

School of Physiotherapy and Exercise Science

The Biomechanics of Running with Load

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**This thesis is presented for the Degree of
Doctor of Philosophy
of
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Author's Declaration

To the best of my knowledge and belief, this thesis contains no material previously published by any other person except where due acknowledgment has been made.

This thesis contains no material which has been accepted for the award of any other degree or diploma in any university. This thesis contains four published papers and two papers prepared for submission on load carriage running. The statements of contribution of co-authors in these papers are presented in the Appendix. The Author Accepted Manuscript version of the manuscripts, where applicable, is included as thesis chapters. This is in compliance with copyright requirements of the respective publishers and Curtin University. For chapters containing published manuscripts, section and subsection numbers have been added to the manuscript's headings and subheadings. In-text references to figures and tables include the thesis chapter number. All references were standardized to the style output of the Journal of Biomechanics. This ensures consistency in formatting throughout the thesis. Every reasonable effort has been made to acknowledge the owners of copyright material. I would be pleased to hear from any copyright owner who has been omitted or incorrectly acknowledged.

Human Ethics: The research presented and reported in this thesis was conducted in accordance with the National Health and Medical Research Council National Statement on Ethical Conduct in Human Research (2007) – updated March 2014. The proposed research studies received human research ethics approval from the Curtin University Human Research Ethics Committee (RD-41-14).

Signature: 

[Bernard Liew]

Date: ...30th June 2017.....

Abstract

Background: Personnel needing to fulfil specific operational tasks, athletes competing in ultra-endurance races, and commuters running to work, often have to run with load. Load carriage has the potential to negatively impact physical performance and increase injury risk potential. Presently, physical conditioning provides the most feasible and immediate solution for the mitigation of load carriage detriments in running. However, current load carriage physical conditioning practices do not target the biomechanical requirements of load carriage gait, especially that of running. This is because, relatively little is known about how load carriage changes running biomechanics.

Objectives: The aims of this doctoral thesis were to comprehensively investigate how load carriage changes running biomechanics, and to develop and test the efficacy of a biomechanically informed load carriage resistance training program for loaded running.

Methods: In the Pilot study ($n = 6$), we designed and validated a marker set suitable for use in backpack running experiments. A purely lateral pelvic marker cluster was developed and validated against an International Society of Biomechanics (ISB) pelvic marker set in overground running. Dependent variables included pelvic segment and hip kinematics in the stance phase. Statistical measures included: Statistical Parametric Mapping (SPM) paired t-test, coefficient of multiple correlation (CMC), and standard error of measurement (SEM). In Study one, we performed a descriptive study on load carriage running involving healthy recreational runners ($n = 30$). Participants performed running at three velocities (3, 4, and 5 m/s) while carrying three loads (0 %, 10 %, and 20 % body weight) in a backpack. Three-dimensional motion capture of running was performed. Dependent variables including stance phase joint work, joint power, and joint kinematics were extracted. Statistical measures included: linear mixed model and SPM Canonical Correlation Analysis.

In Study two, a biomechanically informed “Targeted” resistance program was developed from the knowledge gleaned in Study one and the background literature. “Targeted” training consisted of loaded single leg hopping to augment leg stiffness, loaded countermovement jumps to augment knee power generation, and hip flexor cable pull to augment hip power absorption. These three muscular variables were hypothesized to mediate the reduction of mechanical work and the negative kinematic adaptations to tissue loading potential during load carriage running. This was a two-week preparation, six-week training study with concealed allocation and blinded assessor.

The primary aim of Study two was to understand if the new “Targeted” program was more effective at reducing loaded running mechanical work and increase ankle positive work proportion, compared to a “General” heavy-resistance training program. The secondary aim was to compare if “Targeted” training was more effective at reducing peak ankle and knee flexion angles, and late stance hip adduction angle, compared to “General” training. These are kinematic changes hypothesized to increase tissue loading potential in loaded running.

Outcome assessments included pre- and post-intervention three-dimensional motion capture of loaded (20 % bodyweight) running at 3.5 m/s, loaded (20 % bodyweight) countermovement and squat jumps, and isokinetic strength test at 60°/s of the knee and ankle extensors. Dependent variables included: joint work, leg stiffness, joint kinematics in running, peak jump power, and maximal knee and ankle extensor strength. Statistical measures included linear mixed model and SPM Repeated Measures Analysis of Variance (RM-ANOVA). All kinetic outcomes were expressed as dimensionless units.

Results: In the Pilot study, average CMC values ranged from 0.87 to 0.98 for pelvic and hip angles. SPM paired t-test found slight differences in pelvic and hip angles at approximately 20 % of stance phase. SEM averaged between 0.30° to 0.58° for both marker sets. In Study one, load significantly increased total joint work and total positive work and this effect was greater at faster velocities. Load carriage increased ankle positive work (β coefficient = rate of 6.95×10^{-4} units work per 1 % BW carried), and knee positive ($\beta = 1.12 \times 10^{-3}$ units) and negative work ($\beta = -2.47 \times 10^{-4}$ units), and hip negative work ($\beta = -7.79 \times 10^{-4}$ units). Incremental load magnitude was positively correlated to joint power in the second half of stance. Increasing load magnitude was also positively correlated with alterations in three-dimensional ankle angles during mid-stance (4.0 and 5.0 m/s), knee angles at mid-stance (at 5.0 m/s), and hip angles during late stance (at all velocities). Post hoc analyses indicated that at faster running velocities (4.0 and 5.0 m/s), increasing load magnitude appeared to alter power contribution in a distal-to-proximal (ankle → hip) joint sequence from mid-stance to toe-off. In addition, at faster velocities, increasing load magnitude increased peak knee and ankle flexion angles, and late stance hip adduction angle.

In Study two, all 30 participants completed the trial and were available for final time point analysis. There were no significant time-by-group interactions in total joint work and ankle positive work proportion. There was only a time effect on strength and power variables. Between the “Targeted” and “General” group, respectively, ankle strength improved between 0.23 and 0.34 Nm/kg; knee strength improved between 0.11 and 0.24 Nm/kg; and squat jump power improved between 0.09 and 0.18 dimensionless units. There was a tendency that “General” training (0.19 units) increased countermovement jump power more than “Targeted” training (0.05 units). “Targeted” training reduced running leg stiffness by 2.27 more units (1.88 kN/m) than “General” training. An increase in knee strength was associated with a reduction in total work after “General” training, but an increase in work after “Targeted” training. A post-hoc exploratory analysis found that the reduction in leg stiffness after “Targeted” training was associated with a reduction in loaded running impact force.

For all kinematic waveforms, there was only a time effect. Both training programs did not significantly reduce peak ankle and knee flexion angles. Although both training programs reduced hip adduction angle at late stance, the magnitude of change was $< 1^\circ$. “Targeted” training appeared to increase ankle and knee flexion angular excursion from initial contact to mid-stance, more than “General” training.

Conclusion: The lack of transference of increased strength and power to hypothesized mechanical work and kinematic outcomes, suggests that mechanical work and tissue loading optimization are not the only goals in loaded running. The decrease in leg stiffness after “Targeted” training suggests that impact force management may be an important goal during load carriage running. For exercises to benefit loaded running mechanical work and tissue loading potential, the present training needs to be modified to couple muscular strength gains with successful leg stiffness augmentation. Greater leg stiffness would provide the mechanical leverage necessary for the motor control system to reduce the muscular effort of load carriage running.

The mechanical adaptations after resistance training likely reflects the innate prioritization of different cost functions during loaded running. Some of these functions may include mechanical work, tissue loading, and impact force. Future research on load carriage physical conditioning should investigate how different training regimes can be tailored to augment subject-specific cost functions during load carriage running.

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Dedication

“This most beautiful system of the sun, planets and comets, could only proceed from the counsel and dominion of an intelligent and powerful Being.” - Isaac Newton.
The Principia: Mathematical Principles of Natural Philosophy.

This work that I have undertaken, and the journey I have trodden, would not have been possible without God our Father, His Son Jesus, and His Blessed Mother. The work of this thesis is nothing more than the sum of the innumerable gifts I have received daily - this life, my family, my wife, my child, my education, and my talents. Throughout my life, I have always wondered what talents I have received. Unlike many of my peers who have incredible artistic skills or could perform brilliant math, I do not seem to have any overt talent(s). The introspective journey of this PhD made me realise that the gift of work ethic and curiosity are the talents God have bestowed upon me.

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Table of Contents

Author’s Declaration	ii
Abstract	iii
Acknowledgements	vi
Dedication	viii
Table of Contents	x
List of Figures	xvii
List of Tables	xx
List of Abbreviations	xxii
Chapter 1 Introduction	1
1.1. Evolution of load carriage in humans	1
1.2. Organization and contribution	2
1.3. References	3
Chapter 2 Literature review	8
2.1. Introduction	8
2.2. Definition	8
2.3. Load carriage, injuries and physical performance	9
2.3.1. Injuries	9
2.3.2. Physical performance	10
2.3.3. Summary	11
2.4. Intervention strategies	11
2.4.1. Ergonomic and engineering strategies	12
2.4.2. Physical conditioning	13
2.4.3. Summary	15
2.5. Load carriage running	15
2.5.1. Work-energy requirements in walking and running	16
2.5.2. Injury mechanisms in walking and running	18
2.5.3. Summary	20
2.6. Adaptations of biomechanics in load carriage running	20
2.6.1. Adaptations in load carriage running	20
2.6.2. Summary	22
2.7. Physical conditioning in load carriage – informing the “what” of training	23
2.7.1. Rationale	23
2.7.2. Specificity of muscle groups	23

2.7.3	Specificity of muscular variables trained	24
2.7.4	Specificity of contraction modes	25
2.7.5	Specificity of coaching cues	26
2.7.6	Summary	27
2.8	Thesis landscape	27
2.9	Summary of key points	29
2.10	References	30
Chapter 3	Performance of a lateral pelvic cluster technical system in evaluating running kinematics.....	45
Abstract	45
3.1	Introduction	46
3.2	Methods.....	47
3.2.1	Participants	47
3.2.2	Marker set placement and biomechanical model.....	47
3.2.3	Experimental protocol.....	48
3.2.4	Data processing.....	48
3.2.5	Statistical analysis.....	48
3.3	Results	49
3.4	Discussion	49
3.5	References	51
3.6	Figures.....	54
3.7	Tables	56
3.8	Supplementary material	59
3.8.1	Definition of joint centres and co-ordinate systems.....	61
3.8.2	Differences between ISB's pelvic segment coordinate system (PCS) and that used in this study.....	61
3.8.3	Marker shells design (scaled to actual size)	62
3.9	Conflict of interest and source of funding.....	62
3.10	Acknowledgement	62
Chapter 4	The effects of load carriage on joint work at different running velocities	63
Abstract	63
4.1	Introduction	64
4.2	Methods.....	65
4.2.1	Participants	65
4.2.2	Biomechanical model.....	66
4.2.3	Protocol	66
4.2.4	Data measurement and processing.....	67

4.2.5	Statistical analysis.....	68
4.3	Results.....	68
4.3.1	Spatio-temporal parameters.....	68
4.3.2	Total and net lower limb work.....	68
4.3.3	Positive and negative joint work.....	69
4.3.4	Percentage joint work contribution.....	69
4.4	Discussion.....	69
4.5	Conclusion.....	73
4.6	References.....	73
4.7	Figures.....	77
4.8	Tables.....	79
4.9	Supplementary material.....	82
4.9.1	Participant inclusion/exclusion criteria.....	82
4.9.2	Biomechanical model.....	82
4.10	Conflict of interest and funding.....	86
4.11	Acknowledgement.....	86
Chapter 5	Joint power and kinematics coordination in load carriage running: implications for performance and injury.....	87
Abstract	87
5.1	Introduction.....	88
5.2	Methods.....	89
5.2.1	Study design.....	89
5.2.2	Participants.....	90
5.2.3	Running protocol.....	90
5.2.4	Biomechanical modelling and processing.....	90
5.2.5	Statistical analysis.....	91
5.3	Results.....	92
5.3.1	Effect of load on joint power vector.....	92
5.3.2	Effect of load on joint angle vector.....	93
5.4	Discussion.....	93
5.5	Conclusion.....	96
5.6	References.....	96
5.7	Figures.....	100
5.8	Tables.....	103
5.9	Supplementary material.....	105
5.10	Conflict of interest and funding.....	113
5.11	Acknowledgement.....	113

Chapter 6	Effects of two neuromuscular training programs on running biomechanics with load carriage: a study protocol for a randomised controlled trial	114
Abstract		114
6.1	Introduction	115
6.1.1	Potential adaptive and maladaptive biomechanics.....	116
6.1.2	Rationale	117
6.1.3	Objectives.....	117
6.2	Methods.....	118
6.2.1	Research design	118
6.2.2	Participants, setting and recruitment.....	118
6.2.3	Sample size calculation	118
6.2.4	Randomization, allocation and blinding.....	119
6.2.5	Subjective assessment	119
6.2.6	Three dimensional motion capture - load carriage running protocol	120
6.2.7	Three dimensional motion capture – jumping and hopping protocol	120
6.2.8	Data measurement and processing.....	121
6.2.9	Biomechanical modelling.....	122
6.2.10	Strength assessment (Isokinetic dynamometry)	122
6.2.11	Intervention	123
6.2.11.1	Familiarization phase (all participants) (two weeks).....	123
6.2.11.2	Training phase (six weeks) - Standardized warm up (both groups)	123
6.2.11.3	Generic neuromuscular resistance training group (GT group).....	123
6.2.11.4	Biomechanically informed neuromuscular resistance training group (BIT group)	124
6.2.11.5	Augmented feedback (both groups).....	125
6.2.11.6	Determining initial training loads and progression	125
6.2.11.7	Between group differences in training volume	126
6.2.11.8	Intervention adherence.....	126
6.2.12	Dependent variables and statistics	126
6.3	Discussion	127
6.4	References	128
6.5	Figures.....	135
6.6	Tables	136
6.7	Additional files	139
6.8	Declarations.....	151
6.8.1	Acknowledgements.....	151
6.8.2	Funding	151
6.8.3	Availability of data and materials.....	151

6.8.4	Authors' contributions	151
6.8.5	Competing interests	151
6.8.6	Consent for publication	151
6.8.7	Ethics approval and consent to participate	151
Chapter 7	Resistance training on load carriage mechanical work	152
Abstract	152
7.1	Introduction	153
7.2	Methods.....	154
7.2.1	Participants	155
7.2.2	Procedures	155
7.2.3	Intervention	156
7.2.4	Outcome measure	156
7.2.5	Sample size.....	158
7.2.6	Statistical analyses	158
7.3	Results	159
7.4	Discussion	160
7.4.1	Effectiveness of interventions	161
7.4.2	Modifiability of running mechanical work.....	162
7.5	References	165
7.6	Figures.....	171
7.7	Tables	175
7.8	Supplementary material	179
7.9	Acknowledgement	209
7.10	Conflict of interest and source of funding.....	209
Chapter 8	Loaded running kinematics after resistance training	210
Abstract	210
8.1	Introduction	211
8.2	Methods.....	213
8.2.1	Design, ethics, blinding.....	213
8.2.2	Sample size.....	213
8.2.3	Participants	214
8.2.4	Intervention	214
8.2.5	Assessment	215
8.2.6	Statistical analysis.....	215
8.3	Results	216
8.4	Discussion	216
8.5	References	219

8.6	Figures.....	225
8.7	Tables	233
8.8	Acknowledgement	236
8.9	Conflict of interest and source of funding.....	236
Chapter 9	Discussion	237
9.1	Lessons learnt on load carriage running.....	237
9.1.1	Lesson one: Accurate experimentation requires valid task-specific experimental methods	237
9.1.2	Lesson two: Knowing “what” to train in loaded running.....	238
9.1.3	Lesson three: Optimization of different cost functions in loaded running	239
9.1.4	Lesson four: Leg stiffness may underlie work and tissue loading optimization.....	240
9.2	Limitations of research.....	240
9.3	Future research directions	241
9.4	Conclusion.....	243
9.5	References	243
Appendix A	Short commentary on swing mechanics in loaded running.....	245
A.1	Introduction	245
A.2	Methods.....	245
A.3	Results	246
A.4	Discussion	246
A.5	References	247
A.6	Tables	249
A.7	Figures.....	251
Appendix B	Statement of contribution from co-authors.....	259
Appendix C	Permission to reproduce article in thesis and communicate contents within the public sphere	263
C.1	Correspondence with Curtin Library	263
C.2	Correspondence with Elsevier publisher.....	265
C.3	Correspondence with Biomedical Central publisher.....	266
Appendix D	Study 1	267
D.1	Ethics and amendment	267
D.2	Patient information sheet.....	268
D.3	Informed consent	271
D.4	Video informed consent	273
D.5	Gait collection sheet.....	274
D.6	Advertisement.....	275

Appendix E	Study 2	276
E.1	Ethics and amendment	276
E.2	Patient information sheet.....	277
E.3	Informed consent	282
E.4	Video informed consent	284
E.5	Gait collection sheet.....	285
E.6	Advertisement.....	286
E.7	Pre-test questionnaire.....	287
E.8	Baseline health questionnaire	289

List of Figures

Figure 2-1 Intervention strategies in load carriage	12
Figure 2-2 Flowchart of studies	29
Figure 3-1 Position of lateral pelvic marker clusters (lateral view).	54
Figure 3-2 a. Hip joint angle group mean (SD) in the top row: Joint angles derived from ISB pelvis and pelvic cluster protocol in the sagittal (left), frontal (middle), and transverse (right) plane. SPM paired t-test results in the bottom row of angles in the sagittal, frontal, and transverse plane. The horizontal dotted line represents the critical random field threshold of $t = 0.0167$	55
Figure 3-3 Pelvic clusters	62
Figure 4-1 Dimensionless total positive and negative joint work.....	77
Figure 4-2 Dimensionless positive joint work means and standard deviation (error bars)	77
Figure 4-3 Dimensionless negative joint work means and standard deviation (error bars)	78
Figure 5-1 (a-l). Representative mean (standard deviation) power, angle curves at 5.0 m/s running. (a-c) Joint power, (d-f) Ankle three dimensional angles, (g-i) Knee three dimensional angles, (j-l) Hip three dimensional angles.....	100
Figure 5-2 Canonical correlation analysis (SPM [F]) results of (a) joint power vector, (b) ankle angle vector, (c) knee angle vector, (d) hip angle vector at 5.0 m/s running. Shaded regions represent significant supra-threshold clusters that exceeded random critical threshold.	101
Figure 5-3 Representative post hoc analyses at 5.0 m/s. Shaded regions represent significant supra-threshold clusters that exceeded random critical threshold.....	102
Figure 5-4 Vector field analysis at 3.0 m/s (left column) and 4.0 m/s (right column).....	105
Figure 5-5 Mean (standard deviation) joint power curves, with post-hoc analysis at 3.0 m/s.....	106
Figure 5-6 Mean (standard deviation) joint power curves, with post-hoc analysis at 4.0 m/s.....	107
Figure 5-7 Mean (standard deviation) ankle angle curves at 3.0 m/s.....	108
Figure 5-8 Mean (standard deviation) ankle angle curves, with post-hoc analysis at 4.0 m/s.....	109
Figure 5-9 Mean (standard deviation) knee angle curves at 3.0 m/s and 4.0 m/s	110
Figure 5-10 Mean (standard deviation) hip angle curves, with post-hoc analysis at 3.0 m/s.....	111
Figure 5-11 Mean (standard deviation) hip angle curves, with post-hoc analysis at 4.0 m/s.....	112
Figure 6-1 CONSORT Flow Diagram	135
Figure 7-1 CONSORT flow diagram.....	171
Figure 7-2 Mean (95 % confidence interval as error bars) of spatio-temporal and mechanical work during loaded running.	172

Figure 7-3 Mean (95 % confidence interval as error bars) of isokinetic strength and loaded jumping power. Countermovement jump (CMJ) and squat jump (SJ).....	173
Figure 7-4 Mean (95 % confidence interval as error bars) of loaded running leg stiffness.	174
Figure 7-5 Retro-reflective markers positions	187
Figure 7-6 Running leg length change (m).....	204
Figure 7-7 SPM results of leg length change. Instantaneous leg length only has a tendency towards significance in the “Targeted” pre-post group. This is evidence from the rising slope of the SPM interaction, and the Mean (SD). The linear mixed model results demonstrate that change in leg length in the “Targeted” group was significantly different from the “General” group.....	205
Figure 7-8 Ground reaction force in loaded running.....	206
Figure 7-9 SPM results for ground reaction force. * The magnitude of GRF projected in line with the leg non-significantly increased post training in the “General” group. The results here demonstrate change to leg stiffness in the “Targeted” group was not due to changes in projected GRF.....	207
Figure 7-10 Simple linear regression between leg stiffness and impact peak (G = “General”, T = “Targeted”)	208
Figure 8-1 CONSORT Flow Diagram	225
Figure 8-2 (a-c) Mean (standard deviation) waveform of sagittal plane kinematics over the stance phase of running; (d-f) Mean (standard deviation) waveforms of change in kinematics with time (post curves minus pre) for each group.	226
Figure 8-3 (a-c) Mean (standard deviation) waveform of frontal plane kinematics over the stance phase of running. (d-f) Mean (standard deviation) waveforms of change in kinematics with time (post curves minus pre) for each group.	227
Figure 8-4 (a-c) Mean (standard deviation) waveform of transverse plane kinematics over the stance phase of running; (d-f) Mean (standard deviation) waveforms of change in kinematics with time (post curves minus pre) for each group.....	228
Figure 8-5 Statistical parametric maps of repeated measures anova for ankle X (sagittal plane), Y (frontal plane), Z (transverse plane) angles.....	229
Figure 8-6 Statistical parametric maps of repeated measures anova for knee X (sagittal plane), Y (frontal plane), Z (transverse plane) angles.....	230
Figure 8-7 Statistical parametric maps of repeated measures anova for hip X (sagittal plane), Y (frontal plane), Z (transverse plane) angles.....	231
Figure 8-8 Illustration of foot-ground and backpack-trunk collisions during loaded hopping training.....	232
Figure 9-1 Decision tree for designing load carriage training	242
Figure 9-2 Swing phase joint power SPM results.....	251
Figure 9-3 Swing phase joint power mean waveforms	252
Figure 9-4 Swing phase joint angle (X axis) SPM results.....	253

Figure 9-5 Swing phase joint angle (X axis) mean waveforms	254
Figure 9-6 Swing phase joint angle (Y axis) SPM results.....	255
Figure 9-7 Swing phase joint angle (Y axis) mean waveforms	256
Figure 9-8 Swing phase joint angle (Z axis) SPM results	257
Figure 9-9 Swing phase joint angle (Z axis) mean waveforms.....	258

List of Tables

Table 3-1 Between-protocol (ISB vs Cluster pelvic protocols) coefficient of multiple correlations	56
Table 3-2 Within protocol standard error of measurement (°)	57
Table 3-3 Marker placements	59
Table 3-4 Range of motion (ROM) (°) measured using the ISB pelvis	60
Table 4-1 Mixed effect model of dimensionless joint work parameters (unit), and load (%BW) and velocity (m/s)	79
Table 4-2 Percentage joint contribution to total positive and negative joint work	81
Table 4-3 95 % confidence interval of coefficients between work, load (% BW) and velocity (m/s).....	84
Table 5-1 Vector and post-hoc scalar field analyses of dependent variables	103
Table 6-1 Biomechanical adaptations of load carriage to potentially optimize metabolic cost and minimise injury risk	136
Table 6-2 Guide for determining one repetition maximum in novice weight trainers.	138
Table 6-3 Familiarization phase (2 weeks (4 sessions)).....	139
Table 6-4 Training phase- warm up	141
Table 6-5 Exercises progression table (Biomechanically informed training group).....	142
Table 6-6 Exercises progression table (General training group)	144
Table 6-7 Augmented feedback (AF) cues for biomechanically informed training group	146
Table 6-8 Augmented feedback (AF) cues for general training group.....	147
Table 6-9 Augmented feedback (AF) principle and schedule (Both groups).....	150
Table 7-1 Participant's baseline characteristics.....	175
Table 7-2 Relationship between Δ (post-pre) mediating factors and Δ (post-pre) total joint work	177
Table 7-3 Consensus on Exercise Reporting Template (CERT)	179
Table 7-4 Individual weekly attendance (maximum 3 sessions/week)	184
Table 7-5 “General” group "group average" weekly load.....	185
Table 7-6 “Targeted” group "group average" weekly load	186
Table 7-7 Influence of Time, Group, Time-by-Group interactions on spatio-temporal variables	191
Table 7-8 Influence of Time, Group, Time-by-Group interactions on kinetic variables.....	193
Table 7-9 Influence of Time, Group, Time-by-Group interactions on individual joint work	195
Table 7-10 Influence of Time, Group, Time-by-Group interactions on percentage joint contribution	197
Table 7-11 Influence of Time, Group, Time-by-Group interactions on strength and jump power.....	199

Table 7-12 Influence of Time, Group, Time-by-Group interactions on running leg stiffness	201
Table 7-13 Relationship between leg stiffness and impact peak	203
Table 8-1 Baseline participant characteristics	233
Table 8-2 Mean (standard deviation) of joint angle change with training.....	235
Table 9-1 Effect of speed and load on running swing phase joint work	249

List of Abbreviations

%	Percentage	kN	Kilo Newton
[F]	F statistic	KP	Knowledge of performance
[t]	t statistic	LL	Leg length
°	Degree	m	Metre
3D	Three dimensional	MTU	Muscle tendon unit
AF	Augmented feedback	N	Newtons
ASIS	Anterior superior iliac spine	ns	Non-significant
BM	Body mass	PFJ	Patellofemoral joint
BMI	Body mass index	POSE	Position and orientation
BW	Body weight	PSIS	Posterior superior iliac spine
CERT	Consensus on Exercise Reporting Template	RCT	Randomized controlled trial
CI	Confidence interval	RE	Running economy
CMC	Coefficient of multiple correlation	RFT	Random field theory
CMJ	Countermovement jump	RM	Repetition maximum
COM	Centre of mass	RM-ANOVA	Repeated measures analysis of variance
CONSORT	Consolidated Standards of Reporting Trials	ROM	Range of motion
CS	Coordinate system	RRI	Running related injuries
DOF	Degree of freedom	s	Second
g	Gravitational constant	SD	Standard deviation
GCS	Global coordinate system	SEM	Standard error of measurement
GRF	Ground reaction force	SJ	Squat jump

Hz	Hertz	SL	Single leg
IK	Inverse kinematic	SPM	Statistical parametric mapping
ISB	International Society of Biomechanics	SSC	Stretch shortening cycle
ITT	Intention to treat	STA	Soft tissue artefact
J	Joules	W	Watts
kg	Kilogram	α	alpha
km	Kilometre	β	beta

Chapter 1 Introduction

1.1. Evolution of load carriage in humans

It is difficult to pinpoint when humans first began carrying load. The evolution of bipedalism enabled early hominids to free their hands and carry load for food and tool transportation (Videan and McGrew, 2002), as well as infant carriage (Wall-Scheffler et al., 2007). In fact, load carriage systems may have evolved to reduce the cost of carrying unwieldy objects with the hands (Wall-Scheffler et al., 2007). Despite advances in modern transportation systems, carrying load during locomotion remains an important mode of object conveyance (Kinoshita, 1985). Load carrying during locomotion is utilised in various occupations, such as the military (Knapik et al., 2004) and emergency service personnel (Park et al., 2010), in a variety of organized sports, such as adventure racing and orienteering (Fordham et al., 2004; Lucas et al., 2016; Schutz et al., 2012), and in less organized pursuits, such as hiking and tramping (Lobb, 2004).

A consistent phenomenon in load carriage is that the load is carried during locomotion (i.e. movement from one place to another). For example, soldiers can march more than 15 km a day with carried load (e.g. vests, backpacks, and weapons) of up to 55 kg (Orr, 2010). In addition to walking, military soldiers often have to run, jump, and land whilst carrying load (Clements et al., 2012; Hunt et al., 2016; Spiering et al., 2012). Load carriage is not only present in the military context, but also in the civilian sporting context. Adventure racers and orienteers can be expected to carry loads (e.g. food, water, and clothing) of up to 15 kg while running, cycling, and canoeing (Equipment, 2014; Marais and de Speville, 2004). These races can span between one to ten days (Fordham et al., 2004). In hiking and tramping, participants may carry up to 30 % body weight (BW) on hikes lasting more than eight days (Lobb, 2004). In recent times, load carriage has evolved into a training method to enhance athletic and soldier performance (Knapik et al., 2012; Markovic et al., 2013; Reiman et al., 2010; Swain et al., 2011), a physical assessment method for occupational readiness, such as the pack hike test (Petersen et al., 2010; Phillips et al., 2012), and a tool to attenuate bone loss and improve bone density (Genc et al., 2006; Roghani et al., 2013).

The ubiquity of load carriage in occupation and sport is not without problems. Load carriage has been shown to increase the risk of developing neuro-musculo-skeletal injuries, as well as reduce locomotion performance (Andersen et al., 2016; Billing et al., 2015; Carlton and Orr, 2014; Knapik et al., 1996; Knapik et al.; Knapik et al.; Orr et al., 2015; Teunissen et al., 2007; Treloar and Billing, 2011). This has inspired researchers to develop intervention strategies capable of mitigating the negative influence load carriage has on occupational and sporting health and performance.

Physical conditioning has been used to increase the physical capacity of an athlete needing to perform with load (Knapik et al., 2012). One challenge with designing optimal physical conditioning programs for load carriage, is that individuals who carry load use a diverse range of gait patterns to fulfil occupational (Brown et al., 2014a; Brown et al., 2016; Brown et al., 2014b; Clements et al., 2012) and sporting objectives (Cross et al., 2014). However, most studies on load carriage have largely focused on walking (Liew et al., 2016), with only recent research interest being pursued into more dynamic activities such as running (Brown et al., 2014b; Silder et al., 2015; Xu et al., 2017). Of the gait patterns aside from walking, running with load represents a fundamental task needed in many occupational (Treloar and Billing, 2011), and sporting events (Marais and de Speville, 2004). In addition, an increasing number of civilians are adopting environmentally-friendly means of commuting to work including running (Zander et al., 2014), which necessitates some form of load carriage (e.g. to carry clothing).

An adequate understanding of the biomechanical demands of running with load is a crucial step towards developing occupational and sports specific physical training programs. The relative dearth of research in load carriage running has inspired me to begin the scientific pursuit of knowledge regarding how load carriage influences running biomechanics, and the training required for its mechanical optimization.

1.2 Organization and contribution

The aims of this thesis are to address two gaps in the load carriage literature. First, we will address the relative lack of in-depth biomechanical investigations on load carriage running. Second, we will also address the absence of a biomechanically informed load carriage physical conditioning program for running.

To fulfil these aims, I first conducted a narrative literature review to explore the current state of knowledge with regards to load carriage on physical performance, injury, running mechanics, and physical conditioning. The review also rationalizes the limitations present in the load carriage literature. Next, we conducted a descriptive experiment on load carriage running biomechanics to inform the development of a load carriage running specific resistance training program. Lastly, we conducted and reported the results of a randomized controlled trial, on the effectiveness of a load carriage specific resistance training program on load carriage running biomechanics.

The theoretical construct and methodology employed in this thesis falls solely within the domain of clinical biomechanics. In this thesis, we employed three dimensional inverse dynamics analysis to quantify running kinematics and joint-level muscle energetics. This thesis is an original and significant contribution to the load carriage literature for several reasons. The thesis provides an in-depth investigation of load carriage running biomechanics, which has not received much research interest. Second, this thesis developed a biomechanically informed training program for load carriage running, within a translational framework that has been variedly adopted in the rehabilitation literature (Awad et al., 2016a; Awad et al., 2016b; O'Sullivan and Beales, 2007). Specifically, this framework allows the hypothesizing of candidate training variables based on background mechanistic knowledge, the efficacy of which can be tested in a clinical trial. Third, I tested the efficacy of the new training program using a robust, randomized controlled trial design. Together, this thesis provides solutions to the significant health, economic, and performance burden posed by load carriage to occupational and civilian athletes.

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Chapter 2 Literature review

2.1 Introduction

The investigation of load carriage in human locomotion provides an excellent model to further the understanding of human performance limits and its advancement. Scientists who are interested in the study of load carriage locomotion have long marvelled at the physical attributes of the Himalayan Sherpas in Nepal, and the women of the Luo and Kikuyu tribes in East Africa, Kenya (Heglund et al., 1995). As a consequence of poor transportation infrastructure, these people have developed superior load carrying capabilities in order to carry food, water and equipment for survival. Himalayan Sherpas have been reported to carry between 125 % and 183 % of their body weight (BW) using a head strap to support the load (Bastien et al., 2005). While military recruits double their oxygen consumption rate when carrying a 70 % BW load during walking compared to BW walking (i.e. walking with body weight only), East African women have been reported to increase their oxygen consumption rate by only 50 % carrying similar loads (Maloiy et al., 1986).

This chapter serves several functions. First, it will provide a definition of load carriage and the reasons why as a topic, load carriage research has scientific, clinical, and societal importance. Second, given the importance of load carriage and the problems load carriage poses to an individual, I will highlight the range of interventions that have been designed to ameliorate these problems. Third, this chapter critically explores the body of work that has been conducted in load carriage to date, and the important gaps within the literature. Lastly, I will highlight the aims of this thesis within the gaps identified in the literature.

2.2 Definition

Load can be defined as “a mass or weight supported by something” (Merriam-Webster Online Dictionary, 2016). When load is defined in this context, it is immediately apparent that load can be carried for a variety of reasons under different conditions. Humans must support their own BW due to the influence of the Earth’s gravity. In addition, humans have body mass (BM) which is due to the amount of matter within the body.

Although the notion of load carriage in humans is difficult to conceive of carrying one's own BW and BM – since one cannot mentally parse out the effects of gravity and matter in routine daily life, individuals can still experience the impact of altered BW and BM in many instances. For example, during swimming, the effects of gravity and hence BW is reduced while BM is constant. However, the purpose of this thesis is not to parse out individual contributions of gravity (weight) and matter (mass) on the experiential influence of altered loads borne by humans. For this reason, the remainder of this thesis will focus on the use of the term “load” as inherently embodying these dual physical properties.

The experiential effect of altered loading in humans can be from both internally and/or externally imposed loads. Altered internal loads may include the physical phenomena of obesity or pregnancy (Foti et al., 2000; Fu et al., 2015). Altered external loads may include backpack carriage in school children and military soldiers (Brackley and Stevenson, 2004; Orr, 2010). Within the biomechanics research literature, altered internal loading states such as obesity can be simulated via external loading paradigms, such as backpack carriage (Castro et al., 2013). The ubiquity of backpack carriage as a form of load carriage in schools (Haselgrove et al., 2008), occupation (Brown et al., 2014b), and sport (Marais and de Speville, 2004), means that the focus of this thesis will be on backpack carriage.

2.3 Load carriage, injuries and physical performance

2.3.1 Injuries

The injury burden of load carriage is well documented in the occupational military setting, but less in the sporting and recreational setting (Knapik et al., 1996; Knapik et al., 2004b; Orr et al., 2017; Orr et al., 2015; Orr and Pope, 2016; Orr et al., 2014). The injury burden of load in the military is likely due to the extremely high load magnitude borne, which can reach up to 55 kg (Orr, 2010), and the economic burden of discharging a serviceman because of injuries (Huffman et al., 2008). Injuries in the military can result in more than three million days of limited duty annually (Ruscio et al., 2010). Further, the consequence of an injury is not trivial given that the cost of training each soldier exceeds US \$57 000 (Huffman et al., 2008), and that an injury increases the odds of being discharged from military service by 4.6 times (Larsson et al., 2009).

In the military, overuse lower limb injuries represent up to 48 % of all musculoskeletal injuries (Taanila et al., 2010). Up to 8 % of the 5188 injuries recorded between the years 2009 and 2010 in the Australian Defence Force have been attributed to load carriage (Orr et al., 2015). The injury risk potential of load carriage may reach up to 34 % (Orr et al., 2017) given that a proportion of injuries may remain unreported. The most common site of injury is the lower limb, which makes up 56 % of the reported injuries due to load carriage (Orr et al., 2015).

Soldiers are at greatest risk of sustaining an injury due to load carriage during the trainee phase of their career (Orr et al., 2017; Robinson et al., 2016; Surveillance Snapshot: Illness and injury burdens among U.S. military recruit trainees, 2012, 2013). This is not surprising given that new military recruits are exposed to a sudden increase in the volume and intensity of physical training, the stresses of which may exceed the repair capacity of musculoskeletal tissue (Finestone and Milgrom, 2008). The overall injury risk posed by load carriage on the musculoskeletal system does not differ between male and female soldiers (Orr and Pope, 2016). However, female soldiers are more susceptible to foot and back injuries because of load carriage, as compared to their male counterparts (Orr and Pope, 2016). The sex differences in injury sites could be due to the sex differences in upper body anthropometry – which influences the personal fit of occupational equipment, as well as joint specific muscular strength deficit (Orr and Pope, 2016).

2.3.2 Physical performance

The deleterious influence of load carriage on human physical performance can have a negative impact on operational mission goals and survivability. The consequence of soldiers being slowed down by load carriage can increase the risk of being wounded or killed by enemy fire (Billing et al., 2015), which can have dire consequences on operational mission success (Orr, 2010). Military soldiers are often required to carry a heavy backpack, body armour, weapons, and ammunitions during combat and peace time training (Carlton and Orr, 2014; Orr, 2010). Firefighters often carry heavy oxygen tanks and personal protection equipment (Park et al., 2010), whilst police officers often don body vests and firearms (Ramstrand et al., 2016).

These loads can reach up to 55 kg in military soldiers (Orr, 2010), 20 kg in firefighters (Taylor et al., 2016), and 10 kg in police officers (Carlton et al., 2013). At the mobility level, an early review estimated that physical performance reduces by approximately 1 % per kilogram of external load carried (Knapik et al., 1996). In elite Special Forces soldiers, doubling the backpack weight from 34 kg to 61 kg, increased a 20 km time trial route march from 171 min to 253 min, representing a 48 % decrease in performance (Carlton and Orr, 2014; Knapik et al., 1997). The deleterious impact of load carriage on performance is likely to be exacerbated by marching distance. For example, when soldiers were tested on a shorter 10 km route, doubling load magnitude from 18 to 36 kg only reduced time to completion performance by 23 % (Carlton and Orr, 2014; Harper et al., 1997).

Load carriage also influences the performance of ballistic movement tasks. In a simulated military break contact drill involving soldiers sprinting over 30 m, adding a 21.6 kg load of military gear increased completion time from an unloaded timing of 6.2 s to a loaded timing of 8.2 s (Treloar and Billing, 2011). Although a reduction in performance of 2 s may appear trivial in the civilian setting, a 1 s increment in completion of a 30 m break contact drill can increase a soldier's susceptibility of being contacted by an enemy's fire by 5 % (Billing et al., 2015). Even in highly trained occupational personnel, the negative effect of added load carriage on ballistic movement tasks can be observed. Carrying a load of 22 kg increased completion times in elite policemen by 7 % on a 35 m sprint tactical task, and 10 % on a 20 m mannequin drag task (Carlton et al., 2013).

2.3.3 Summary

Load carriage negatively impinges on injury risk and physical performance. This negative impact has motivated the development of intervention strategies to ameliorate the negative impact load carriage has on injury risk and physical performance.

2.4 Intervention strategies

Intervention strategies undertaken to ameliorate the deleterious impact of load carriage on injury and physical performance can be broadly classified into ergonomic, engineering, and physical conditioning strategies (Figure 2-1).

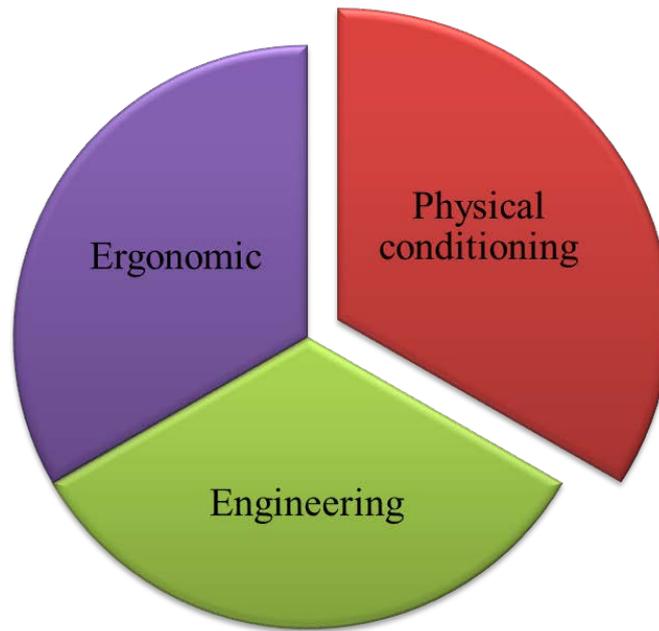


Figure 2-1 Intervention strategies in load carriage

2.4.1 Ergonomic and engineering strategies

Ergonomic strategies usually involve principles of “fitting the task to the individual”. In the context of military personnel, one such strategy typically involves imposing a load magnitude limit on a soldier’s total load carriage for a mission. Load carriage weight recommendations within the military are usually referenced in the form of a “doctrine” (Orr and Pope, 2015). However, despite the presence of load carriage recommendations, load magnitudes carried by soldiers over the decades are increasing (Orr, 2010). Load carriage recommendations are necessarily broad and vague, which ensures relevance of the recommendations for all military personnel. However, without clear guidelines, there is a lack of sufficient guidance for military commanders in imposing individual specific load limits (Orr and Pope, 2015). This highlights the need for ergonomic approaches to be supplemented by other strategies in ameliorating the deleterious impact of load carriage (Nindl et al., 2016).

Engineering strategies usually involve the principle of “fitting the individual to the task”. These strategies are commonly centred around developing wearable devices that provide assistive joint torques (Mooney et al., 2014; Panizzolo et al., 2016). Although potentially beneficial, current devices have only been tested in the laboratory with relatively small sample sizes (Mooney et al., 2014), been tested only in load carriage walking (Mooney et al., 2014; Panizzolo et al., 2016), and are bulky in material design (Cherry et al., 2016). Within a controlled laboratory setting, exoskeletons have been shown to reduce the energy cost of load carriage walking. For example, an autonomously powered ankle exoskeleton could reduce the energy cost of load carriage (23 kg vest) walking by an average of 8 % compared to loaded walking without the exoskeleton (Mooney et al., 2014). In addition, a recently developed soft exoskeleton suit reduced the energy cost of load carriage walking (30 % BW backpack) by 5 % relative to loaded walking with no suit (Panizzolo et al., 2016). However, not all participants benefited from its use. Mooney and colleagues documented that one out of the seven participants that made up the study’s sample, experienced an increase in energy cost with the use of the exoskeleton (Mooney et al., 2014).

