defines how the system affects the amplitude of the input signal (in absolute terms) for any given frequency, and ϕ_{FRF} defines how the system affects the phase of the input signal for any given frequency.

2.1.2. Measurement

The H_{FS} was composed of a high precision force sensor (PCB Piezotronics, 218A) fixed on the head of a modified hammer, so-called impact hammer. The G_{FS} were embedded into a ground-installed force platform (BP600900TT, AMTI, USA). The T_{FS} were embedded into a treadmill-installed force platform (DBCEEWI,

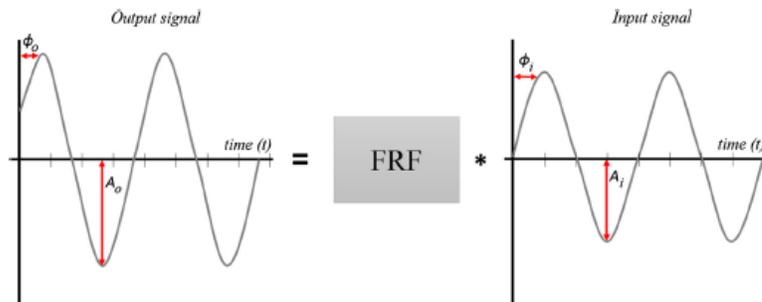


Fig. 1. Response of a linear time-invariant system to a sinusoidal input (right). The steady state output (left) depends on the characteristics of the system (FRF).

AMTI, USA). The impact hammer has been calibrated using a known mass and accelerometer (Waltham and Kotlicki, 2009) and connected to a 2 channel charge amplifier (Rion, UV-16). The devices were synchronized using Nexus data acquisition system (Oxford Metrics Ltd, Oxford, UK) at a sample frequency of 2000 Hz. The H_{FS} has a flat response up to 1000 Hz (Appendix A), therefore it provides an accurate measure of the force applied to the platforms. The ratio between the output from platform force sensors and the H_{FS} shows how the measurement is affected by the dynamic behavior of the system. When the response is 1 N/N, it means that the force measured by both instruments perfectly match.

Using the hammer we generated a set of 20 vertical impacts at five locations on each platform (four corners and the platform center). The average magnitude of the impacts was 100.2 ± 39.7 N, which is the linear range of the force platform (0–8800 N) meaning that the measured FRF is valid for any force below 8800 N. The FRF linearity was validated with a coherence function which was above 0.90 between 5 and 200 Hz (Randall, 2008). Data were exported to Matlab (Math Works Inc., USA) for FRF analysis, averaging the 20 impacts to achieve adequate coherence function between 0 and 100 Hz. In order to evaluate the dynamic behavior of the treadmill, the FRF was computed from the force signals of force platforms and hammer using the so-called H1 estimator (Rocklin et al., 1985), which reduces the effect of the measurement noise in the force platforms signal, therefore:

$$FRF(\omega) = \frac{P_{FH}}{P_{HH}} \quad (5)$$

where P_{FH} is the cross-spectrum between the force platform and the hammer signals, and P_{HH} is the auto-spectrum of the H_{FS} signal (Randall, 2008). Amplitude and phase were then evaluated to investigate the occurrence of the first mode of vibration (i.e. natural frequency).

2.2. Stage 2

2.2.1. Predictive model

The FRF of the measurement devices (e.g. force platform on the treadmill) represents, in the frequency domain, how a force measurement is distorted at every frequency by the dynamic behavior of the measurement device (e.g. natural frequency of the structure). An ideal measurement device would have a flat FRF throughout its frequency range which means that there would be no amplification nor delay between the real input (e.g. applied force) and reading (e.g. measured force).

Effect of the amplification and delay on the measurement can be assessed in the time domain using a predictive model. To do so, the first step was to transform the FRF into the time domain using the inverse Fast Fourier transform (Randall, 2008). The transformed FRF is known as the Impulse Response Function (IRF). The

reading on the measurement device, $y_{sp}(t)$, in response to a certain input, $x(t)$, can be predicted by convolving the IRF with x :

$$y_{sp}(t) = IRF(t) * x(t) \triangleq \int IRF(\tau)x(t - \tau)d\tau \quad (6)$$

where τ is a time lag integration variable.

The accuracy of the treadmill and ground-installed force-platforms measurements can be assessed by comparing the predicted response of both measurement devices for different inputs. We selected three archetypal signals that represent the vertical component of typical ground reaction force vectors (VGRF) generated by humans when running (data collected in a previous experiment). These archetypes had distinct impact transients associated with low, medium, and high loading (Fig. 2).