2.4.2 Physical conditioning

Physical conditioning has been recognised as pivotal in ensuring optimal physical performance and occupational readiness (Friedl et al., 2015; Nindl et al., 2016), and may provide the most immediate and feasible solution to the burden of load carriage. All of the studies investigating physical conditioning programs for load carriage have been conducted within the military (Coyle et al., 2010; Friedl et al., 2015; Knapik et al., 2012; Nindl et al., 2016). No studies to my knowledge have explored optimal physical conditioning programs for load carriage in civilian cohorts. This is surprising given the growing popularity of ultra-endurance and adventure races, where loads are also carried (Lucas et al., 2016; Schutz et al., 2012). In addition, given the high injury rates of newly enlisted military servicemen, pre-enlistment physical conditioning of civilians may assist in reducing injury rates of military recruits (Molloy et al., 2012).

Physical conditioning programs reported to improve load carriage physical performance have consisted of various combinations of aerobic training, upper and lower body resistance training, and occupational task training with load carriage (Knapik et al., 2012). Collectively, these programs improved load carriage physical performance by effect sizes (Cohen d value) ranging from 0.79 (95 % CI 0.16 to 1.42) to 1.69 (95 % CI 1.04 to 2.32), with training periods spanning 8 to 24 weeks (Knapik et al., 2012). However, the literature suggests that the resistance training component ($d = 0.75$) elicits a greater effect on load carriage physical performance improvements, compared with the aerobic component ($d = 0.29$) (Knapik et al., 2012).

Task specificity has been recognised as an important consideration in the design of physical conditioning programs for military personnel (Knapik et al., 2009). In the United States military, a standardized functional training program which incorporates exercises that mimic the occupational task demands of soldiers was superior at reducing injury risk, and increasing passing rates on a military fitness test, compared to a general physical training program (Knapik et al., 2005; Knapik et al., 2003). However, the benefits of the standardized functional training program were not present in all military cohorts (Knapik et al., 2004a).

It is possible that the vast range of exercises and assessment batteries used in present military training studies, may mask important benefits of some exercises to specific functional task performance and injury risk (Harman et al., 2008; Hendrickson et al., 2010; Knapik et al., 2005; Knapik et al., 2004a; Knapik et al., 2003). For example, physical training in the military have used muscle contraction patterns varying from isoinertial (e.g. in a squat), isometric (e.g. in a stiff-legged deadlift), to stretch-shortening contractions (e.g. in a box jump) (Harman et al., 2008; Hendrickson et al., 2010). Some contraction patterns may be more beneficial to load carriage running (e.g. stretch-shortening), while others may be more beneficial to other functional tasks (e.g. deadlift training for lift-and-carry assessment).

Although currently implemented exercises for soldier physical conditioning may bear global similarity to the gross movement patterning of occupationally relevant tasks (Knapik et al., 2012; Knapik et al., 2009), it is argued that these exercises lack specific biomechanical task similarities (e.g. inter-muscular coordination). For example, it is known that a significant amount of forward propulsion energy during load carriage walking comes from the ankle at late stance (Huang and Kuo, 2014). In addition, the precise timing between power generation by the trailing limb and power absorption by the leading limb significantly influences the metabolic cost of BW walking (Soo and Donelan, 2012). Task specific biomechanical knowledge may have important training implications when designing load carriage physical conditioning programs.

2.4.3 Summary

Three strategies work synergistically to improve an individual's load carriage physical performance and reduce injury risk: ergonomic, engineering, and physical conditioning. Of the three strategies, physical conditioning remains the most immediate and feasible form of intervention that has the potential to improve performance and reduce injuries. Most of the resistance training programs used to improve load carriage performance have not been explicitly informed from specific biomechanical demands of load carriage gait patterns, especially in running.

2.5 Load carriage running

For physical conditioning to be broadly effective for occupational and sporting athletes who must perform with load, the biomechanical demands of different load carriage gait patterns must be understood and trained. Load carriage is commonly performed while walking (Krupenevich et al., 2015), running (Brown et al., 2014b), jumping and landing (Brown et al., 2016), changing of directions (Brown et al., 2014a), crawling (Pandorf et al., 2002), and climbing (Pandorf et al., 2002). Despite the varied gait patterns routinely adopted while carrying load, most of the research on load carriage has focused on walking (Liew et al., 2016), and a significant gap exists in the study of other locomotion patterns, such as running.

This gap may exist for two reasons. First, it may have been perceived that locomotion patterns such as running are rarely performed with load. This is untrue given that running is very common during load carriage (Brown et al., 2014b; Clements et al., 2012; Cross et al., 2014; Simpson et al., 2017). Second, there may be a perception that the influence of load carriage on running can be inferred from walking studies. From a logical inferential perspective, physical conditioning designed to improve load carriage walking performance and injury risk, may similarly be expected to improve loaded running performance and injury risk. This section will emphasize the distinctive features between BW walking and running gait patterns, which makes load carriage running biomechanical requirements different from loaded walking requirements. This emphasizes the importance of investigating loaded running biomechanics to inform physical conditioning training development.

2.5.1 Work-energy requirements in walking and running

Muscles are the largest consumer of chemical energy during gait (Arellano and Kram, 2014; Grabowski et al., 2005). Muscles undergoing contraction that result in a change in muscle fibre length perform mechanical work which requires chemical energy (Hill, 1938). In addition, muscles undergoing isometric contractions perform no mechanical work but still consume chemical energy (Hill, 1938). Hence, it is useful to conceptualize the total energy requirement of muscle contractions as that due to the cost of workless contractions and the cost of work contractions (Latash and Zatsiorsky, 2016). Alternatively, Kram and Taylor (Kram and Taylor, 1990) proposed the cost of force hypothesis in running where the energy cost is inversely proportional to the stance period for force generation. This hypothesis is a more general theory of the energy cost incurred during muscle contractions, without distinguishing whether contractions are work or workless. When load is added to both running and walking, energy requirements increase to sustain constant velocity, as more muscle contractions are required (Griffin et al., 2003; Teunissen et al., 2007). The mechanistic reasons for this energy increase differ from running to walking, and can be better understood when differences in BW gaits are first explored.

In BW walking, most of the chemical energy consumed by muscles occurs during double-phase support, where a transition in step occurs (Donelan et al., 2002). During this phase, a portion of the body's energy is lost during the inelastic collision of the lead limb with the floor. Muscle contractions by the trailing limb redirect the centre of mass (COM) velocity vector from a forward-downward direction at initial contact, to a forward-upward direction during push-off (Donelan et al., 2002). Mechanical work is performed on the COM in the forward horizontal direction and vertical direction. In addition, workless contractions are needed for BW support even in this phase. A previous investigation reported that when walking with BW, 50 % of the net energy cost can be attributed to the forward horizontal mechanical work performed by muscles during the double-support phase (Gottschall and Kram, 2003). However, it is likely that the energy cost during the double-support phase in BW walking is greater than the reported proportion of 50 %. This is because energy needed to perform vertical mechanical work on the COM and to perform workless contractions for BW support during this phase was not quantified.

In contrast to walking, most of the chemical energy consumed by muscles in BW running occurs during single-support (Arellano and Kram, 2014). It has been reported that 70 % of the net energy cost associated with BW running is due to muscles generating vertical ground reaction force (GRF) (Arellano and Kram, 2014; Teunissen et al., 2007). Only less than 10 % of running's energy cost is associated with leg swing (Arellano and Kram, 2014). Part of the vertical GRF accelerates the body mass in the vertical direction beyond gravity (i.e. perform vertical mechanical work on COM), whilst part of it supports BW against gravity (workless contractions). It has been often argued that mechanical work during single-support in running can be recovered by stored elastic energy (Arellano and Kram, 2014). However, hysteresis in elastic elements of the muscle-tendon unit means that not all the stored elastic energy can be recovered to perform work (Lichtwark and Wilson, 2005). Compared with walking (Griffin et al., 2003), the energy cost associated with supporting BW is higher in running. This is due to a greater volume of muscle activation from an increased external GRF lever arm to individual joints' centre in running compared to walking (Biewener et al., 2004). The energy cost associated with generating horizontal anterior-posterior GRF in single-support approximates 40 % of the net energy cost of BW running (Arellano and Kram, 2014; Chang and Kram, 1999). However, there is a close coupling between vertical and horizontal GRF (Chang et al., 2000), such that the energy cost of generating vertical and horizontal GRF are not independent, but overlapping. When the interactive nature of vertical and horizontal GRF is considered, the energy cost of the single-support phase amounts to 80 % of the total energy cost of BW running (Arellano and Kram, 2014).

During load carriage running, a 30 % BW load increased net energy cost by 38 % relative to BW running (Teunissen et al., 2007). Most of the increased energy consumption in load carriage running is due to increased muscle contractions to support an increased total weight (Teunissen et al., 2007), and to perform vertical and horizontal mechanical work on the body and load. During load carriage walking, most of the increased energy consumption is due to the increased mechanical work performed by muscles to redirect the velocity vector of an increased mass during the double-support phase (Huang and Kuo, 2014; Krupenevich et al., 2015).

2.5.2 Injury mechanisms in walking and running

There are three broad mechanisms by which load carriage can increase the risk of neuro-musculo-skeletal injuries. First, carrying a load increases the risk of developing injuries that would otherwise be absent in gait without load carriage, such as brachial plexus neuropathy (Andersen et al., 2016). Second, load carriage can also increase the risk of developing overuse injuries that are intrinsic with BW gait (Andersen et al., 2016). For example, overuse injuries such as tibial stress fracture are highly prevalent in long distance BW running (Duckham et al., 2015), and load carriage may further exacerbate tibial bone loading (Xu et al., 2016). Lastly, load carriage can alter gait kinematics in such a way that tissue loading is exacerbated.

Running and walking have distinctive intrinsic injury potentials. For example, the prevalence of overuse lower limb injuries in recreational BW running can reach 20 % (Lopes et al., 2012). In contrast, the prevalence of overuse lower limb injuries in recreational BW walking is unknown, likely a consequence of a low prevalence. The primary reason for this may be due to intrinsic differences in the in-vivo tissue loads experienced during running and walking (Burr et al., 1996; Lafortune, 1991; Saxby et al., 2016; van den Bogert et al., 1999). For example, the compressive strain experienced at the tibia in BW running is 160 % greater than that experienced in BW walking (Burr et al., 1996). This difference in tissue loads is in turn a consequence of different dynamics between gait patterns. For example, running results in greater vertical GRF impact peak than walking (Chan et al., 2013), even at the same metabolic intensity (Swain et al., 2016). The difference in impact peak is due to the quicker landing velocity of the shank and foot segments during running than walking (Chi and Schmitt, 2005). A greater impact peak demands greater damping muscular activity (Boyer and Nigg, 2006), which in turn generates greater rate of tibial stress loading (Meardon et al., 2015). This could potentially explain the greater tibial loading experienced during load carriage running, compared to walking (Xu et al., 2016; Xu et al., 2017).

Differences in tissue loading between running and walking could be due to intrinsic differences in joint-level muscle activities and energetics needed to sustain constant velocity during load carriage (Xu et al., 2016; Xu et al., 2017). When loads of up to 20 % BW were carried in walking, the ankle, knee, and hip joint reaction forces increased by 15 %, 23 %, and 17 % respectively (Xu et al., 2016). When running, a 15 kg (20 % BW) symmetrical loaded vest increased peak patellofemoral and tibiofemoral joint contact forces by 4 % and 5 %, respectively (Willy et al., 2016b). This was lower than the 10 % increase in general knee joint reaction force with a 20 % BW load during running reported in another study (Xu et al., 2017). An important distinction between joint contact force (Willy et al., 2016b) and joint reaction force (Xu et al., 2017), is that the former includes forces imposed by muscle contractions, in addition to the loads imposed by measured accelerative inertial forces (i.e. force used to accelerate the inertia of bone segments). It may be that the addition of load to running does not increase muscle activities and forces as much as the accelerative inertial forces, as compared to walking. Surprisingly, the increase in knee joint reaction force during load carriage was greater in walking than in running (Willy et al., 2016b; Xu et al., 2016; Xu et al., 2017). One reason may be that the increase in knee joint torque was greater for the same load carried in walking, compared to running (Xu et al., 2016; Xu et al., 2017).

Load carriage may also elevate lower limb overuse injury risk due to maladaptive gait kinematics, such that the total tissue load becomes greater than that imposed by load carriage itself. The potential for gait mechanics to shift from optimal in BW, to sub-optimal in load carriage is likely greater in running than in walking. This is because muscles utilize a greater proportion of its force generation capacity in BW running than in walking (Arnold et al., 2013; Kulmala et al., 2014). As a result, it is easier for a muscle's force generating capacity to be exceeded during running with load carriage, compared to walking. This can result in aberrant running kinematics manifesting during load carriage running, increasing the risk of injuries. For example in BW running, a reduced force capacity of the trunk and hip frontal plane muscles in weight-bearing activities increases knee abduction angle (Cronstrom et al., 2016), the latter being associated with an elevated risk of developing anterior knee pain (Myer et al., 2010). It is not difficult to envisage that aberrant hip frontal plane kinematics can be exacerbated by load carriage, since external loads intrinsically increase the frontal plane external lever arm of the load to the hip (Neuman and Hase, 1994). This may potentially increase hip adduction angle during loaded running, and increase the stress on the knee joint complex, elevating the risk of developing anterior knee pain (Neal et al., 2016; Powers, 2010).

2.5.3 Summary

This section highlights the two features that distinguish load carriage running from walking. In load carriage running, single-support phase contributes to the greatest energy cost. This is different from load carriage walking where most of the energy is consumed during the double-support phase. The potential for injury in load carriage is also very different in running as compared to walking. This is due to intrinsic tissue loading differences inherent to different gait patterns. In addition, the potential for load carriage to cause aberrant gait kinematics is higher in running, than in walking, as running intrinsically utilizes a greater proportion of the muscles' force generation capacity compared to walking.

2.6 Adaptations of biomechanics in load carriage running

During constant velocity gait, the addition of load alters gait biomechanics. These biomechanical alterations can inform the design of physical conditioning training for load carriage. Biomechanical changes can be considered as either positive or negative adaptations to load. Positive adaptations reflect biomechanical strategies that aid in load carriage physical performance and tissue loading management. Negative adaptations reflect an inability to cope with the imposed load. By failing to adapt to load, negative biomechanical adaptations could exacerbate the negative influence of load carriage on physical performance, and further increase the magnitudes and asymmetrical patterns of stress on musculoskeletal tissue.

2.6.1 Adaptations in load carriage running

A smaller amount of muscle force would reduce load carriage energy consumption, and in-vivo tissue loading in loaded running. While carrying a 30 % BW load and running on a force plate instrumented treadmill at 3 m/s, vertical GRF increased by 12 % compared to BW running (Chang et al., 2000; Teunissen et al., 2007). Carrying a 30 % BW load also increased peak braking force by 6 % to 14 % and peak propulsive horizontal GRF by 18 % (Chang et al., 2000; Teunissen et al., 2007). During load carriage running, an alteration in vertical GRF was coupled with a proportional alteration in the horizontal anterior-posterior GRF (Chang et al., 2000). This coupling is thought to minimize the external lever arm of the resultant GRF to individual joints' centre (Chang et al., 2000). This would mean that a smaller amount of muscle force is needed to run with load, than if the external GRF lever were to be large (Chang et al., 2000).

An increase in leg stiffness may prevent excessive increases in joint torques during loaded running. This may in turn benefit loaded running energetics (Ackerman and Seipel, 2016) and tissue loading. Leg stiffness (or more accurately termed “quasi-stiffness” (Latash and Zatsiorsky, 1993)) represents the relationship between the GRF and change in lower limb displacement, and is thought to be an indirect global measure of lower limb “spring-like” behaviour (Brughelli and Cronin, 2008). During incremental load carriage using a weighted body vest of 10 % to 30 % BW and running at 3.3 m/s on a treadmill, leg stiffness increased from 35 % BW/LL (expressed as a percentage of body weight and inverse of leg length) during BW, to 44 % BW/LL at 30 % BW loaded running (Silder et al., 2015).

Accurate derivation of leg stiffness in running requires that the GRF variable used represents the portion of the GRF that is projected in line with the lower limb (Coleman et al., 2012). A closer alignment between the GRF vector and leg vector would increase leg stiffness, and may represent an energetically more optimal gait pattern (Moore, 2016). Unfortunately, the leg stiffness measured in Silder and colleagues (2015) did not project the GRF onto the lower limb. Instead, the vertical GRF was used which may overestimate leg stiffness magnitude in loaded running.

In a cohort of male military personnel, carrying an average composite load magnitude of 6 kg (light), 20 kg (medium) and 40 kg (heavy) whilst running overground at 3.5 m/s, load carriage of up to 40 kg (\approx 50 % BW) did not increase peak knee flexion angle (Brown et al., 2014b). This may be adaptive as greater knee flexion angle requires greater counteracting torques. However, a 30% BW load was found to increase peak knee flexion angle on treadmill running (Silder et al., 2015). These kinematic differences with load could be attributed to the study of experienced load carriers in Brown and colleagues (2014b), but novice load carriers in Silder and colleagues (2015). Surprisingly, given the attention provided to the investigation of transverse and frontal plane kinematics in determining injury risk in BW running (Esculier et al., 2015; Neal et al., 2016; Powers, 2003; Willy et al., 2012), the effects of load carriage on three dimensional joint angles have not been investigated in running.

The investigations of load carriage in running joint kinetics have focused on the knee and hip, but less on the ankle (Brown et al., 2014b; Xu et al., 2017). This is a significant gap given that the ankle extensors contribute significantly to weight support and forward propulsion in BW running (Hamner and Delp, 2013; Hamner et al., 2010). A heavy load magnitude (40 kg) increased internal peak knee extensor moment, and a medium load (20 kg) increased internal peak hip extensor moment, relative to a light load (6 kg) (Brown et al., 2014b). This could represent increased muscle forces to support an increased total weight (Hamner and Delp, 2013; Hamner et al., 2010). However, the effect of incremental load magnitude on ankle extensor moment was not reported in a previous study (Brown et al., 2014b), and only reported in a small study of four participants (Xu et al., 2017). Although loads of up to 30 % BW have been shown to increase ankle extensor moment in running, the small size precluded statistical inference (Xu et al., 2017).

During load carriage in running, similar to walking (Huang and Kuo, 2014), muscles must generate more power (and work) to sustain constant velocity. The total power needed to sustain constant velocity in load carriage can either be derived from a general increase in power performed by all the joints' muscles, and/or a shifting of power performed from one joint to another joint (Farris and Sawicki, 2012). Brown and colleagues (2014b) did not find a redistribution of power between joints during load carriage running. However, only inter-joint proportions of power were reported and it is unknown if load carriage would increase individual joint's absolute power (Brown et al., 2014b). In addition, measures of inter-joint power proportionality were based on measures of average stance power rather than instantaneous power (Brown et al., 2014b). Average power during gait does not consider the gait cycle-dependent effect of load carriage on muscle force output, since muscle fibre tension-velocity properties vary within the gait cycle of running (Arnold et al., 2013).

2.6.2 Summary

Load carriage changes running biomechanics. An adequate understanding of the positive and negative biomechanical changes to load carriage running, provides an avenue to design task specific physical conditioning programs. Current load carriage running biomechanical studies have focused largely on sagittal plane mechanics of the knee and hip, and at a single running velocity. It is unclear how load carriage changes the three-dimensional kinematics and kinetics of running, and its possible interaction with different running velocities.

2.7 Physical conditioning in load carriage – informing the “what” of training

2.7.1 Rationale

Translating background knowledge into foreground intervention relies on a framework that can identify biomechanical task features that need to be augmented (adaptive changes) and those that need to be mitigated (maladaptive changes) (O'Sullivan and Beales, 2007). Thus far, this framework has not been adopted in load carriage physical conditioning. To design a biomechanically informed load carriage training program for running, background mechanistic knowledge can be used to inform three training parameters, which includes: what muscle groups, muscle variables (hypertrophy, strength, power), and muscle contraction modes are needed in loaded running.

In this section, I will provide the evidence for an improved efficacy of training, when the three aforementioned training parameters are incorporated into training regimes. Load carriage can be understood as an experimental paradigm where the physical capacity of an individual is deficient relative to the task. In other words, carrying 200 N in load may be similar to weakening the muscle by 200 N. As such, it may be valuable to explore the effectiveness of this translational framework in athletic, as well as patient populations, where physical capacity relative to task demands is altered.

2.7.2 Specificity of muscle groups

Training task specific muscle groups and its timely activation may benefit load carriage running performance and injury risk, but have yet to feature in current physical training programs. As previously mentioned, load carriage increased mid-stance knee extensor moment and push-off ankle extensor moment in loaded running (Brown et al., 2014b; Xu et al., 2017), which may aid in impact shock absorption and forward propulsion, respectively. A biomechanically informed training program has been shown to be superior to general training in neurological patients. Stroke (cerebrovascular accident) is one of the most prevalent neurological conditions in the world that result in significant disability (Barclay et al., 2015). One of the most common forms of disability after stroke is reduced walking velocity (Dunn et al., 2015; Kramer et al., 2016), compared to healthy adults. A consistent mechanical cause for a reduced walking velocity in this cohort is a lack of ankle extensor torque and its timely activation during push-off in the paretic limb (Allen et al., 2014; Farris et al., 2015; Hall et al., 2011). This biomechanical deficit has been similarly shown in patients running after a traumatic brain injury (Williams et al., 2013).

One study which utilized this finding conducted a randomized controlled trial on chronic post stroke adults, comparing a treatment that was biomechanically informed (fast walking plus functional electrical stimulation to the ankle plantarflexors at push-off), to two other treatment arms (fast walking only and self-selected velocity walking). The most significant change after the intervention was that the between-time improvements in energy cost of walking (both fast and comfortable velocity) was greatest in the biomechanically informed treatment arm, over the other two treatment arms (Awad et al., 2016a; Awad et al., 2016b).

2.7.3 Specificity of muscular variables trained

Both power and strength oriented resistance training programs may benefit load carriage running performance, although the superiority of which remains uncertain. It was reported that higher countermovement jump power and higher one repetition maximum (1RM) squat magnitude were significantly associated with quicker timing on a load carriage 30 m military straight-line sprint and a zig-zag obstacle course (Mala et al., 2015). Individuals who were regular participants of resistance training, as compared to irregular participants, had greater jump power and 1RM squat strength, and quicker load carriage sprint completion times (Mala et al., 2015). In a BW countermovement jump, positive jump power relies heavily on knee extensor power (Raffalt et al., 2016), which collectively suggests that strength and power of the knee extensors are candidate training variables for load carriage running.

Current load carriage resistance training programs have adopted hypertrophy, endurance, strength, or power oriented training parameters either in isolation or in combination with each other (Knapik et al., 2012). There is some evidence that a power or a strength oriented training program could improve load carriage running performance more than a program focusing on other variables. However, prospective evidence for the superiority of power oriented training over strength oriented training in load carriage performance is lacking. For example, in a study which compared power oriented whole body resistance training to whole body hypertrophy training, there was a tendency towards statistical significance for quicker completion times on a two-mile load carriage run in the former compared to the latter (Kraemer et al., 2001). However, in a separate study which used 5RM training loads (pure strength oriented), two-mile load carriage running performance improved by 4 % from pre-training (Kraemer et al., 2004), while a program which used a periodised training load of 6RM to 12RM (mixed strength and hypertrophy oriented) improved on the same two mile loaded run by 11 % from pre-training (Hendrickson et al., 2010).

2.7.4 Specificity of contraction modes

It is possible that modification of contraction modes from slow-isoinertial to rapid stretch-shortening cyclic (SSC) mode (i.e. plyometric training) could improve the efficacy of resistance exercises on load carriage running performance and injury risk, although this has not been specifically investigated. Most of the exercises adopted in load carriage physical training have used a relatively slow isoinertial contraction mode (Hendrickson et al., 2010; Knapik et al., 2012; Kraemer et al., 2001; Kraemer et al., 2004). In contrast, muscles of the lower limb typically function in rapid SSC contraction mode in running (Lai et al., 2014). This may be especially important in load carriage given that load has been reported to increase leg stiffness during running (Silder et al., 2015).

Plyometric exercises, either in isolation or in combination with other contraction modes, have been shown to consistently improve BW running performance and economy (Alcaraz-Ibanez and Rodriguez-Perez, 2017; Balsalobre-Fernandez et al., 2016; Denadai et al., 2016; Giovanelli et al., 2017). There is also some evidence for the superiority of plyometric exercises (improvement by 4.83 %) compared to heavy-isoinertial exercises (3.65 %) in BW running economy, although this was not statistically significant (Denadai et al., 2016). Plyometric exercises may benefit running not only by improving maximal muscular strength and power (Beattie et al., 2017; Mikkola et al., 2011; Spurrs et al., 2003), but also improving leg and muscle-tendon unit stiffness (Beattie et al., 2017; Spurrs et al., 2003). An increase in local and global stiffness properties can increase elastic energy recycling (Lai et al., 2014), optimize a muscle fibre's length-tension-velocity relationship (Roberts and Azizi, 2011), and reduce external joint torque by aligning the GRF vector closer to the joint's centre (Chang et al., 2000).

The effect of different contraction modes on task specific functional improvements may also be dependent on an individual's baseline neuromuscular status (Radnor et al., 2017; Vissing et al., 2008). For example, plyometric exercises have been shown to produce greater improvements in maximal running acceleration, squat jump height and reactive strength index (ratio of jump height and stance duration) in pre-maturation adolescent compared to post-maturation adolescent (Radnor et al., 2017). The greater effect of plyometric training in pre-maturation adolescent may be due to the lower baseline neuromuscular coordination (Radnor et al., 2017). A lower baseline neuromuscular coordination may also explain why muscle power gains in a novice strength-trained adult cohort were greater after plyometric training than isoinertial training (Vissing et al., 2008).

2.7.5 Specificity of coaching cues

Coaching feedback cues, in the form of verbal, tactile, auditory, or a combination of cues have been shown to be effective in improving physical performance (Morgan et al., 1994) and preventing acute and overuse injuries (Davis and Futrell, 2016; Storberget et al., 2017). In load carriage running, verbally cueing an individual to increase their step rate by 7.5 % above their baseline step rate resulted in a 9% reduction in peak patellofemoral joint contact force (Willy et al., 2016b). In BW running, participants who engaged in three weeks of audio-visual feedback during treadmill running, had greater reductions in oxygen consumption during running, when compared to the same treadmill running without feedback (Morgan et al., 1994). One of the mechanisms by which coaching cues enhances physical performance and reduces injury risk is by augmenting the learning of more optimal kinematic movement patterns during sports-specific motor tasks (Benjaminse et al., 2015). For example, individuals who land during dynamic activities with excessive knee extension angle and moment, and with excessive knee valgus angle and moment are at a higher risk of sustaining a non-traumatic anterior cruciate ligament injury, as compared to those who land with more optimal angles and moments (Quatman et al., 2010). Coaching cues as simple as a verbal instruction could facilitate the learning of landing movement patterns that increased knee flexion angle and reduce valgus angle (Benjaminse et al., 2015).

Presently, studies that have used coaching cues in running (body weight or load carriage) have not integrated it within a resistance training program (Davis and Futrell, 2016; Willy et al., 2016a; Willy et al., 2016b), but instead embedded it solely within a gait retraining framework. It is possible that coaching cues can be added to a biomechanically informed resistance training program, to augment tasks-specific performance and injury risk reduction. For example, verbally instructing participants to perform a depth jump with the cue to “jump as high as possible” resulted in greater propulsive impulse during the concentric phase, compared to cueing to “jump as quickly as possible” (Louder et al., 2015). It is possible that by utilizing cues that augment power generation during each repetition of jump training, this could result in greater cumulative training stimulus to the lower limb muscles, resulting in greater lower limb muscular power improvements (Asadi and Ramirez-Campillo, 2016; Moreno et al., 2014).

2.7.6 Summary

This section highlights the importance of understanding the specificity of muscle groups, muscle variables and muscle contraction modes in the design of a biomechanically informed training program for loaded running. A training program informed from the biomechanical requirements of a motor task may better augment the training's efficacy at improving task relevant clinical outcomes.

2.8 Thesis landscape

In the load carriage literature, two significant gaps exist. One is the relative lack of understanding of how load carriage changes running biomechanics. Second, load carriage physical conditioning programs have not been informed from biomechanical knowledge of load carriage running. Hence, this doctoral thesis adds to the body of knowledge regarding running mechanics optimization during load carriage. Three studies were performed to address these gaps.

When performing biomechanical analysis of load carriage, the load may occlude reflective markers placed on the posterior pelvis, thus compromising the fidelity of kinematic data during load carriage biomechanics studies. Chapter 3 reports the findings of a Pilot study that was conducted to establish an alternate biomechanical marker set that could be used in subsequent load carriage experiments. Given the relative lack of studies in load carriage running biomechanics, Chapter 4 and Chapter 5 report the findings of Study one on running mechanics and its changes under varying combinations of load magnitude and running velocity. The stance phase of running was the focus of this thesis given that the energetic penalty of running is largely determined from single-support mechanics. In addition, known injury mechanical risk factors of BW running also occur largely within the single-support phase. The methods employed in Study one deal with two limitations in contemporary gait biomechanics research: the use of the "fractions" approach in quantifying mechanical work (Grenier et al., 2012; Latash and Zatsiorsky, 2016), and the sole use of discrete time point statistics (Pataky et al., 2015).

It has been commonly argued that minimal chemical energy is consumed by muscles to perform mechanical work in running (Arellano and Kram, 2014). One argument in support of this is that mechanical work performed over an arbitrary number of complete gait cycles at constant running velocity is zero. The apparent zero-work paradox occurs when the method used for calculating mechanical work is the “fractions” approach. In this approach, mechanical work is defined as work performed on the COM (“external” work), and work performed on segments about the COM (“internal”) (Aleshinsky, 1986). The “fractions” approach assumes two situations: 1) that energy is completely recycled, and 2) that complete inter-segmental energy compensation occurs (Aleshinsky, 1986). However, energy recycling is typically incomplete, and inter-segmental energy compensation is also incomplete (Williams and Cavanagh, 1983). Although the “fractions” approach has its limitations, it serves as a simple model to understand the energetic demands of gait and load carriage. More advanced understandings of the mechanical work demand of load carriage can be gleaned using inverse dynamics (“joint work”) and musculoskeletal modelling (“muscle and tendon work”). The latter approaches are known as the “source” approach as they focus on the work done by muscles rather than the work done on the COM (Aleshinsky, 1986). Inverse dynamics is the method used in this thesis.

Whilst Chapter 4 focuses on discrete outcome analysis, Chapter 5 focuses on waveform statistical analysis. Discrete outcome analysis alone may mask clinically important effects of load carriage on gait mechanics. Given that a muscle’s force generating capacity varies throughout the gait cycle, a combined discrete and waveform approach is argued to yield a more complete understanding of the biomechanical demands of load carriage running. This will provide greater knowledge to better inform the design of a load carriage specific physical conditioning program.

Chapter 6 reports the protocol of a Randomized Controlled Trial (RCT) adopted in Study two, that was designed to improve load carriage running mechanics. Importantly, it describes the rationale for exercise selection based on the findings reported in Chapter 4 and Chapter 5 of Study one, and that from Chapter 2. In the RCT, the effectiveness of a biomechanically informed “Targeted” training program was compared to a “General” resistance training program, in changing loaded running biomechanics. Chapter 7 and Chapter 8 report the results of the RCT, where Chapter 7 focuses on the results of strength, power, loaded running stiffness and mechanical work. Chapter 8 focuses on outcomes of loaded running kinematics.

Lastly, Chapter 9 summarizes the findings of the studies that were undertaken for this thesis, the limitations of the studies conducted and opportunities for further research.

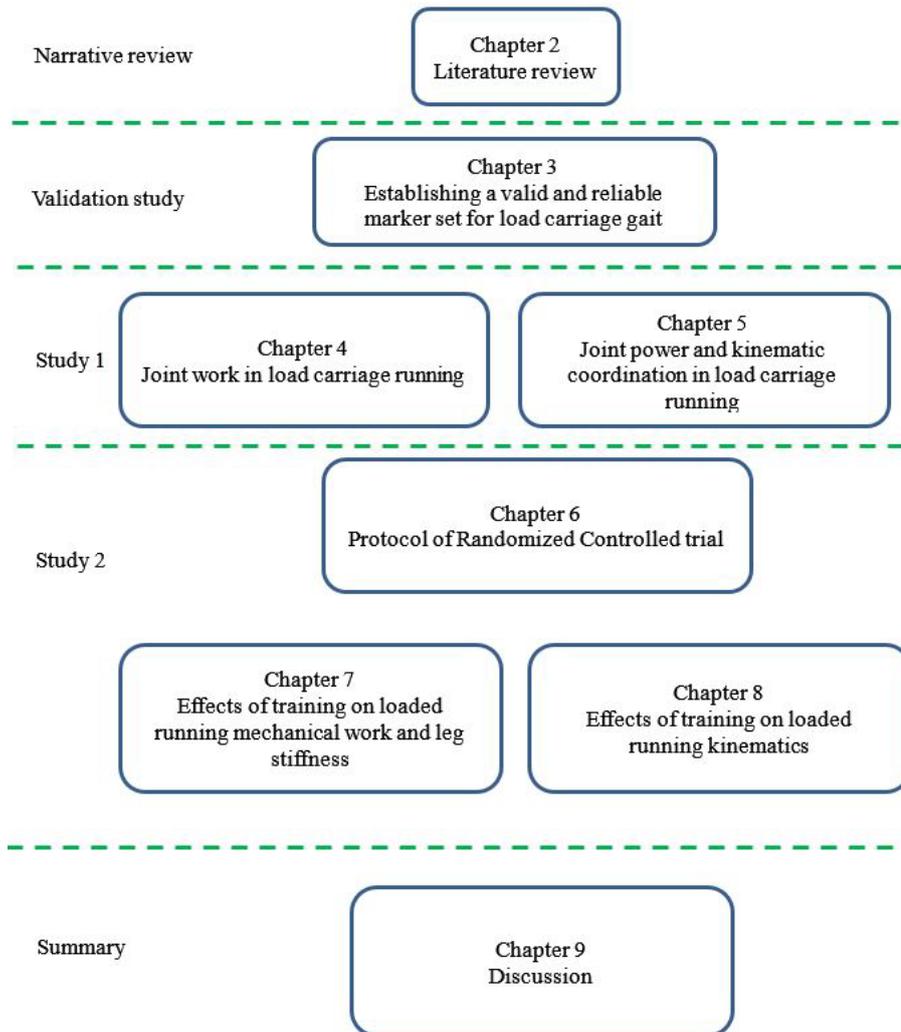


Figure 2-2 Flowchart of studies

2.9 Summary of key points

- Load carriage is increasingly being adopted for running tasks due to the expansion of load carriage activities into civilian ultra-endurance and adventure sports.
- Load carriage negatively influences physical performance and injury risks.
- Load carriage can increase injury risk due to increase tissue loading from the load itself; and it can also increase injury risk by exacerbating sub-optimal gait mechanics.
- The primary reason for reduced physical performance is due to the greater energy cost of load carriage gait. Muscles are the largest consumer of chemical energy, which implicates altered biomechanics as the determinant of an increased energy cost with load carriage.

- Biomechanical variables of the stance phase are the most relevant for the energy cost and injury mechanisms of running. For this reason, the focus of this thesis is the investigation of stance phase mechanics in load carriage running. Swing phase mechanics are presented in Appendix A to provide a complete descriptive report of load carriage running in Study one.
- Biomechanical changes to load carriage can be classified as positive (adaptive) or negative (maladaptive). The former reflects strategies aimed at augmenting physical performance and tissue loading management. The latter reflect failed biomechanical responses with load that could exacerbate the risk of overuse injuries and the deterioration of physical performance.
- Classifying biomechanical changes as adaptive or maladaptive assist in translating background mechanistic knowledge into foreground intervention design.
- There is indirect evidence that a biomechanically informed training program results in greater treatment efficacy, as compared to a non-specific program.
- Biomechanically informed training in load carriage running can only be designed with sufficient background understanding of load carriage running biomechanics. Presently, load carriage running biomechanics is an under investigated area of research.
- Consequently, this thesis aims to address two significant gaps in the load carriage literature. First, we will address the lack of in-depth biomechanical investigations on load carriage running. Second, we will also address the absence of a biomechanically informed load carriage physical conditioning program, specifically focused on optimizing loaded running mechanics.

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Chapter 3 Performance of a lateral pelvic cluster technical system in evaluating running kinematics

Synopsis: Biomechanical pelvic marker sets used in contemporary running research will be either visually occluded by the backpack from the cameras, and/or be physically displaced. To be able to perform valid experimentations on backpack running, I designed and validated a new lateral pelvic marker cluster set. When compared against the traditional International Society of Biomechanics pelvic marker set, the new lateral pelvic cluster produced comparable hip and pelvic kinematics. This provided me the confidence to pursue subsequent backpack running biomechanical research. This study has been published as “Liew BX, Morris S, Robinson MA, Netto K. Performance of a lateral pelvic cluster technical system in evaluating running kinematics. *J Biomech.* 2016 Jun 14;49(9):1989-93. doi: 10.1016/j.jbiomech.2016.05.010.”. The statement of primary contribution of the first author and the permission to reproduce the material can be found in the Appendix.

Abstract

Valid measurement of pelvic and hip angles during posterior load carriage gait task requires placement of pelvic markers which will not be occluded or physically displaced by the load. One solution is the use of pure lateral pelvic clusters to track the pelvis segment. However, the validity of this method has not been compared against pelvic marker systems recommended by the International Society of Biomechanics (ISB) during high impact tasks, such as running. The purpose of this study was to validate the lateral tracking pelvic clusters against the ISB pelvis during running. Six participants performed overground running at a self-selected running speed with shoes. Three dimensional motion capture and synchronised in-ground force plates were used to determine lower limb joint angles and gait events respectively. Two biomechanical models were used to derive pelvic segment and hip joint angles. The ISB pelvis used the anterior and posterior iliac spines as anatomical and tracking markers, whilst the other model used lateral pelvic clusters as tracking markers. The between participant averaged coefficient of multiple correlation suggested good to excellent agreement between the angle waveforms generated from the two marker protocols. In addition, both marker protocols had similar sensitivity in detecting three dimensional pelvic and hip joint angles during the stance phase. This study suggests that in the event posterior load carriage is involved in running gait, pelvic and hip kinematics can be measured by the use of lateral pelvic clusters.

Key Words: Kinematics; Gait; Validity; Running; Biomechanics

3.1 Introduction

The International Society of Biomechanics (ISB) recommends modelling and tracking the pelvic segmental coordinate system using the anterior and posterior superior iliac spine markers (ASIS, PSIS) (Wu et al., 2002). In recent years, posterior pelvic clusters have been used to track the pelvis in tasks involving significant hip flexion (Borhani et al., 2013; Vogt et al., 2003) to overcome the ASIS markers being occluded or displaced (Hara et al., 2014). However, posterior pelvic clusters may not be suitable in studies involving posterior load carriage, as these markers can be displaced or occluded during motion capture (Dames and Smith, 2015). One solution is to use lateral pelvic clusters during these motor tasks (Benedetti et al., 1998; Bruno and Barden, 2015; McClelland et al., 2010).

A concern on the use of lateral pelvic clusters is the presence of significant soft tissue artefact (STA), as this region contains greater soft tissue compared to the posterior pelvis (Schwenzer et al., 2010). In contrast, markers positioned on bony prominences are less likely to be affected by STA. STA influence has been shown to be greater during running (Dumas et al., 2014) compared to walking (Peters et al., 2010). However, previous studies have only evaluated the performance of lateral pelvic markers on walking (Benedetti et al., 1998; Bruno and Barden, 2015; McClelland et al., 2010), which may not be translated into running. In addition, previous studies using lateral pelvic markers used a composite lateral and posterior and/or anterior pelvic marker system (e.g. ASIS-iliac crest markers) (Bruno and Barden, 2015; Kisho Fukuchi et al., 2010; McClelland et al., 2010). This composite pelvic tracking system may have disadvantages during load carriage tasks involving significant hip flexion range, where both ASIS and PSIS markers can be occluded. Given that there is an increasing interest in the influence of posterior load carriage in running (Brown et al., 2014; Silder et al., 2015), there is a need to first validate the use of purely lateral pelvic clusters to track the pelvis during body weight running (i.e. running with no external load).

Given the emerging interest in load carriage running biomechanics, there is a need to validate a new pelvic marker tracking protocol, which does not require markers placed on pelvic bony prominences. Hence, the primary aim of this study was to validate the use of purely lateral pelvic clusters against the traditional ISB pelvic marker protocol, as a means to track the pelvis during running, when deriving variables of pelvic and hip angles.

3.2 Methods

3.2.1 Participants

Six healthy runners [five males, one female; mean (SD) age = 25.5 (4.0) years old; height = 1.72 (0.14) m; weight = 64.9 (12.5) kg; BMI = 21.6 (1.1)] participated in this study. Ethics approval was attained from the Curtin University Human Research Committee (RD-41-14) and written consent was sort from participants prior to study commencement.

3.2.2 Marker set placement and biomechanical model

Reflective markers were adhered to the pelvic, thigh, leg, and foot segments (Diamond et al., 2014). Two biomechanical models were created, the only difference being the method of tracking the pelvic segment (Supplementary material for details). For the ISB pelvis model, two ASIS and two PSIS markers were used to define the pelvic segment coordinate system (CS), and to track its motion. The origin of the pelvis anatomical coordinate system was defined as the mid-point between the ASIS markers. The (x-y) plane of the segment coordinate system is defined as the plane passing through the right and left ASIS markers, and the mid-point of the right and left PSIS markers. The x-axis was defined from the ORIGIN towards the Right ASIS. The z-axis was orthogonal to the (x-y) plane. The y-axis was the cross product of the x-axis and z-axis. The pelvic CS was defined using Visual 3D (version 5.0, C-Motion Inc., Germantown, USA) default pelvic CS, and does not follow the ISB's pelvic CS (Supplementary material for differences). For the pelvic cluster model, ASIS and PSIS markers were used to only model the pelvis, with all six markers on two lateral clusters (each cluster on each side of the pelvis) acting as tracking markers (Figure 3-1). The dimension of each cluster triad was identical for all participants (Supplementary material). All clusters positioned on the pelvis, thigh and shank were attached via double sided tape and rigid sports tape.