2.3. Stage 3

2.3.1. Application and evaluation of a stiffening frame

The treadmill-installed force platforms are supported by a framework structure of steel beams (Fig. 3). The rectangular shape of the treadmill frame lays upon four feet posted at the corners. To stiffen the long axis of the frame and increase the natural frequency, we positioned two wooden support bearers under each long side of the treadmill frame (Fig. 3, appendix B). To evaluate the bias of the new system, TWFS response was modelled and tested using the three archetypal signals as input. Bias is reported as root mean squared error (RMSE). The natural frequency didn't shift between tests and the coherence function was close to one, which suggests that the supports behave linearly throughout all the tests.

3. Results

3.1. Treadmill frequency response

Fig. 4 presents the amplitude (a) and phase shift (b) features of the FRFs produced from the hammer test on the three measurement systems: G_{FS} , T_{FS} , and TW_{FS} .

For the amplitude, a FRF < 1 implies there is an underestimation of the signal at that frequency, whereas a FRF > 1 implies that there is an overestimation at that frequency. For instance, at 30 Hz the ratio between the applied force and the measured one is 1.6, which means the measured force at 30 Hz is 37% greater than what it is in reality (i.e. the force applied by the hammer). At 32 Hz there is a 10% increase with respect to 30 Hz. Thus, between 32 ms and 33 ms of the loading phase, the measured signal will show a 10% increase in the first peak force that does not exist in reality. At 40 Hz (ratio 0.68) the measurement by the T_{FS} will underestimate the force by 47%.

The T_{FS} FRF presents two peaks at 32 Hz and 55 Hz; whereas the G_{FS} shows the relatively flat response that is expected from a gold-standard force measurement device (Fig. 4a). After applying woo-

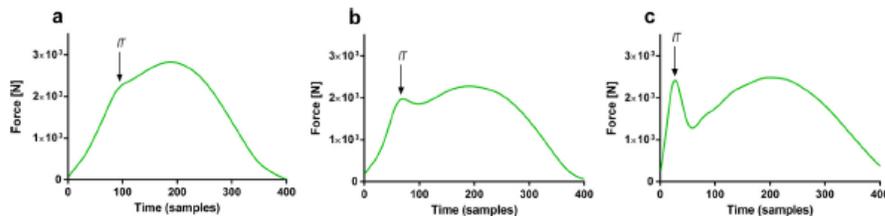


Fig. 2. GRF archetypal signals with different impact transient properties. The intensity of the loading is low (a), moderate (b) and high (c); IT indicates the Impact Transient.

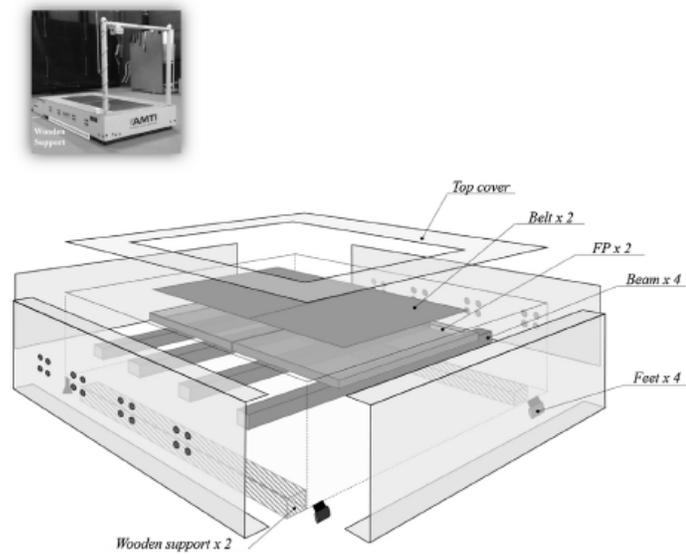


Fig. 3. Structural components of the instrumented treadmill. Wooden supports were added underneath the lateral sides of the treadmill frame to improve overall stiffness of the device. Treadmill was resting on the wooden supports instead of on the four legs during the experiment.*

den bearers to the treadmill, the first natural frequency shifted from 32 to 36 Hz. For the phase, T_{FS} shows two main shifts at the two natural frequencies (32 and 55 Hz) and TW_{FS} has also a phase shift in correspondence of its first natural frequency (36 Hz). In contrast, the G_{FS} shows no phase shift among the analyzed frequencies.