3.2.3 Experimental protocol

Trajectories and ground reaction force were captured using 18 motion capture cameras (Vicon T-series, Oxford Metrics, UK) (250 Hz), and synchronised in-ground force plates (AMTI, Watertown, MA) (1000 Hz), and stored using manufacturer supplied software (Vicon Nexus v2.1.1, Oxford Metrics, UK). All participants performed at least 10 over ground self-paced running trials (10 metre run up before, and 10 metres run off after the force plate). Initial contact and toe off was determined by a 20 N threshold in ground reaction force. All six participants wore their own running shoes.

3.2.4 Data processing

Visual 3D was used for post processing. Trajectories were filtered using a zero-lag, fourth order Butterworth (12 Hz) (Sinclair et al., 2014). Trajectories were normalized to 101 data points between initial contact and toe-off. Pelvic segment (relative to “virtual lab” axes) and hip joint angles were quantified using a ZYX and XYZ Cardan rotation sequence (Baker, 2001). A ZYX sequence was used for the pelvis as it produces pelvic rotation angles that more closely relates to clinical understanding of pelvic movement (Baker, 2001).

3.2.5 Statistical analysis

Overall between-protocol reliability was assessed using the coefficient of multiple correlations (CMC) (Ferrari et al., 2010), performed in Matlab (version 14a, MathworksInc.,USA). CMC is routinely used to provide a metric that summarizes the average waveform similarity between two marker protocols (Ferrari et al., 2010).

A one dimensional SPM paired t-test was performed in Python 2.7 (Canopy 1.5.2, Enthought Inc., Austin, USA) (Pataky et al., 2013). A statistical parametric map (SPM [t]) was created using the paired difference in mean angle waveforms (between the two protocols) at each normalized time point in the stance phase. Significance level was set at alpha (α) 0.0167, which was $\alpha = 0.05$ corrected for three comparisons per joint. Statistical inference was undertaken using random field theory (Adler and Taylor, 2007). Since biomechanical signals are routinely using one-dimension (e.g. time varying joint angles), and since there was no a priori expectation of when in a gait cycle differences in derived angles would occur between marker sets, statistical parametric mapping (SPM) enables a more robust statistical inference testing between the two protocols over an entire gait cycle (Pataky et al., 2013).

The “standard error of measurement (SEM)” (Hopkins, 2000) of discrete joint angles between the two protocols (Hopkins, 2000) was calculated at 20 % and 80 % of the stance phase. A post-hoc decision was made to identify the sensitivity at these two gait phases, as visual inspection suggested that the standard deviation of angle waveforms for each marker protocol were large at these two time points. SEM was calculated by dividing the standard deviation of the angle metric across gait trials by the square root of the number of gait trials (Hopkins, 2000).

3.3 Results

The mean (SD) running speed was 3.07 (0.19) m s⁻¹. Average CMC values varied from 0.87 for pelvic tilt to 0.98 for hip transverse rotation (Table 3-1). SPM paired t-test found that for the hip frontal plane, the ISB pelvis resulted in significantly greater hip adduction between 21 % to 29 % of the right stance phase (P = 0.00313) (Figure 3-2a). For pelvis frontal plane, the ISB pelvis also resulted in significantly greater pelvic obliquity compared to the cluster protocol between 14 % to 26 % of the right stance phase (P = 0.00073) (Figure 3-2b). The sensitivity for hip and pelvic angle varied from an average of 0.30° to 0.58° depending on the participant, marker protocol, joint angle and phase of gait (Table 3-2).

3.4 Discussion

Relatively few studies involving posterior load carriage in gait have reported pelvic and hip angles (Smith et al., 2006). This may be due to the difficulty in positioning markers on pelvic bony prominences during these tasks. Given the emerging interest in load carriage running research (Brown et al., 2014; Silder et al., 2015), this study demonstrated that pelvic and hip kinematics derived using purely lateral tracking pelvic clusters was comparable with that using the traditional ISB pelvis, in running.

This study reported relatively high between marker protocols CMC values, indicating that the pelvic and hip angles derived using both marker protocols correlated well. Our CMC values were higher than previously reported results, when posterior pelvic clusters were compared against the traditional ISB pelvic model, when considering participants with a BMI of < 24 kg/m² (Borhani et al., 2013). One reason could be that bilateral pelvic clusters were used, which increased the inter-marker separation distance in the medio-lateral axis, which minimizes propagation of error from markers to the bone position and orientation (POSE) (Cappozzo et al., 1997). In contrast, posterior pelvic marker clusters are usually spaced closely over the sacrum (Borhani et al., 2013), which increases the error propagating to bone POSE estimation (Cappozzo et al., 1997).

CMC values were greater for hip compared to pelvic angles, due to the dependence of CMC values on joint excursion (Ferrari et al., 2010). Participant four and five had relatively lower CMC values for pelvic tilt, and participant three had relatively lower CMC value for hip flexion (Table 3-1). Visual inspection of these individual's waveform data suggests that there was a small temporal phase lag in angles derived from the two marker protocols, which could contribute to a reduced magnitude of CMC values. A previous investigation also reported slightly greater phase lag in pelvic tilt angles in walking between the ISB pelvis method and a composite pelvis tracking method (ASIS-iliac crest markers) (Bruno and Barden, 2015). Variation between participants could also be due to variations in manual identification of iliac bony prominences, which could differ between participants by as much as 20 mm (della Croce et al., 1999). This could influence derived joint angle magnitudes, which could in turn influence the magnitude of calculated CMC values.

Despite high CMC values, significant differences were detected at approximately loading response of stance. CMC negates within-cycle variability of biomechanical signals and does not provide a measure of statistical inference. The lateral pelvic clusters resulted in smaller hip adduction and pelvic obliquity compared to the ISB pelvis. Angles derived from the lateral pelvic clusters may be more sensitive to soft-tissue artefact from high hip abductor muscle activity, especially during the high impact phase of initial contact to loading response of running (Chumanov et al., 2012). However, no study to our knowledge has quantified the relationship between hip muscle activity and pelvic STA. Despite the significance, these differences were small (hip difference at 25 % = 0.73°; pelvic difference at 20 % = 1.11°) (McGinley et al., 2009), and likely clinically acceptable. The sensitivity of each marker protocol was similar in magnitude, for all cardinal planes and all participants. Although the sensitivity of the lateral pelvic clusters used in this study was greater compared to previous research using a composite lateral pelvic protocol (Bruno and Barden, 2015), the significance of this finding is unclear, given that clinically meaningful changes in kinematics have not been defined.

A limitation of this study was the use of a relatively homogenous dimensioned participant group (mean BMI of 21.6). Although a previous study which used a composite lateral pelvic protocol validated it in participants with a range of BMI (23-43), only walking was investigated. Using lateral tracking pelvic clusters in an overweight population during running requires a specific validation study. In conclusion, purely lateral tracking clusters may be used as an alternate form of tracking the pelvis in load carriage running studies, where posterior pelvic markers may not be feasible.

3.5 References

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3.6 Figures

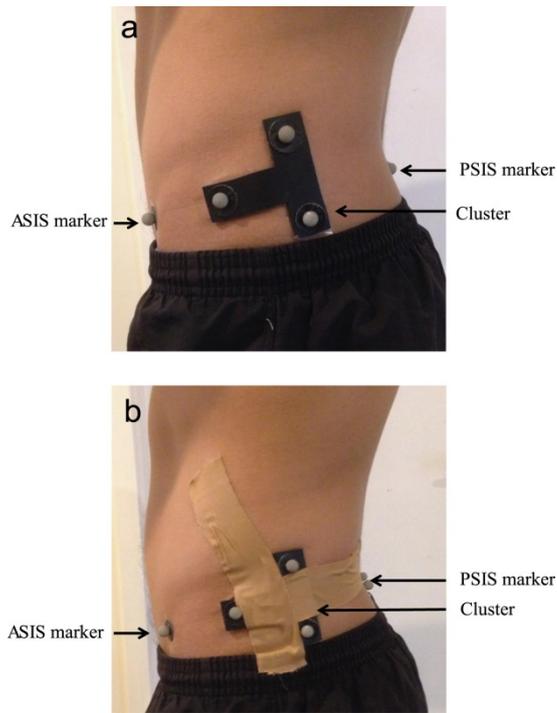


Figure 3-1 Position of lateral pelvic marker clusters (lateral view).

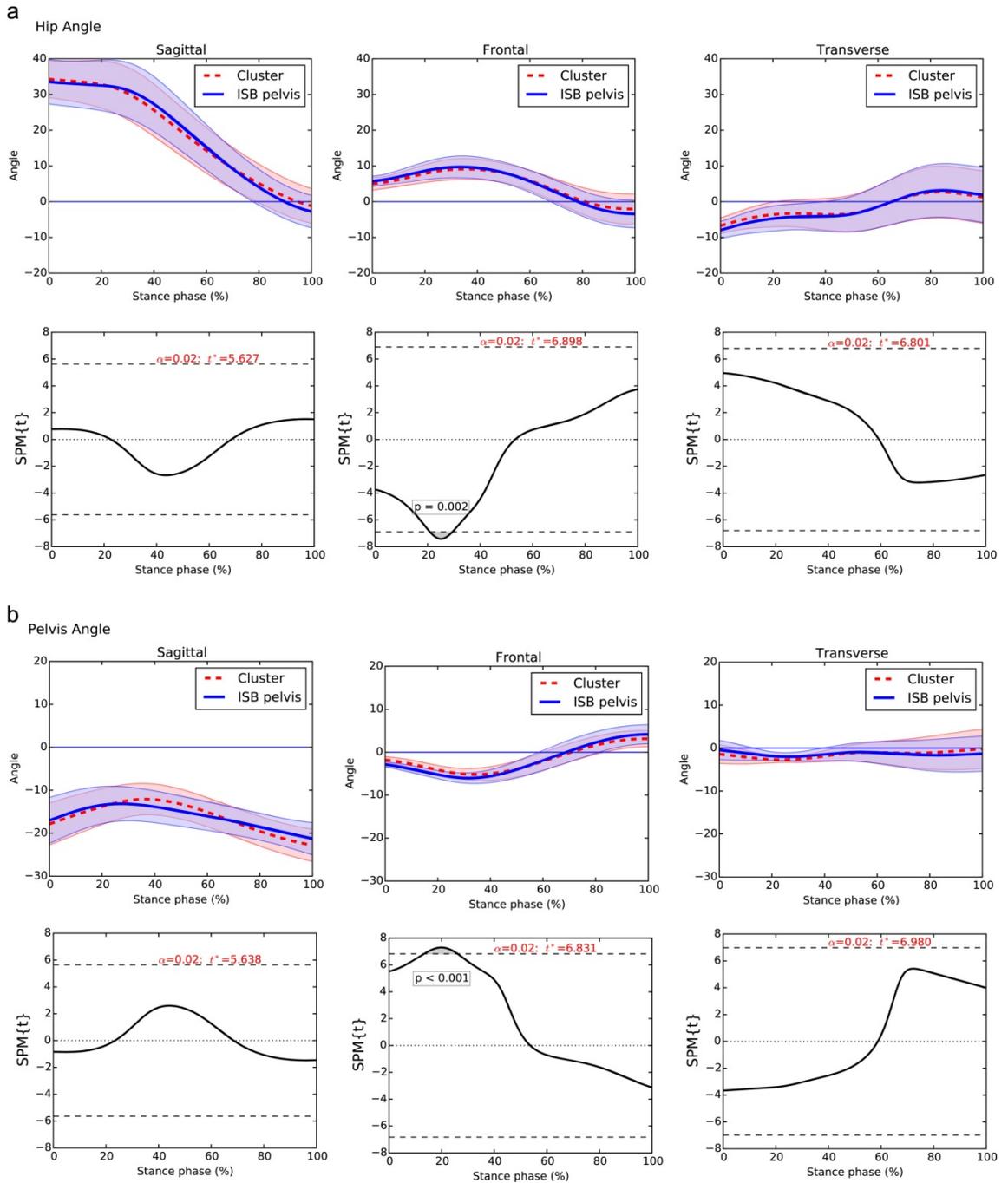


Figure 3-2 a. Hip joint angle group mean (SD) in the top row: Joint angles derived from ISB pelvis and pelvic cluster protocol in the sagittal (left), frontal (middle), and transverse (right) plane. SPM paired t-test results in the bottom row of angles in the sagittal, frontal, and transverse plane. The horizontal dotted line represents the critical random field threshold of $t = 0.0167$.

b. Pelvis segment angle group mean (SD) in the top row: Joint angles derived from ISB pelvis and pelvic cluster protocol in the sagittal (left), frontal (middle), and transverse (right) plane. SPM paired t-test results in the bottom row of angles in the sagittal, frontal, and transverse plane. The horizontal dotted line represents the critical random field threshold of $t = 0.0167$.

3.7 Tables

Table 3-1 Between-protocol (ISB vs Cluster pelvic protocols) coefficient of multiple correlations

Participant	Pelvis			Hip		
	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
1	0.87	0.97	1.00	0.99	0.99	1.00
2	0.98	0.98	0.97	1.00	0.99	0.99
3	0.90	0.95	0.81	0.75	0.99	0.96
4	0.75	0.92	0.87	0.99	0.83	0.96
5	0.73	0.99	0.84	0.92	0.99	0.99
6	1.00	1.00	1.00	0.91	0.99	0.98
Average	0.87	0.97	0.91	0.93	0.96	0.98
Sagittal plane: Hip = flexion/extension, Pelvis = anterior/posterior tilt; Frontal plane: Hip = abduction/adduction, Pelvis = obliquity; Transverse plane: Hip = rotation, Pelvis = rotation.						

Table 3-2 Within protocol standard error of measurement (°)

Subject (% gait cycle)	ISB model (°)						Cluster (°)					
	Hip			Pelvis			Hip			Pelvis		
	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
1 (20 %)	0.60	0.50	0.56	0.35	0.4	0.99	0.67	0.46	0.52	0.46	0.34	0.97
1 (80 %)	0.56	0.21	0.20	0.38	0.33	0.73	0.61	0.18	0.20	0.45	0.30	0.71
2 (20 %)	0.28	0.09	0.28	0.18	0.18	0.31	0.28	0.10	0.29	0.19	0.17	0.30
2 (80 %)	0.24	0.14	0.24	0.17	0.09	0.17	0.25	0.14	0.23	0.15	0.09	0.16
3 (20 %)	0.70	0.84	0.24	0.47	0.63	0.89	0.62	0.84	0.23	0.45	0.62	0.88
3 (80 %)	0.66	0.35	0.23	0.30	0.35	0.64	0.68	0.36	0.17	0.42	0.38	0.67
4 (20 %)	0.58	0.18	0.28	0.29	0.28	0.33	0.61	0.18	0.27	0.31	0.24	0.34
4 (80 %)	0.38	0.14	0.27	0.26	0.12	0.49	0.39	0.14	0.27	0.26	0.13	0.48
5 (20 %)	0.56	0.34	0.38	0.46	0.38	0.68	0.55	0.34	0.40	0.46	0.37	0.68
5 (80 %)	0.47	0.20	0.29	0.37	0.42	0.48	0.50	0.19	0.29	0.40	0.38	0.51
6 (20 %)	0.31	0.35	0.17	0.23	0.22	0.56	0.36	0.30	0.15	0.27	0.18	0.58
6(80 %)	0.19	0.27	0.35	0.26	0.30	0.71	0.19	0.28	0.35	0.25	0.28	0.72
Average	0.46	0.30	0.29	0.31	0.31	0.58	0.48	0.29	0.28	0.34	0.29	0.58

Sagittal plane: Hip = flexion/extension, Pelvis = anterior/posterior tilt; Frontal plane: Hip = abduction/adduction, Pelvis = obliquity; Transverse plane: Hip = rotation, Pelvis = rotation.

3.8 Supplementary material

Table 3-3 Marker placements

Segment	Marker	Description	Present in both Static / Dynamic trials
Pelvis			
ISB model (traditional)	RASI, LASI	Right and Left Anterior Superior Iliac Spine	Both
	RPSI, LPSI	Right and Left Posterior Superior Iliac Spine	Both
Cluster model (new)	RASI, LASI	Right and Left Anterior Superior Iliac Spine	Both
	RPSI, LPSI	Right and Left Posterior Superior Iliac Spine	Both
	R/L ILA1, ILA2, ILA3	Right and Left clusters of lateral pelvic markers	Both
Thigh (Common to both models)	R/L LFC, MFC	Right and Left lateral and medial femoral epicondyles	Static
	R/L TH1, TH2, TH3	Right and Left triad of thigh clusters. Long axis of shell runs along long axis of lateral thigh. Short axis of shell wraps around the anterior thigh	Both
Shank (Common to both models)	R/L TIBF	Right and Left anterior surface of shank along longitudinal axis of shank	Static
	R/L TB1, TB2, TB3	Right and Left triad of shank clusters. Long axis of shell runs along long axis of lateral shank. Short axis of shell wraps around the anterior shank	Both
	R/L LMAL, MMAL	Right and Left lateral and medial malleoli	Static
Foot (Common to both models)	R/L CAL	Right and Left middle of calcaneus	Both
	R/L RMT1	Right and Left 1st metatarsal head	Both
	R/L RMT5	Right and Left 5th metatarsal head	Both
Abbreviations: R- Right; L-Left			

Table 3-4 Range of motion (ROM) (°) measured using the ISB pelvis

Participant	Pelvis			Hip		
	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
1	8.80	8.78	5.69	35.46	15.92	3.73
2	8.58	10.32	6.24	37.13	13.03	19.03
3	8.90	9.40	4.60	34.32	13.06	6.24
4	7.52	10.03	3.32	40.39	6.36	9.56
5	10.23	12.99	4.81	33.73	14.52	19.03
6	5.95	10.78	7.80	39.13	17.75	14.83
Mean (SD)	8.33 (1.45)	10.38 (1.46)	5.41 (1.54)	36.69 (2.67)	13.44 (3.91)	12.07 (6.54)

3.8.1 Definition of joint centres and co-ordinate systems

The z-axis of the thigh ACS in both models was oriented as the line joining the hip and knee joint centre with its positive direction proximal. The y-axis was orthogonal to the frontal plane with its positive direction anterior. The frontal plane was formed by the lateral epicondyle, hip and knee joint centre. The x-axis was then defined as the cross-product of z- and y-axes with its positive direction lateral.

The z-axis of the shank ACS was the line passing from the ankle joint to the knee joint centre. The x-axis was orthogonal to the plane defined by the ankle and knee joint centre, and the tibial tubercle, with its positive direction lateral. The y-axis was then defined as the cross-product of the z- and x-axes with its positive direction anterior.

In the foot ACS, the z-axis was defined as the line joining the mid-point between the two malleoli, and the mid-point between the two foot markers (1st and 5th MTP). The y-axis was orthogonal to the plane defined by the two malleoli and two foot markers (1st and 5th MTP). The x-axis was the cross-product of the z- and y-axes with its positive direction lateral. Rotation of the foot ACS was not performed in this study, as ankle angles were not the variables of interest.

We also created a virtual lab with the z-axis defined as the vertical axis (positive proximal), y-axis as the anterior posterior axis (positive anterior), and the x-axis defined as the medio-lateral axis (positive lateral). The virtual lab was created as segmental angles were made relative to the virtual lab.

3.8.2 Differences between ISB's pelvic segment coordinate system (PCS) and that used in this study

First, ISB defines the origin of the pelvis at the hip joint centre, while this study defined the pelvic origin as the mid-point between the two ASISs markers. Second, the nomenclature of the axis recommended by ISB had the Y-axis as the vertical axis, Z-axis as the medio-lateral axis, and X axis as the anterior-posterior axis (following the right hand rule). In contrast, this study used Visual 3D's default axis naming convention: Z-axis as vertical axis, Y-axis as anterior-posterior axis, and X-axis as medio-lateral axis (following the right hand rule).

3.8.3 Marker shells design (scaled to actual size)

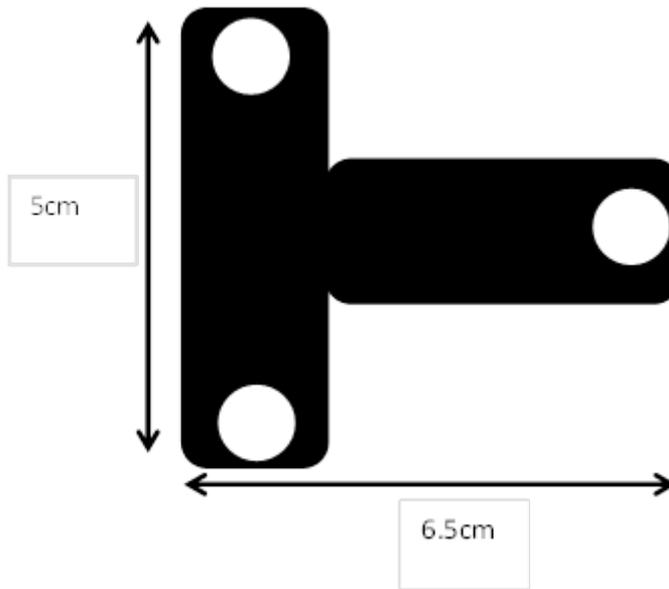


Figure 3-3 Pelvic clusters

3.9 Conflict of interest and source of funding

No funds were received in support of this work. No benefits in any form have been or will be received from a commercial party related directly or indirectly to the subject of this manuscript. Mr. Bernard Liew is currently supported by an institutional doctoral scholarship.

3.10 Acknowledgement

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Chapter 4

The effects of load carriage on joint work at different running velocities

Synopsis: A biomechanically informed program requires knowledge of task specific muscle groups for training. Muscles need to consume chemical energy to perform mechanical work during running. When load is added, running velocity inherently decreases likely as a strategy to reduce energy expenditure. To sustain constant velocity, muscles must perform more mechanical work. This study investigates the change in mechanical work when running under varying load magnitudes at varying running velocities. The results of this study suggest that all joint-level muscle groups are important to load carriage running. Of smaller importance to the adaptation of load carriage running is the shift in work from the hip to the ankle. These suggest that a biomechanically informed training for loaded running needs to involve all muscle groups, rather than a focus on a single muscle group. This study has been published as “Liew BX, Morris S, Netto K. The effects of load carriage on joint work at different running velocities. *J Biomech.* 2016 Oct 3;49(14):3275-3280. doi: 10.1016/j.jbiomech.2016.08.012.”. The permission to reproduce the material can be found in the Appendix.

Abstract

Running with load carriage has become increasingly prevalent in sport, as well as many field-based occupations. However, the “sources” of mechanical work during load carriage running are not yet completely understood. The purpose of this study was to determine the influence of load magnitudes on the mechanical joint work during running, across different velocities. Thirty-one participants performed overground running at three load magnitudes (0 %, 10 %, 20 % body weight), and at three velocities (3, 4, 5 m/s). Three dimensional motion capture was performed, with synchronised force plate data captured. Inverse dynamics was used to quantify joint work in the stance phase of running. Joint work was normalized to a unit proportion of body weight and leg length (one dimensionless work unit = 532.45 Joules). Load significantly increased total joint work and total positive work and this effect was greater at faster velocities. Load carriage increased ankle positive work (β coefficient = rate of 6.95×10^{-4} unit work per 1 % BW carried), and knee positive ($\beta = 1.12 \times 10^{-3}$ unit) and negative work ($\beta = -2.47 \times 10^{-4}$ unit), and hip negative work ($\beta = -7.79 \times 10^{-4}$ unit). Load carriage reduced hip positive work and this effect was smaller at faster velocities. Inter-joint redistribution did not contribute significantly to increased total work within the spectrum of load and velocity investigated. Hence, the ankle joint contributed to the greatest extent in work production, whilst that of the knee contributed to the greatest extent to work absorption when running with load.

4.1 Introduction

Load carriage has been investigated in occupational, ergonomic, injury and performance settings (Knapik et al., 2004), with current research focused on the effects of load magnitude on the biomechanical and metabolic demands while walking (Huang and Kuo, 2014). However, recent studies have investigated load carriage during running, jump-landing, and side-step cutting tasks, which highlights the growing importance of research into more dynamic tasks during load carriage (Brown et al., 2014a; Brown et al., 2016; Brown et al., 2014b). Running with load is not only relevant to military personnel (Brown et al., 2014a), but also to non-military personnel as an increasing number of people are commuting to work on foot (Zander et al., 2014), and participating in ultra-endurance races (Hoffman and Wegelin, 2009). All of these pursuits routinely require load carriage running (Marais and de Speville, 2004). With the growing prevalence of load carriage running as an activity, more research into the biomechanics of loaded running is needed (Brown et al., 2014b; Silder et al., 2015). This knowledge is essential for the future development of pre-conditioning programs for individuals needing to run with load.

An important area of research in many biomechanical studies is quantifying and explaining the “source” of mechanical work (or its time derivative: power) (Aleshinsky, 1986) and its alterations during load carriage (Brown et al., 2014b; Huang and Kuo, 2014). For example, the increased metabolic cost of walking during load carriage (Huang and Kuo, 2014), has been attributed to an increased ankle positive work during push-off (Huang and Kuo, 2014). Although an increase in metabolic cost during load carriage was observed in running (Teunissen et al., 2007), a recent investigation reported that the percentage average positive power of the hip, knee, and ankle joints during the stance phase of running did not alter with load carriage (Brown et al., 2014b). Differences in the effects of load on joint work between walking and running could be due to inherent differences in joint-level work distribution between varied gait patterns (Farris and Sawicki, 2012; Schache et al., 2015). In addition, a lack of change observed in the study could be that analysis was performed only on relative joint average power contributions, with no absolute metrics being reported (Brown et al., 2014b). Relative joint power provides no indication of the absolute effect of load carriage on joint work in running.

When load is added, greater mechanical work is performed to sustain gait velocity (Huang and Kuo, 2014). The increase in mechanical work with added load in running was only investigated in the stance phase of running (Brown et al., 2014b). Previous research into unloaded running (running with no external load) at 3.0 to 4.0 m/s reported that leg swing contributed only 7 % of the net metabolic cost of unloaded running (Arellano and Kram, 2014). This dominant stance phase effect of load carriage on joint work was also demonstrated in walking (Huang and Kuo, 2014). Greater mechanical work during load carriage running may involve an increase in joint work magnitude in all three joints, and/or inter-joint work redistribution (Farris and Sawicki, 2012). The effect of load on running joint work may also differ depending on running velocity, which is an important but yet poorly investigated area, given that load running may occur at varying velocities (e.g. 3.0 m/s to 5.0 m/s is common in ultramarathon runners) (Senefeld et al., 2016).

To this end, no studies have adequately investigated the joint-level mechanical work involved in load carriage running at varying velocities. Hence, the primary aim of this study was to determine the effect of load magnitude on lower limb joint work across three running velocities. It was hypothesized that across the range of velocities, total joint work would increase with load magnitude and that individual joint contributions to total work would also increase with load.

4.2 Methods

4.2.1 Participants

Thirty-one healthy participants (16 males and 15 females) enrolled in this study [mean standard deviation (sd) age = 30.8 (5.9) years old; height = 1.70 (0.08) m; mass = 66.4 (10.8) kg; hours run per week = 3.73 (2.86) hrs. All participants provided written informed consent prior to participation and the study was approved by the Curtin University Human Research Ethics Committee (RD-41-14) prior to commencement.

4.2.2 Biomechanical model

Anatomical markers were taped to the following anatomical landmarks: bilateral first and fifth metatarsal heads, posterior mid-calcaneus, superior and inferior aspects of mid-calcaneus, medial and lateral malleoli, medial and lateral femoral condyle, anterior superior iliac spines (ASIS), and posterior superior iliac spines (PSIS). Tracking marker clusters were placed bilaterally on the lateral surface of the pelvis, thigh, and shank. Anatomical markers were removed after calibration trials. Hip joint centres were calculated using a regression equation (Bell et al., 1989). Knee and ankle joint centres were calculated as the midpoint between the medial and lateral femoral condyles, and between the medial and lateral malleoli respectively (Pohl et al., 2010). Segment inertial and geometric properties were based on Dempster et al. (1955) (Dempster, 1955) and Hanavan et al. (1964) (Hanavan, 1964) respectively.

Inverse kinematic (IK) modelling was performed using Visual 3D (C-motion, Germantown, MD) (Lu and O'Connor, 1999). The pelvis segment relative to the global coordinate system (GCS) had six degrees of freedom (DOF), whilst the thigh relative to pelvis, shank relative to thigh, and foot relative to shank link each had three DOF (three rotational mobilizers) (Robinson et al., 2014). Segment weighting factors of four were given to the pelvis and foot segments, three for the shank, and two for the thigh. The position and orientation (POSE) of the model was calculated for each frame using a weighted least-squared Levenberg-Marquardt global optimization algorithm (Robinson et al., 2014). Details of the biomechanical lower limb model can be found in the Supplementary material.

4.2.3 Protocol

All participants wore their personal running shoes. For the experiment, participants were given 10 to 15 metres (m) before the first force platform (three non-camouflaged platforms in series totalling a length of 3 m in the line of progression) (AMTI, Watertown, MA) to achieve the desired steady velocity, and a 10 metre deceleration distance after the last force platform. Participants were verbally instructed to run at three prescribed velocities (3.0 m/s, 4.0 m/s, 5.0 m/s), over three load conditions (0 %, 10 %, 20 % body weight (BW)).

The order of presentation was randomized by generating 31 different sequences of load-velocity condition using a random sequence generator (<https://www.randomizer.org/>). Sand bags filled to the prescribed weight, were carried in a backpack (CAMELBAK, H.A.W.G.® NV, 14 litre), which was secured via adjustable chest strap and waist belt. Timing gates (SMARTSPEED Pro, Fusion Sport Pty Ltd, Australia) positioned on both sides of the force plates 5 m apart, were used to measure running velocity. A familiarization period (maximum five minutes) was given before each condition. Five successful running trials for each velocity-load condition were required. A successful trial was defined by the attainment of the prescribed velocity within a $\pm 10\%$ velocity variance window and when no visible targeting of the force platform was made. Each trial was interspersed with a 30 s standing rest break.

4.2.4 Data measurement and processing

Kinematic data were captured using an 18 camera motion capture system (Vicon T-series, Oxford Metrics, UK) (250 Hz). Ground reaction force (GRF) was measured using synchronised force plates (2000 Hz). Data were captured and stored using manufacturer supplied software (Vicon Nexus, v2.1.1, Oxford Metrics, UK). Data processing was performed in Vicon Nexus and Visual 3D. Marker trajectories and GRF were filtered at 18 Hz (4th order, zero-lag, Butterworth) (Robinson et al., 2014; van den Bogert and de Koning, 1996).

Positive and negative joint work was calculated by integrating the positive and negative power of the ankle, knee, and hip joints of the right leg, over the stance phase. Initial contact and toe-off were defined by a GRF threshold of 20 N. Total positive and negative lower limb work were calculated by summing individual joint's positive and negative work over the stance phase, respectively. The net lower limb work in stance was calculated by taking the algebraic sum of the positive and negative joint work. The total lower limb work in stance was calculated by taking the absolute sum of the positive and negative joint work. Joint work was expressed as dimensionless units by normalizing base factors of body mass, M (kg), gravitational constant g (9.81 m/s^2), and leg length, L (m) [MLg (mean (SD) scaling factor = 532.45 (111.25))] (Huang and Kuo, 2014).

4.2.5 Statistical analysis

Descriptive statistics were calculated for the dependent variables for each repeated condition. All statistics were performed in R (R Core Team, 2015) within RStudio (Version 0.98.1062, RStudio, Inc.). Linear mixed models were used to evaluate the linear relationships between the dependent variables, and the predictor variables (velocity and load magnitude) (West et al., 2014). Velocity and load were treated as fixed continuous variables, while a random intercept and random slope effect of load for each participant was included. The load by velocity interaction fixed effect was removed from the model when an F-test, based on the restricted maximum likelihood (REML) estimate, revealed non-significance at $P < 0.05$.

4.3 Results

4.3.1 Spatio-temporal parameters

Stride duration (i.e. time between right to right initial contact) was significantly related to velocity ($t(245) = -14.22, P < 0.001$) and load ($t(245) = -9.03, P < 0.001$) (Table 4-1). The effect of load magnitude on stance duration was dependent on running velocity ($\beta_{\text{Load:Speed}}: t(245) = -4.03, P < 0.001$). Increasing load increased stance duration but this increase was attenuated by increasing velocity. Increasing load increased step length ($\beta_{\text{Load:Speed}}: t(245) = -4.99, P < 0.001$) and stride length ($\beta_{\text{Load:Speed}}: t(235) = -4.02, P < 0.001$), but this increase was attenuated at faster velocities (Table 4-1).

4.3.2 Total and net lower limb work

Increasing load reduced total joint positive work ($\beta_{\text{Load:Speed}}: t(245) = 2.73, P = 0.0068$) and increased total joint work ($\beta_{\text{Load:Speed}}: t(245) = 2.29, P = 0.023$) in running, but the reduction in total positive work was smaller at faster velocities, whilst the increase in total work was magnified at faster velocities (Table 4-1, Figure 4-1). Increasing load magnitude ($t(246) = -8.21, P < 0.001$) and increasing running velocity ($t(246) = -17.86, P < 0.001$) independently increased the magnitude of total negative work (Table 4-1, Figure 4-1). There was no significant relationship between load magnitude and velocity on net joint work (Table 4-1).

4.3.3 Positive and negative joint work

Increasing load magnitude increased ankle ($t(246) = 7.55, P < 0.001$) as well as knee joint positive work ($t(246) = 10.84, P < 0.001$) (Table 4-1, Figure 4-2). Further, increasing running velocity increased ankle ($t(246) = 21.20, P < 0.001$) and knee joint positive work ($t(246) = 9.04, P < 0.001$). Increasing load reduced hip positive work ($\beta_{\text{Load:Speed}} : t(245) = 2.32, P = 0.021$), but this effect was attenuated at faster running velocities (Table 4-1, Figure 4-2). Increasing load magnitude resulted in increased negative work performed at the ankle ($t(246) = -6.82, P < 0.001$), knee ($t(246) = -2.56, P = 0.011$), and hip joint ($t(246) = -5.06, P < 0.001$) (Table 4-1, Figure 4-3). Increasing velocity also significantly increased the magnitude of negative work performed at the ankle ($t(246) = -11.54, P < 0.001$), knee ($t(246) = -10.00, P < 0.001$), and hip joint ($t(246) = -11.18, P < 0.001$) (Table 4-1, Figure 4-3).

4.3.4 Percentage joint work contribution

Increasing load reduced the percentage hip ($\beta_{\text{Load:Speed}} : t(245) = 2.36, P = 0.02$) and increased the ankle ($\beta_{\text{Load:Speed}} : t(245) = -2.18, P = 0.03$) positive work contribution, but this effect was reduced at faster velocities (Table 4-1 and Table 4-2). Increasing load ($t(246) = 10.12, P < 0.001$) and increasing velocity ($t(246) = -3.05, P = 0.002$) both increased percentage knee positive work (Table 4-1 and Table 4-2). With respect to percentage negative work, increasing load increased the hip ($t(246) = 3.80, P < 0.001$) and knee ($t(246) = -6.40, P < 0.001$) contribution, and increasing velocity increased the hip ($t(246) = 4.81, P < 0.001$) and knee ($t(246) = -6.52, P < 0.001$) contribution (Table 4-1 and Table 4-2). There was no effect of load and velocity on percentage ankle negative work (Table 4-1 and Table 4-2).

4.4 Discussion

This present investigation evaluated the effects of varying load magnitude on absolute joint work while running across three velocities. The main hypothesis of this study was supported in that load carriage running required greater total joint work, compared to unloaded running. The secondary hypothesis was not supported in that load carriage did not increase the magnitude of positive and negative work at all three joints.

This study provides the first evidence that running with load carriage resulted in greater total joint work. In addition, this study also found that load carriage increased absolute total negative and positive joint work during the stance phase of running. The increase in total negative work during constant velocity load carriage running means that there was an increase in energy absorption by the lower limb during the stance phase (Winter, 1983). Part of this negative work occurs during the first half of stance, as the ankle and knee extensors eccentrically contract to decelerate the body, reducing both the potential and kinetic energy of the body segments plus the carried load (Williams and Cavanagh, 1983; Winter, 1983). Greater negative work performed during the first half of stance in load carriage running, as compared to unloaded running, could be due to a greater vertical displacement of the total centre of mass (body plus load) as a result of greater lower limb compression in mid-stance (Silder et al., 2015), coupled with the need to decelerate a greater total weight (body plus load).

The increase in hip negative work as a result of load carriage is largely driven by an increase in hip negative power in late stance (Liew et al., 2016). Joint moment analysis of our data (not reported) pointed to an earlier onset of eccentric internal hip flexor moment during the second half of stance during load carriage running, as compared with unloaded running, which is consistent with findings from a previous investigation (Brown et al., 2014b). Muscle activity of the hip flexor – iliopsoas, has been previously reported to occur during the late stance of unloaded running (Andersson et al., 1997), and contribute significantly to hip flexor moment during this phase (Dorn et al., 2012).

The iliopsoas not only induces a hip flexion moment, but it also exerts an internal trunk flexor moment, which would be required to counteract the externally induced trunk extensor moment caused by carrying a posterior load while running (Brown et al., 2014b). By increasing its muscle activity during load carriage running, the iliopsoas may be involved in absorbing energy of the decelerating thigh segment (which is undergoing thigh extension), as it enters the swing phase, and transferring and increasing the energy of the proximal trunk segments as it enters the flight phase. Our study indicates an increase in total positive work as a result of load carriage. The increased total positive work associated with load carriage in running would be needed to restore the system's potential and kinetic energy as it enters the flight phase (Williams and Cavanagh, 1983).

Consistent with previous findings, in this study net joint work of all three joints within the stance phase of unloaded running shows a bias towards net positive work (Heise et al., 2011). A net positive work in the stance phase of unloaded running is usually attributed to uncaptured soft-tissue (passive) negative work (inverse dynamics assumes a rigid body), especially during the impact phase of running (Riddick and Kuo, 2016). Interestingly, net joint work did not alter with the addition of load in running, providing indirect evidence that soft-tissue negative work did not alter during the stance phase. Using the difference in mechanical work derived by the centre of mass approach and inverse dynamics approach as a surrogate measure of soft-tissue work (Riddick and Kuo, 2016), a previous study on load carriage walking also reported very small alterations in soft-tissue negative work performed when load was added (Huang and Kuo, 2014). It is unknown if soft-tissues would increase its contribution to negative work, with heavier load in running, given that previous research documented that participants carrying a heavy load of up to 57 % BW landed from a jump task with reduced knee flexion (Brown et al., 2016). A stiffer landing with reduced knee flexion could shift work absorption from active “sources” to passive soft tissues (Zelik and Kuo, 2012).

According to the findings of this study, the increased total joint work during load carriage running can be attributed largely to an increased joint work magnitude, and to a much smaller extent, inter-joint work redistribution. Importantly, the ankle still generated the greatest positive work and the knee still absorbed the largest negative work during the stance phase of load carriage running.

However, a novel finding in this study was that hip positive power actually decreased with load carriage. The ankle and knee joints demonstrate an alternating flexion to extension angular change in the stance phase of running (Winter, 1983), which means that both negative and positive joint work can increase in magnitude during load carriage running. In contrast, the hip only extends during the stance phase of running (Sinclair et al., 2013). Assuming that hip power largely arises from sagittal plane hip moments (Schache et al., 2011), the increase in hip flexor moment with load carriage means that hip joint power must become increasingly more negative, at the expense of positive work generation.

The results of this study demonstrated that there was a very small change in percentage joint work redistribution during load carriage running compared to unloaded running. During unloaded running of 3.0 to 5.0 m/s, a previous study reported that the ankle, knee and hip joints contributed approximately 60 %, 30 %, and 10 % of stance phase positive work, respectively, whilst these three joints contributed approximately 35 %, 45 %, and 20 % of stance phase negative work, respectively (Schache et al., 2015). In our study, the contribution of the knee joint to total negative work reduced by a rate of 2 % per 10 % BW load carried. Load carriage also increased knee positive work contribution at a rate of 2.4 % per 10 % BW carried. Even at the heaviest load of 20 % BW investigated, individual joint percentage contribution was largely comparable to unloaded running at the same velocity (Schache et al., 2015), indicating that inter-joint work redistribution played a small role when running with loads of up to 20 % BW.

Inter-joint work redistribution may, however alter when running with extremely heavy load. Assuming that mechanical work increases linearly with load magnitude, using a previously investigated load magnitude of 57 % BW (Brown et al., 2014b), coupled with our regression model (Table 4-1), running with an extremely heavy load could increase percentage knee positive work contribution by up to 14 %. Interestingly, percentage joint contribution to average positive power during the stance phase of load carriage running did not alter in a previous study (Brown et al., 2014b). This could be due to the use of average joint power as an outcome variable in the study by Brown et al. (Brown et al., 2014b). A previous study in walking reported that load carriage resulted in gait-phase specific effects, rather than average effects, on joint power waveforms (Huang and Kuo, 2014).

Extrapolating our findings to load greater than 20 % BW should be made with caution. A previous study in walking found that linear increase in leg stiffness may be reached at 40 % BW load (Caron et al., 2015). However, no study to our knowledge has reported similar load magnitude boundaries in running. It is possible that total joint work could increase to a plateau with increasing load, resulting in a transition to a new gait pattern where mechanical work demands are reduced. It is also possible that inter-joint work redistribution may be involved to a greater extent with heavier loads, similar to the inter-joint work redistribution observed when running velocity increased beyond a boundary of 5.0 m/s (Dorn et al., 2012; Schache et al., 2015). An important limitation of using the joint work approach in running, is that it assumes no inter-segmental energy transference (Aleshinsky, 1986), and no elastic energy recovery is permitted in the model (Aleshinsky, 1986). Future investigations should consider the use of forward dynamic musculoskeletal models to more accurately understand individual muscle-tendon work demands during load carriage running.

4.5 Conclusion

Carrying load while running increased the magnitude of all three lower limb joints' negative work, while it only increased positive work at the ankle and knee joint. This study also found that inter-joint work redistribution played a less significant role compared to global alterations in all joints' work magnitudes, during the stance phase of load carriage running, when carrying loads of up to 20 % BW. The ankle joint contributed to the greatest role in work production, whilst the knee contributed to the greatest role in work absorption in running with and without load carriage of up to 20 % BW. This information will be useful for the development of future performance-enhancing aids for use in real world, field-based load carriage tasks.