3.2. Effect of improved treadmill stiffness

Table 1 lists the level of agreement between the three archetypal signals and the model-predicted VGRF signals derived from the FRF. The degree of overlap between the measured and archetypal signals for the three different types of impact intensity and force sensor type is shown in Fig. 5. The measurement error of the G_{FS} increases as loading intensity increases while, the lowest error for the T_{FS} was at Medium load (52.5 N) and the highest value was at High loading (127.8 N), representing a 243% relative increase. TW_{FS} follows a similar trend to T_{FS} . The largest difference between T_{FS} and TW_{FS} was in High loading condition with a reduction in RMSE of 48%. Overall the TW_{FS} displays less error (-22%) compared to the T_{FS} . The modified frame reduced the error in the variables related to the impact transient, such as average loading rate (ALR) and impact peak. The TW_{FS} exhibits an error 3-times lower in the ALR (a reduction of 68 percentage points), and an error 5-times lower in the impact peak (a reduction of 80 percentage points; see Table 1).

Fig. 5(a)–(c) shows the three archetypal signals (a – low; b – medium; c – high) compared against the predicted force reading for the G_{FS} , T_{FS} and TW_{FS} . Fig. 5(d)–(f) represents the raw error for each condition. Main error for the T_{FS} is in the first half of stance at high loading with an evident oscillatory behavior that decays over time. TW_{FS} consistently overestimates the force measurement in early stance and underestimates it from mid stance forward. G_{FS} almost perfectly measures force applied in any loading condition.

4. Discussion

The general aim of this study was to evaluate the force measurement bias from a typical T_{FS} by comparing it against a 'gold standard' G_{FS} . The force reading from the G_{FS} is precise across a range of analyzed frequencies (1–100 Hz), whilst the signal from the T_{FS} has some measurement bias. Any applied force to the T_{FS} that is above 10 Hz will either over- or under-estimate the true magnitude of the applied force and this measurement error will depend on the frequency content of the applied force.

The measurement error of the treadmill followed a different trend compared to the ground-installed force platform. While the G_{FS} showed a consistent increase with the loading intensity, the T_{FS} was inconsistent between these three archetypal signals. This is explained by the number and position of the treadmill's natural frequencies. The G_{FS} has a very high first natural frequency (>500 Hz), while the treadmill has two natural frequencies at approximately 32 and 55 Hz. Therefore, as the frequency content of the applied force increases with increased loading intensity, it is adjacent to the first natural frequency at Low loading, it sits between the two natural frequencies at Medium loading and it is adjacent to the second natural frequency at High loading. As the application of wood support bearers does not eliminate the natural frequencies, the trend is similar for the TW_{FS} .

The first natural frequency of the treadmill was identified at 59 Hz prior to shipping (Appendix C). This suggests that the measured first natural frequency (32 Hz) was either not identified by the manufacturer, or the testing conditions were different. For instance, the soft elastic floor covering the ground (Mondo®) in our laboratory creates a compliant substrate of the treadmill-floor interface, which may have changed modes in the frequency bandwidth of interest. To further investigate the reasons for these discrepancies, a full modal analysis of the treadmill including several degree of freedom must be performed in different laboratory

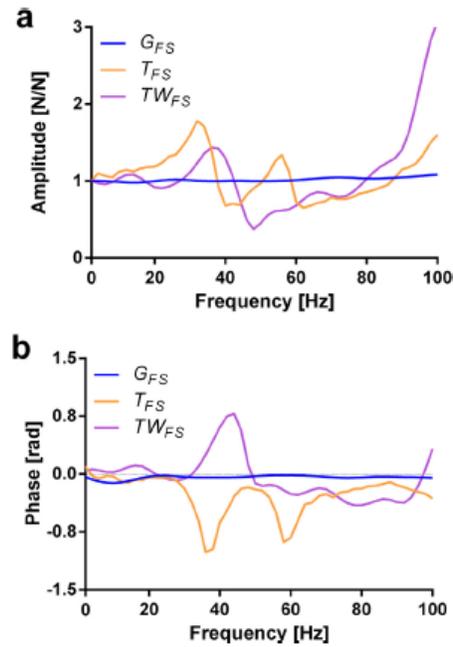


Fig. 4. Frequency Response Function test displayed in the Amplitude (a) and phase (b) domain. FRF outcomes of the three hammer tests are over-ground sensor (G_{FS} , blue), treadmill sensor (T_{FS} , orange), and treadmill with wood sensor (TW_{FS} , purple). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

environments (e.g. floor structure, and mounting conditions). This type of systematic study would highlight how the dynamic behaviors of the system depend on its boundary conditions and establish general guideline for instrumented-treadmill installation.

The position where the measurements are made could also affect the number of natural frequencies appearing in the frequency response function. If the excitation or the measurement has been made on a 'node' of a mode shape, the natural frequency of this mode doesn't appear on the FRF. As the tests presented in this paper were conducted at the point where the runner most commonly hits the platforms (i.e. its center), we ensured that all the relevant natural frequencies were measured. After modelling the FRF for the G_{FS} , T_{FS} and the adapted TW_{FS} , we then compared their output force measurement with archetypal signals. While the G_{FS} seems to be more consistent in measurement error between loading intensities, the T_{FS} behaves differently depending on the type of VGRF profiles (Fig. 5): it may be the case that the frequency content of the input signal is actually increasing as the loading profile of the VGRF increases. VGRF with high loading profile has a frequency content close to a resonance frequency of the treadmill, therefore the measured force signal is amplified. Instead, when the VGRF curve becomes smoother the frequency content changes – reduce – moving away from a resonance frequency; as a result, the signal is minimally amplified due to the structural damping.

Due to the low natural frequencies of the treadmill, the T_{FS} VGRF profile degenerates, leading to errors in measures of gait particulars associated with the impact transient (Table 1). For instance, the recorded signals by the T_{FS} show that there can be errors in impact transient parameters of up to 12%. Accurate measurement of impact transient parameters is important for clinical evaluation of running performance and risk of injuries (Davis et al., 2004; Milner et al., 2006). Moreover, results from running retraining studies (Crowell et al., 2010) aiming to reduce the impact transient may be affected by the dynamic behavior of the instrumented treadmill. The measurement bias could be either systematic or random – because it is dependent upon frequency; hence if a person applies different load intensities the observed

Table 1

Root mean squared error (RMSE) is reported as a measure of bias. The error of over-ground force platform sensor (G_{FS}), treadmill-installed force platform sensor (T_{FS}), and adapted treadmill (TW_{FS}) are reported for low loading (Low), medium loading (Med) and high loading profiles (High). The average (AVG) is also reported. RMSE is reported as raw values [N], percentage of peak force, and percentage of mean force. Average loading rate (ALR) and Impact peak are reported as percentage change from the archetypal VGRF signals. ALR was computed between 20 and 90% of impact peak.

	Loading pattern			Avg
	Low	Med	High	
RMSE [N]				
G_{FS}	3.9	7.0	8.4	6.4
T_{FS}	56.7	52.5	127.8	79.0
TW_{FS}	68.4	54.9	60.7	61.3
RMSE % peak force				
G_{FS}	0.1	0.3	0.3	0.3
T_{FS}	2.0	2.3	5.2	3.2
TW_{FS}	2.4	2.4	2.4	2.4
RMSE % mean force				
G_{FS}	0.2	0.5	0.5	0.4
T_{FS}	3.5	3.5	7.2	4.7
TW_{FS}	4.2	3.6	3.4	3.7
ALR (%)				
G_{FS}	-2.0	-3.8	-1.3	2.4
T_{FS}	1.8	12.3	3.7	5.9
TW_{FS}	-1.5	3.4	0.8	1.9
Impact peak (%)				
G_{FS}	-0.4	0.0	0.4	0.3
T_{FS}	4.1	4.8	9.2	6
TW_{FS}	1.1	1.3	1.1	1.2