4.6 References

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4.7 Figures

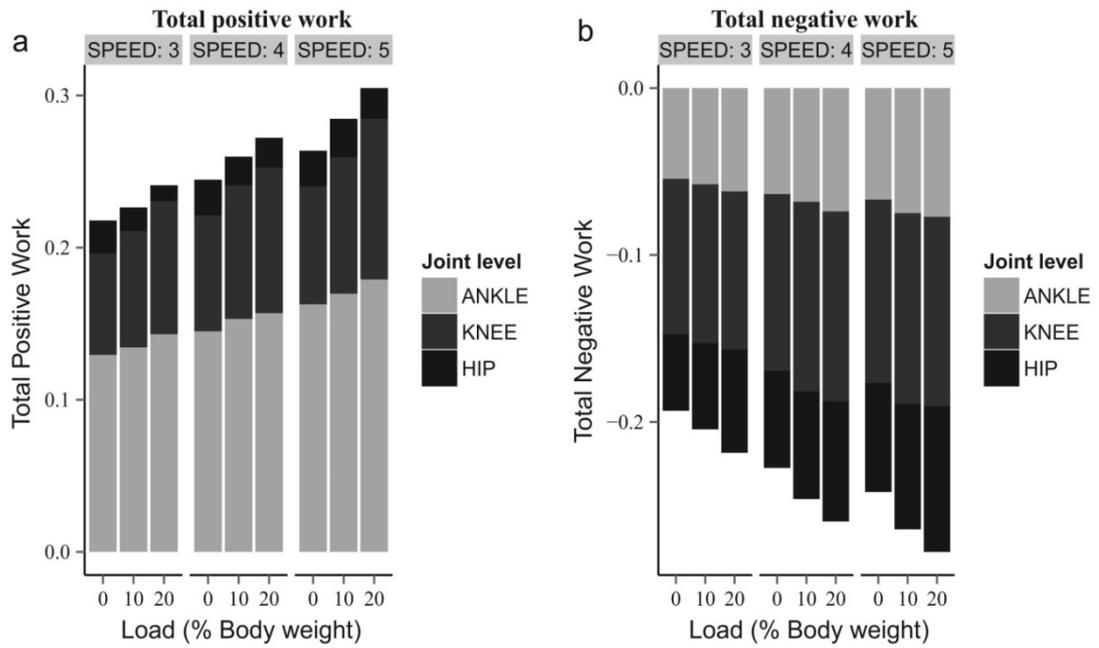


Figure 4-1 Dimensionless total positive and negative joint work

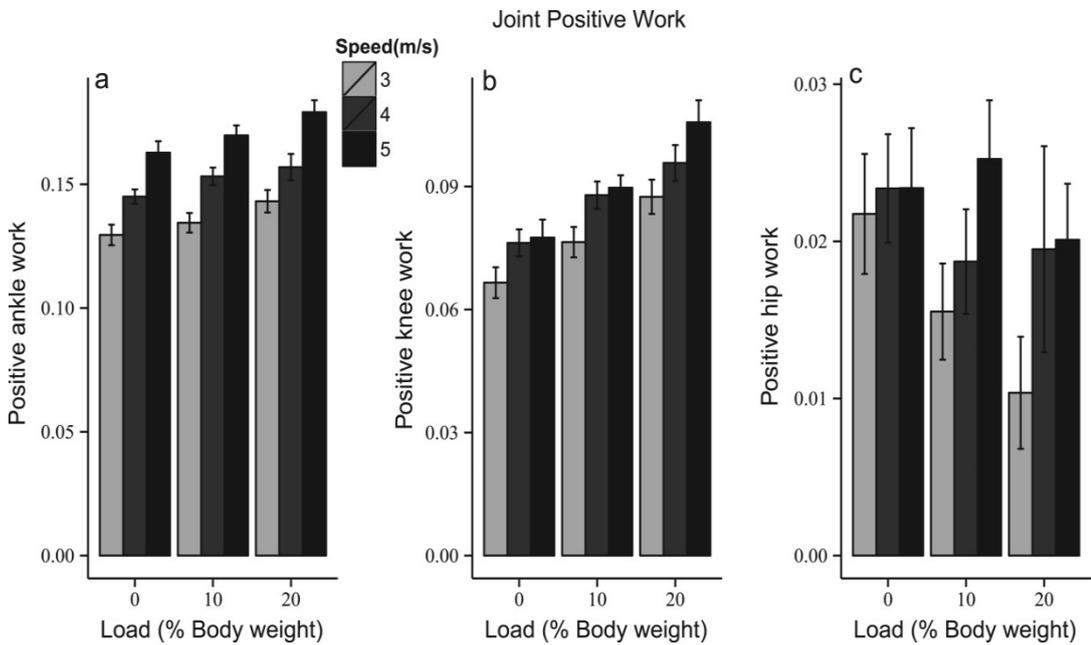


Figure 4-2 Dimensionless positive joint work means and standard deviation (error bars)

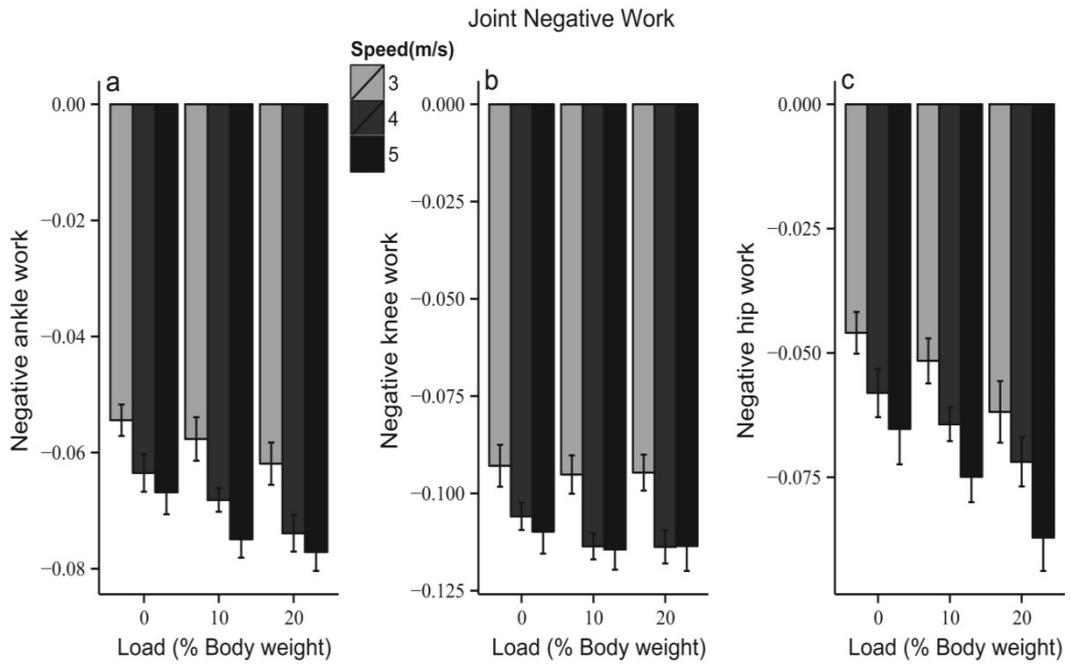


Figure 4-3 Dimensionless negative joint work means and standard deviation (error bars)

4.8 Tables

Table 4-1 Mixed effect model of dimensionless joint work parameters (unit), and load (%BW) and velocity (m/s)

Variables	Intercept	β_1 (%Load)	β_2 (Velocity)	β_3 (Load:Velocity)
Ankle positive work (W_{ankle}^+)	7.63×10^{-2}	6.95×10^{-4}	1.73×10^{-2}	NA
Ankle negative work (W_{ankle}^-)	-3.19×10^{-2}	-4.69×10^{-4}	-7.48×10^{-3}	NA
Knee positive work (W_{knee}^+)	4.60×10^{-2}	1.12×10^{-3}	6.87×10^{-3}	NA
Knee negative work (W_{knee}^-)	-6.30×10^{-2}	-2.47×10^{-4}	-1.03×10^{-2}	NA
Hip positive work (W_{hip}^+)	1.72×10^{-1}	-1.22×10^{-3}	1.40×10^{-3}(ns)	2.29×10^{-4}
Hip negative work (W_{hip}^-)	-1.20×10^{-2}	-7.79×10^{-4}	-1.12×10^{-2}	NA
% ankle positive work	57.11	1.87×10^{-1}(ns)	8.10×10^{-1}	-6.7×10^{-2}
% ankle negative work	28.33	6.94×10^{-3}(ns)	5.05×10^{-2}(ns)	NA
% knee positive work	33.20	2.43×10^{-1}	-7.07×10^{-1}	NA
% knee negative work	53.17	-2.06×10^{-1}	-1.57	NA
% hip positive work	10.36	-5.39×10^{-1}	-2.87×10^{-1}(ns)	9.45×10^{-2}
% hip negative work	18.49	1.93×10^{-1}	1.52	NA
Total positive work (W_{total}^+)	1.48×10^{-1}	-2.69×10^{-4}(ns)	2.37×10^{-2}	4.42×10^{-4}
Total negative work (W_{total}^-)	-1.05×10^{-1}	-1.51×10^{-3}	-2.98×10^{-2}	NA
Total joint work (W_{stance}^{total})	2.67×10^{-1}	2.28×10^{-4}(ns)	4.97×10^{-2}	7.07×10^{-4}
Net joint work (W_{stance}^{net})	2.47×10^{-2}(ns)	-6.91×10^{-5}(ns)	-8.53×10^{-4}(ns)	NA
Stance duration	3.63×10^{-1}	1.98×10^{-3}	-3.69×10^{-2}	-2.62×10^{-4}
Cycle duration	8.80×10^{-1}	-1.51×10^{-3}	-4.77×10^{-2}	NA

Step length	4.76×10^{-1}	4.25×10^{-3}	2.26×10^{-1}	-1.98×10^{-3}
Stride length	8.82×10^{-1}	6.53×10^{-3}	4.67×10^{-1}	-3.21×10^{-3}
<p>β = beta coefficient; ns= non-significant; NA = not applicable to final model due to non-significant interaction. Example: Total positive work at 3.0 m/s carrying 20 % BW equates to: $0.148 - 0.000269 (20) + 0.0237 (3) + 0.000442 (20)(3) = 0.240$ units.</p>				

Table 4-2 Percentage joint contribution to total positive and negative joint work

Velocity (m/s)	Load (% BW)	% Ankle W_{pos}	% Ankle W_{neg}	% Knee W_{pos}	% Knee W_{neg}	% Hip W_{pos}	% Hip W_{neg}
3	0	59.9	24.5	30.4	48.1	9.7	23.4
3	10	59.7	28.5	33.5	46.7	6.8	24.8
3	20	59.7	28.7	36.0	43.4	4.3	27.9
4	0	59.5	28.5	31.2	46.7	9.4	24.8
4	10	59.2	28.2	33.6	46.2	7.1	25.6
4	20	57.7	29.1	34.9	43.6	7.3	27.3
5	0	61.7	28.7	29.6	45.1	8.7	26.2
5	10	59.7	28.9	31.6	43.0	8.8	28.0
5	20	58.9	28.3	34.6	40.7	6.5	31.0

BW = body weight; % percentage; W_{pos} = Positive joint work; W_{neg} = Negative joint work

4.9 Supplementary material

4.9.1 Participant inclusion/exclusion criteria

Participants were included if they were between 18 to 45 years old and, currently involved in regular running, jumping, and/or hopping based activities, with a cumulated total time of at least one hour per week, within the past 12 months. Participants were excluded if they had any medical disorders that affected their load carrying running ability, presented with any musculoskeletal injuries within the last three months of testing, had a lower limb surgical history within the past year, those with current episodes of pain, and females who were pregnant.

4.9.2 Biomechanical model

The origin of the pelvis anatomical coordinate system was defined as the mid-point between the ASIS markers. The (x-y) plane of the segment coordinate system was defined as the plane passing through the right and left ASIS markers, and the mid-point of the right and left PSIS markers. The x-axis (medial-lateral) was defined from the ORIGIN towards the Right ASIS. The z-axis (distal-proximal) was perpendicular to the (x-y) plane. The y-axis (posterior-anterior) was defined as the cross product of the x-axis and z-axis.

The z-axis (distal-proximal) of the thigh ACS in both models was oriented as the line joining the knee and hip centre with its positive direction pointing superiorly. The y-axis (posterior-anterior) was orthogonal to the frontal plane with its positive direction anterior. The frontal plane was formed by the lateral femoral condyle, hip and knee joint centre. The x-axis (medial-lateral) was then defined as the cross-product of the z- and y-axes with its positive direction lateral.

The z-axis (distal-proximal) of the shank ACS was the line passing from the ankle joint to the knee joint centre. The y-axis (posterior-anterior) was orthogonal to the plane defined by the lateral malleoli, ankle and knee joint centre with its positive direction anterior. The x-axis (medial-lateral) was the cross-product of the z- and y-axes with its positive direction lateral.

In the kinetic foot ACS, the z-axis was defined as the line joining the mid-point between the medial and lateral malleoli and the mid-point between the 1st and 5th MTP. The y-axis was orthogonal to the plane defined by the two malleoli and two foot markers (1st and 5th MTP). The x-axis was the cross-product of the z- and y-axes with its positive direction lateral. The kinetic foot was used for kinetic calculations.

In the kinetic foot ACS, the y-axis was defined as the line joining the mid-point between the superior and inferior calcaneal markers, and the mid-point between the 1st and 5th MTP. The y-axis was orthogonal to the plane defined by the two malleoli and two foot markers (1st and 5th MTP). The x-axis was the cross-product of the z- and y-axes with its positive direction lateral.

A virtual foot segment (vFoot1) was created and used for kinematic calculations, which enabled the use of the static calibration trial as a reference posture. The z-axis was defined as the line joining the mid-point between the superior and inferior calcaneal markers (MID_CALA), and the mid-point between the 1st and 5th MTP (MID_MT). The y-axis was defined as the cross product between the z-axis and the plane formed between the first two points, and the ankle joint centre. Lastly, the x-axis was defined as the cross-product between the z-axis and y-axis. The virtual foot ACS was then rotated so that the y-axis was pointing anteriorly, z-axis was pointing superiorly.

A second virtual foot segment (vFoot2) was created to enable accurate centre of pressure calculations. First, a virtual laboratory was created with the z-axis directed superiorly, y-axis directed anteriorly, and x-axis directed lateral. MID_CALA and MID_MT were projected onto the virtual laboratory. vFoot2 was created with the z-axis defined as the projected MID_CALA and MID_MT. The y-axis was defined as the cross product between the z-axis and the plane formed between the first two points, and the ankle joint centre. Lastly, the x-axis was defined as the cross-product between the z-axis and y-axis. The virtual foot ACS was then rotated so that the y-axis was pointing anteriorly, z-axis was pointing superiorly.

Subject 29 (model): The pelvis was constructed differently for this participant, due to the presence of abdominal adiposity, precluding accurate placement of the ASIS markers. Bilateral markers were placed on the apex of the iliac crest, with two markers placed on the greater trochanter. The origin of the pelvis anatomical coordinate system was defined as the mid-point between the iliac crest markers. The x-axis (medial-lateral) was defined from the ORIGIN towards the Right iliac crest marker. The z-axis was defined as the cross product between the x-axis, and the frontal plane formed by the two iliac crest and two greater trochanter markers. The y axis was defined as the cross product between the x and z-axes. The hip joint centre was defined by a radius of 18 cm measured using a sliding calliper from the greater trochanter to the hip joint centre. The remaining segments were constructed with the same method as the remaining participants

Table 4-3 95 % confidence interval of coefficients between work, load (% BW) and velocity (m/s)

Variables	Modelled variables	Intercept	β_1 (% Load)	β_2 (Velocity)	β_3 (Load:Velocity)
Ankle positive work	W_{ankle}^+	6.72 to 8.54×10^{-2}	5.14 to 8.76×10^{-4}	1.57 to 1.73×10^{-2}	NA
Ankle negative work	W_{ankle}^-	- 4.10 to -2.28×10^{-2}	-6.05 to -3.34×10^{-4}	-8.76 to -6.20×10^{-3}	NA
Knee positive work	W_{knee}^+	3.85 to 5.40×10^{-2}	9.34×10^{-4} to 1.35×10^{-3}	5.30 to 8.26×10^{-3}	NA
Knee negative work	W_{knee}^-	-7.32 to -4.98×10^{-2}	-4.23×10^{-4} to -5.51×10^{-5}	-1.30×10^{-2} to -8.70×10^{-3}	NA
Hip positive work	W_{hip}^+	7.34×10^{-3} to 2.71×10^{-2}	-2.02×10^{-3} to -4.24×10^{-4}	-7.45×10^{-4} to 3.55×10^{-3}	3.47×10^{-5} to 4.23×10^{-4}
Hip negative work	W_{hip}^-	-2.26×10^{-2} to 2.99×10^{-4}	-1.10×10^{-3} to -4.83×10^{-4}	-1.34×10^{-2} to -9.38×10^{-3}	NA
% ankle positive work	-	54.0 to 60.2	-5.97×10^{-4} to 4.34×10^{-3}	1.16×10^{-3} to 1.50×10^{-2}	-1.27×10^{-3} to -6.58×10^{-5}
% ankle negative work	-	24.2 to 32.5	-6.39×10^{-4} to 7.78×10^{-4}	-4.82×10^{-3} to 5.83×10^{-3}	NA
% knee positive work	-	54.0 to 60.2	-5.97×10^{-4} to 4.34×10^{-3}	1.16×10^{-3} to 1.50×10^{-2}	NA
% knee negative work	-	49.7 to 56.7	-2.69×10^{-3} to -1.42×10^{-3}	-2.04×10^{-2} to -1.09×10^{-2}	NA

% hip positive work	-	6.7 to 14.0	-8.61×10^{-3} to -2.16×10^{-3}	-1.11×10^{-2} to 5.41×10^{-3}	1.55×10^{-4} to 1.74×10^{-3}
% hip negative work	-	14.6 to 22.4	9.31×10^{-4} to 2.93×10^{-3}	8.99×10^{-3} to 2.14×10^{-2}	NA
Total positive work	W_{total}^+	1.28 to 1.67×10^{-1}	-1.57×10^{-3} to 1.03×10^{-3}	1.95 to 2.78×10^{-2}	1.23 to 7.62×10^{-4}
Total negative work	W_{total}^-	-1.23×10^{-1} to -8.63×10^{-2}	-1.88 to -1.15×10^{-3}	-3.30 to -2.65×10^{-2}	NA
Total joint work	W_{stance}^{total}	2.29 to 3.05×10^{-1}	-2.24×10^{-3} to 2.70×10^{-3}	4.19 to 5.76×10^{-2}	9.91×10^{-5} to 1.34×10^{-3}
Net joint work	W_{stance}^{net}	9.35×10^{-3} to 4.03×10^{-2}	-2.86 to 1.92×10^{-4}	-6.30 to 4.46×10^{-3}	NA
Stance duration	Stance	3.51 to 3.74×10^{-1}	1.46 to 2.51×10^{-3}	-3.93 to -3.46×10^{-2}	-3.90 to -1.34×10^{-4}
Cycle duration	Cycle time	8.51 to 9.08×10^{-1}	-1.84 to -1.18×10^{-3}	-5.43 to -4.11×10^{-2}	NA
Step length	Step length	4.06 to 5.46×10^{-1}	1.07 to 7.43×10^{-3}	2.07 to 2.45×10^{-1}	-2.76 to -1.20×10^{-3}
Stride length	Stride length	7.54×10^{-1} to 1.01	1.79×10^{-4} to 1.29×10^{-2}	4.34 to 5.01×10^{-1}	-4.79 to -1.64×10^{-3}
β = beta coefficient; NA = not applicable to final model due to non-significant interaction					

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Chapter 5

Joint power and kinematics coordination in load carriage running: implications for performance and injury

Synopsis: A biomechanically informed program requires knowledge of the phase-specific and angle-specific muscular demands of loaded running. To investigate the time-varying influence of load on running mechanics, I used a recently introduced waveform statistical method, termed “Statistical Parametric Mapping”. Rather than a constant increase in power with load, running with load demanded transient increased “bursts” of muscle power at specific periods of the gait cycle. Phase-specific effects of load on joint power were at mid-stance for the ankle, push-off for the knee, and late stance for the hip. This means that resistance training designed for loaded running needed to augment force generation/absorption capacity within a brief period, by incorporating plyometric exercises. These results also informed the pertinent muscle contraction modes during training for each joint-level muscle group, from a rapid eccentric-concentric transition mode for the ankle, concentric mode for the knee, and eccentric mode for the hip. The kinematic analyses also assisted in the development of coaching cues needed to ensure optimal movement patterns when training under load. This study has been published as “Liew BX, Morris S, Netto K. Joint power and kinematics coordination in load carriage running: Implications for performance and injury. *Gait Posture*. 2016 Jun;47:74-9. doi: 10.1016/j.gaitpost.2016.04.014.” The permission to reproduce the material can be found in the Appendix.

Abstract

Investigating the impact of incremental load magnitude on running joint power and kinematics is important for understanding the energy cost burden and potential injury-causative mechanisms associated with load carriage. It was hypothesized that incremental load magnitude would result in phase-specific, joint power and kinematic changes within the stance phase of running, and that these relationships would vary at different running velocities. Thirty-one participants performed running while carrying three load magnitudes (0 %, 10 %, 20 % body weight), at three velocities (3, 4, 5 m/s). Lower limb trajectories and ground reaction forces were captured, and global optimization was used to derive the variables. The relationships between load magnitude and joint power and angle vectors, at each running velocity, were analysed using Statistical Parametric Mapping Canonical Correlation Analysis.

Incremental load magnitude was positively correlated to joint power in the second half of stance. Increasing load magnitude was also positively correlated with alterations in three dimensional ankle angles during mid-stance (4.0 and 5.0 m/s), knee angles at mid-stance (at 5.0 m/s), and hip angles during toe-off (at all velocities). Post hoc analyses indicated that at faster running velocities (4.0 and 5.0 m/s), increasing load magnitude appeared to alter power contribution in a distal-to-proximal (ankle→hip) joint sequence from mid-stance to toe-off. In addition, kinematic changes due to increasing load influenced both sagittal and non-sagittal plane lower limb joint angles. This study provides a list of plausible factors that may influence running energy cost and injury risk during load carriage running.

5.1 Introduction

Running related sports which require load carriage (e.g. ultra-marathon) have become increasingly popular over the past two decades (Cejka et al., 2014; Hoffman et al., 2010). However, compared with walking research into load carriage (Huang and Kuo, 2014) running mechanics has received little attention (Brown et al., 2014; Silder et al., 2015), with prior investigations focusing mainly on military applications (Brown et al., 2014; Knapik et al., 2004). Within the military setting, overuse lower limb injuries are commonly associated with heavy load carriage which may involve both walking and running (Orr et al., 2015). However, epidemiological evidence of a detrimental effect of load on non-military athletes is lacking. Research into loaded running in civilians is required to increase our understanding of the impact of load carriage on running energy cost (Teunissen et al., 2007) and injury risks (Orr et al., 2015).

The mechanics of running without external load (termed unloaded running) are well understood. Prior to mid-stance, the knee and ankle extensors absorb power to decelerate the body segments and support body weight (BW) (Hamner and Delp, 2013; Winter, 1983). During push-off, power to accelerate the body into flight is largely driven by the ankle extensors (Hamner and Delp, 2013; Winter, 1983). This temporal coordination in power flow likely reflects muscle coordination patterns which provide the required energy in running while minimising metabolic cost (Arnold et al., 2013; Hamner and Delp, 2013).

Interestingly, it has been reported that increasing load magnitude in running does not alter the proportional contribution of hip, knee and ankle when considering average positive power (Brown et al., 2014). However, when considering phase-specific gait effects, a previous study in walking reported that load carriage did influence joint power (Huang and Kuo, 2014). In running, it is yet unknown how each joint contributes to the total power across the stance phase, when load is carried. In addition, since previous studies have found different joint power contributions at different velocities (Schache et al., 2015), the effect of load on running joint power control may vary at different velocities.

An in-depth analysis of three dimensional (3D) joint angles is needed in this area, as changes to joint kinematics alter joint power contributions (Heiderscheit et al., 2011) and soft-tissue strain patterns (Lee et al., 2003; Tateuchi et al., 2015). For example, small alterations in knee flexion angle (e.g. 4°) has been shown to increase knee joint positive work by 2.5 J/kg (Heiderscheit et al., 2011) and increase the magnitude of knee joint load (Bonacci et al., 2013). Current studies have only investigated the effects of load on sagittal plane kinematics in running at relatively slow velocities (Brown et al., 2014; Silder et al., 2015). However, load carriage exerts significant non-sagittal plane torque on the body (LaFiandra et al., 2002), which when coupled with insufficient muscle capacity, may result in deviations of non-sagittal plane kinematics and create asymmetrical soft-tissue stresses (Lee et al., 2003). Since loaded running occurs across a range of velocities and joint internal loads increase at faster running velocities (de David et al., 2015), investigating the effect of load in 3D whilst running at a range of velocities is needed.

With an increasing involvement of people in running sports requiring load carriage, detailed research into the effect load carriage has on running mechanics is warranted to facilitate the management of these athletes. Statistical Parametric Mapping (SPM) has been used to perform hypothesis testing on biomechanical time-series data (Pataky et al., 2013), which provides a more robust statistical method for understanding the phase-specific effects of load on running mechanics. Thus, the aim of this study is to determine the phase-specific effect of running with three different loads across three different velocities on joint power and 3D kinematics over the stance phase.

5.2 Methods

5.2.1 Study design

A repeated measures design was adopted where participants performed a single testing session, which occurred in Curtin University's biomechanics laboratory

5.2.2 Participants

16 male and 15 female participants enrolled [mean (standard deviation (SD)) age = 30.8 (5.9) years old; height = 1.70 (0.08) m; mass = 66.4 (10.8) kg; distance ran per week = 39.2 (26.4) km; hours ran per week = 3.73 (2.86) hrs]. Nine participants had at least one year experience in frequent load carriage (>10 % BW, at least six separate occasions within a year) during sports and/or as a requirement of their occupation. Twenty two participants had no prior experience in frequent load carriage. All participants provided signed informed consent prior to study enrolment. Ethical approval for this study was provided by Curtin University Human Research Ethics Committee (RD-41-14).

5.2.3 Running protocol

Participants wore their personal running shoes and completed a warm-up before the experiment. Participants ran across a 20 metre runway, embedded with three consecutive force platforms (3 metre lengthwise) (AMTI, Watertown, MA), while carrying three load conditions (0 %, 10 %, 20 % BW) across three velocities (3.0 m/s, 4.0 m/s, 5.0 m/s). Timing gates (SMARTSPEED Pro, Fusion Sport Pty Ltd, Australia) were placed five metre apart on either side of the force plates, whilst a 15 metre run up was given to enable each participant to achieve the desired velocity before running across the force plates. Thirty-one sequences of load-velocity condition were generated using a random sequence generator (<https://www.randomizer.org/>), to minimize the influence of testing order on our dependent variables. Load carriage was achieved through varying the volume of sand (in sandbags) carried in a backpack (CAMELBAK, H.A.W.G.® NV, 14 litre). The backpack was fitted snugly to the participants' trunk with waist and chest straps. Each condition required five successful running attempts, each within a ± 10 % variation of the prescribed velocity and with no visible alteration in running gait pattern to target the force plate. At least 30 seconds rest was provided between each running attempt and a 5 minute rest between each running condition.

5.2.4 Biomechanical modelling and processing

The position and orientation of the right lower limb segments was calculated using an inverse kinematic (IK) lower limb model created in Visual 3D (C-motion, Germantown, MD), using the Levenberg-Marquardt algorithm (Robinson et al., 2014). A standard lower limb marker set protocol was used, which had been previously described (Liew et al., 2016).

The hip joint centre was defined using a regression equation (Bell et al., 1989), whilst the knee and ankle joint centres were defined as the mid-point of the femoral epicondyles and malleoli (Liew et al., 2016), respectively. For the IK model, the hip, knee, and ankle joints were constrained to have three rotational degrees of freedom (DOF), whilst that of the pelvis segment had six DOF. A segmental weight of two was given to the thigh, three to the shank, and four to the pelvic and foot segments (Robinson et al., 2014).

Marker trajectories were captured at 250 Hz using an 18 camera motion analysis system (Vicon T-series, Oxford Metrics, UK), while ground reaction force (GRF) was recorded at 2000 Hz using the force platforms. Gap filling was performed in Vicon Nexus (v2.1.1, Oxford Metrics, UK). Raw marker trajectories and force data were filtered using a low pass, zero-lag, 4th order Butterworth filter at 18 Hz (Robinson et al., 2014) for inverse dynamics. Trajectories were filtered at a higher frequency for inverse dynamics, to match the force data filtering frequency in order to avoid joint moment artefacts (Kristianslund et al., 2012). However, this frequency of 18 Hz resulted in excessive “noise” in the kinematic waveforms. Hence, raw trajectories data was filtered at 12 Hz for kinematic analysis (Sinclair et al., 2014). A Cardan XYZ rotation sequence was used to calculate 3D joint angles (Cole et al., 1993). Joint angles were expressed in an orthogonal frame in the proximal segment using the right hand rule (Liew et al., 2016). This meant that positive values along the x-axis (medio-lateral axis) represented hip flexion, knee extension, and ankle dorsiflexion; positive values along the y-axis (postero-anterior axis) represented hip adduction, knee adduction, and ankle inversion; and positive values along the z-axis (vertical axis) represented hip and knee internal rotation, and ankle adduction. Instantaneous joint angles and power trajectories of the right hip, knee, and ankle were computed only during the stance phase of running. A threshold of 20 N in ground reaction force was used to determine initial foot contact and toe-off. Joint power was normalized by the scaling factor of $ML^{0.5}g^{1.5}$ (mean (SD)) scaling factor = 1845.10 (338.36), with base factors of gravitational constant g (9.81 m/s^2), leg length, L (m), and body mass, M (kg) (Huang and Kuo, 2014).

5.2.5 Statistical analysis

All analyses were performed using `spm1d` package (v0.3) (www.spm1d.org), installed in Python 2.7, and implemented in Enthought Canopy 1.5.4 (Enthought Inc., Austin, USA) (Pataky, 2012). Canonical correlation analysis was performed to determine the magnitude of the correlation between the predictor variable (load magnitude), and the dependent vector variables (Pataky et al., 2013).

As the current implementation of SPM in `spm1d` only allows for a univariate predictor variable (Pataky, 2012), Canonical correlation analyses was performed at each running velocity. In order to determine significance, field smoothness was derived from time-varying gradients of the residuals (Friston et al., 2007). Next, given the calculated smoothness, Random Field Theory (RFT) was used to determine a critical threshold that maintained an alpha rate of 0.05 (Pataky, 2012). Hence, a critical threshold of 0.05 was set for joint power, and a Bonferroni corrected threshold of 0.0167 (0.05/3) was set for each of the three joint angles. Post-hoc scalar field analysis was performed on each vector component, only when significance was achieved at the vector-field level. For scalar field analysis, SPM linear regression t-statistic was performed on each vector components. A Bonferroni corrected threshold of 0.0167 (0.05/3) was used to account for three post-hoc comparisons for each vector field. The time normalized domain reported was in increments of 0.5 %, from 0 % indicating initial contact to 100 % indicating toe off.

5.3 Results

The mean (SD) waveforms at 5.0 m/s are shown in Figure 5-1, and the figures of other velocities can be found in the Supplementary material. Complete graphical representation of all primary and post-hoc analyses for velocities of 3.0 and 4.0 m/s can be found in the Supplementary material.

5.3.1 Effect of load on joint power vector

Vector field analyses on joint power indicated significant relationships with incremental load magnitude during the mid to second half of stance phase at all three running velocities (Figure 5-2 for analyses at 5.0 m/s, Table 5-1 for all velocities). At all velocities, post-hoc scalar field analyses indicated that with increasing load magnitude, knee positive and hip negative power both increased in magnitude (Figure 5-3a-c for analyses at 5.0 m/s, Table 5-1). Ankle negative power increased in magnitude with increasing load magnitude, only at 4.0 and 5.0 m/s (Table 5-1). Post-hoc analyses indicated that load influenced ankle power just after mid-stance, knee power in late stance, whilst hip power at toe-off (Table 5-1).

5.3.2 Effect of load on joint angle vector

Vector field analysis resulted in one significant supra-threshold cluster at the ankle, each at 4.0 m/s and 5.0 m/s after mid-stance; one cluster at the knee at 5.0 m/s after mid-stance; one cluster at the hip joint at each of the three running velocities during toe-off (Figure 5-2 for analyses of 5.0 m/s, Table 5-1). At the hip, it was observed that late stance vector field significance was driven largely by an increase in hip adduction angle associated with increasing load magnitude (Figure 5-3k for analyses at 5.0 m/s, Table 5-1). At the knee, increasing load was associated with an increase in knee flexion angle during the region of mid-stance only at 5.0 m/s (Figure 5-3g for analyses at 5.0 m/s, Table 5-1). At the ankle, significance at the vector field level was driven by an increase in ankle dorsiflexion associated with increasing load only at 4.0 and 5.0 m/s (Figure 5-3d for analyses at 5.0 m/s, Table 5-1).

5.4 Discussion

In this study, we were interested about the effects of load on running joint power and kinematics. From mid-stance to toe-off in unloaded running, lower limb muscles provide power for weight support and propulsion (Hamner and Delp, 2013). During this phase, power is transferred to the trunk, as it gains potential and kinetic energy going into the flight phase (Williams and Cavanagh, 1983). From our vector field analyses, we were able to identify different joint power contribution patterns at different running velocities, as a result of load carriage. Joint power changes during load carriage were more apparent at faster velocities, compared to slower velocities consistent with previous research (Brown et al., 2014). In addition, there was phase-specific 3D kinematic changes in running pattern during load carriage, which has not been previously reported (Brown et al., 2014). This highlights the importance of performing 3D analysis on joint angles during load carriage running, particularly at higher velocities.

At velocities of 4.0 and 5.0 m/s, increasing load altered the power contribution in a distal-to-proximal (ankle → hip) joint sequence from mid-stance to toe-off (Figure 5-1). Interestingly, increasing load was associated with a small increase in ankle dorsiflexion (by $< 5^\circ$) whilst running at 4.0 and 5.0 m/s, during approximately 52 % to 80 % of stance (Table 5-1). These findings suggest that an increase in eccentric ankle extensor activity with load, at faster velocities, was needed to eccentrically “brake” the dorsiflexing shank (Neptune et al., 2001), removing energy from the shank and transferring to the foot (Robertson and Winter, 1980). This energy would then be transferred through the joints to proximal segments, to provide increased vertical force to support an increased total weight (Neptune et al., 2001). Hence, the increased negative power provided by the ankle extensors, contributed to the earliest source of increased proximal energy flow, at faster running velocities, associated with load carriage. This increase in ankle negative power, coupled with an increase in ankle dorsiflexion angle, while necessary to support an increased total weight in running, may increase mechanical stress and strain on the Achilles tendon (Almonroeder et al., 2013). Over many steps, this may predispose an individual running with load to an increased risk of developing Achilles tendon injuries, due to a greater cumulated loading of the tendon (Rooney and Derrick, 2013).

After mid-stance, increased knee power generation by the knee extensors with load carriage, functions to increase the segmental power of the thigh as it extends from a position of knee flexion. It can be observed that the greatest increase in knee power generation with load occurred at 5.0 m/s, which was also associated with a significant increase in mid-stance knee flexion angle (Figure 5-3g). Previous studies on load carriage running have not identified significant alterations in peak knee flexion angle (Brown et al., 2014), and peak leg length compression (Silder et al., 2015), at lower velocities of 3.3 to 3.5 m/s, which is consistent with our finding. This increase in peak knee flexion angle and knee extensor power generation likely represent sub-optimal running mechanics at fast velocity under load, which may be associated with an increased energy cost (Heiderscheit et al., 2011). In addition, although the increase in peak knee flexion with a 20 % BW load was small ($< 5^\circ$), a previous study has found that a change in peak knee flexion angle of 2° to 3° was associated with a change in patellofemoral joint stress by 2.4 MPa, which could significantly influence cumulative knee joint loads (Bonacci et al., 2013). Therefore, the increased work at the knee during mid stance is likely to be more energetically costly and potentially damaging.

At toe-off, an increased hip negative power with load absorbs power from the thigh segment, with simultaneous power transfer to the trunk segments (Williams and Cavanagh, 1983). This pattern of hip power absorption was seen in our findings and was largest at 3.0 m/s, and smallest at 5.0 m/s (Supplementary material). It can be observed that hip extension angles achieved during toe-off were greater in loaded relative to unloaded running, albeit not reaching statistical significance, with this difference greatest at 3.0 m/s compared to 5.0 m/s (Supplementary material). Since the data reported was time normalized, the greater effect of load on hip extension angle at slower velocities compared with faster velocities may be because of the proportionately longer stance phase duration at a slow velocity. For example, results from a previous study indicated that stance phase duration increased from 34 % of cycle duration (i.e. initial contact to initial contact) [3.0 m/s, unloaded] to 52 % of cycle duration [3.0 m/s, 20 % BW load] (Liew et al., 2016). In contrast in our study, the percentage increase in stance phase duration rose from 28% of cycle duration [5.0 m/s, unloaded], to 31 % of cycle duration [5.0 m/s, 20 % BW load], with this small change being enough to influence the mechanics of hip extension (Liew et al., 2016).

This study found specific, albeit small ($< 5^\circ$), alterations in joint kinematics as a result of load carriage. Consistent positive relationships between increasing load and increasing hip adduction angle at toe-off was observed (Table 5-1). Although running occurs largely in the sagittal plane, forces applied to the body act in three dimensions. At toe-off, the hip was in 2° to 4° abduction in an unloaded running, which deviated to $< 2^\circ$ of hip adduction with a 20 % BW load (Figure 5-3, Supplementary material). This small increase in hip adduction at toe-off could be due to reduced activity of the hip abductor muscle group (Hamner et al., 2010), thus reducing the counteracting torque needed to resist a greater frontal plane external adduction moment, which is associated with load carriage. Deviations in the hip frontal plane during running, has often been implicated as a risk factor for sustaining overuse knee injuries, due to the asymmetrical loading of knee joint structures (Dierks et al., 2008).

It is acknowledged that the kinematic changes associated with load carriage appear small, yet a clinically important threshold has not been defined. However, kinematic differences of this magnitude have often been reported in cross-sectional studies between injured and healthy runners (Donoghue et al., 2008; Esculier et al., 2015). More importantly, kinematic alterations with load during an acute bout of running may be exacerbated with prolonged running, and should not be discounted as potentially relevant changes.

A limitation of this study was its cross-sectional design, using inverse-dynamics methods, which does not allow causal inference to be made in relation to load carriage running energy cost and injury rates. However, this study was able to generate a list of plausible factors that may inform future training programs of athletes needing to run with load.

5.5 Conclusion

Load carriage during running resulted in a temporally sequenced set of distal-to-proximal increase in joint power from mid-stance to toe-off. This likely contributes to the increased power needed to support an increased weight and provide propulsive forces during load carriage running. In addition, load carriage was associated with a small increase in ankle dorsiflexion and knee flexion in mid-stance, and hip adduction at toe-off, which depended on running velocity. These kinematic alterations could increase soft-tissue loads and the energy cost of running with load. Prospective investigations are warranted to see if load carriage running energy cost and injury risk could be altered when these mechanical factors are addressed in physical conditioning programs.