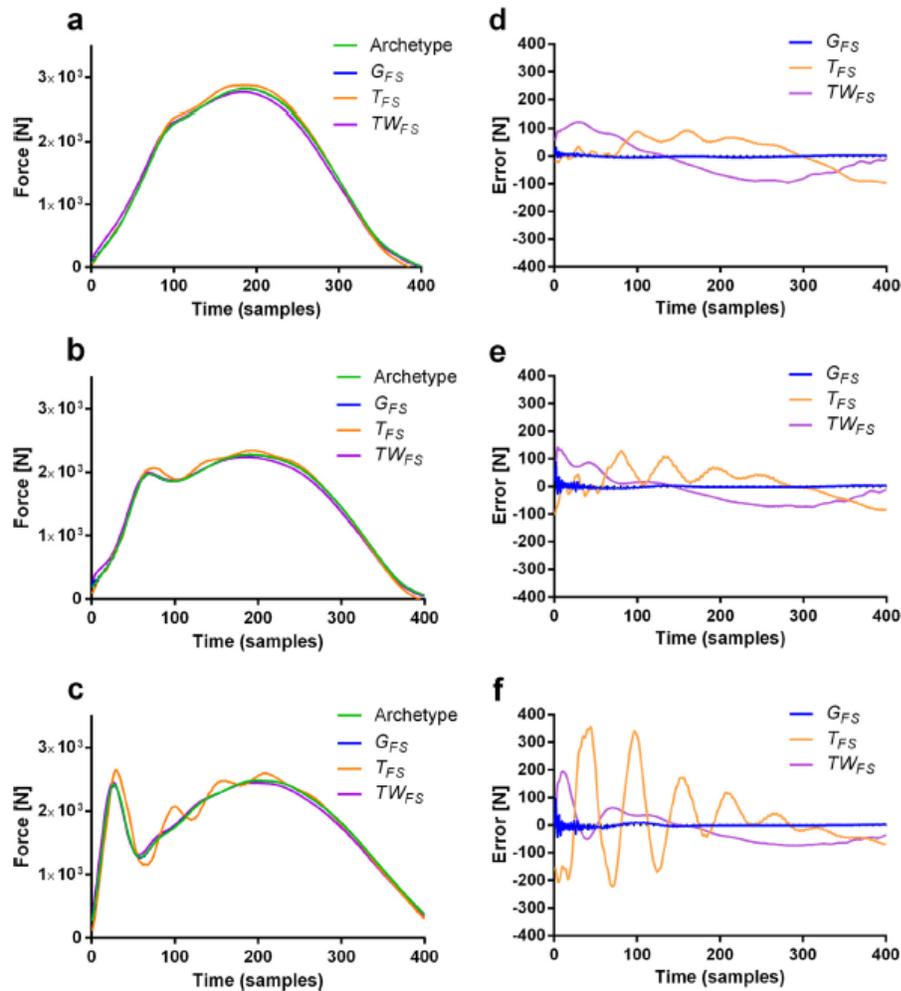


Fig. 5. Archetypal VGRF signals from over-ground running with low loading (a), medium loading (b), and high loading (c). Archetypal VGRF signal (green) is compared against over-ground model-prediction (G_{FS} , blue), treadmill model-prediction (T_{FS} , orange), and new treadmill configuration (with wood bearers) model-prediction (TW_{FS} , purple). Error for each model is reported for low loading (d), medium loading (e), and high loading (f). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

error could vary (under/over) between foot contacts within a trial. Therefore, pre-post intervention differences may be partially contributed by the bias associated with the dynamic (vibratory) behavior of the treadmill. For many future studies using instrumented treadmills, researchers could evaluate the confidence they have in their data by using the FRF and IRF method. Indeed this is performed by manufacturers prior to shipping, however, this evaluation also needs to be conducted in the lab setting.

It is worth noticing that measurement errors – related to the dynamic behavior of the treadmill – will pass undetected when error evaluation techniques are employed with conventional static calibrations (Gill and O'Connor, 1997; Hsieh et al., 2011). The

results from the dynamic validation method performed in this study demonstrates the effect that a T_{FS} can have on the data quality within a biomechanics lab, and raises the necessity to include such an evaluation procedure as regular practice prior to the reporting of data. The evaluation of the modified TW_{FS} is indicative of why a T_{FS} should be tested in its specific environment and condition. The application of supports underneath the body of the treadmill showed an overall improvement of the ratio between input (hammer) and output (force platform), reducing the measurement error of the VGRF. Although the natural frequency has been increased slightly (from 32 Hz to 36 Hz), the reduction of the error is remarkable. For instance, at 30 Hz the ratio decreased

from 1.60 to 1.15, reducing the 37% artificial increase in force recording to just 13%. When comparing the amount of measurement bias (RMSE) and the change in loading variables across the different loading conditions, the modified TW_{FS} shows a smaller average error (Table 1). Although a benchmark of an acceptable error limit will vary according to derived parameters, we can consider a level of error equivalent to that of the ground embedded force platform as the gold standard benchmark. Achieving this will require improvement in two areas: (i) mathematical models of the frequency response, and (ii) engineering a stiffening frame comparable to a ground embedded force platform. A mathematical model will minimize the effect of systematic error; while an improved frame structure will increase resonance frequency and provide a more reliable measurement of high frequency forces.

Indeed, the effect of systematic artifact will have a greater impact on certain users and their analyses, while others might find these levels acceptable. For example, the ground reaction force orientation may be sufficiently altered to affect joint kinetic parameters, particularly the hip joint moments (where a combination of both kinematic and kinetic errors would exist). In another context, the appeal of using instrumented treadmills is that they accommodate analyses that require long continuous data sets. However, analyses that quantify time-series behavior of gait parameters (e.g. (Dingwell et al., 2010; Hausdorff et al., 1996)) should be cautious when considering similar analyses on gait parameters measured from instrumented treadmills, particularly impact transient.

An alternative method to avoid sensor natural frequency related error is to use a digital low-pass filter. Commonly, in running studies, force signals are low-pass filtered with a cut-off frequency of 50 Hz (Baggaley et al., 2017; Cheung and Rainbow, 2014; Kulmala et al., 2013) with some using 100 Hz (Hobara et al., 2012). As the frequency content of the force signal recorded during running can reach frequencies up to 50 Hz (Blackmore et al., 2016; Shorten and Mientjes, 2011), any cut-off frequency lower than 50 Hz will necessarily delete part of the true signal. In our case, as the first natural frequency started affecting the signal at 10 Hz, a lower cut-off frequency (i.e. 6 Hz) would be needed to remove the amplification effect caused by the treadmill dynamic behavior, however, it will also smooth every sharp change in the signal (i.e. rising portion of the GRFv). Therefore, when applying a low-pass filter to the force signal, the user should appreciate the effect of three influential factors: (1) the natural frequency of the treadmill; (2) the typical frequency content of the force signal being recorded (i.e. influence of different types of impact); and (3) the type of bias that the treadmill's dynamic behavior has on the force signal. In this study we showed how to address those issues with a rather simple test. Results will give confidence not only on the validity of the force signal, but also on the adequacy of low-pass filter cut-off frequency.

The main limitation of this study is the generalizability of our results. As the laboratory environment affects the natural frequency, the error found and solution proposed is only applicable to our treadmill. However, with this study we highlight the need of ensuring appropriate system quality check and report of measurement associated error which should be a priority for any biomechanical laboratory. Although our method was able to raise the natural frequency of the treadmill, it improved force reading accuracy without suppressing the bias. However, the procedure presented highlights that an evaluation of T_{FS} measurements performed in the frequency domain provide sensitive characteristics of the force signal that can expose any presence of systematic error – this form of measurement error would otherwise be undetected through time domain procedures. Such an evaluation should always be performed *in situ*, that is, in the specific environment and condition in which the treadmill is used, and results should accompany any reported data for quality assurance.

Conflict of interest statement

The authors have no personal financial conflict of interests related to this study.

Appendix A. Supplementary material

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.jbiomech.2018.10.025>.

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APPENDIX D Questionnaire



RUNNING QUESTIONNAIRE

1. How long have you been running?
2. What is your preferred event/distance?
3. Are you currently training for a particular race? Yes No
4. Do you run for: Fitness Recreation Competition
5. Describe a typical week including :
 - a. How many days a week do you run?
 - b. The type of training runs you do?

Tempo Runs Interval Training Hills Long Runs

Other (Describe)
 - c. How long is each run (kilometers, miles, or time)?
6. What surfaces do you generally run on?

Sidewalk Asphalt Grass Track

Trails Gravel Treadmill

Other
7. What is your race pace (if known)?
8. What shoes are you currently running in?
9. How often do you change your shoes?

10. Do you wear custom or off-the-shelf orthotics? If so, for what reason?

11. Do you focus on running form as you run? Yes No

If yes, how?

12. Have you made any changes to your running technique or training regime (intervals, hills, speed, surface, shoes, cross training activities, running form, or others)? Yes No

If yes, describe?

13. Do you participate in any other activities or exercise (gym, yoga, stretching)? Yes No

If yes, describe?

MEDICAL HISTORY

1. Do you currently have any pain, soreness or injuries? Yes No

If yes, describe?

2. Have you sustained any previous injuries including upper body injuries? Yes No

If so, describe and indicate how these have been managed?

3. Do you have any general health issues we should know about:

Diabetes Cardiac Conditions Breathing Disorders Dietary Issues

Others that you feel may be relevant

Name:

Preferred Phone Number:

Email Address:

PLEASE EMAIL COMPLETED QUESTIONNAIRE TO: alessandro.garofolini@vu.edu.au