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5.7 Figures

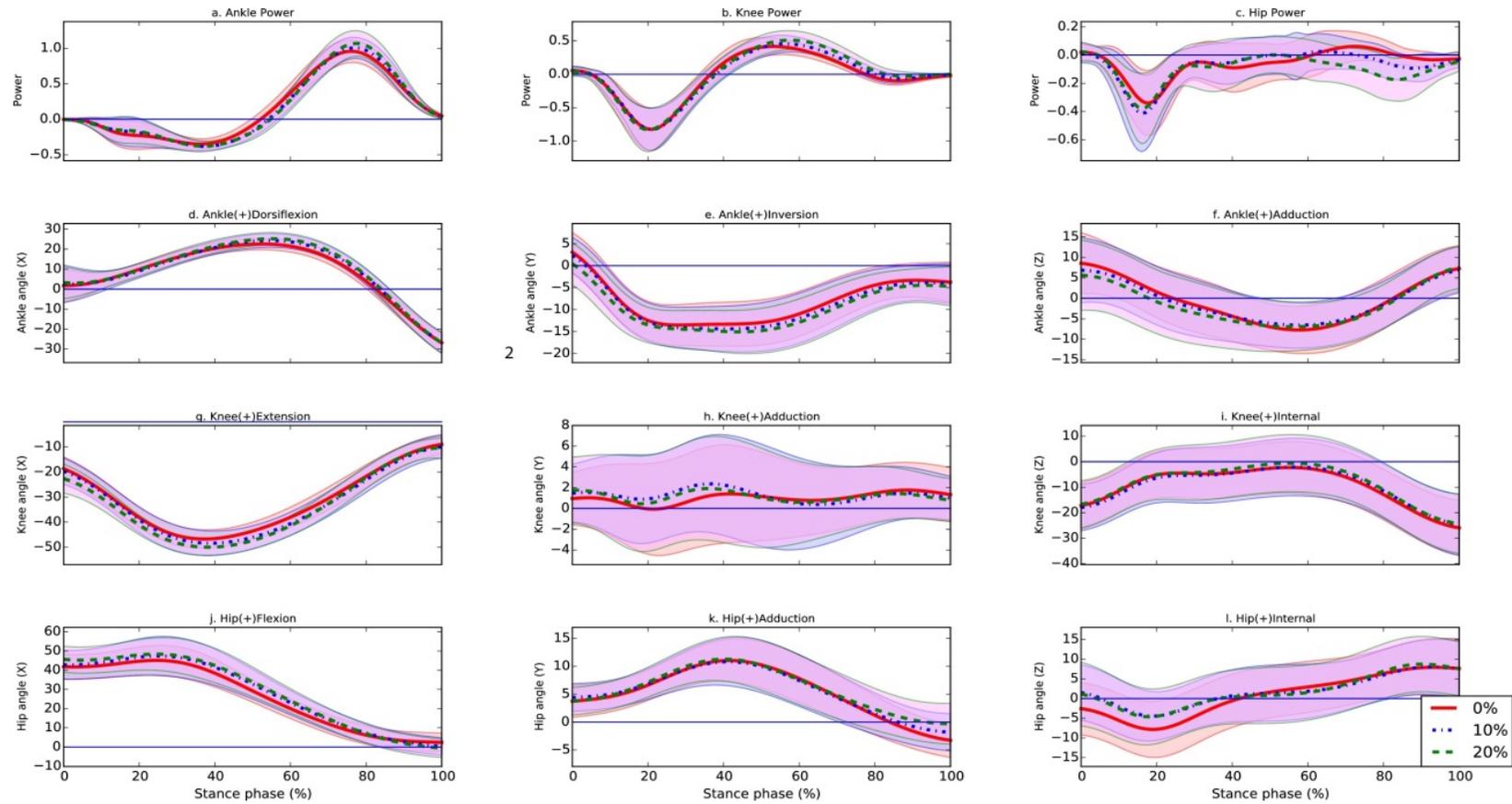
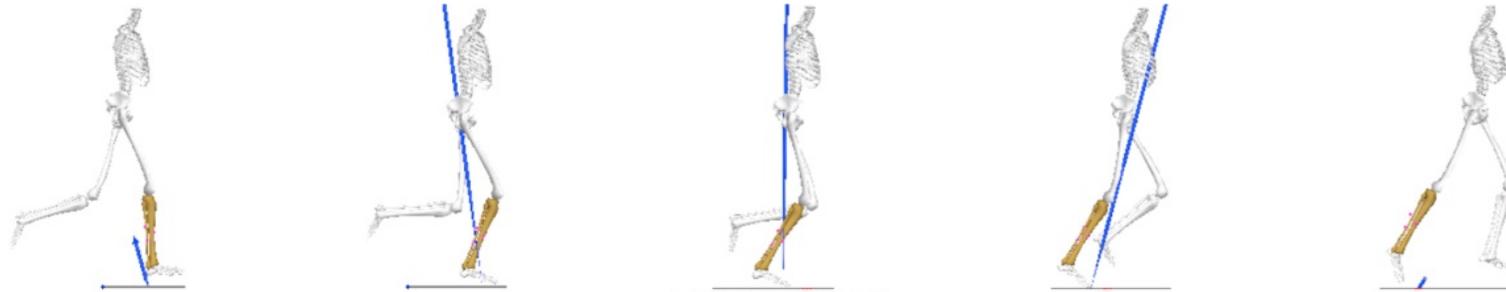


Figure 5-1 (a-l). Representative mean (standard deviation) power, angle curves at 5.0 m/s running. (a-c) Joint power, (d-f) Ankle three dimensional angles, (g-i) Knee three dimensional angles, (j-l) Hip three dimensional angles.



a. Joint Power Vector

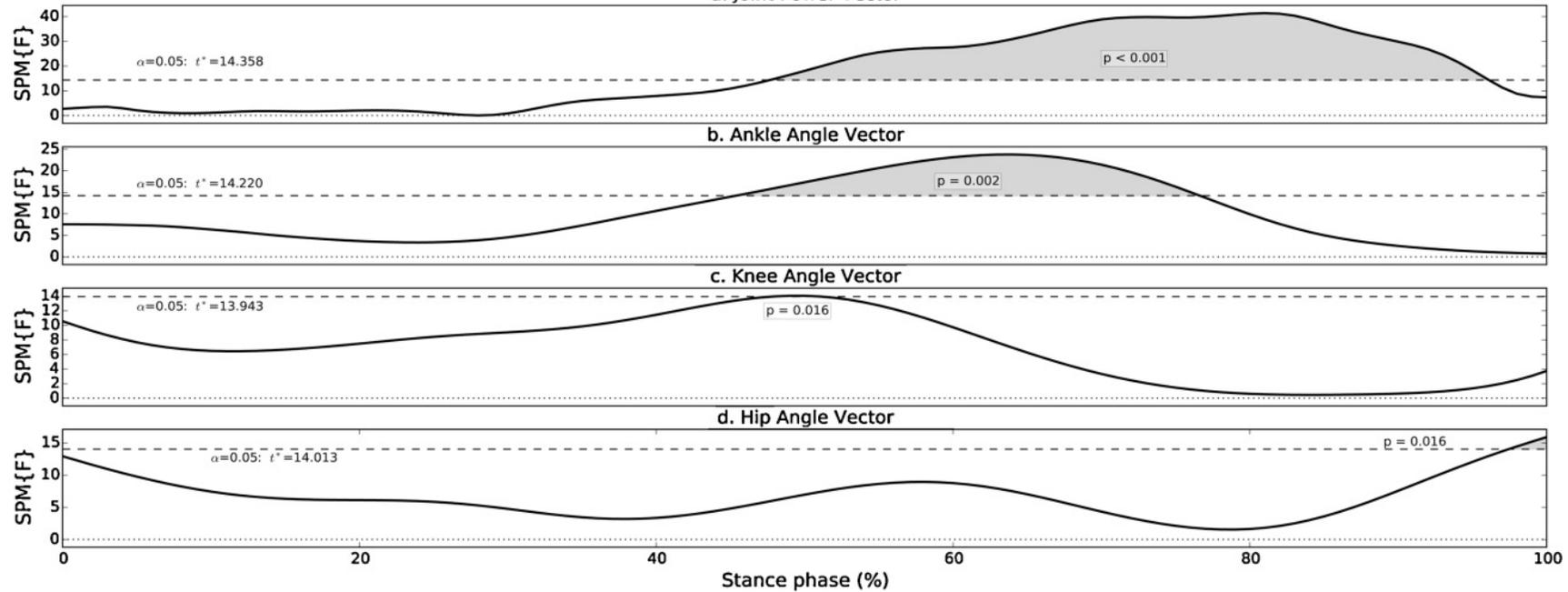


Figure 5-2 Canonical correlation analysis (SPM [F]) results of (a) joint power vector, (b) ankle angle vector, (c) knee angle vector, (d) hip angle vector at 5.0 m/s running. Shaded regions represent significant supra-threshold clusters that exceeded random critical threshold.

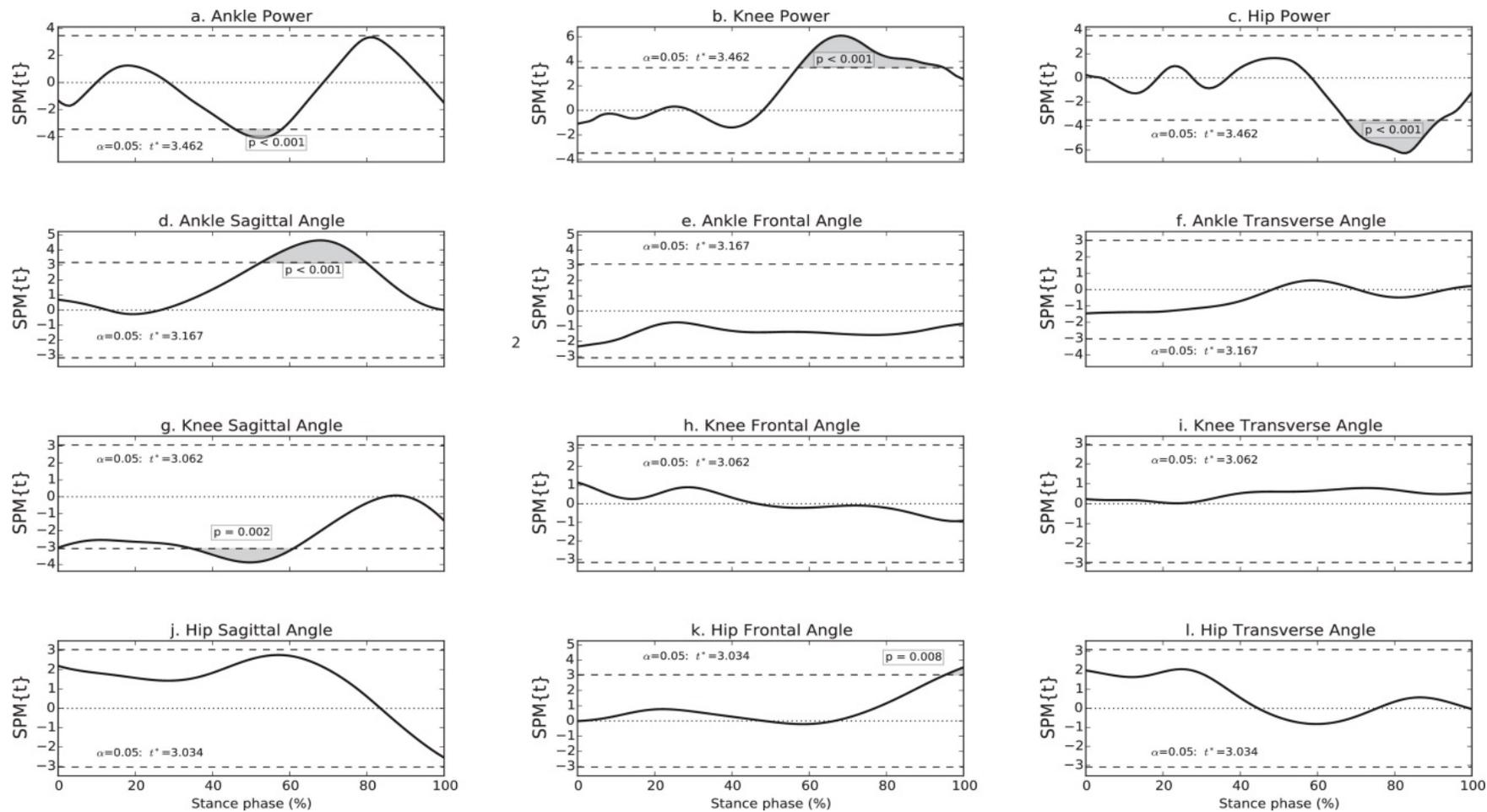


Figure 5-3 Representative post hoc analyses at 5.0 m/s. Shaded regions represent significant supra-threshold clusters that exceeded random critical threshold.

5.8 Tables

Table 5-1 Vector and post-hoc scalar field analyses of dependent variables

Vector Field	Velocity (m/s)	Significance	Stance phase (0 % = IC, 100 % = TO)	Post-hoc scalar field	Relation	Stance phase (0 % = IC, 100 % = TO)
Joint power	3.0	S	49.5 % to 93.5 %	Ankle Power	NS	-
				Knee Power	Positive	57.5 % to 94.5 %
				Hip Power	Negative	57.0 % to 89.0 %
	4.0	S	44.0 % to 93.0 %	Ankle Power	Negative	42.5 % to 58.5 %
				Knee Power	Positive	59.0 % to 71.0 %
				Hip Power	Negative	76.0 % to 86.0 %
	5.0	S	48.0 % to 96.0 %	Ankle Power	Negative	45.5 % to 57.5 %
				Knee Power	Positive	57.5 % to 94.5 %
				Hip Power	Negative	67.5 % to 91.5 %
Ankle angles	3.0	NS	-	Sagittal	NA	-
				Frontal	NA	-
				Transverse	NA	-
	4.0	S	55.0 % to 73.0 %	Sagittal	Positive	62.5 % to 76.5 %
				Frontal	NS	-
				Transverse	NS	-
	5.0	S	45.5 % to 76.5 %	Sagittal	Positive	52.5 % to 79.5 %
				Frontal	NS	-
				Transverse	NS	-
Knee angles	3.0	NS	-	Sagittal	NA	-
				Frontal	NA	-

				Transverse	NA	-
	4.0	NS	-	Sagittal	NA	-
				Frontal	NA	-
				Transverse	NA	-
	5.0	S	48.0 % to 51.0 %	Sagittal	Negative	34.5 % to 60.5 %
				Frontal	NS	-
				Transverse	NS	-
Hip angles	3.0	S	93.0 % to 100.0 %	Sagittal	NS	-
				Frontal	Positive	92.0 % to 100.0 %
				Transverse	NS	-
	4.0	S	98.0 % to 100.0 %	Sagittal	NS	-
				Frontal	Positive	94.0 % to 100.0 %
				Transverse	NS	-
	5.0	S	98.0 % to 100.0 %	Sagittal	NS	-
				Frontal	Positive	96.0 % to 100.0 %
				Transverse	NS	-
<p>IC = initial contact; TO = toe-off; S = significant; NS = not significant; NA = not applicable (Used when primary level vector field analysis did not reach significance. Post-hoc scalar field was not pursued or not interpreted); Relation = positive indicates as load magnitude increases, there is an increase in the positive magnitude of dependent variables; Relation = negative indicates as load magnitude increases, there is an increase in the negative magnitude of dependent variables</p>						

5.9 Supplementary material

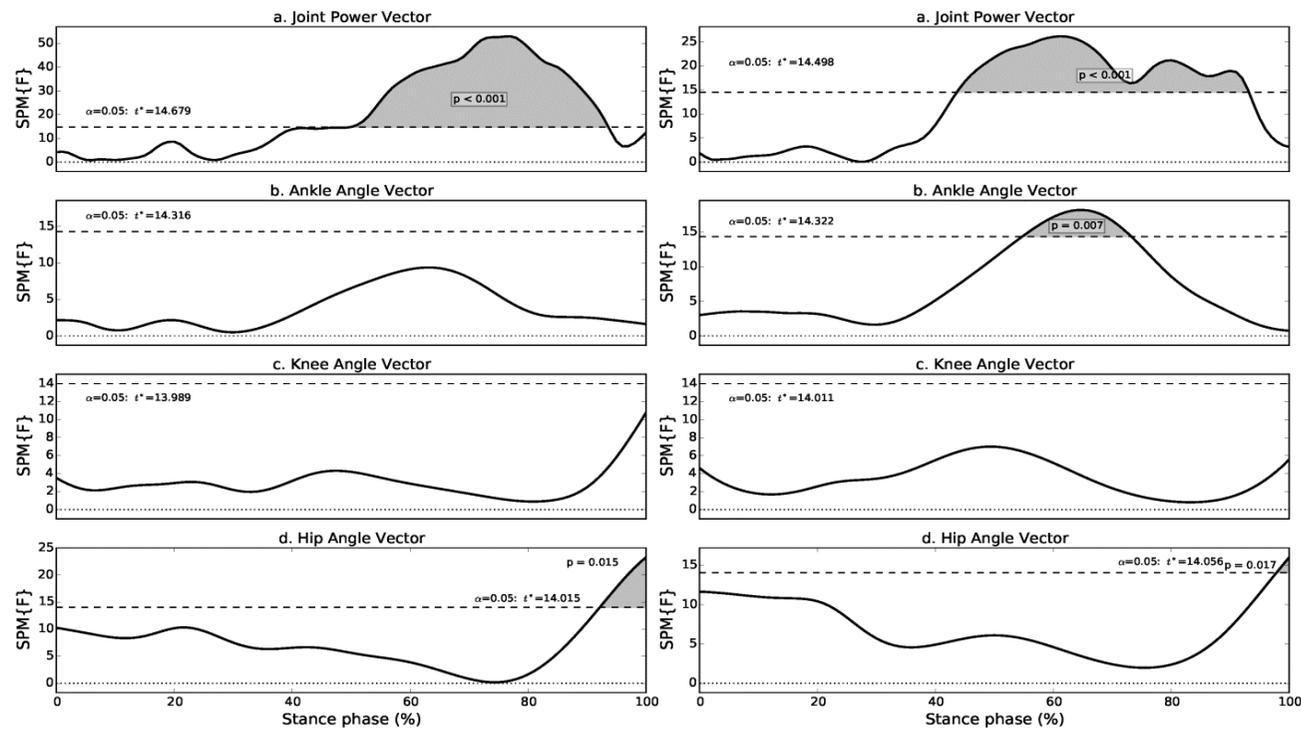


Figure 5-4 Vector field analysis at 3.0 m/s (left column) and 4.0 m/s (right column)

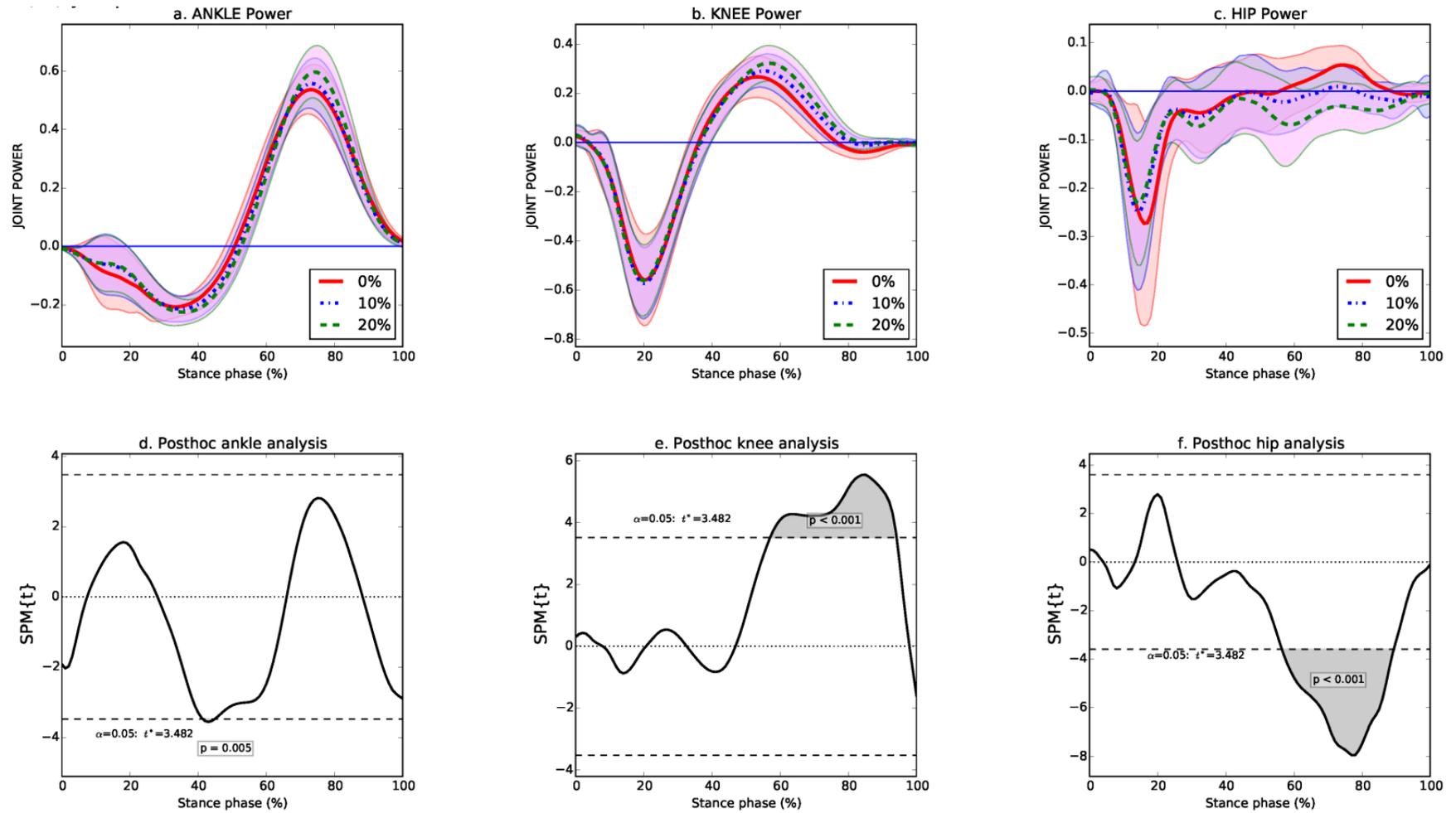


Figure 5-5 Mean (standard deviation) joint power curves, with post-hoc analysis at 3.0 m/s

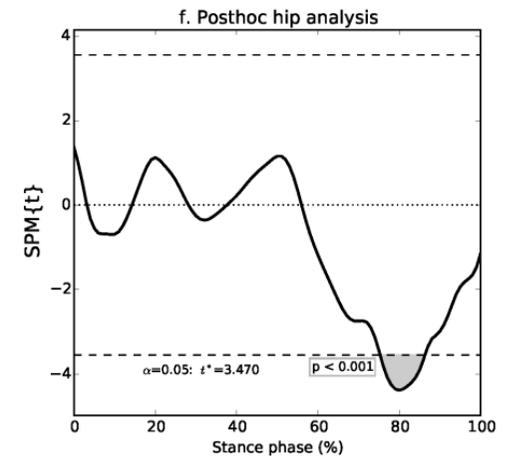
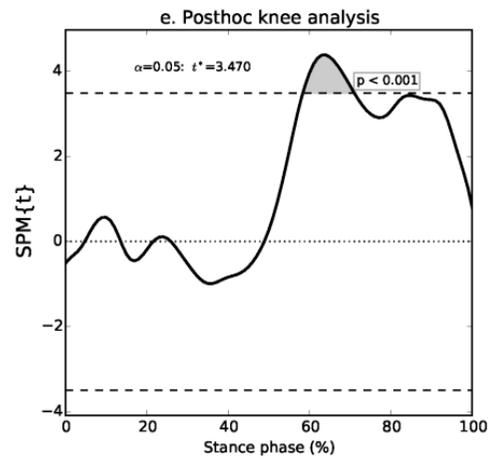
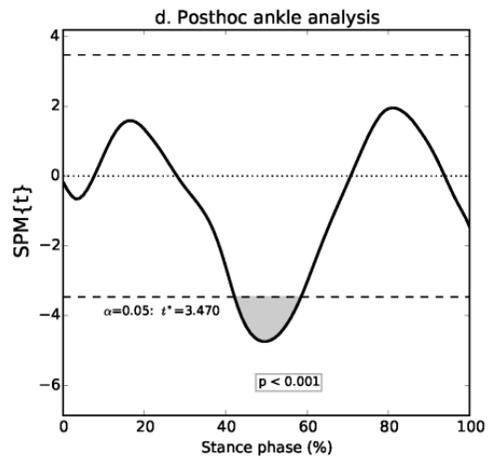
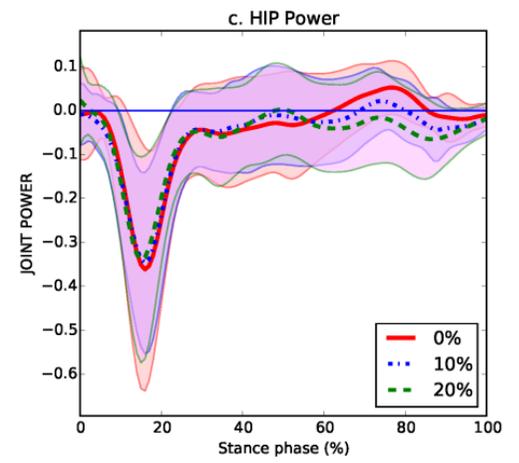
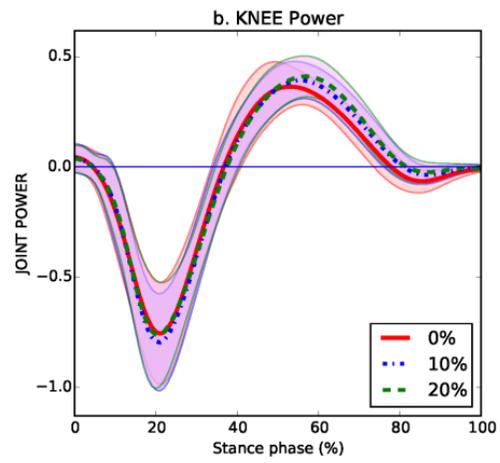
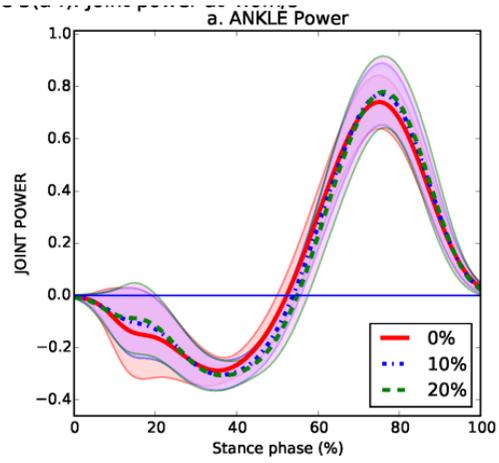


Figure 5-6 Mean (standard deviation) joint power curves, with post-hoc analysis at 4.0 m/s

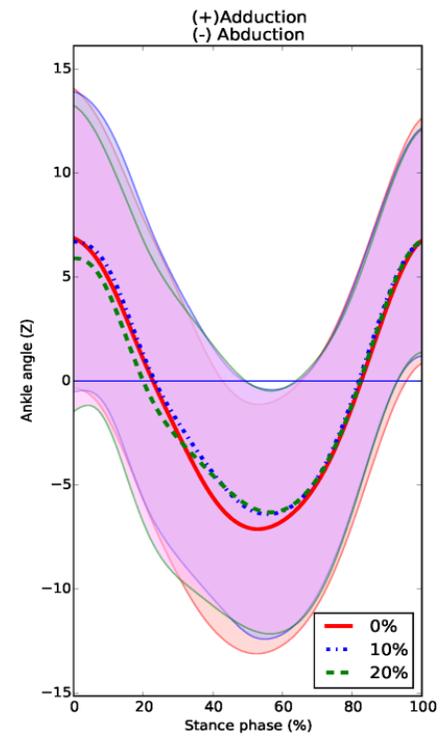
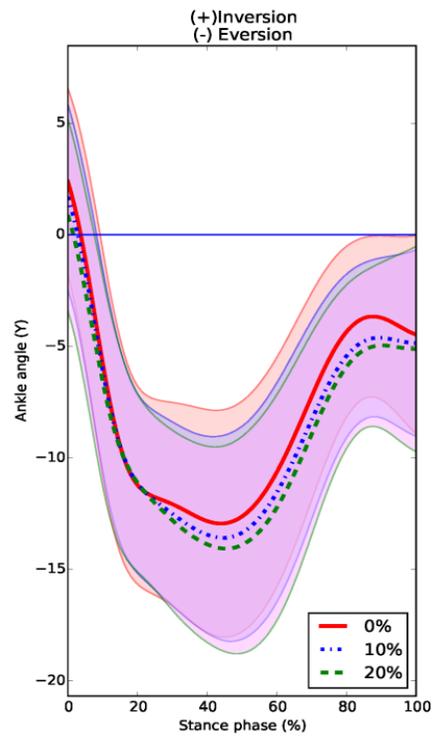
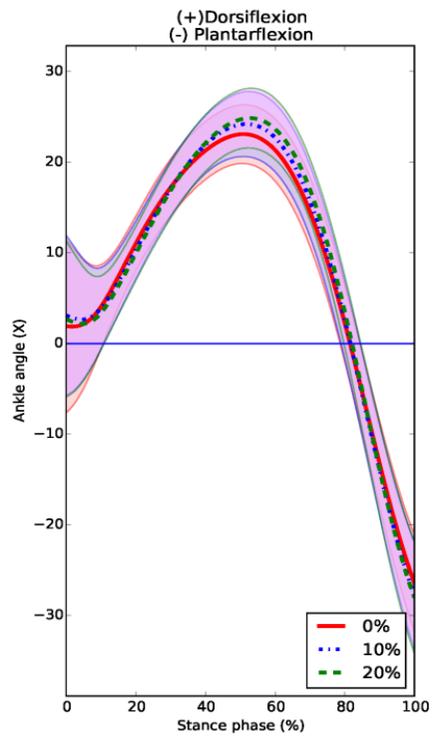


Figure 5-7 Mean (standard deviation) ankle angle curves at 3.0 m/s

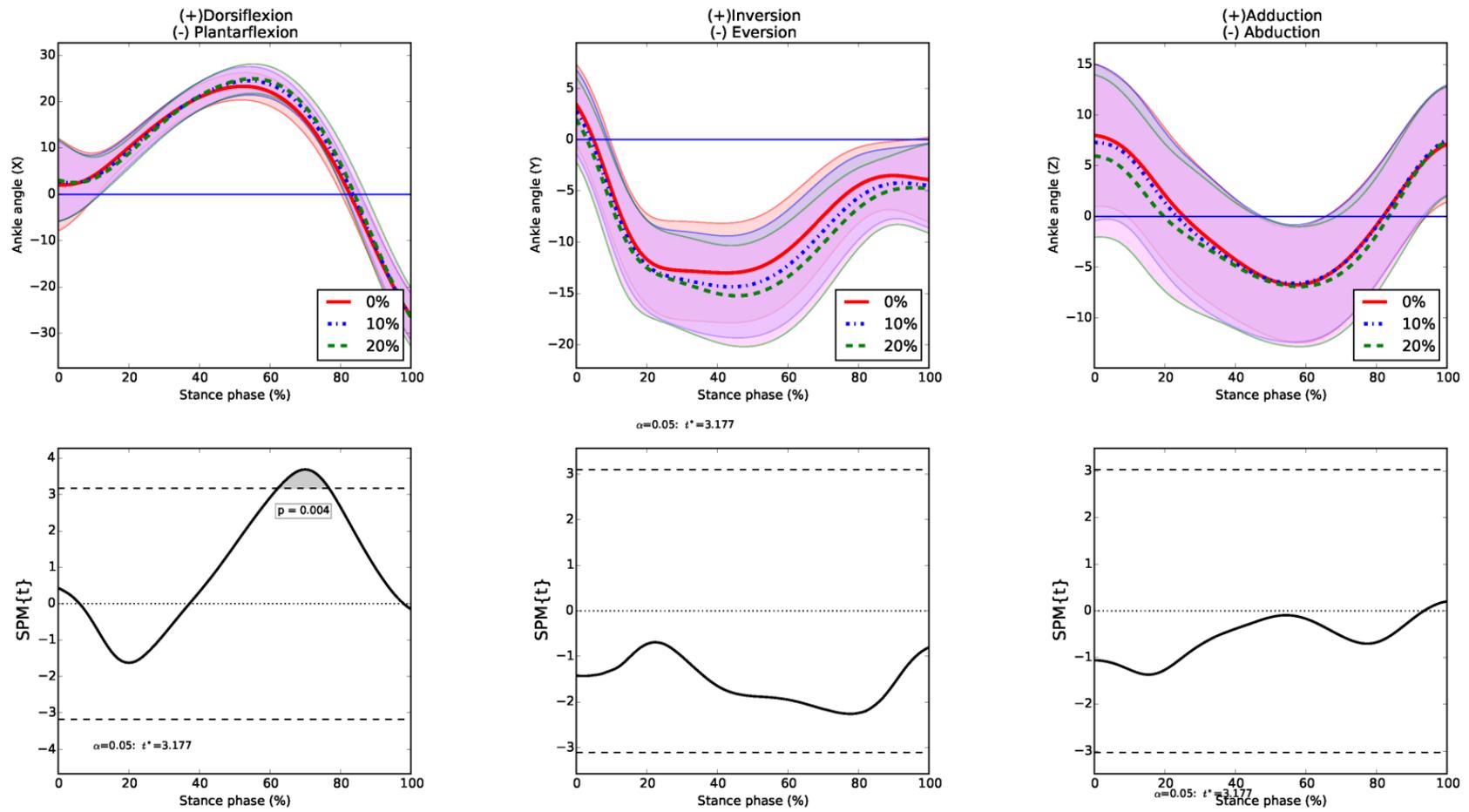


Figure 5-8 Mean (standard deviation) ankle angle curves, with post-hoc analysis at 4.0 m/s

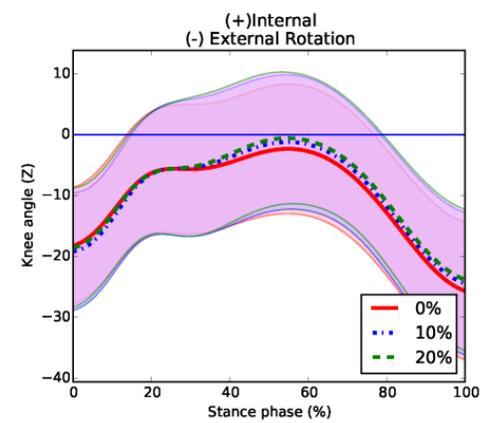
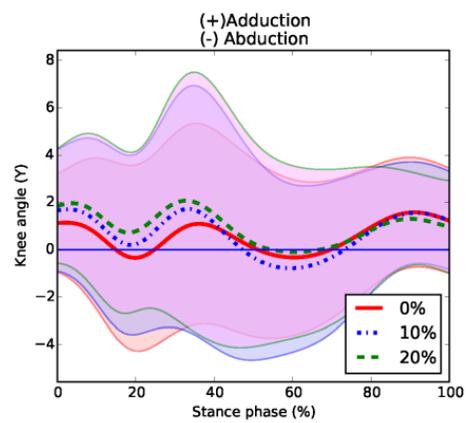
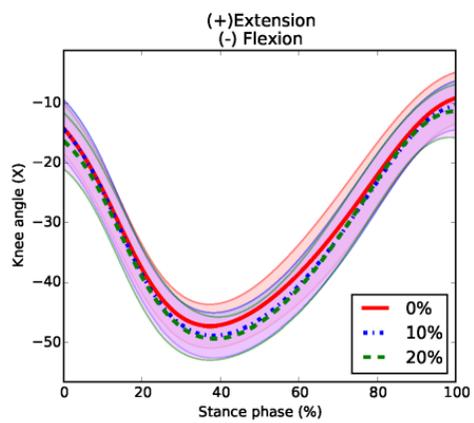
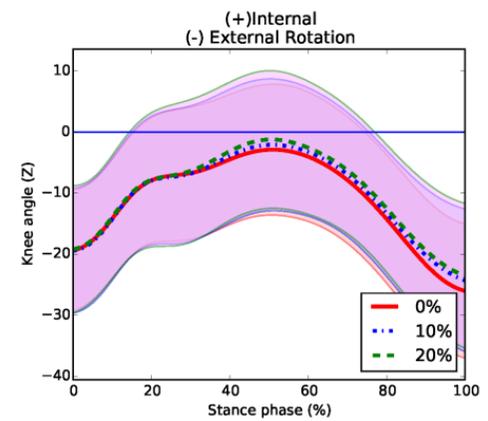
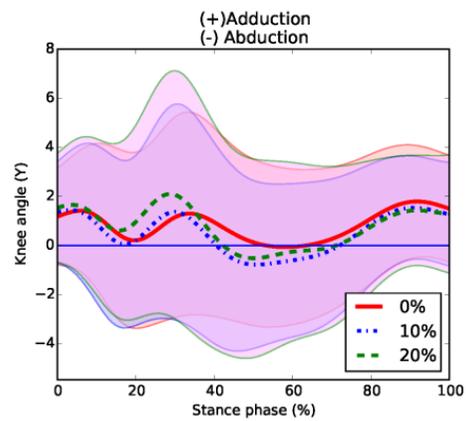
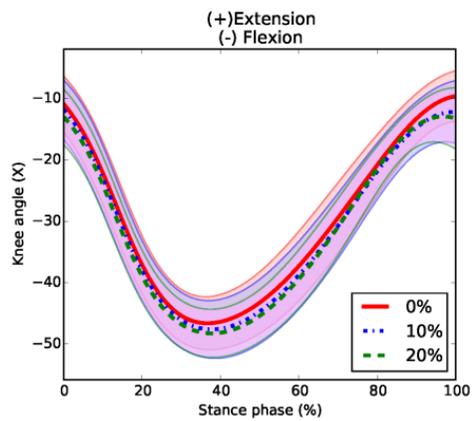


Figure 5-9 Mean (standard deviation) knee angle curves at 3.0 m/s and 4.0 m/s

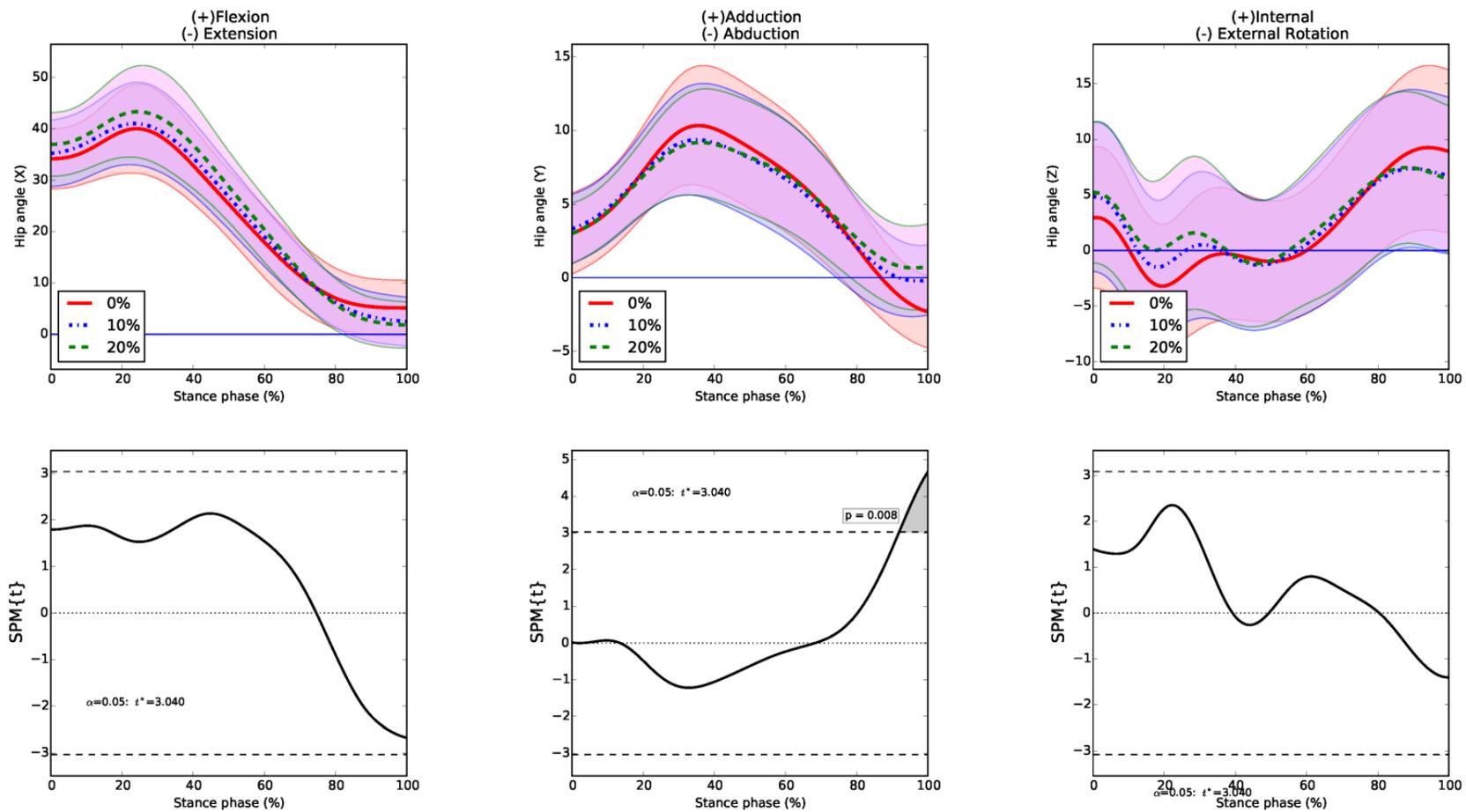


Figure 5-10 Mean (standard deviation) hip angle curves, with post-hoc analysis at 3.0 m/s

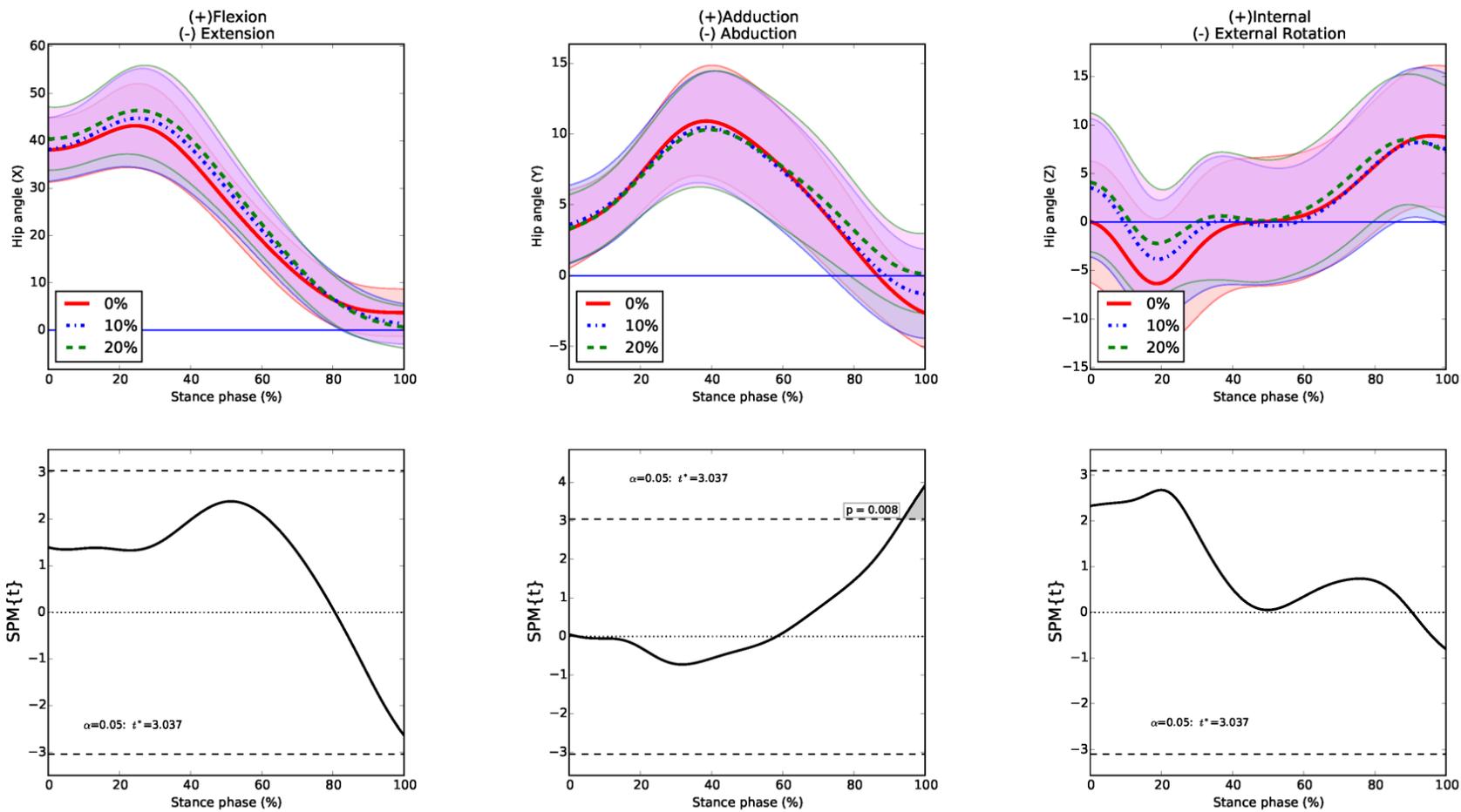


Figure 5-11 Mean (standard deviation) hip angle curves, with post-hoc analysis at 4.0 m/s

5.10 Conflict of interest and funding

The authors declare no competing interests.

5.11 Acknowledgement

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Chapter 6 Effects of two neuromuscular training programs on running biomechanics with load carriage: a study protocol for a randomised controlled trial

Synopsis: Designing interventions from background biomechanical knowledge requires knowing what factors to enhance and what factors to mitigate. Augmenting mechanical factors that could increase loaded running velocity could in turn reduce mechanical work expenditure at similar velocities. The mitigation of excessive non-sagittal plane kinematics and joint flexion angles may benefit loaded running injury risk potential. These kinematic changes could also reduce mechanical work expenditure by increasing the proportion of muscle work used for forward propulsion. We designed a training program, termed “Targeted” training, consisting of three exercises targeting three muscular variables: single legged hopping to augment leg stiffness; countermovement jumps to augment knee power generation; hip flexor cable pull to augment hip power absorption. This study has been published as “Liew BX, Morris S, Keogh JW, Appleby B, Netto K. Effects of two neuromuscular training programs on running biomechanics with load carriage: a study protocol for a randomised controlled trial. BMC Musculoskelet Disord. 2016 Oct 22;17(1):445”. The statement of primary contribution of the first author and the permission to reproduce the material can be found in the Appendix.

Abstract

BACKGROUND: In recent years, athletes have ventured into ultra-endurance and adventure racing events, which tests their ability to race, navigate, and survive. These events often require race participants to carry some form of load, to bear equipment for navigation and survival purposes. Previous studies have reported specific alterations in biomechanics when running with load which potentially influence running performance and injury risk. We hypothesize that a biomechanically informed neuromuscular training program would optimize running mechanics during load carriage to a greater extent than a generic strength training program.

METHODS: This will be a two group, parallel randomized controlled trial design, with single assessor blinding. Thirty healthy runners will be recruited to participate in a six weeks neuromuscular training program. Participants will be randomized into either a generic training group, or a biomechanically informed training group. Primary outcomes include self-determined running velocity with a 20 % body weight load, jump power, hopping leg stiffness, knee extensor and triceps-surae strength. Secondary outcomes include running kinetics and kinematics. Assessments will occur at baseline and post-training.

DISCUSSION: To our knowledge, no training programs are available that specifically targets a runner's ability to carry load while running. This will provide sport scientists and coaches with a foundation to base their exercise prescription on.

TRIAL REGISTRATION: ANZCTR (ACTRN12616000023459) (14 Jan 2016).

6.1 Introduction

Ever since the “Battle of Marathon” between Greece and Persia was recorded (Grogan, 1981), the accomplishment of running a 42 km marathon is seen as the ultimate achievement for a distance runner. However, the last two decades has seen both recreational and elite level runners striving for distances well beyond a standard marathon. Interest and participation in ultra-endurance races (Cejka et al., 2014), multi-stage racing events (Joslin et al., 2014), and off-terrain trail and adventure races have risen (Knechtle et al., 2015) as individuals seek new ways to challenge human limits. These races not only test a runner’s speed and endurance, but also their ability to navigate and survive over undulating terrains and harsh environments (Joslin et al., 2014). Navigation and survival requires routine access to specialized equipment and sustenance. This requirement necessitates athletes to compete with externally carried loads (Marais and de Speville, 2004). Few studies have considered the role of load carriage on the potential impact on a runner’s health and performance (Liew et al., 2016b). In addition, no studies have considered if runners can be trained to adapt to external loads in running.

Load carriage in running poses two fundamental problems to athletes and occupational personnel: 1) an increased injury rate, 2) and increased metabolic energy expenditure that may reduce performance (Knapik et al., 2004). In the adult population, the increased overuse injury rate associated with load carriage has largely been investigated in the military setting, where load carriage biomechanics have been predominantly investigated while walking (Knapik et al., 2004). A previous study reported that 8 % of the 5000 injuries reported in the Australian Defence Force from January 2009 to December 2010, were related to heavy load carriage (Orr et al., 2015). Of these injuries, 56 % affected the lower limb and were classified as muscular stress related (Orr et al., 2015).

Although no causative studies have been performed, it is likely that load carriage while running may exacerbate the already high incidence of running related injuries (Lopes et al., 2012). In addition, when an individual runs with load, the energy demand involved in maintaining constant running speed is increased (Teunissen et al., 2007). Minimising the reduction in running speed associated with load carriage is important for the survivability of military personnel, the performance of athletes, and the overall efficiency of movement in recreational runners (Carlton and Orr, 2014; Solomonson et al., 2015).

The risk of injury and reduced performance associated with load carriage in running, points to the need for a preconditioning program for these athletes. There is convincing evidence that resistance based neuromuscular training programs are effective at reducing running related injuries (RRI) during body weight (BW) running (i.e. running with no external load) (Lauersen et al., 2014), and improving BW running performance (Barnes and Kilding, 2015). However, current training programs have been developed using BW running research (Barnes et al., 2014; Clansey et al., 2014), rather than loaded running research. The only studies that have attempted to define best training practices for load carriage gait have been performed in the military setting (Knapik et al., 2012). A limitation in existing training studies has been that exercise prescription has not been explicitly informed from biomechanical studies of load carriage gait. Rather, training was of a generalised nature, targeting the large muscle groups of the lower limb (Knapik et al., 2012). The type of exercises and mode of contractions used for preconditioning programs should be specific to the gait pattern required for athletes, and be based on prior knowledge of biomechanical adaptations during load carriage.

6.1.1 Potential adaptive and maladaptive biomechanics

Studies using computed muscle control and induced acceleration analysis have identified the integrated roles of lower limb muscles in BW running. Collectively, the functions of these muscles are to provide a vertical force to accelerate/decelerate body weight, and horizontal forces to accelerate/decelerate inertial mass (Hamner and Delp, 2013; Hamner et al., 2010). When additional load is imposed on a runner, greater vertical and horizontal forces are needed to accelerate and decelerate an increased total weight and total inertial mass, respectively.

Biomechanical changes to running with load are classified as adaptive if they enable an increase in baseline motor function (Table 6-1). For example, an increase in ankle power absorption in mid stance with load may be adaptive as it transfers power away from proximal segments to the foot (Liew et al., 2016b). An increase in eccentric ankle plantar flexor strength indicates greater potential for ankle power absorption, and so may enable faster running velocities to be maintained during load carriage.

6.1.2 Rationale

Load carriage in running is increasingly common in running related sports. The ability to positively and predictably adapt to the imposed load when running necessitates an evidence-based training program. Existing training studies for load carriage performance in the military setting cannot be immediately applied to load carriage running, as most studies investigated performance in walking. This is because running and walking involve different movement dynamics, making the extrapolation of results from walking studies problematic when applied to running. For example, the hip contributes approximately 20 % of total positive power in the stance phase of BW walking, but less than 10 % of total positive power in the same phase of BW running (Schache et al., 2015). Second, studies that have investigated ways to improve load carriage performance have adopted a non-randomized design (Knapik et al., 2012). Reported effect sizes of benefit in intervention studies were larger in trials without a randomized design compared to one with a randomized design (Savovic et al., 2012). Lastly, studies on load carriage do not appear to specifically target the known neuromuscular demands of load carriage gait patterns (Knapik et al., 2012). Therefore, the purpose of this investigation is to compare the effects of a biomechanically informed neuromuscular training program to a generic standard best-practice resistance training program on changes in the biomechanics of running with load.

6.1.3 Objectives

To compare changes in (1) self-determined running velocity with and without load carriage, (2) lower limb running kinematics and kinetics, (3) jumping power and hopping stiffness, (4) and isokinetic knee and ankle extensor strength in healthy adult runners participating in a biomechanically informed training program compared to a generic resistance training program. This generic resistance training program may be seen as the current “gold-standard” program based on current best evidence (Knapik et al., 2012).

6.2 Methods

6.2.1 Research design

The study is a single blinded, parallel-grouped randomized controlled trial which will be designed and reported according to the Consolidated Standards of Reporting Trials (CONSORT) statement (Figure 6-1) (Moher et al., 2001). This study is registered with the Australian New Zealand Clinical Trials Registry (ACTRN: ACTRN12616000023459). The Curtin University Human Research Ethics Committee (RD-41-14) has approved this study protocol. All participants will provide written informed consent prior to study inclusion.

6.2.2 Participants, setting and recruitment

Runners with a variety of training experience residing in Western Australia will be invited to participate. All assessments and intervention will be conducted within Curtin University, Perth, Australia. Participants between 18 and 60 years old who are in good general health, and have been running or participating in running-related sports with a minimum cumulated total of 4 km/week or 45 minutes/week over the past 12 months, will be recruited. Exclusion criteria include: the presence of any disorders that could affect their gait and load carrying ability; medical conditions that preclude heavy resistance training and strenuous running; presence of a training-loss running related injury within the last three months (Kluitenberg et al., 2016); current running related pain (except blisters or muscle soreness) (Kluitenberg et al., 2016); lower limb surgery within the past 12 months; and females who are pregnant.

6.2.3 Sample size calculation

This study was powered on the effects of a core stability program on changes to hopping leg stiffness (Dupeyron et al., 2013). Sample size was planned based on a two way, repeated measures ANOVA, using the Hotelling-Lawley Trace to test for an intervention by time interaction (Guo et al., 2013). Previous studies on leg stiffness reported a standard deviation of 3600 N/m (Dupeyron et al., 2013), and a correlation between repeated measures of 0.80 (Pruyn et al., 2013). For a desired power of 0.80, and a Type 1 error rate of 0.05, 24 participants are needed to detect a between group mean difference of 3000 N/m. In order to account for a 20 % dropout over the six week intervention period, 30 participants will be tested.

6.2.4 Randomization, allocation and blinding

Prior to randomization, participants will be stratified into two groups based on their gender. Previous studies have identified gender differences in BW running mechanics and different associative relationships between running mechanics and economy (Barnes et al., 2014; Phinyomark et al., 2014). Permuted block randomization will be performed within each stratum, using two different block sizes (two blocks of four and three blocks of eight) (Doig and Simpson, 2005). Each block consists of either four or eight group assignments (half of the assignments to one of two groups), to ensure that at the end of each block, the number of participants allocated to either intervention group will be balanced. The sequence within each block, and the order of all five blocks per stratum are randomized. The randomization sequence will be generated in using an online random sequence generator used in a previous study (Baltich et al., 2014). When a participant has provided signed-informed consent, an external allocator not involved in the experiment, will sequentially draw an envelope (lowest numbered to highest) from either of two containers, depending on the stratum. The allocator will write the participant's name, identifier number, date, and allocator's signature on the envelope, which will be ink-printed onto the treatment allocation card via carbon paper. The envelope's seal will be broken and the participant and the trainer will be informed about the allocated intervention group. Participant and trainer blinding will not be feasible due to the nature of prescribed intervention. Outcome assessor (for biomechanical and strength assessments) will be blinded to the allocation of participants to intervention groups.

6.2.5 Subjective assessment

The following data will be collected at baseline: 1) participant's demographics, 2) self-reported running training, load carriage, and strength training history; 3) self-reported medical status (including the Physical Activity Readiness Questionnaire); 4) self-reported running overuse injury history (Clarsen et al., 2013; Kluitenberg et al., 2016); 5) self-reported running race performance history; 6) and use of any orthotics during running or training.

6.2.6 Three dimensional motion capture - load carriage running protocol

Participants will be carrying a backpack (CAMELBAK, H.A.W.G.® NV, 14 litre) fitted with chest strap and hip belt secured snugly. Participants will be wearing their personal running attire and running shoes for all assessments. Participants will run over ground, in a straight line at two velocities: 1) self-determined velocity, and 2) 3.5 m/s, over two external load conditions (0 % and 20 % BW). A lead up distance of at least 20 m to the edge of the first force plate and tail off distance of 10 m after the edge of the last force plate will be given to enable sufficient distance for acceleration and deceleration. A velocity of 3.5 m/s was predetermined as it closely represented the running velocity of the fastest annual men in 24 hour ultra-marathons (Zingg et al., 2013). Sand bags will be filled to 20% BW and secured within the backpack. A load of 20 % BW will be used as this represents a common relative load borne by tactical athletes during periods of running (Carlton and Orr, 2014). Indirect evidence from recommended backpack volume during ultra-endurance and adventure racers, suggest that a 20 % BW load is a reasonable approximate to actual load magnitude carried (Marais and de Speville, 2004). The order of load-velocity testing will be completely randomized using an online random sequence generator (Liew et al., 2016b). Timing gates (SMARTSPEED Pro, Fusion Sport Pty Ltd, Australia) will be positioned on both sides of the force plates (AMTI, Watertown, MA) (2000 Hz) 5 m apart, to monitor running velocity. Familiarisation trials will be given to practice the required running velocity. Participants will be required to complete a minimum of five successful running trials for each condition. Each trial would be interspersed with a 30 s rest break. A successful trial is defined when the right limb contacts the middle of the force platform without alteration of running pattern, within ± 10 % of the prescribed speed. A minimum of five minutes rest will be given after each condition.

6.2.7 Three dimensional motion capture – jumping and hopping protocol

Squat jumps (SJ), countermovement jumps (CMJ), and single leg vertical hopping (SL hop) will be performed on the force plates. For both the SJ and CMJ, the assessor will demonstrate and instruct technique, and participants will have to practise these jump techniques until performance is stable. For all tests, participants will be required to fix their arms at 90° abduction (“T” pose), to limit the influence of upper body on jump performance (McErlain-Naylor et al., 2014).

For the SJ, participants will perform a maximal concentric vertical jump from an initial squat depth of approximately 90° knee angle (visual estimation). For the CMJ, participants will descend into a squat depth of approximately 90° knee angle (visual estimation), followed without pause by a rapid maximal vertical jump and landing in a comfortable squat position (Gheller et al., 2015). When the assessor is satisfied with the practice performances, three maximal SJ trials and three maximal CMJ trials will be performed and recorded. Both SJ and CMJ will be performed with and without an external 20 % BW load. Each trial will be interspersed with a 30s rest and each test interspersed with a one minute rest to avoid fatigue. SL hopping will be performed for both legs separately over four conditions: 1) self-paced hopping frequency (BW), 2) self-paced hopping frequency with 20 % BW carriage, 3) hopping at a frequency of 3.0 Hz (BW), and 4) hopping at a frequency of 3.0 Hz with 20 % BW. Hopping frequency will be set using a handheld digital metronome. Each hop condition will last 10 s, with a one minute break interspersed (Hobara et al., 2013). Trials will be repeated if hopping frequency is not maintained. The order of testing will be randomized using an online random sequence generator

6.2.8 Data measurement and processing

An 18 camera motion capture system (Vicon T-series, Oxford Metrics, UK) (250 Hz), synchronised to three consecutive in-ground force plates (three metre long in direction of progression) will be used to collect motion and force data for running, hopping and maximal jump tasks. Data will be captured and stored using manufacturer supplied software (Vicon Nexus, v2.3, Oxford Metrics, UK). Data processing will be performed in Vicon Nexus and Visual 3D (C-motion, Germantown, MD).

Marker trajectories and ground reaction forces will be filtered at identical cut-off values, for use in inverse-dynamics calculations (van den Bogert and de Koning, 1996). For joint angles, raw marker trajectories will be filtered at a separate cut-off frequency. The choice of cut-off frequencies for motion and force data will be based on past research (van den Bogert and de Koning, 1996). A Cardan XYZ rotation sequence will be used to calculate 3D joint angles (Cole et al., 1993), and both kinematics and kinetics will be expressed in an orthogonal frame in the proximal segment (Schache and Baker, 2007). Data will be computed during the stance and swing period of running. A threshold of 20 N in ground reaction force will be used to determine initial contact and toe-off. Kinetic variables will be normalized using base factors of gravitational constant g (9.81 m/s^2), leg length, L (m), and body mass, M (kg) (Huang and Kuo, 2014).

6.2.9 Biomechanical modelling

Individual retro-reflective markers will be attached to anatomical bony landmarks, and marker cluster-shells to limb segments of the thorax, pelvis, bilateral thigh, shank, and foot segments (Hamner and Delp, 2013). An eight segment, 27 degrees of freedom (DOFs) model will be constructed from a static standing trial (Hamner and Delp, 2013). The position and orientation of each segment will be calculated using an inverse kinematic (IK) algorithm in Visual 3D (Lu and O'Connor, 1999).

6.2.10 Strength assessment (Isokinetic dynamometry)

An isokinetic dynamometer (HUMAC NORM, Computer Sports Medicine Inc., Stoughton, MA) will be used for strength testing, according to manufacturer's guidelines. For the ankle plantar flexor assessment, participants will lay supine (inclination at 30° above horizontal) with the hip and knee positioned at 60° and 80° of flexion, respectively (Webber and Porter, 2010). Based on the manufacturer's manual, the dynamometer's axis will be positioned in line with the lateral malleolus of the test limb. Next, the input arm penetration depth and foot plate penetration depth will be adjusted to approximate the dynamometer's axis to the ankle axis visually. The testing ankle range will be set from 10° dorsiflexion to 30° plantarflexion. For knee extension testing, participants will be seated in the machine with the hip flexed to 80°. The axis of rotation will be aligned to the femoral condyles with the knee flexed at 90°. Testing range will be set from 0° (complete knee extension) to 90° knee flexion.

For all testing, appropriate stabilization of segments will be applied using Velcro straps according to manufacturer's testing guidelines, and gravity correction mode will be used (de Araujo Ribeiro Alvares et al., 2015). Participants will first perform a standardised warm up protocol, consisting of 10 knee extension/flexion repetitions and ankle plantar flexion/dorsiflexion repetitions (90°/s), at a submaximal effort (Baroni et al., 2013). Concentric knee extensor and ankle plantar flexor torque will be assessed at an angular velocity of 60°/s. For each muscle assessment, two sets of six maximal concentric contractions will be performed, with a between set rest period of 1 minute. Between test and between side rest periods of 3 minutes will be provided.

6.2.11 Intervention

6.2.11.1 Familiarization phase (all participants) (two weeks)

All participants will first be enrolled into a two week preparatory training phase prior to baseline testing and randomization. During this phase, all participants will perform the same set of exercises (see Additional file 1 “Familiarization phase”). This preparatory phase will control for the effect of motor learning on improvements in performance on the assessments (Amarante do Nascimento et al., 2013; Feltner and MacRae, 2011; Glass, 2008). Participants will be encouraged not to alter any of their personal training regimes throughout the entire program. Self-reported training for the period of the intervention will be recorded by the participants in a supplemented training diary log book. Variables to report for self-resistance training include external mass magnitude, sets, and repetitions. Variables to report for cardiovascular training include duration, and type of exercise undertaken.

6.2.11.2 Training phase (six weeks) - Standardized warm up (both groups)

Participants from both groups will begin each exercise session with a 15 minute warm up consisting of four active, dynamic stretches consisting of 1) lunge, 2) “Good Morning” hamstrings, 3) squats, and 4) calf raises off a step (see Additional file 1 “warm up”). Each dynamic stretch will be performed using only the BW as resistance. Each dynamic stretch will involve two sets of ten repetitions (Yessis, 2009).

6.2.11.3 Generic neuromuscular resistance training group (GT group)

The principle governing this training program is that participants perform progressively heavier isoinertial (constant external mass) resistance training, at intensities from approximately 80 % progressing to approximately 88 % of one repetition maximum (1 RM). This program will involve three training sessions per week for six weeks (total 18 sessions). Inter-set rest duration of up to three minutes will be provided (de Salles et al., 2009). Exercises will include isoinertial bilateral leg press (Cybex® Plate Loaded Squat Press, Cybex International, Inc.), unilateral calf raises (Cybex® Plate Loaded Seated Calf, Cybex International, Inc.), and lunge (Cybex® Plate Loaded Smith Press, Cybex International, Inc.).

For the leg press, foot placement will be shoulder width, and the depth of foot placement on the plate will be such that at 90° of knee flexion, the tip of the toes is in line with the knee and shoulders. For the calf raises, the foot will be positioned at the level of the 1st metatarsophalangeal head (i.e. “ball” of the foot). For the lunge, the length of foot placements will be determined as a position that would enable approximately 90° knee flexion of the lead leg at the lowest position of the lunge. The foot of the trailing leg will be positioned such that the trailing knee is slightly posterior to the hip at the lowest position of the lunge. These exercises were selected as they represented generic lower limb exercises used in current load carriage training studies (Knapik et al., 2012). The intensity, repetitions, sets, rest duration, and description of each exercise will be gradually built up over the six weeks, and is described in the Additional file 1 (see “Exercises progression table (General training group)”).

6.2.11.4 Biomechanically informed neuromuscular resistance training group (BIT group)

The principle of this training program is that key neuromuscular requirements of load carriage running are targeted by specific neuromuscular exercises. This program will involve three training sessions per week for six weeks (total 18 sessions), and will involve SL hopping, CMJ, and hip flexor pull (Cybex® Bravo Pull, Cybex International, Inc.). For the plyometric component (hop and CMJ), intensity will be varied using a weighted vest. A maximal mass of 20 % BM will be added, as previous studies have demonstrated a reduction in peak power with heavier loads (Suzovic et al., 2013). A previously published review indicated that sessions incorporating more than 50 foot contacts per session resulted in the most benefit for jump performance (de Villarreal et al., 2009).

In order to maintain peak power application for the CMJ and hip flexor pull, a cluster set method (multiple sets of two to three repetitions with 10 s inter-set breaks) will be used (Moreno et al., 2014). The hip flexor pull will involve a range of 5° hip extension to 90° hip flexion (visual estimation), and a single hand-hold support will be used to maintain balance. For the CMJ, a depth of 80° to 90° knee flexion will be visually estimated and used for all participants. For SL hopping, participants will be encouraged to generate hopping power from the ankle joint, with the knee kept in a relatively “isometric”, slightly flexed posture. SL hopping will not involve a cluster-set method as the exercise intrinsically involves continuous, repetitive cycles of fast stretch-shortening cycle. The intensity, repetitions, sets, rest duration, and description of each exercise will be gradually built up over the six weeks, and is listed in the Additional file 1 (see “Exercises progression table (Biomechanically informed training group)”).

6.2.11.5 Augmented feedback (both groups)

Augmented feedback (AF) during all exercises for both groups will be provided to participants, using the principles of motor skill learning (Schmidt and Lee, 2014) (see Additional file). This is to enhance the learning and retention of optimal exercise performance in both groups, with the intention that sub-optimal lower limb kinematics with load may be corrected post-intervention.

First, AF that directs an individual's attention to the consequence of a movement (i.e. external focus of attention) has been shown to result in better motor learning and retention. Second, AF will be provided before (demonstration and instruction), during (mirror feedback and physical/verbal guidance), and after (knowledge of performance) each set of exercises in the initial stages, progressing to feedback delivered only after each exercise set. Feedback based on knowledge of performance (KP) will be provided in a prescriptive sense (i.e. what you should perform) at the initial stages, progressing to descriptive sense (i.e. what was performed) in the later stages of training. This is to allow participants to self-formulate correctional motor strategies in the later stages. The frequency of AF will be reduced from occurring at every set in the early stage of learning, to the last set of an exercise in the later stages. Previous research has shown that introducing a time delay, from motor task completion to feedback delivery, especially when participants self-evaluate their performance during this time lag, improves motor skill learning (Schmidt and Lee, 2014).

6.2.11.6 Determining initial training loads and progression

During the familiarization phase, for all isoinertial exercises a 10 RM will be utilised. A 1 RM load will then be derived from a 10RM load using a regression table for novice strength trainers (Table 6-2) (Baker, 2008). Load intensity will progress from 80 % of estimated 1 RM (equivalent to a 10RM load) in the first two weeks, to 84 % of estimated 1RM in the next two weeks (equivalent to an 8RM load), to 88 % of estimated one RM (equivalent to a 6RM load) in the final two weeks. The number of repetitions performed per set will be two repetitions less than the repetition maximum (Sampson and Groeller, 2015). Estimated 1 RM load for each exercise will be adjusted by a weekly increase of approximately 2.5 % to account for progression in strength. The rate of progression was based on a previous study on time course for strength gains, which demonstrated approximately 20 % increase in measured 1 RM in six weeks (Abe et al., 2000). For the hopping and CMJ, a weekly increase in load carried (approximately 5 % BW per week) will be used, until a limit of 20 % BW is reached.

6.2.11.7 Between group differences in training volume

Training volume as quantified by the number of sets, repetitions, and load magnitude will not be exactly matched between groups. This could mediate any potential between group intervention effects (Krieger, 2009). However, the aim of this study is to test two ecologically realistic models of training on load carriage running mechanical outcomes.

6.2.11.8 Intervention adherence

Participants attending $\geq 70\%$ of all training sessions ($\geq 13/18$ sessions) will be classified as high adherence, whereas those attending $< 70\%$ will be classified as low adherence.

Adherence will be calculated from attendance records in each participant's exercise training records. Adherence to prescribed neuromuscular training has been previously reported to be an important effect modifier in these programs (Sugimoto et al., 2014; Sugimoto et al., 2012). Efforts to increase participant adherence include, weekly mobile text (short messaging service) reminders and an exercise diary.

6.2.12 Dependent variables and statistics

For the SJ and CMJ, peak power using inverse dynamics and the force plate approach will be derived (Jandacka et al., 2014). For SL hopping, leg stiffness at each condition will be derived. For the self-paced running tasks with and without load carriage, average self-paced running velocity will be derived over a complete stride. Discrete variables of individual joint positive and negative work, total and net joint work for stance and swing phase of all running trials will be derived. Spatio-temporal variables of stance and swing duration, stride length, and cadence will be derived for all running trials. Time series of the three dimensional joint angles, moments, and powers of all three joints will be extracted for all running trials. Three-dimensional leg stiffness in running will be calculated in a three-step process from an adapted method in a previous study (Coleman et al., 2012). First, a three dimensional leg length will be defined as the vector from the hip joint centre to the centre of pressure. Second, the component of the resultant three dimensional ground reaction force (GRF) projected onto the leg vector will be calculated (taking the dot product of the GRF vector with the unit vector of the leg). Lastly, leg stiffness will be derived using the ratio of projected GRF (at the time of peak resultant GRF) to the change in leg length (between initial contact to peak GRF). For strength analysis, average peak concentric torque and power, and absolute peak concentric torque and power, of the knee extensors and ankle plantar flexors will be extracted.

Descriptive statistics (mean and standard deviation) will be calculated for baseline demographics of participants. Between groups difference in baseline demographics will be calculated using t-test or non-parametric test where appropriate. Analysis will be based on an intention-to-treat (ITT) using the multiple imputation method (Powney et al., 2014). A repeated measures linear mixed model with time, group, and their interaction as fixed effects, and participants clustered within groups as random effects will be used to analyse our discrete dependent variables (West et al., 2014). For the linear mixed model, significance will be set at $\alpha = 0.05$. Descriptive statistics and linear mixed modelling will be performed in R software within RStudio (Version 0.98.1062, RStudio, Inc.). Statistical testing between group and within group mechanical wave form data (kinematics, kinetics) will be analysed using Statistical Parametric Mapping (SPM). Statistical significance will be inferred using Random Field Theory (RFT), with appropriate Bonferroni correction applied to retain a family-wise error rate of $\alpha = 0.05$. SPM will be performed using the latest version of spm1d package (www.spm1d.org), installed in Python 2.7, and implemented in Enthought Canopy 1.5.4 (Enthought Inc., Austin, USA).

6.3 Discussion

Carrying some form of external load is becoming increasingly ubiquitous in running related sports, such as adventure racing and ultra-endurance events. Running mechanics alterations with load carriage could represent adaptive or maladaptive mechanics. Mechanical changes like increased joint power may represent attempts at maintaining constant running velocity, support an increased weight, maintain postural control and/or attenuate excessive impact shocks. In addition, some mechanical changes are likely to represent a failed capacity of lower limb muscles to cope with the additional load, that result in a reduction in running performance and an increased risk of future injuries. The long term sequela not only has an effect at an individual level, but could affect long term sporting participation and health care costs. In addition, runners may have to compromise running economy and running velocity when load carriage is involved if lower limb muscles are not tuned to the specific neuromuscular demands. Velocity decrements as a result of load carriage would result in compromised survivability in combat soldiers, and reduced performance in competing running athletes. This study will provide preliminary evidence of the potential efficacy of a targeted neuromuscular training program or a best-practice strength training program on improvements in strength, stiffness, running velocity and biomechanics during load carriage running.

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6.5 Figures

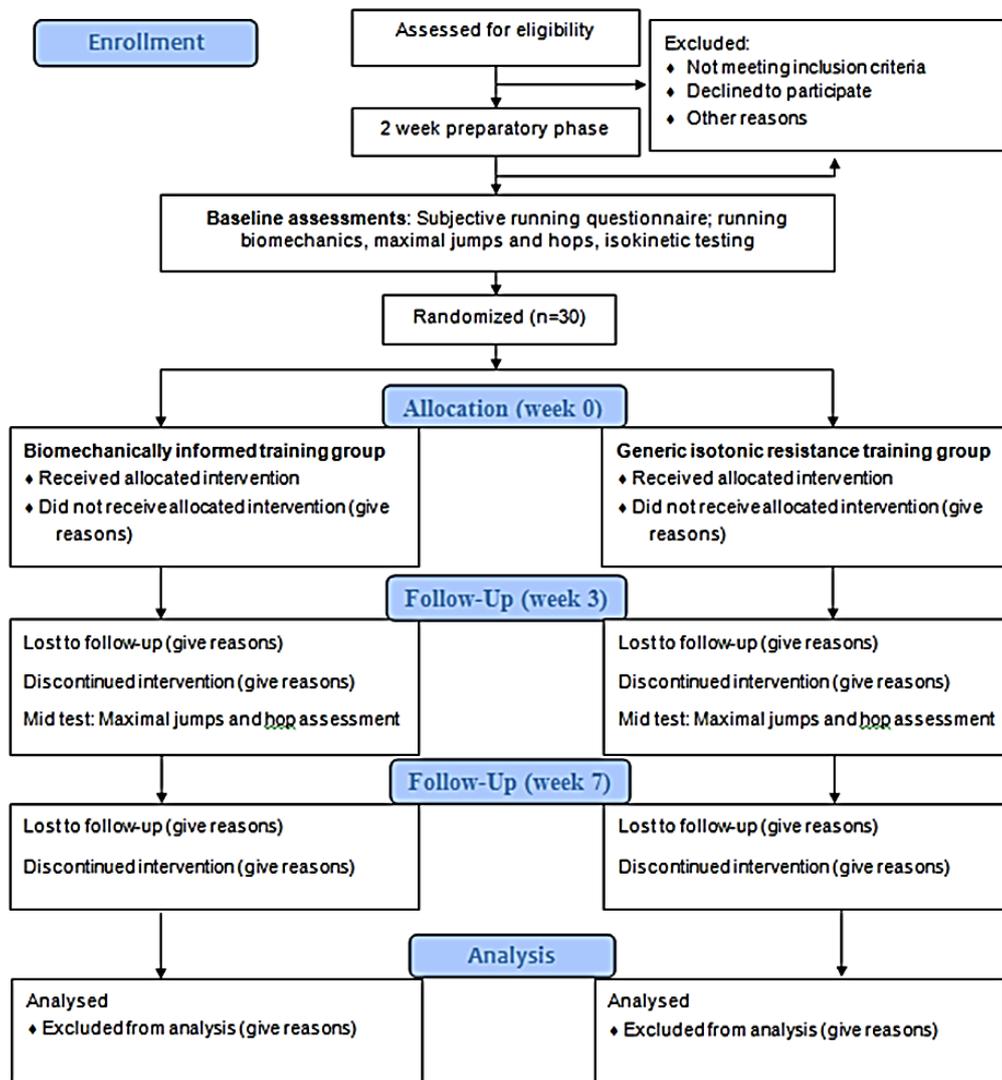


Figure 6-1 CONSORT Flow Diagram

6.6 Tables

Table 6-1 Biomechanical adaptations of load carriage to potentially optimize metabolic cost and minimise injury risk

• Potential positive adaptation	• Biomechanical changes with load	• Potential negative adaptation
<ul style="list-style-type: none"> • Transfer energy from foot to proximal segments (Siegel et al., 2004) • ↑ Energy stored as elastic energy (Lai et al., 2014) 	<ul style="list-style-type: none"> • ↑ Ankle negative power mid-stance (Liew et al., 2016b) 	<ul style="list-style-type: none"> •
<ul style="list-style-type: none"> • Accelerates leg into extension to ↑ energy transferred to proximal segments (Siegel et al., 2004) 	<ul style="list-style-type: none"> • ↑ Knee positive power late stance (Liew et al., 2016b) 	<ul style="list-style-type: none"> •
<ul style="list-style-type: none"> • ↑ Hip extension deceleration of trailing thigh segment for preparation into hip flexion swing (Dorn et al., 2012) • Transfers energy from trunk to trailing stance limb to prepare into swing (Siegel et al., 2004) 	<ul style="list-style-type: none"> • ↑ Hip negative power late stance (Liew et al., 2016b) 	<ul style="list-style-type: none"> •
<ul style="list-style-type: none"> • ↑ Elastic energy recovery (Brughelli and Cronin, 2008) • Avoid excessive vertical COM excursion and maintain ground reaction force alignment to stance limb (Caron et al., 2013; Moore et al., 2015) 	<ul style="list-style-type: none"> • ↑ Leg stiffness (Silder et al., 2015) 	<ul style="list-style-type: none"> •

<ul style="list-style-type: none"> • Architecture of triceps-surae muscle tendon unit makes it an efficient force generator (Lai et al., 2014) 	<ul style="list-style-type: none"> • Small role for inter-joint work redistribution (Liew et al., 2016a) 	<ul style="list-style-type: none"> •
<ul style="list-style-type: none"> • 	<ul style="list-style-type: none"> • ↑ Hip adduction late stance (Liew et al., 2016b) 	<ul style="list-style-type: none"> • Asymmetrical loading on knee soft tissues (Dierks et al., 2008)
<ul style="list-style-type: none"> • 	<ul style="list-style-type: none"> • ↑ Knee and ankle flexion mid-stance (Liew et al., 2016b) 	<ul style="list-style-type: none"> • ↑ COM vertical excursion (Silder et al., 2015) • ↑ Patellofemoral joint compression pressure and ↑ Achilles tendon compression (Bonacci et al., 2013; Rooney and Derrick, 2013)
<ul style="list-style-type: none"> • ↑ = Increase; ↓ = Decrease 		

Table 6-2 Guide for determining one repetition maximum in novice weight trainers.

% 1RM	100	96	94	92	90	88	86	84	82	80	78	76	74	72	70	68	66	64	62	61
Number Repetitions	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20
Reconvert factor	-	1.04	1.06	1.08	1.11	1.13	1.16	1.19	1.22	1.25	1.28	1.31	1.35	1.39	1.43	1.47	1.52	1.56	1.61	1.63

6.7 Additional files

Table 6-3 Familiarization phase (2 weeks (4 sessions))

Session	SL hop (Body Weight)	CMJ (Body Weight)	Hip flexor pull	Leg press	Calf raise	Lunge
1.1	2 sets of 20 hops	3 sets of 3reps	1 set of 10 reps (15RM load) 1 set of 10 reps (15RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (15RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (15RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (15RM load)
1.2	2 sets of 20 hops	3 sets of 3reps	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)
2.1	2 sets of 20 hops	3 sets of 3reps	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1 set of 10 reps (10RM load)
2.2***	2 sets of 20 hops	3 sets of 3reps	1 set of 10 reps (10RM load)	1 set of 10 reps (15RM load) 1sets of 10 reps (12RM load) 2sets of 10 reps (10RM load)	1 set of 10 reps (10RM load)	1 set of 10 reps (10RM load)

Note	Used as warm up and technique phase	Used as warm up and technique	Machine	Machine	Machine	Machine
SL = Single leg; CMJ = Countermovement jumps; RM = repetition maximum; reps = repetitions						

Table 6-4 Training phase- warm up

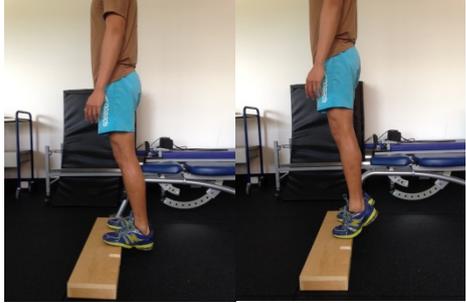
<p>Lunge X 2 sets 10 repetitions per side</p>	 A side-view photograph of a man in a brown t-shirt and light blue shorts performing a lunge. He is standing on his right leg with his left leg stepped back and bent at the knee. His arms are crossed in front of his chest.
<p>Good Morning X 2 sets 10 repetitions</p>	 Two side-view photographs of a man in a brown t-shirt and light blue shorts performing a Good Morning exercise. The left image shows him standing upright with his hands behind his head. The right image shows him leaning forward at the hips, with his torso nearly parallel to the floor.
<p>Squats X 2 sets 10 repetitions</p>	 Two side-view photographs of a man in a brown t-shirt and light blue shorts performing a squat. The left image shows him standing upright with his hands behind his head. The right image shows him in a squatting position with his hips below his knees and his back nearly parallel to the floor.
<p>Bilateral calf raises X 2 sets 10 repetitions</p>	 Two side-view photographs of a man in a brown t-shirt and light blue shorts performing bilateral calf raises. He is standing on a wooden block with both feet. The left image shows him standing flat-footed on the block. The right image shows him on his toes, with his heels raised off the block.

Table 6-5 Exercises progression table (Biomechanically informed training group)

Exercise/ Week (session)	SL hopping (Sets.Time.Frequency)	Total % BW performed for hops	CMJ (Sets.Reps)	Total % BW performed for CMJ	Hip flexor pull (Sets.Reps.Load)	Total reps per limb
1.1	2.20s.2.2 Hz	110	10.2	100	8.2.10RM	124
1.2	2.20s.2.2 Hz	110	5.3	100	8.2.10RM	119
1.3	2.20s.2.2 Hz	110	10.2	100	8.3.10RM	132
Total week 1	264		55		56	375
2.1	2.20s.2.2 Hz	110	10.2	105	10.2.10RM	128
2.2	2.20s.2.2 Hz	110	5.3	105	10.2.10RM	123
2.3	3.20s.2.2 Hz	110	10.2	105	10.3.10RM	182
Total week 2	308		55		70	433
3.1	2.20s.2.2 Hz	115	10.2	110	8.2.8RM	124
3.2	2.20s.2.2 Hz	115	5.3	110	8.2.8RM	119
3.3	4.20s.2.2 Hz	115	10.2	110	8.3.8RM	220
Total week 3	352		55		56	463
4.1	2.20s.2.2 Hz	120	10.2	115	10.2.8RM	128
4.2	2.20s.2.2 Hz	120	5.3	115	10.2.8RM	123
4.3	4.20s.2.2 Hz	120	10.2	115	10.3.8RM	226
Total week 4	352		55		70	477

5.1	2.20s.3 Hz	120	10.2	120	8.2.6RM	156
5.2	2.20s.3 Hz	120	5.3	120	8.2.6RM	151
5.3	4.20s.3 Hz	120	10.2	120	8.3.6RM	284
Total week 5	480		55		56	591
6.1	3.20s.3 Hz	120	10.3	120	10.2.6RM	220
6.2	2.20s.3 Hz	120	10.2	120	10.2.6RM	160
6.3	3.20s.3 Hz	120	10.2	120	10.3.6RM	240
Total week 6	480		70		70	620
Total weeks 1-6						2959 reps
	Inter set rest = 3 mins Inter exercise rest of 3 to 5 mins		Inter set rest = 10 s Inter exercise rest of 3 to 5 mins		Inter set rest = 3 mins Inter exercise rest of 3 to 5 mins	
SL = Single leg; CMJ = Countermovement jumps; RM = repetition maximum; reps = repetitions; % BW = percentage body weight; mins = minutes						

Table 6-6 Exercises progression table (General training group)

Exercise/ Week (session)	Leg press (Sets.Reps.Load)	Calf raises (Sets.Reps.Load)	Lunge (Sets.Reps.Load)	Total reps per limb
1.1	2.8.10RM	2.8.10RM	2.8.10RM	48
1.2	2.8.10RM	2.8.10RM	2.8.10RM	48
1.3	3.8.10RM	2.8.10RM	2.8.10RM	56
Total week 1	56	48	48	152
2.1	3.8.10RM	2.8.10RM	2.8.10RM	56
2.2	2.8.10RM	2.8.10RM	2.8.10RM	48
2.3	3.8.10RM	3.8.10RM	3.8.10RM	72
Total week 2	64	56	56	176
3.1	2.6.8RM	3.6.8RM	3.6.8RM	48
3.2	2.6.8RM	2.6.8RM	3.6.8RM	42
3.3***	3.10.10RM***	2.6.8RM	1.6.8RM	48
Total week 3	54	42	42	138
4.1	2.6.8RM	2.6.8RM	2.6.8RM	36
4.2	4.6.8RM	3.6.8RM	2.6.8RM	54
4.3	4.6.8RM	4.6.8RM	3.6.8RM	66
Total week 4	60	54	42	156

5.1	3.4.6RM	3.4.6RM	3.4.6RM	36
5.2	3.4.6RM	2.4.6RM	2.4.6RM	28
5.3	4.4.6RM	3.4.6RM	3.4.6RM	40
Total week 5	40	32	32	104
6.1	4.4.6RM	4.4.6RM	4.4.6RM	48
6.2	2.4.6RM	2.4.6RM	2.4.6RM	24
6.3***	3.10.10RM***	3.4.6RM	3.4.6RM	54
Total week 6	54	36	36	126
Total				852
	Inter set rest of 3 mins Inter exercise rest of 3 to 5 mins			
RM = repetition maximum; reps = repetitions; % BW = percentage body weight; mins = minutes				

Table 6-7 Augmented feedback (AF) cues for biomechanically informed training group

Exercises	Optimal technique	Feedback cues
Leg press	<p>Sit erect with back pressed against back of seat</p> <p>Feet should width apart, hips slighted external rotated. Tip of toes, knee, and should be in a straight line at 90° knee flexion.</p> <p>Push footplate away by extending the hips and knees.</p> <p>Do not lock knees</p> <p>Bring footplate back till 90° knee flexion, and smoothly transition back to full extension.</p>	<p>Verbal:</p> <ol style="list-style-type: none"> 1. “Keep knees over toes”
Calf raise	<p>Set erect, and position foot such that 1st MTP head is at the edge of the foot plate.</p> <p>Push up from the balls of the feet to raise the heel as high as possible.</p> <p>Avoid lifting thigh and leaning backwards when pushing up.</p>	<p>Verbal:</p> <ol style="list-style-type: none"> 1. “Tighten foot arch as you push up” 2. “Keep trunk tall”
Lunge	<p>Set anterior-posterior foot width such that 90° knee flexion of lead limb at lowest point of descent.</p> <p>Lower hips down till forward limb’s thigh parallel to floor.</p> <p>Flex trail limb’s knee, without touching floor.</p> <p>Raise hips by pushing off with both limbs while exhaling to return to start position.</p> <p>Maintain erect trunk.</p>	<p>Verbal:</p> <ol style="list-style-type: none"> 1. “Trunk and hips stay tall” 2. “Keep knee over toes”

Table 6-8 Augmented feedback (AF) cues for general training group

Exercises	Optimal technique	Feedback cues
Single leg hop	Bilateral shoulder and pelvis level in single leg stance Hip, knee, and ankle joint centre should be in the same sagittal plane Avoid excessive knee flexion	Verbal: <ol style="list-style-type: none"> 1. “Trunk and hips stay tall” 2. “Spring from your ankle”; “stay tall at the knees” 3. “Keep landing soft” 4. “Spring up as fast as you can “ Visual: <ol style="list-style-type: none"> 1. Use of mirror during exercise Auditory: <ol style="list-style-type: none"> 1. Ensure soft landing

<p>Countermovement jump</p>	<p>Preparatory and eccentric phase: Feet approximately shoulder width apart Adequate trunk, hip, knee, ankle flexion on descent Adequate shoulder extension Amortization: Transit from eccentric to concentric phase fast with no pause Concentric phase: Full trunk, hip, knee, ankle extension during flight No lateral lean of trunk Full shoulder flexion Landing: Soft landing Landing position same as preparatory position Should land as close as starting position with little AP and lateral translation</p>	<p>Verbal:</p> <ol style="list-style-type: none"> 1. "Feet and knees shoulder width apart"; 2. "Maintain weight over centre of foot"; 3. "Keep landing soft" 4. "Spring up as fast as you can", "drive with your shoulders" 5. "Stay tall in the air" 6. "Avoid kissing the knees" <p>Visual:</p> <ol style="list-style-type: none"> 1. Use of mirror during exercise 2. Check foot position and compare to original start position <p>Auditory:</p> <ol style="list-style-type: none"> 1. Ensure soft landing 2. Ensure only one contact sound
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<p>Hip flexor pull</p>	<p>Stand on one leg maintaining vertical trunk, level pelvis alignment</p> <p>From a position of hip extension, flex the opposite hip and knee to 90 degree flexion</p> <p>Return opposite limb to a position of hip extension</p> <p>Maintain global alignment and postural control</p> <p>Limit use of trunk flexion and lateral flexion during exercise</p>	<p>Verbal:</p> <ol style="list-style-type: none"> 1. "Trunk and hips stay tall"; 2. "Pull up with your thigh" 3. "Grip the floor with your standing feet"
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Table 6-9 Augmented feedback (AF) principle and schedule (Both groups)

Weeks	Focus of attention	Period of AF	Type of KP	Frequency of AF	Delay in AF
Familiarization	External	Before movement (Observation and demonstration); During movement (visual, verbal, physical guidance); After movement (KP verbal)	Descriptive and Prescriptive	Every set of an exercise	Almost instantaneous
1	External	Before movement (Observation and demonstration); During movement (visual, verbal, physical guidance); After movement (KP verbal)	Prescriptive	Every alternate set of an exercise	Delay ~ 5 to 10 secs
	↑ ↓	↑ ↓	↑ ↓	↑ ↓	↑ ↓
6	External	After movement (KP verbal)	Descriptive	Last set of an exercise (summary feedback)	Delay + concurrent subjective performance estimations
KP: Knowledge of performance					

6.8 Declarations

6.8.1 Acknowledgements

Not applicable.

6.8.2 Funding

No funds were received in support of this work. No benefits in any form have been or will be received from a commercial party related directly or indirectly to the subject of this manuscript. Mr Bernard Liew is currently supported by an institutional doctoral scholarship.

6.8.3 Availability of data and materials

Not applicable.

6.8.4 Authors' contributions

BL, SM, and KN conceived of the study and participated in its design and coordination. JK and BA provided in depth contributions to the design of the resistance training program and the assessment used in this study. BL is responsible for the coordination of the study, and is involved with the assessment of all participants. BL was responsible for training the trainers in conducting the resistance training program. All five authors were responsible for providing statistical supervision for this study. All five authors were involved in drafting the protocol, and contributed to and approved the final manuscript.

6.8.5 Competing interests

The authors declare that they have no competing interests.

6.8.6 Consent for publication

Written informed consent was provided by the participant for publication of their image in the Additional file of this manuscript.

6.8.7 Ethics approval and consent to participate

This study protocol design follows the Declaration of Helsinki and is approved by the Curtin University Human Research Ethics Committee (RD-41-14). All participants will provide written informed consent prior to study inclusion.

Chapter 7

Resistance training on load carriage mechanical work

Synopsis: In Chapter 4, we demonstrated that load carriage increased mechanical work in running. This could explain why load carriage increases running energy cost. A reduced mechanical work and a greater contribution of work coming from the ankle may represent an energetically more optimal loaded running pattern. Both “Targeted” and “General” training programs were similarly effective at increasing lower limb strength and jumping power. However, these muscular gains did not translate to significant changes in loaded running mechanical work and its distribution.

Surprisingly, only the “Targeted” training program reduced running leg stiffness. This suggests that “Targeted” training drove the optimization of other cost functions in loaded running, such as impact force management. Exercises designed to augment mechanical work optimization needs to couple muscular force gains with successful leg stiffness augmentation. In addition, resistance training may need to be supplemented with gait retraining to facilitate running coordination strategies that increase the proportion of “effective” muscle work used for forward body propulsion. This study has been prepared as an article for submission.

Abstract

Purpose: This study aimed to compare the effectiveness of a load carriage “Targeted” against a non-specific “General” training program to change loaded running mechanical work and its distribution, lower limb stiffness, power, and strength.

Methods: This was a parallel grouped, randomized controlled trial design with concealed allocation and assessor blinding, using 30 healthy adult recreational runners. Participants were randomly allocated to either a six week “General” training program (bilateral leg press, unilateral calf raises, and lunge), or a “Targeted” training program (countermovement jumps, single legged hopping, and hip flexor pull). Pre-post assessments included three dimensional motion capture of loaded (20 % body weight) running at 3.5 m/s, countermovement and squat jumps, and isokinetic strength testing. Primary outcomes included stance phase total joint work and proportion of ankle positive work in loaded running. Secondary outcomes included loaded running leg stiffness, peak jump powers, and ankle and knee extensor concentric strength.

Results: All 30 participants completed the trial. There were no significant time-by-group interactions in the primary outcomes. There were significant time effects on strength and power variables. Ankle strength improved between 0.23 (“Targeted”) and 0.34 Nm/kg (“General”); knee strength improved between 0.11 (“Targeted”) and 0.24 Nm/kg (“General”); countermovement jump power improved between 0.05 (“Targeted”) and 0.19 dimensionless units (“General”); and squat jump power improved between 0.09 (“Targeted”) and 0.18 dimensionless units (“General”). “Targeted” training reduced running leg stiffness by 2.27 more units (1.88 kN/m) than “General” training.

Conclusions: Although resistance training improved lower limb concentric strength and power, loaded running mechanical work was not significantly altered. The lack of transference of increased strength and power to mechanical work outcomes, suggests that mechanical work optimization is not the only goal in loaded running. Evidence related to the decrease in leg stiffness in the “Targeted” group suggests that impact force management may be one goal of load carriage running.

Key Words: Load carriage; Running; Mechanical work; Resistance training; Plyometric training; Jump; Strength

7.1 Introduction

Athletes participating and competing in ultra-endurance distance race events and commuters who run to work often must run with load (Lucas et al., 2016; Marais and de Speville, 2004; Schutz et al., 2012; Zander et al., 2014). Load carriage increases the mechanical work of running (Brown et al., 2014; Liew et al., 2016b; Liew et al., 2016c), which could explain the increase in energy cost (Simpson et al., 2017; Teunissen et al., 2007), and decrease in performance (Simpson et al., 2017).

The capacity to run with less mechanical work at a given velocity, and have a larger work contribution coming from the ankle, is considered an energetically optimal running pattern (Heise et al., 2011; Williams et al., 2013; Williams and Cavanagh, 1987). The importance of reducing mechanical work and managing its distribution during load carriage has inspired the development of mechanical assistive devices (Mooney et al., 2014; Panizzolo et al., 2016). Even though mechanical work and its distribution can be positively altered with assistive devices (Mooney et al., 2014; Panizzolo et al., 2016), it is unknown if these variables can be modified using physical training, although others have begun thinking about the possibility (Beijersbergen et al., 2017; Williams et al., 2016).

Training to improve leg stiffness, knee power generation, and hip power absorption could mediate a reduction in mechanical work and alter its distribution during loaded running (Liew et al., 2016a). This could explain why training for these variables improved body weight (BW) (i.e. with no external load) running performance (Balsalobre-Fernandez et al., 2016; Barnes et al., 2015; Deane et al., 2005; Denadai et al., 2016; Spurrs et al., 2003; Turner et al., 2003). Increased leg stiffness reduces the external lever arm of the ground reaction force (GRF) vector to joint centres (Chang et al., 2000; Wang et al., 2015), which reduces internal joint torque requirements. An increase in knee power generation could allow earlier extension of the thigh-shank segments in running, which could increase the forward accelerative potential of the ankle plantarflexors (Bobbert and van Soest, 2001; Steele et al., 2013). An increase hip power absorption in late stance could enhance the energy flow from the proximal trunk-pelvic segments and the contralateral swing limb, into the ipsilateral stance limb to aid in push off (Czerniecki et al., 1996).

This randomized controlled trial was undertaken to compare the effects of two training programs on mechanical work and its distribution in loaded running. Given the three mechanical factors that could mediate changes in mechanical work and its distribution, the intervention group (termed “Targeted” group) focused on increasing leg stiffness, knee power generation, and hip power absorption. The comparison group (termed “General” group) performed heavy-resistance leg exercises, which presently represents the “best” training practice for load carriage in walking (Knapik et al., 2012). The primary hypotheses for this study were that “Targeted” training would reduce total joint work and increase ankle joint work proportion during the stance phase of load carriage running at equivalent velocities, more than the “General” training. Our secondary hypotheses were that “Targeted” training would increase leg stiffness, jump power, knee and ankle concentric strength more than “General” training. In addition, we performed several analyses that explored the relationship between leg stiffness, jumping power, and concentric strength, and changes in total joint work for each group.

7.2 Methods

This study was an assessor blinded, randomized controlled trial, the protocol of which has been published (Liew et al., 2016a). The reporting of this trial follows the recommended CONSORT guidelines. The reporting of the Consensus on Exercise Reporting Template (CERT) guidelines (Slade et al., 2016) is included in the supplementary material (see supplementary: Table 7-3 to Table 7-6).

7.2.1 Participants

Thirty healthy adult recreational runners between 18 to 60 years old were invited via public advertisements to join the study conducted at Curtin University, Australia, between December 2015 and April 2016. Curtin University's Human Research Ethics Committee approved this study (RD-41-14). Participants had to be actively engaged in running or running-related sports with a minimum cumulated total duration of 45 minutes per week to be considered for inclusion. Given that the requirement of a previous study on novice runners was a minimum cumulated distance of 5 km per week (Baltich et al., 2014), a relatively average running velocity of 3 m/s would guarantee a minimum weekly duration of 28 minutes. The threshold of 45 minutes was determined to maximise the number of runners eligible for this study. Load carriage experience was not a requirement for inclusion as this study's focus was on the capability of resistance training to alter mechanical work in a laboratory model of load carriage running, rather than on its immediate ecologic applicability. The main exclusion criteria included: 1) medical conditions which precluded safe performance of running, jumping, and hopping with load carriage; 2) conditions which precluded safe performance of heavy resistance exercises; 3) running related injuries currently and within the past three months; 4) surgeries within the past year; and 5) females who are pregnant. Written informed consent was sought prior to study enrolment.

7.2.2 Procedures

An online random number generator was used to produce the randomization sequence, which was stratified by sex (Liew et al., 2016a). For each block, sequentially numbered envelopes (lowest to highest) were created, which was sealed and made opaque. After each participant completed the familiarization phase and baseline assessment, an independent allocator not involved in the study selected the next lowest numbered envelope from the block which matched the participant's sex. The allocator wrote the participants' details (name, unique number identifier and allocation date) on the envelope which was transferred to the assignment card via carbon paper. Next, the allocator opened the envelope and wrote the participant's group allocation onto an electronic list. This list was only available to the trainers delivering the intervention. Blinding of trainers and participants was not possible due to the physical nature of both interventions. Only the researcher who assessed the outcomes (B.L.) was blinded to the intervention group assignment.

7.2.3 Intervention

In the first phase (common to all participants), participants performed two weeks (twice per week) of familiarization training comprising of six exercises that formed the entire repertoire of both intervention groups (Liew et al., 2016a). In the second phase which lasted six weeks, participants were allocated into either the “General” or “Targeted” group. Participants in the “General” group performed heavy resistance, isoinertial training involving lunges, bilateral leg press, and seated unilateral calf raises, as these are commonly performed in military physical conditioning (Knapik et al., 2012). Training intensity started from 80 % of the baseline estimated 1RM during the first week, progressing to 88 % of the weekly adjusted 1RM during the sixth week (Liew et al., 2016a). The number of repetitions per set progressed from eight at week one, to four at week six. Participants in the “Targeted” group performed countermovement jump (CMJ) to enhance knee power generation; single legged (SL) hopping to enhance leg stiffness; standing single-legged hip flexor cable isoinertial pull to enhance hip power absorption. For the CMJ and SL hopping, training intensity started with BW progressing to wearing a loaded vest of 20 % BW at week six. For SL hopping, the frequency was fixed at 2.2 Hz up to week four, but increased to 3.0 Hz from week five to six. For the hip flexor pull, external load mass was adjusted from 80 % of the baseline measured 1RM during the first week, progressing to 88 % of the weekly adjusted 1RM during the sixth week (Liew et al., 2016a).

7.2.4 Outcome measure

All assessments were carried out immediately after the familiarization phase (baseline) and immediately after the six weeks training program (Liew et al., 2016a).

Three-dimensional (3D) motion capture: Participants performed repeated overground running at a fixed velocity of 3.5 m/s ($\pm 10\%$), whilst carrying a loaded backpack (fixed at a load of 20 % of force plate measured BW at each testing session) (Liew et al., 2016a). A run up distance of 20 m was given prior to the first force plate, and a tail off distance of 10 m was given prior to the last force plate. Participants performed a minimum of five successful over-ground running trials at 3.5 m/s. This velocity was selected as it represented the average running velocity of the fastest male competitor in 24 hour ultra-marathons (Zingg et al., 2013). Jump power was assessed by maximal vertical squat jumps (SJ) and CMJ with the same loaded backpack.

Passive reflective markers were attached to anatomical landmarks, and cluster-shells (of four markers) were applied to limb segments of the thorax, pelvis, bilateral thigh, shank, and foot (see in supplementary: Figure 7-5, Biomechanical modelling procedures). An eight-segment model with 27 degrees of freedom was created and tracked using an inverse kinematic algorithm within Visual 3D (C-motion, Germantown, MD) (Liew et al., 2016b). Marker trajectories were captured using an 18-camera motion capture system (Vicon T-series, Oxford Metrics, UK, 250 Hz), whilst ground reaction force (GRF) was collected using three synchronized in ground force plates (AMTI, Watertown, MA, 2000 Hz). Marker trajectories and GRF data were filtered at similar frequencies of 18 Hz for running (Liew et al., 2016b), and 8 Hz for jumping (Raffalt et al., 2016). The stance phase was defined between initial contact and toe off using a 20 N threshold (Liew et al., 2016b).

Strength assessment: Isokinetic concentric strength testing of the knee extensors and ankle plantarflexors was performed on a dynamometer (HUMAC NORM, Computer Sports Medicine Inc., Stoughton, MA), collecting data at 100 Hz and set up according to the manufacturer's guidelines. Both the right and left limbs were assessed. Knee extensor testing was performed in an 80° seated position, whilst ankle plantarflexor testing was performed in a 30° inclined position. For each muscle group tested, participants first performed one warm-up set (10 repetitions) of concentric-concentric contractions at 90°/s, followed by one minute of rest in-situ. Next, two sets of six maximal concentric-concentric contractions was performed at 60°/s, interspersed with one minute of seated rest in-situ. Between muscle group and side rest periods of three minutes were provided (Liew et al., 2016a).

Dependent variables: Descriptive variables include stride length and stance duration. Primary variables include total joint work in stance (absolute sum of positive and negative work of the hip, knee, and ankle) (Prilutsky et al., 1996), and percentage positive work performed by the ankle in stance. Secondary variables include running leg stiffness, and its skeletal and muscular subcomponents, SJ and CMJ peak positive total joint power, ankle and knee maximal extensor torque.

Leg stiffness was calculated using a modified kinematic-kinetic method (Liew et al., 2016a). Leg stiffness was deconstructed into muscular and skeletal components (see in supplementary: Leg stiffness calculation method) (Wang et al., 2015). The term muscular leg stiffness broadly encompasses the total nonlinear contribution of active (contractile elements) and passive (non-contractile serial and parallel elastic elements) components that provides resistive forces to leg length changes. Skeletal leg stiffness represents the resistance to movement provided by the inter-segmental skeletal alignment. In a purely extended position where the GRF passes directly through every joint centre, leg stiffness is afforded entirely by the skeletal component.

Joint power, work and leg stiffness were scaled to a percentage of a participant's body mass (M) and standing leg length (L) measured at each static calibration, and gravitational constant (g) where appropriate (power scaling factor: $M \cdot g^{1.5} \cdot L^{0.5}$, work scaling factor: MgL , stiffness scaling factor: Mg/L) (Liew et al., 2016b). One dimensionless power unit represents a mean (standard deviation) of 1947.5 (364.8) W, one work unit represents 565.9 (118.3) J, and one stiffness unit represents 828.9 (113.1) N/m.

7.2.5 Sample size

Given that leg stiffness augmentation is an important component of training, this study was powered on a previously reported change to vertical stiffness after a core stabilization program (Dupeyron et al., 2013). With a standard deviation of 3600 N/m and a correlation between repeated measures of 0.80, 24 participants were needed to detect a between group difference of 3000 N/m, at a power of 0.80 ($\alpha = 0.05$). Thirty participants were recruited to allow for a 20 % drop out proportion.

7.2.6 Statistical analyses

A linear mixed model with time (pre and post), group ("Targeted", "General"), and their interaction as fixed effects, was used to analyse the dependent variables. This was performed in R software (v 3.2.5) within RStudio (v0.99.902, RStudio, Inc.), using the "nlme" and "nlmeU" packages (Galecki and Burzykowski, 2013; Pinheiro et al., 2016). For all statistical models, different variances were allowed for each time (pre and post). A correlation structure of "compound symmetry" was used for each participant. A random intercept was used for each participant. Analysis was performed on an intention-to-treat basis. Since there were no drop outs and no cross-over in intervention, an intention-to-treat analysis is equivalent to a per-protocol analysis.

For the exploratory analyses, a simple linear regression was performed to analyse several relationships between the change of the dependent variable (total joint work), the independent variable (Group) and the change in candidate predictor variables (ankle and knee concentric strength, CMJ and SJ power, running muscular and skeletal leg stiffness). All significances were set at $\alpha = 0.05$.

7.3 Results

Thirty two out of 50 potentially interested participants met the inclusion criteria and participated in the first phase of training. One participant developed sudden 1st metatarsophalangeal joint pain (not due to the training) during this period and dropped out pre-randomization. Another participant declined to continue with phase two of the training for personal reasons. Thus, 30 participants were randomized into either the “General” or the “Targeted” training (Figure 7-1, Table 7-1). All participants completed the training program and were available for final endpoint analysis (Figure 7-1). The mean (standard deviation) percentage of load carried was 19.4 % (0.7 %) BW and velocity was 3.50 (0.11) m/s.

Descriptive variables: There were no Time, Group, or Time-by-Group effects on stance duration and stride length (Figure 7-2a, b, and see supplementary Table 7-7).

Primary variables: There were no Time, Group, or Time-by-Group effects on total joint work and percentage ankle positive work in the stance phase (Figure 7-2c, d, and see supplementary Table 7-8). The lack of statistical influence of our independent variables was also found in the constituent individual joint work and percentage work (supplementary Table 7-9, Table 7-10).

Secondary variables: For maximal ankle plantarflexor and knee extensor torque, there was a significant main effect of Time on the dependent variables but no Group or Time-by-Group interaction (Figure 7-3a, b, and see supplementary Table 7-11). Training improved maximal ankle torque by 0.34 Nm/kg (“General”) and 0.23 Nm/kg (“Targeted”), and maximal knee torque by 0.24 Nm/kg (“General”) and 0.11 Nm/kg (“Targeted”) (Figure 7-3a, b).

The main effect of Time on peak total joint power in the CMJ was significant, with a tendency towards a significant Time-by-Group interaction, where CMJ peak joint power increased by 0.19 units (370W) (“General”) and 0.05 units (97 W) (“Targeted”) post training (Figure 7-3c, and see supplementary Table 7-11). The main effect of Time on peak joint power in the SJ was significant, where SJ peak joint power increased by 0.18 units (351 W) (“General”) and 0.09 units (175 W) (“Targeted”) post training (Figure 7-3d, and see supplementary Table 7-11). There was no Group or Time-by-Group interaction for SJ power.

For running leg stiffness, there was a Time-by-Group training effect (Figure 7-4a). Leg stiffness during load carriage running reduced by 2.27 more units (1.88 kN/m) after training in the “Targeted” group, compared to the “General” group (see supplementary Table 7-12). In the “Targeted” group, there was a non-significant tendency that the skeletal component of leg stiffness reduced post training (Figure 7-4b, and see supplementary Table 7-12). However, there was no effect of the independent variables on the muscular component of leg stiffness (Figure 7-4c, and supplementary Table 7-12).

Post-hoc exploratory analyses: Of all the candidate mediating factors, only knee extensor torque was a statistically significant predictor to total joint work. A 0.1 Nm/kg increase in knee torque resulted in a reduction of total work by 0.007 units (4 J) in “General” group, but an increased total work by 0.006 units (3 J) in “Targeted” group (Table 7-2). Running muscular leg stiffness had a tendency towards significance with respect to its relationship with total joint work (Table 7-2). A 0.1 unit (82 N/m) reduction in muscular leg stiffness resulted in a reduction of total work by 0.001 units (0.6 J) in “General” group and reduction by 0.0008 units (0.4 J) in “Targeted” group (Table 7-2).

7.4 Discussion

Performing loaded running with less mechanical work or shifting the work to the ankle plantarflexors may underlie a more energetically optimal loaded running pattern. In this study, we compared the ability of two different resistance training programs to change loaded running mechanical work and its distribution. The findings of this study did not support our primary hypotheses, in that load carriage running mechanical work and its distribution did not statistically change after either resistance training program. Contrary to our secondary hypotheses, improvements in lower limb concentric strength and SJ power were similar in both groups, with a tendency towards greater CMJ improvements after “General” training compared to “Targeted” training. In addition, there was an unexpected finding that running leg stiffness was reduced after “Targeted” training and not after “General” training.

7.4.1 Effectiveness of interventions

Both training programs resulted in meaningful and significant increases to concentric strength and jump power, despite not influencing mechanical work and its distribution. The magnitude of improvements to muscular concentric strength and jumping power were comparable to previous studies. In the present study, jumping power after “Targeted” training improved by 5 % during the SJ, and 3 % during CMJ from baseline. A previous jump training study achieved slightly larger improvements in loaded (30 % BW load) CMJ and SJ vertical power of 9 % (Markovic et al., 2013). The slightly greater gains in jump power in the previous study could have been due to a slightly longer training period (8 weeks), the use of greater load magnitudes (30 % BW) during jump training, and greater jump training volume (1404 jumps) (Markovic et al., 2013). In the current study, the training period was shorter and involved a lighter load (0 % to 20 % BW), with a much lower jump training volume (345 jumps). A longer, 10 week (20 sessions) isoinertial training program increased knee extensor torque by 9 % (Walker et al., 2016), which was less than the effect observed in this study (11 % in “General”). The slightly smaller improvement in the previous study could be attributed to the inclusion of experienced strength-trained participants with relatively high baseline knee extensor torque (Walker et al., 2016).

Surprisingly, “Targeted” but not “General” training resulted in a reduction in loaded running leg stiffness, when the former’s objective was to increase stiffness. This finding was additionally surprising given that “Targeted” training concomitantly increased strength and power. These are variables whose improvements have typically been accompanied by increases to leg stiffness (Cormie et al., 2010a; Korff et al., 2009; Secomb et al., 2015). In fact, it appears that “General” training could have been the more effective method at augmenting leg stiffness, even though the increase was small and not statistically significant (Figure 7-4a). This could have been due to “General” training improving eccentric force capacity more than “Targeted” training. Even though eccentric strength was not measured, there was a tendency for “General” training to increase CMJ power more than “Targeted” training, but not SJ power. Indirectly, this points to greater dynamic eccentric force improvements after “General” over “Targeted” training (McGuigan et al., 2006).

The relatively greater improvements in dynamic eccentric force capacity could mean that “General” training better targeted muscular deficits in our cohort, compared with “Targeted” training. The present cohort was generally inexperienced in resistance training (Table 7-1). A previous study found that power based jump training tended to benefit individuals with a greater baseline muscular strength, compared to those with weaker baseline strength (Cormie et al., 2010b). In addition, active recreational adults have been shown to demonstrate a greater force-deficit, rather than a velocity-deficit during BW vertical jumps (Cuk et al., 2016). This could mean that “General” training provided a greater matched stimulus for muscular force gains compared with “Targeted” training (Jimenez-Reyes et al., 2016). By providing a closer matched training stimulus, greater muscular gains could be achieved within a short training period of six weeks after “General” compared to “Targeted” training.

7.4.2 Modifiability of running mechanical work

Despite the training improving physiological measures of power and strength, these gains were not translated into functional mechanical work improvements. This lack of positive change may be due to a lack of modification in running inter-segmental coordination. From a Dynamical Systems perspective, it would mean that “Targeted” training parameters did not facilitate the change in the appropriate “order” parameters needed to change inter-segmental coordination patterns (van Emmerik et al., 2016).

Improve loaded running mechanical work may occur only when a greater proportion of muscle work is used to perform “effective work” on the centre of mass (COM) for forward body propulsion (Bobbert and van Soest, 2001). Muscles are joint rotation actuators and so perform work on segments that result in changes in angular velocity (Bobbert and van Soest, 2001). In order for the COM to be displaced forward in running, the change in angular velocity in a series of linked body segments must be coordinated (Bobbert and van Soest, 2001).

Muscle mechanical work may reduce if the mechanics of load carriage running changes, such that more muscle work is used for “effective work”. The exploratory analyses revealed that a 0.1 Nm/kg increase in knee extensor concentric strength in the “Targeted” group would result in 0.006 units (3 J) increase in total joint work, but the same strength gain in the “General” group would result in 0.007 (4 J) reduction in total joint work (Table 7-2). Even though power and strength represent a continuum of a muscle’s force-velocity profile (Cormie et al., 2011), it appears that strength rather than power played a more important functional role in this cohort. However, a gain in concentric strength does not guarantee an energetic “benefit” to loaded running.

The energetic “benefit” of a gain in knee extensor concentric strength may occur only when “ineffective” work reduces. The concentric strength gained in the “Targeted” group occurred with a significant reduction in skeletal leg stiffness (Figure 7-4), but a slight increase in stiffness in the “General” group (Figure 7-4). Reduced skeletal leg stiffness in running would increase “ineffective” muscle work used to rotate limb segments (Chang et al., 2000; Wang et al., 2015). This may be similar to the reduced forward acceleration potential of muscles in individuals walking in crouched pattern (Steele et al., 2013; Steele et al., 2010). Although not statistically significant, “Targeted” training increased total joint work by 0.006 units (3 J) per stance, but “General” training reduced total work by 0.0006 units (0.3 J) per stance (Table 7-8 of supplemental). Given that for every 1 J of mechanical work aided by an exoskeleton reduced 1.8 J of metabolic cost per step in load carriage walking (Panizzolo et al., 2016), over many steps (e.g. 17 000 steps in a marathon with a stride length of 2.4 m) the energetic benefit of coupling strength gains and stiffness gains may be significant.

From the results, an interesting question remains as to why an increase in strength and power was coupled with a decrease in leg stiffness, which only occurred after “Targeted” training. One reason could be that “Targeted” training drove the optimization of cost functions other than mechanical work (Miller and Hamill, 2015), especially functions that would benefit from a reduction in leg stiffness. One cost function that could be optimized may be impact force minimization (Miller and Hamill, 2015). This was supported by a post-hoc exploratory simple linear regression between leg stiffness and impact peak (see Table 7-13 and Figure 7-10 in supplementary). The percentage of variance explained by leg stiffness on impact force was highest after “Targeted” training. Given that leg stiffness after “General” and “Targeted” training was 31.344 and 30.194 units, this indicates a greater reduction in impact force by 12 N after “Targeted”, compared to “General” training.

Training which places high impact force on an individual may drive mechanical adaptations which benefit impact force minimization (Hunter and Marshall, 2002; Taube et al., 2012). Individuals in the “Targeted” group performed jump and hop training with loads. This could have resulted in a “learnt” kinematic response which reduces landing impact force (Hunter and Marshall, 2002; Taube et al., 2012). In contrast, participants in the “General” group did not practice coping with landing impact force with load carriage, potentially explaining why running leg stiffness was not altered in the “General” group.

To optimize loaded running mechanical work, the current “Targeted” training program needs to be modified, and supplemented by other training modalities. First, exercise selection may need to be tailored to an individual’s baseline muscular status (Cormie et al., 2010b; Jimenez-Reyes et al., 2016). Novice strength trainees may benefit from an initial period of heavy-resistance training to allow more rapid muscular force gains. Second, “Targeted” plyometric exercises which mimic the available contact time of running may induce a more favourable adaptation to loaded running leg stiffness. By shortening the contact time available in exercises, eccentric and concentric force generation demands would more closely match that of loaded running. Third, exercises in the “Targeted” group may need to be supplemented by gait retraining. An increase in step rate by as little as 5 %, reduced total joint work in BW running by 5.6 J/kg (Heiderscheidt et al., 2011), likely by reducing work done on vertical COM excursion.

One limitation of this study was that the 6-week training duration may have been too short for adequate muscular gains and for functional improvements to manifest. However, previous studies have reported improvements to BW running performance after six weeks of resistance training (Denadai et al., 2016; Spurr et al., 2003; Turner et al., 2003). It is possible that for this cohort, “Targeted” training may require a longer training period to achieve a comparable muscular gain with “General” training. It is also possible that muscular and functional gains may change at different rates, the latter requiring a greater training period. The lack of assessment of oxygen consumption in loaded running precludes making conclusions of either training programs on changes to loaded running mechanical efficiency. However, the increased energy cost associated with load carriage has been reported to be more related to an increase in mechanical work, rather than a deterioration in mechanical efficiency (Huang and Kuo, 2014).

In conclusion, six weeks of load carriage “Targeted” or “General” resistance training did not influence running mechanical work and its distribution. However, both training regimes were effective in improving lower limb concentric strength and power. Gains in lower limb concentric strength may be transferred into task specific mechanical work reductions when running leg stiffness increases, such that a greater proportion of muscle work is used for “effective” work. Future studies should investigate the modification and supplementation of currently adopted resistance training program, with loaded running gait retraining, on the effects of loaded running mechanical work and its distribution.

7.5 References

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7.6 Figures

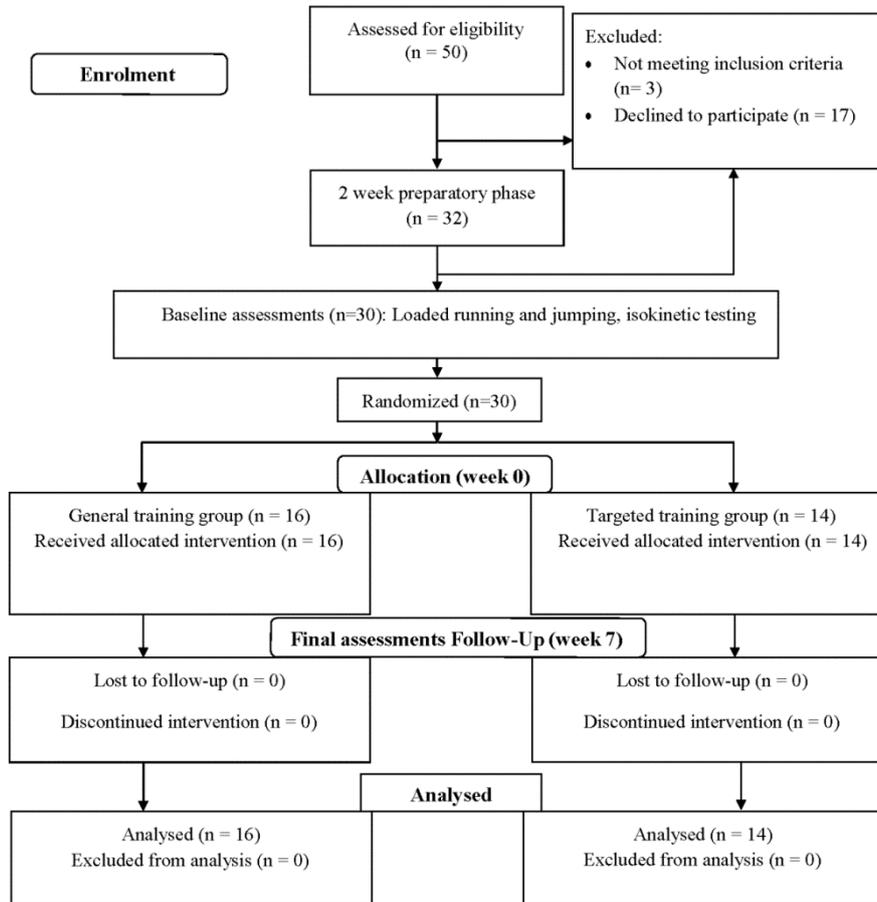


Figure 7-1 CONSORT flow diagram

Figure 2.

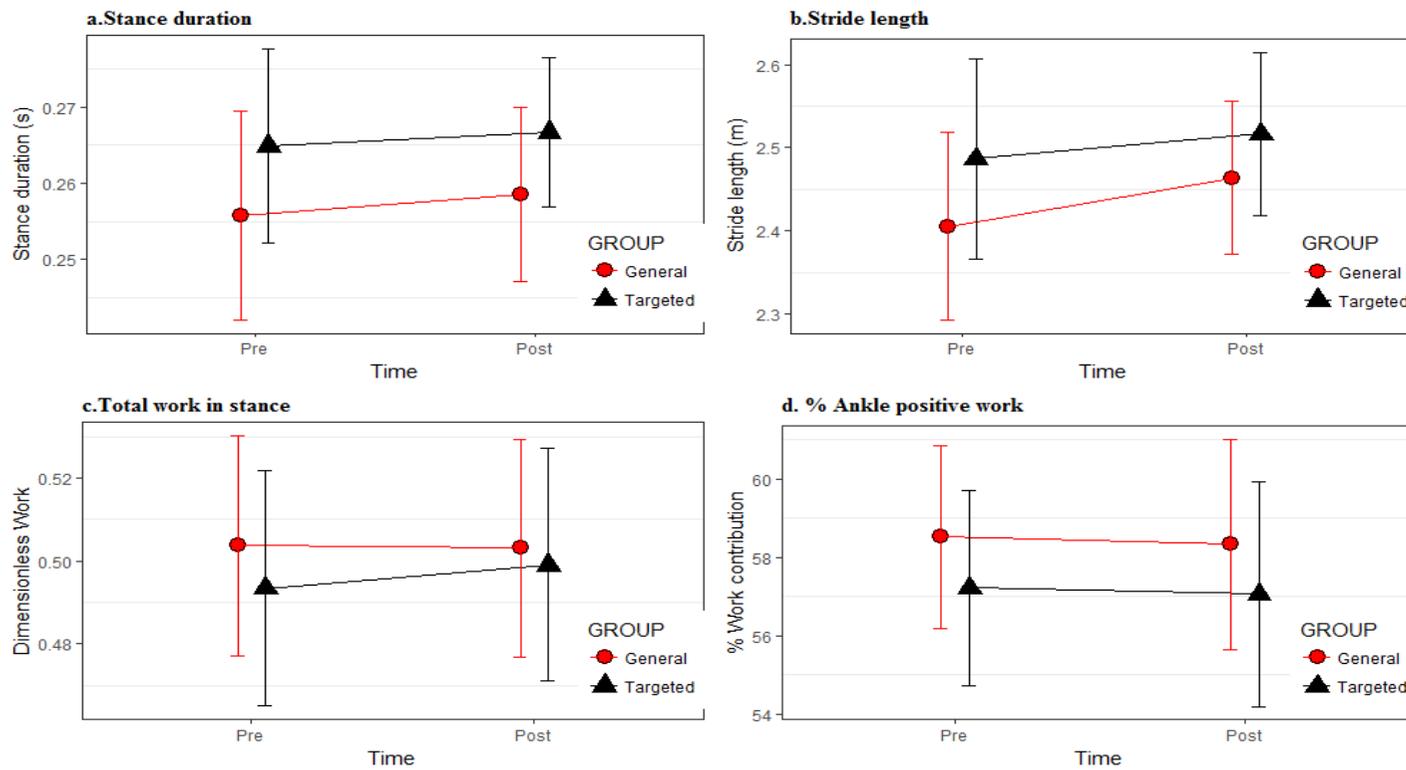


Figure 7-2 Mean (95 % confidence interval as error bars) of spatio-temporal and mechanical work during loaded running.

Figure 3.

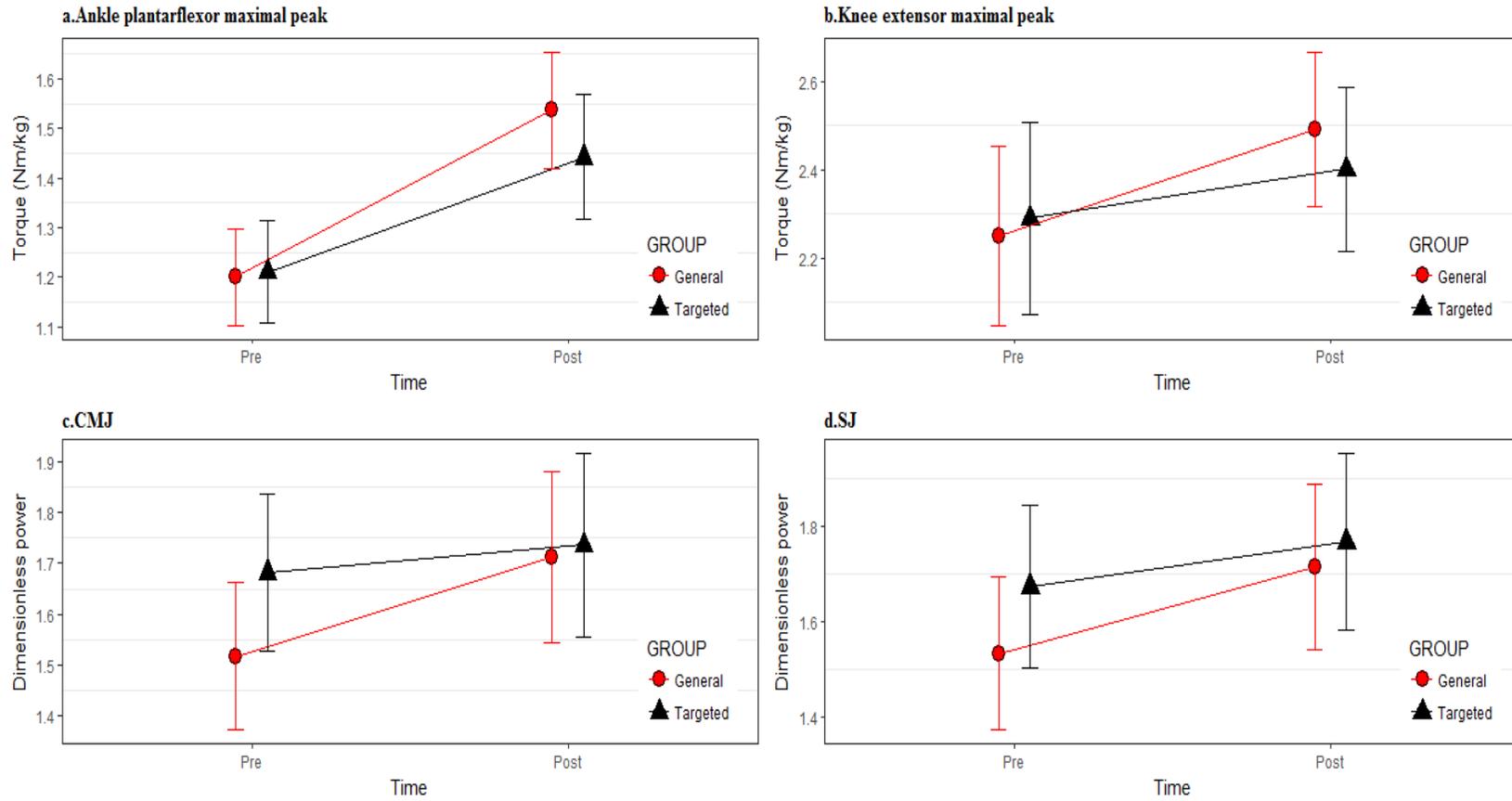


Figure 7-3 Mean (95 % confidence interval as error bars) of isokinetic strength and loaded jumping power. Countermovement jump (CMJ) and squat jump (SJ).

Figure 4.

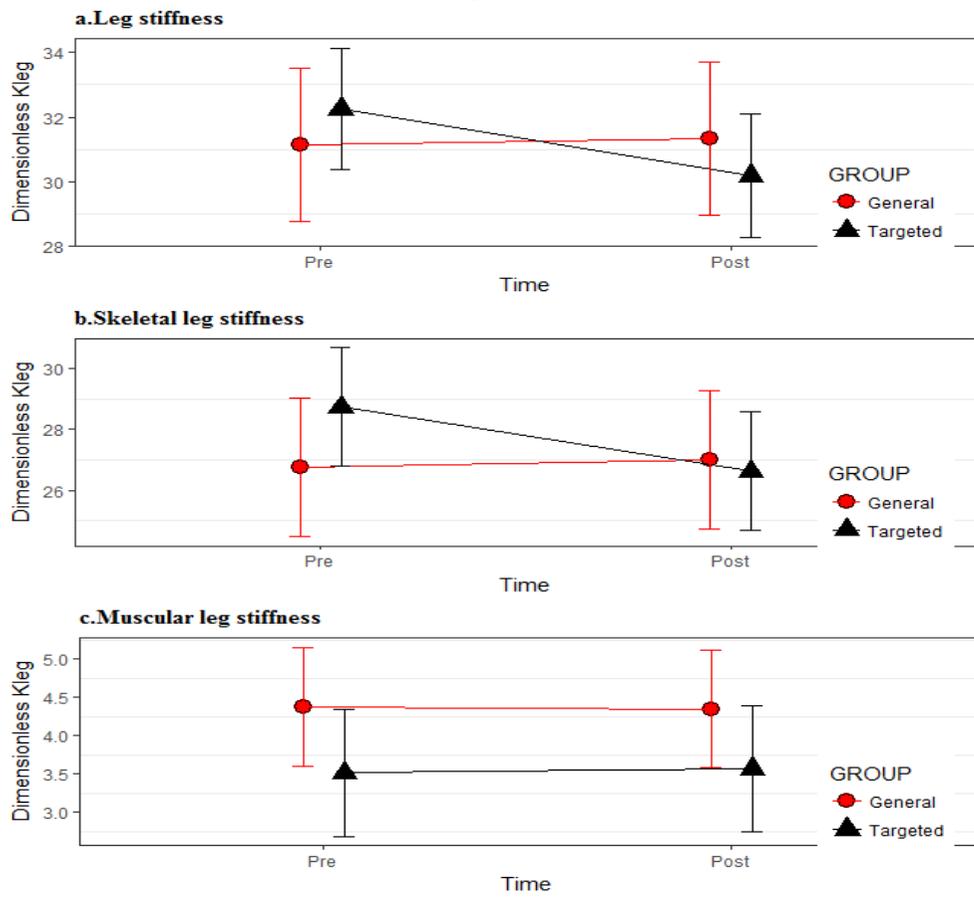


Figure 7-4 Mean (95 % confidence interval as error bars) of loaded running leg stiffness.

7.7 Tables

Table 7-1 Participant's baseline characteristics.

	“General” (n = 16)	“Targeted” (n = 14)	p value
Age (years)	30.8 (8.8)	29.4 (9.9)	0.691
Body mass (kg)	67.8 (14.0)	69.2 (10.3)	0.750
Height (cm)	172.4 (8.8)	171.7 (6.4)	0.811
Sex (Male:Female)	8:8	7:7	1.00
Resistance training frequency over past 6 weeks (times/week)	1.0 (0.8)	1.7 (1.6)	0.146
Resistance training frequency over past 12 months (times/week)	1.2 (1.2)	3.0 (3.3)	0.067
Resistance training experience (B:I:A)#	12:4:0	6:5:3	0.085
Running experience (years)	9.6 (5.9)	9.3 (6.3)	0.903
Running frequency over past 6 weeks (times/week)	2.4 (1.3)	3.0 (1.5)	0.235
Running frequency over past 12 months (times/week)	2.1 (1.4)	3.4 (3.4)	0.007
Running distance over past 6 weeks (km/week)	12.0 (13.7)	21.8 (20.2)	0.138
Running distance over past 12 months (km/week)	12.4 (18.6)	18.6 (13.3)	0.301
Foot contact pattern baseline (RFS:MFS:FFS)	9:5:2	13:1:0	0.064
Foot contact pattern post training (RFS:MFS:FFS)	10:3:3	13:1:0	0.140
Load running experience* (Y:N)	13:3	12:2	1.00

Attendance (maximum of 18)	16.7 (1.8)	16.9 (1.8)	0.801
<p>Abbreviations: DF – degree of freedom; B – beginner; I – intermediate; A- advanced; Y – Yes; N – No; RFS – rear foot strike; MFS – mid-foot strike; FFS – forefoot strike</p> <p>#Beginner = Currently not resistance training or started but ≤ 2 months with a frequency of ≤ 1 session/week; Intermediate = Currently doing resistance training and started within the last 2 to 6 months with a frequency of 2 to 3 sessions/week; Advanced = Currently doing resistance training with ≥ 1 years' experience with a frequency of ≥ 4 sessions/week</p> <p>*Experience (Y) = running for a minimum of 30 minutes at least once per month with a load of ≥ 10 % of your weight, for at least 6 times per year</p>			

Table 7-2 Relationship between Δ (post-pre) mediating factors and Δ (post-pre) total joint work

Total joint work change (dimensionless unit)	Intercept	Group (Target = 1, General = 0)	“Mediator”	Group by “Mediator” interaction (Target = 1, General = 0)
Mediator	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Ankle torque (Nm/kg)	-1.13×10^{-02} (-5.02×10^{-02} to 2.76×10^{-02})	3.08×10^{-03} (-5.52×10^{-02} to 6.13×10^{-02})	3.18×10^{-02} (-6.87×10^{-02} to 1.32×10^{-01})	2.81×10^{-02} (-1.65×10^{-01} to 2.21×10^{-01})
t value (P value)	-0.597 (0.556)	0.109 (0.914)	0.651 (0.521)	0.299 (0.767)
Knee torque ^a (Nm/kg)	-1.23×10^{-02} (-3.83×10^{-02} to 1.37×10^{-02})	2.55×10^{-02} (-7.35×10^{-03} to 5.83×10^{-02})	4.86×10^{-02} (-3.06×10^{-02} to 1.28×10^{-01})	$-1.16 \times 10^{-01**}$ (-2.18×10^{-01} to -1.46×10^{-02})
t value (P value)	-0.976 (0.338)	1.595 (0.123)	1.261 (0.219)	-2.350 (0.027)
CMJ power (dimensionless unit)	-1.45×10^{-03} (-2.83×10^{-02} to 2.54×10^{-02})	2.97×10^{-03} (-3.10×10^{-02} to 3.69×10^{-02})	4.47×10^{-03} (-9.38×10^{-02} to 1.03×10^{-01})	7.31×10^{-02} (-7.00×10^{-02} to 2.16×10^{-01})
t value (P value)	-0.112 (0.912)	0.180 (0.858)	0.093 (0.826)	1.050 (0.303)
SJ power (dimensionless unit)	-2.22×10^{-03} (-2.82×10^{-02} to 2.37×10^{-02})	2.95×10^{-03} (-3.12×10^{-02} to 3.71×10^{-02})	9.01×10^{-03} (-8.80×10^{-02} to 1.06×10^{-01})	4.32×10^{-02} (-9.03×10^{-02} to 1.77×10^{-01})
t value (P value)	-0.176 (0.862)	0.178 (0.860)	0.191 (0.850)	0.665 (0.512)

Running skeletal leg stiffness (dimensionless unit)	3.77×10^{-4} (-1.89×10^{-02} to 1.97×10^{-02})	6.15×10^{-03} (-2.51×10^{-02} to 3.74×10^{-02})	-4.14×10^{-03} (-1.27×10^{-02} to 4.45×10^{-03})	4.54×10^{-03} (-6.02×10^{-03} to 1.51×10^{-02})
t value (P value)	0.040 (0.968)	0.405 (0.689)	-0.991 (0.331)	0.883 (0.385)
Running muscular stiffness ^b (dimensionless unit)	-3.82×10^{-04} (-1.69×10^{-02} to 1.62×10^{-02})	3.08×10^{-03} (-2.12×10^{-02} to 2.74×10^{-02})	7.23×10^{-03} (-1.21×10^{-02} to 2.66×10^{-02})	2.79×10^{-02} * (-2.14×10^{-03} to 5.79×10^{-02})
t value (P value)	-0.047 (0.963)	0.261 (0.796)	0.768 (0.450)	1.909 (0.067)

Abbreviations: M – body mass; g – gravitational constant of 9.81 m/s^2 ; L – leg length; Kleg – leg stiffness; β = Beta coefficient; CI – confidence interval; DOF – degree of freedom; ** **significant**; * non-significant tendency

1 dimensionless work unit = 565.9 (118.3) J

1 dimensionless power unit = 1947.5 (364.8) W

1 dimensionless stiffness unit = 828.9 (113.1) N/m

Example model:

^a A 0.1 Nm/kg increase in knee torque resulted in a reduction of total work by 0.007 units (4 J) in “General” group, but increased total work by 0.006 units (3 J) in “Targeted” group.

^b A 0.1 unit (82 N/m) reduction in muscular leg stiffness resulted in a reduction of total work by 0.001 unit (0.6 J) in “General” group and reduction by 0.0008 units (0.4 J) in “Targeted” group.

7.8 Supplementary material

Table 7-3 Consensus on Exercise Reporting Template (CERT)

Description	Details
Detailed description of the type of exercise equipment	<p><u>“General” group</u></p> <p>Isoinertial bilateral leg press (Cybex® Plate Loaded Squat Press, Cybex International, Inc.)</p> <p>Unilateral calf raises (Cybex® Plate Loaded Seated Calf, Cybex International, Inc.).</p> <p>Lunge (Cybex® Plate Loaded Smith Press, Cybex International, Inc.).</p> <p><u>“Targeted” group</u></p> <p>Hip flexor pull (Cybex® Bravo Pull, Cybex International, Inc.).</p> <p>Countermovement jump and hopping – Weighted vest (20 kg power vest, IronEdge, © Iron365 2016)</p>
Detailed description of the qualifications, expertise and/or training	<p>Training was implemented by two primary trainers, and four secondary trainers. The primary trainers Mr. Nour Faiz Aqil Yacob and Mr. Jason Hu both hold a Bachelor of Science (Exercise, Sports and Rehabilitation Science) (BSc (EXSPRHB)) degree from Curtin University. They are both accredited exercise scientist with ESSA (Exercise and Sports Science Australia). The four secondary trainers were supervised by the primary trainers. Mr. Nicholas Callaghan, Ms. Hannah Watt, Ms. Giorgia Alford were final (3rd) year BSc (EXSPRHB) students. The hours spent in this study contributed to the overall hours needed in their final year clinical placement unit. Ms. Tess Moynihan was a final (4th) year BSc (Physiotherapy) student. Mr. Yacob, Mr. Hu and Ms. Moynihan were employed as casual employees of Curtin University for the period of training.</p>

Describe whether exercises are performed individually or in a group	All exercises were scheduled by the primary investigator (Bernard Liew). A training session (usually a 40 minute period) consisted of one or more individual participants performing their required work out in the gym. The number of participants simultaneously training at a given session depended on the availability of the participants to come to the university at that given time slot, the number of participants at a given session (maximum of 6 participants) and the number of trainers available. The trainer to participant ratio varies from 1:1 to 1:4.
Describe whether exercises are supervised or unsupervised; how they are delivered	All exercises were supervised by the trainers.
Detailed description of how adherence to exercise is measured and reported	Adherence was noted by a physical attendance sheet filled up by the trainers appointed for a participant training session. Since this was entirely a supervised training program, adherence can be confirmed by physical attendance.
Detailed description of motivation strategies	<p>Verbal encouragement was provided by the trainers</p> <p>Participants performed two of the three exercises with a linear position transducer (GymAware, Kinetic Performance, Canberra, ACT, Australia). In the “General” group the exercises included the leg press and lunge, whilst in the “Targeted” group the exercises included CMJ and hip flexor pull. The calf raises and hopping precluded the use of GymAware as the displacement was too small to be detected. Visual feedback of individual repetition peak power was provided by GymAware via a Bluetooth feed into a GymAware App (v2.11) installed in iPad 2 (Wi-Fi & 3G 32GB, Apple Inc.).</p>

<p>Detailed description of the decision rule(s) for determining exercise progression</p>	<p><u>“General” group (Table)</u></p> <p>Decision rule for progression was fixed. Load intensity progressed from 80 % of estimated 1RM (equivalent to a 10RM load) in the first two weeks, to 84 % of estimated 1RM in the next two weeks (equivalent to an 8RM load), to 88 % of estimated 1RM (equivalent to a 6RM load) in the final two weeks. The estimated 1RM load for each exercise was proposed to be adjusted by a weekly increase of approximately 2.5 % to account for a weekly progression in strength. If the trainer observed that the training load was still significantly difficult for the first set of the first session of a new training week, the estimated 1RM load was not increased.</p> <p><u>“Targeted group” (Table)</u></p> <p>Decision rule for progression was fixed. Load intensity for the hip flexor pull followed the principles of “General” group, since it was isoinertial training. For the single leg (SL) hopping and CMJ, external load increments of 5 % body weight (BW) was made weekly, till an upper limit of 20 % BW was reached. For the SL hopping from week 5 to 6, an increase in hopping frequency was made from 2.2 Hz to 3 Hz.</p>
<p>Detailed description of how the exercise program was progressed</p>	<p>Details as above.</p>
<p>Detailed description of each exercise to enable replication</p>	<p>See supplementary material of study protocol (6.8).</p>
<p>Detailed description of any home programme component</p>	<p>There was no home programme component.</p>
<p>Describe whether there are any non-exercise components</p>	<p>There was no non-exercise component.</p>
<p>Describe the type and number of adverse events that occur during exercise</p>	<p>The only side-effect was the occurrence of delayed onset of muscle soreness that occurred after the training sessions, largely within the first two weeks. All symptoms resolved within 48 hours after the training session. There were no adverse events that occurred during the exercise and after the training program.</p>

Describe the setting in which the exercises are performed	Strength and conditioning laboratory within the School of Physiotherapy and Exercise Science, Curtin University, Perth, Australia.
Detailed description of the exercise intervention	See supplementary material of study protocol (6.8).
Describe whether the exercises are generic (one size fits all) or tailored	Depending on the group allocation, all participants performed the same type of exercises. Only the training intensity was individualized to the starting 1RM of the participants.
Detailed description of how exercises are tailored to the individual	As above.
Describe the decision rule for determining the starting level	<p>For all isoinertial exercises (leg press, calf raises, lunges, hip flexor pull), the starting 1RM load was derived from a 10 RM load using a regression table for novice strength trainers. The 10 RM load to be used for the first week of training was determined from four familiarization sessions over a two-week period. For the first session of the familiarization period, a 15 RM load was estimated for each exercise. For the leg press, the estimated load was calculated to be 1 to 1.3 times body mass (kg). For the lunge, the estimated load was calculated to be 0.1 times body mass (kg). For the hip flexor pull and calf raises, clinical experience of the trainers was used to judge the starting 15 RM load. Since only sub-maximal repetitions were used (10 repetitions), at the end of the antecedent set, participants were inquired on how many more repetitions they felt they could accomplish. If ≤ 5 more repetitions were possible, the weight was kept constant. If participants felt 6-7 more repetitions were possible, the weight was increased by ≈ 2.5 kg. If 8-9 more were possible, and increment of ≈ 5 kg was made. A reduction of weight of ≈ 2.5 to 5 kg was made if 10 repetitions could not be accomplished.</p> <p>For the SL hopping and CMJ, the starting external load was fixed to be 10 % BW and 0 % BW respectively.</p>
Describe how adherence or fidelity is assessed/measured	Adherence and fidelity to the prescribed load intensity, repetitions, and sets for every training session was written in a weekly exercise sheet by the trainers.

Describe the extent to which the intervention was delivered as planned	Individual attendance (see Table 7-4, Table 7-5, Table 7-6) and individual loads (see attaching Excel documents) can be seen.
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Table 7-4 Individual weekly attendance (maximum 3 sessions/week)

SUBJ	GROUP	Wk1	Wk2	Wk3	Wk4	Wk5	Wk6	Total
1	G	3	3	3	3	3	3	18
2	G	2	3	3	3	3	3	17
4	G	3	2	3	3	2	3	16
5	T	3	3	3	3	3	3	18
6	T	3	3	3	3	3	3	18
7	G	2	3	2	3	3	2	15
8	G	2	3	2	3	2	2	14
9	T	2	3	3	3	2	3	16
10	T	3	3	3	3	3	3	18
11	T	2	2	2	2	2	2	12
12	G	3	2	3	3	3	3	17
14	T	3	3	2	3	2	3	16
15	G	3	3	3	3	3	3	18
16	T	3	3	3	3	3	3	18
17	G	3	3	3	3	3	3	18
18	T	3	3	3	2	3	3	17
19	T	3	3	3	3	3	3	18
20	G	2	2	2	2	2	2	12
21	G	3	3	3	3	3	3	18
22	T	3	3	3	3	3	3	18
23	T	3	3	2	3	3	3	17
24	G	3	3	3	3	3	3	18
25	G	3	3	2	3	3	3	17
26	G	3	3	3	3	3	3	18
27	T	3	3	3	3	3	3	18
28	T	3	3	3	3	3	3	18
29	T	2	2	3	2	3	2	14
30	G	3	3	3	3	3	3	18
31	G	3	3	3	3	3	3	18
32	G	2	3	3	2	3	2	15

SUBJ = subject; Wk = week; G = "General" group; T = "Targeted" group

Table 7-5 "General" group "group average" weekly load

Weeks	Exercise	Total repetitions	Mass (kg)	Relative mass (% body mass)
1	Calf raises	1696	16.02	0.23
1	Leg press	848	160.63	2.39
1	Lunge	1696	43.44	0.65
2	Calf raises	1712	16.02	0.23
2	Leg press	984	161.25	2.40
2	Lunge	1712	43.44	0.65
3	Calf raises	1248	17.64	0.26
3	Leg press	912	176.79	2.64
3	Lunge	1200	46.64	0.70
4	Calf raises	1656	18.05	0.26
4	Leg press	912	199.84	2.99
4	Lunge	1296	47.20	0.71
5	Calf raises	976	19.57	0.28
5	Leg press	604	210.39	3.14
5	Lunge	976	49.81	0.75
6	Calf raises	1088	19.59	0.28
6	Leg press	818	207.63	3.11
6	Lunge	1088	50.52	0.76

Table 7-6 “Targeted” group "group average" weekly load

Weeks	Exercise	Total repetitions	Mass (kg)	Relative mass (% body mass)
1	Countermovement jump	815	70.60	1.00
1	Hip flexor pull	1648	14.97	0.21
1	Single leg hopping	7744	77.96	1.10
2	Countermovement jump	750	74.26	1.05
2	Hip flexor pull	1660	14.97	0.21
2	Single leg hopping	8360	77.96	1.10
3	Countermovement jump	815	77.94	1.10
3	Hip flexor pull	1708	17.23	0.25
3	Single leg hopping	10384	81.34	1.15
4	Countermovement jump	783	81.52	1.16
4	Hip flexor pull	1960	17.53	0.25
4	Single leg hopping	9768	85.18	1.21
5	Countermovement jump	793	85.04	1.20
5	Hip flexor pull	1584	19.35	0.28
5	Single leg hopping	13440	85.19	1.21
6	Countermovement jump	1020	85.19	1.21
6	Hip flexor pull	2050	19.35	0.28
6	Single leg hopping	14160	85.19	1.21

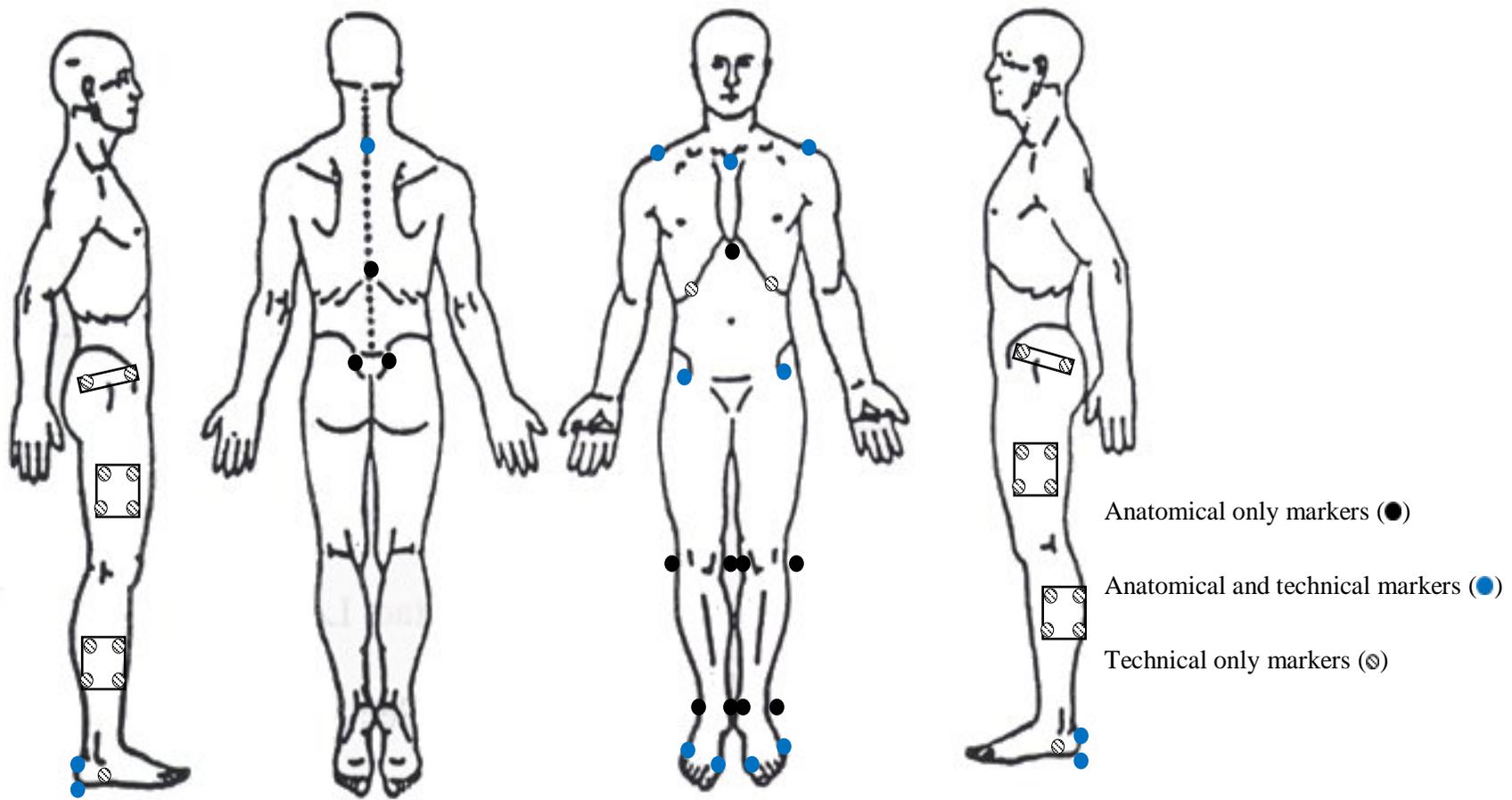


Figure 7-5 Retro-reflective markers positions

Biomechanical modelling procedures: The origin of the pelvic segment coordinate system (CS) was defined as the mid-point between the mid-point of the two PSIS markers and the mid-point of the two ASIS markers. The length of the pelvis was set as the length between the origin and the mid-point between the two (right and left) hip joint centres. The (x-y) plane of the pelvic segment coordinate system was defined as the plane passing through the right and left ASIS markers, and the mid-point of the right and left PSIS markers. The x-axis (medial-lateral) was defined from the pelvic origin towards the right ASIS marker, with the positive direction to the right. The z-axis was orthogonal to this (x-y) plane, with the positive direction pointing superiorly. The y-axis was then calculated to be the cross product of the x-axis and z-axis, with the positive direction pointing anteriorly.

For the thorax segment coordinate system, several virtual landmarks were created. First, a landmark (MID_IJC7) was created as the mid-point between the clavicle marker and the C7 marker. Second, a landmark (MID_PXTV8) was created as the mid-point between the sternum marker and the T10 marker. Third, a landmark (THORAX_X) was created by projecting a landmark from the plane created by C7- MID_PXTV8- clavicle marker, by 0.1 m in the x-axis (medial-lateral axis towards the right), a distance of 50 % of the length between C7 and clavicle marker in the y-axis (anteroposterior axis), and a distance of zero in the z-axis (vertical). The origin of the thorax coordinate system was at the origin of the pelvic coordinate system. The z-axis (distal-proximal) of the thorax was oriented as the line joining the origin to MID_IJC7. The frontal plane was formed by the thorax origin, MID_IJC7, and THORAX_X. The y-axis (posterior-anterior) was orthogonal to the frontal plane with its positive direction anterior. The x-axis (medial-lateral) was defined as the cross-product of the z- and y-axes with its positive direction pointing to the right.

The z-axis (distal-proximal) of the thigh CS in both models was oriented as the line joining the knee and hip centre with its positive direction pointing superiorly. The y-axis (posterior-anterior) was orthogonal to the frontal plane with its positive direction anterior. The frontal plane was formed by the lateral femoral condyle, hip and knee joint centre. The x-axis (medial-lateral) was then defined as the cross-product of the z- and y-axes with its positive direction pointing to the right.

The z-axis (distal-proximal) of the shank CS was the line passing from the ankle joint to the knee joint centre. The y-axis (posterior-anterior) was orthogonal to the plane defined by the lateral malleoli, ankle and knee joint centre with its positive direction anterior. The x-axis (medial-lateral) was the cross-product of the z- and y-axes with its positive direction pointing to the right.

In the kinetic foot CS, the z-axis was defined as the line joining the mid-point between the medial and lateral malleoli and the mid-point between the 1st and 5th MTP. The y-axis was orthogonal to the plane defined by the two malleoli and two foot markers (1st and 5th MTP). The x-axis was the cross-product of the z- and y-axes with its positive direction pointing to the right. The kinetic foot was used for kinetic calculations.

A virtual foot segment (vFoot1) was created and used for kinematic calculations, which enabled the use of the static calibration trial as a reference posture. The z-axis was defined as the line joining the mid-point between the superior and inferior calcaneal markers (MID_CALA), and the mid-point between the 1st and 5th MTP (MID_MT). A sagittal plane was formed by MID_CALA, MID_MT, and the ankle joint centre (mid-point between lateral and medial malleoli). The y-axis was orthogonal to the defined sagittal plane. Lastly, the x-axis was defined as the cross-product between the z-axis and y-axis. The virtual foot CS (vFoot1) was then rotated so that the x-axis was pointing to the right, y-axis was pointing anteriorly, and z-axis was pointing superiorly.

A second virtual foot segment (vFoot2) was created to enable accurate centre of pressure calculations. First, a virtual laboratory was created with the z-axis directed superiorly, y-axis directed anteriorly, and x-axis directed lateral. MID_CALA and MID_MT were projected onto the virtual laboratory. vFoot2 was created with the z-axis defined as the projected MID_CALA and MID_MT. A sagittal plane was formed between the projected MID_CALA, projected MID_MT, and the ankle joint centre (mid-point between lateral and medial malleoli). The y-axis was orthogonal to the sagittal plane. Lastly, the x-axis was defined as the cross-product between the z-axis and y-axis. The virtual foot CS (vFoot2) was then rotated so that the x-axis was pointing to the right, y-axis was pointing anteriorly, and z-axis was pointing superiorly.

Leg stiffness calculation method: First, a 3D virtual leg was created using the hip joint centre and the centre of pressure (COP) of the stance limb. Second, 3D GRF was projected onto the leg vector by taking the inner dot product of the GRF vector with the unit vector of the leg. Lastly, leg stiffness was calculated using the ratio of the peak projected GRF with the change in leg length (leg length at initial contact minus leg length at time of peak projected GRF).

Leg stiffness was deconstructed into muscular and skeletal components (equations 1 and 2), using a previously published method [1]:

$$K_{legm} = K_{leg} \sin \theta_{peak} \text{ (equation 1)}$$

$$K_{legs} = K_{leg} \cos \theta_{peak} \text{ (equation 2)}$$

where K_{leg} represents leg stiffness, K_{legm} = muscular leg stiffness, K_{legs} = skeletal leg stiffness, θ_{peak} represents the angle between two vectors at the time of peak projected GRF. Vector one represents the leg (COP to HJC). Vector two represents the shank (COP to knee joint centre).

[1] Wang TM, Huang HP, Li JD, Hong SW, Lo WC, Lu TW. Leg and Joint Stiffness in Children with Spastic Diplegic Cerebral Palsy during Level Walking. PLoS ONE 2015;10:e0143967.

Table 7-7 Influence of Time, Group, Time-by-Group interactions on spatio-temporal variables

	Intercept (General, Pre)	Time (Post = 1, Pre = 0)	Group (Target = 1, General = 0)	Time x Group (TimePost:GroupTarget = 1)
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Stance duration (s)	2.56×10^{-1} (2.42×10^{-1} to 2.70×10^{-1})	2.73×10^{-3} (-2.58×10^{-3} to 8.05×10^{-3})	9.14×10^{-3} (-1.03×10^{-2} to 2.85×10^{-2})	-9.25×10^{-4} (-8.70×10^{-3} to 6.85×10^{-3})
t-value	36.7758724	1.0226594	9.64×10^{-1}	-2.36×10^{-1}
P value	0.000	0.309	0.343	0.814
Stride length (m)	2.41 (2.29 to 2.52)	5.84×10^{-2} (-4.32×10^{-2} to 1.60×10^{-1})	8.15×10^{-2} (-8.71×10^{-2} to 2.50×10^{-1})	-2.88×10^{-2} (-1.78×10^{-1} to 1.20×10^{-1})
t-value	42.77	1.18	9.90×10^{-1}	-3.97×10^{-1}
P value	0.000	0.249	0.331	0.695

Abbreviations: s – seconds; m – metres; β = Beta coefficient; CI – confidence interval; DOF – degrees of freedom; ** **significant**; * non-significant tendency

Example model:

Stance duration = 0.25 + Time (0.0027) + Group (0.009) + Time:Group (-0.0009)

Baseline stance duration for Target group = 0.25 + (0) (0.0027) + (1) (0.009) + (0) (-0.0009) = 0.259 s

Post intervention stance duration for Target group = 0.25 + (1) (0.0027) + (1) (0.009) + (1) (-0.0009) = 0.2608 s

Table 7-8 Influence of Time, Group, Time-by-Group interactions on kinetic variables

	Intercept (General, Pre)	Time (Post = 1, Pre = 0)	Group (Target = 1, General = 0)	Time x Group (TimePost:GroupTarget = 1)
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Total joint work (dimensionless unit)	5.04×10^{-1} (4.77×10^{-1} to 5.30×10^{-1})	-5.84×10^{-4} (-1.72×10^{-2} to 1.60×10^{-2})	-1.03×10^{-2} (-5.05×10^{-2} to 2.99×10^{-2})	6.24×10^{-3} (-1.81×10^{-2} to 3.05×10^{-2})
t-value	37.54	-6.99×10^{-2}	-5.24×10^{-1}	5.10×10^{-1}
P value	0.000	0.944	0.604	0.611
% Ankle positive	58.53 (56.18 to 60.88)	-1.90×10^{-1} (-2.02 to 1.64)	-1.30 (-4.85 to 2.24)	3.47×10^{-2} (-2.65 to 2.72)
t-value	49.52	-2.06×10^{-1}	-7.53×10^{-1}	2.57×10^{-2}
P value	0.000	0.837	0.458	0.980

Abbreviations: M – body mass; g – gravitational constant of 9.81 m/s^2 ; L – leg length; Kleg – leg stiffness; β = Beta coefficient; CI – confidence interval; DOF – degrees of freedom; ** **significant**; * non-significant tendency

1 dimensionless work unit = 565.9 (118.3) J

Example model:

Total joint work = $0.5 + \text{Time} (-0.0005) + \text{Group} (-0.01) + \text{Time:Group} (0.006)$

Baseline Total joint work for Target group = $0.5 + (0) (-0.0005) + (1) (-0.01) + (0) (0.006) = 0.49 = 277 \text{ J}$

Post intervention Total joint work for Target group = $0.5 + (1) (-0.0005) + (1) (-0.01) + (1) (0.006) = 0.4955 = 280 \text{ J}$

Table 7-9 Influence of Time, Group, Time-by-Group interactions on individual joint work

	Intercept (General, Pre)	Time (Post = 1, Pre = 0)	Group (Target = 1, General = 0)	Time x Group (TimePost:GroupTarget = 1)
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Ankle positive (dimensionless unit)	1.50×10^{-1} (1.40×10^{-1} to 1.60×10^{-1})	9.16×10^{-4} (-5.77×10^{-3} to 7.60×10^{-3})	-7.87×10^{-3} (-2.37×10^{-2} to 7.94×10^{-3})	2.72×10^{-3} (-7.06×10^{-3} to 1.25×10^{-2})
t-value	28.45	2.72×10^{-1}	-1.02	5.53×10^{-1}
P value	0.00	0.786	0.317	0.581
Ankle negative (dimensionless unit)	-6.73×10^{-2} (-7.52×10^{-2} to -5.94×10^{-2})	9.21×10^{-4} (-4.02×10^{-3} to 5.87×10^{-3})	9.34×10^{-3} (-2.58×10^{-3} to 2.13×10^{-2})	-3.07×10^{-3} (-1.03×10^{-2} to 4.17×10^{-3})
t-value	-16.92	3.70×10^{-1}	1.60	-8.44×10^{-1}
P value	0.000	0.712	0.120	0.401
Knee positive (dimensionless unit)	9.35×10^{-2} (8.59×10^{-2} to 1.01×10^{-1})	-1.31×10^{-3} (-7.77×10^{-3} to 5.16×10^{-3})	-1.69×10^{-3} (-1.32×10^{-2} to 9.78×10^{-3})	5.23×10^{-3} (-4.24×10^{-3} to 1.47×10^{-2})
t-value	24.43	-4.01×10^{-1}	-3.03×10^{-1}	1.10
P value	0.000	0.689	0.765	0.275

Knee negative (dimensionless unit)	-1.08×10^{-1} (-1.19×10^{-1} to -9.64×10^{-2})	-2.26×10^{-3} (-8.95×10^{-3} to 4.43×10^{-3})	-8.45×10^{-3} (-2.53×10^{-2} to 8.35×10^{-3})	3.84×10^{-3} (-5.95×10^{-3} to 1.36×10^{-2})
t-value	-19.19	-6.72E-01	-1.03	7.79×10^{-1}
P value	0.000	0.503	0.312	0.438
Hip positive (dimensionless unit)	1.29×10^{-2} (8.45×10^{-3} to 1.73×10^{-2})	3.30×10^{-3} (-2.20×10^{-3} to 8.80×10^{-3})	1.44×10^{-3} (-5.23×10^{-3} to 8.12×10^{-3})	-3.50×10^{-3} (-1.16×10^{-2} to 4.55×10^{-3})
t-value	5.78	1.19	4.43×10^{-1}	-8.64×10^{-1}
P value	0.000	0.236	0.661	0.390
Hip negative (dimensionless unit)	-7.25×10^{-2} (-8.07×10^{-2} to -6.44×10^{-2})	4.84×10^{-3} (-1.52×10^{-3} to 1.12×10^{-2})	1.28×10^{-3} (-1.12×10^{-2} to 1.38×10^{-2})	-2.55×10^{-3} (-1.19×10^{-2} to 6.76×10^{-3})
t-value	-17.63	1.51	2.10×10^{-1}	-5.45×10^{-1}
P value	0.000	0.134	0.835	0.587
Abbreviations: M – body mass; g – gravitational constant of 9.81 m/s^2 ; L – leg length; β = Beta coefficient; CI – confidence interval; DOF – degrees of freedom; DOF for Intercept, Time, Group, Time x Group = 88, 88, 28, 88. ** significant ; * non-significant tendency				
1 dimensionless work unit = 565.9 (118.3) J				

Table 7-10 Influence of Time, Group, Time-by-Group interactions on percentage joint contribution

	Intercept (General, Pre)	Time (Post = 1, Pre = 0)	Group (Target = 1, General = 0)	Time x Group (TimePost:GroupTarget = 1)
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
% Ankle negative	27.45 (24.23 to 30.68)	-3.00×10^{-2} (-2.32 to 2.26)	-3.63 (-8.50 to 1.24)	1.00 (-2.35 to 4.36)
t-value	16.90	-2.61×10^{-2}	-1.53	5.95×10^{-1}
P value	0.000	0.979	0.138	0.553
% Knee positive	36.62 (33.78 to 39.45)	-9.21×10^{-1} (-2.69 to 8.50×10^{-1})	2.91×10^{-1} (-3.59 to 4.17)	1.44 (-1.15 to 4.03)
t-value	25.67	-1.03	1.54×10^{-1}	1.10
P value	0.00	0.304	0.879	0.272
% Knee negative	43.33 (40.10 to 46.56)	1.42 (-3.06×10^{-1} to 3.15)	3.95 (-9.22×10^{-1} to 8.82)	-1.90 (-4.43 to 6.32×10^{-1})
t-value	26.67	1.64	1.66	-1.49
P value	0.000	0.106	0.108	0.140

% Hip positive	4.85 (3.34 to 6.37)	1.11 (-1.24 to 3.46)	1.01 (-1.27 to 3.30)	-1.48 (-4.91 to 1.96)
t-value	6.37	9.40×10^{-1}	9.07×10^{-1}	-0.85
P value	0.000	0.219	0.383	0.264
% Hip negative	29.22 (26.36 to 32.07)	-1.39 (-3.20 to 4.16×10^{-1})	-3.15×10^{-1} (-4.62 to 4.00)	8.96×10^{-1} (-1.75 to 3.54)
t-value	20.33	-1.53	-1.50×10^{-1}	6.72×10^{-1}
P value	0.000	0.130	0.882	0.503
Abbreviations: β = Beta coefficient; CI – confidence interval; DOF – degrees of freedom; DOF for Intercept, Time, Group, Time x Group = 88, 88, 28, 88. ** significant ; * non-significant tendency				

Table 7-11 Influence of Time, Group, Time-by-Group interactions on strength and jump power

	Intercept (General, Pre)	Time (Post = 1, Pre = 0)	Group (Target = 1, General = 0)	Time x Group (TimePost:GroupTarget = 1)
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Ankle PF maximal peak torque (Nm/kg)	1.200 (1.102 to 1.298)	3.363 x 10^{-1**} (2.522 x 10⁻¹ to 4.204 x 10⁻¹)	1.072 x 10 ⁻² (-1.371 x 10 ⁻¹ to 1.585 x 10 ⁻¹)	-1.051 x 10 ⁻¹ (-2.282 x 10 ⁻¹ to 1.804 x 10 ⁻²)
t-value	24.336	7.946	0.149	-1.696
P value	0.000	0.000	0.883	0.093
Knee Ext maximal peak torque (Nm/kg)	2.251 (2.047 to 2.454)	2.419 x 10^{-1**} (1.094 x 10⁻¹ to 3.744 x 10⁻¹)	4.037 x 10 ⁻² (-2.668 x 10 ⁻¹ to 3.476 x 10 ⁻¹)	-1.315 x 10 ⁻¹ (-3.255 x 10 ⁻¹ to 6.250 x 10 ⁻²)
t-value	21.970	3.628	2.692 x 10 ⁻¹	-1.347
P value	0.000	0.001	0.790	0.181
CMJ peak joint power (dimensionless unit)	1.517 (1.369 to 1.665)	1.949 x 10^{-1**} (9.327 x 10⁻² to 2.965 x 10⁻¹)	1.655 x 10 ⁻¹ (-5.070 x 10 ⁻² to 3.817 x 10 ⁻¹)	-1.416 x 10 ^{-1*} (-2.903 x 10 ⁻¹ to 7.210 x 10 ⁻¹)

t-value	21.042	3.928	1.568	-1.949
P value	0.000	0.001	0.128	0.061
SJ peak joint power (dimensionless unit)	1.533 (1.369 to 1.696)	1.813 x 10⁻¹** (7.050 x 10⁻² to 2.920 x 10⁻¹)	1.403 x 10 ⁻¹ (-9.899 x 10 ⁻² to 3.796 x 10 ⁻¹)	-8.702 x 10 ⁻² (-2.492 x 10 ⁻¹ to 7.511 x 10 ⁻²)
t-value	19.204	3.352	1.201	-1.099
P value	0.000	0.002	0.240	0.281

Abbreviations: PF – plantarflexor; Ext – extensor; CMJ – Countermovement jump; SJ – Squat jump; β = Beta coefficient; CI – confidence interval; DOF – degrees of freedom; DOF for Intercept, Time, Group, Time x Group = 88, 88, 28, 88. **** significant**; * non-significant tendency

1 dimensionless power unit = 1947.5 (364.8) W

Example model:

Ankle PF maximal peak torque = 1.2 + Time (0.336) + Group (0.0107) + Time:Group (-0.105)

Baseline Ankle PF maximal peak torque for Target group = 1.2 + (0) (0.336) + (1) (0.0107) + (0) (-0.105) = 1.21 Nm/kg

Post intervention Ankle PF maximal peak torque for Target group = 1.2 + (1) (0.336) + (1) (0.0107) + (1) (-0.105) = 1.44 Nm/kg

Table 7-12 Influence of Time, Group, Time-by-Group interactions on running leg stiffness

	Intercept (General, Pre)	Time (Post = 1, Pre = 0)	Group (Target = 1, General = 0)	Time x Group (TimePost:GroupTarget = 1)
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Kleg (dimensionless unit))	31.14 (28.77 to 33.51)	2.04×10^{-1} (-1.29 to 1.70)	1.12 (-2.01 to 4.24)	-2.27** (-4.46 to -7.60 x 10⁻²)
t-value	26.10	2.72×10^{-1}	7.32×10^{-1}	-2.06
P value	0.000	0.787	0.470	0.043
Kleg length change (dimensionless unit)	8.90×10^{-2} (8.18×10^{-2} to 9.62×10^{-2})	6.88×10^{-4} (-1.46×10^{-3} to 2.84×10^{-3})	-4.94×10^{-3} (-1.60×10^{-2} to 6.11×10^{-3})	$5.03 \times 10^{-3**}$ (1.85×10^{-3} to 8.21×10^{-3})
t-value	24.19	6.29×10^{-1}	-9.16×10^{-1}	3.11
P value	0.000	0.530	0.367	0.002
Kleg GRF (dimensionless unit)	2.69 (2.59 to 2.78)	$3.33 \times 10^{-2**}$ (5.83×10^{-3} to 6.08×10^{-2})	-3.31×10^{-2} (-1.81×10^{-1} to 1.14×10^{-1})	-3.39×10^{-2} (-7.93×10^{-2} to 1.65×10^{-3})
t-value	54.74	2.38	-4.61×10^{-1}	-1.88
P value	0.000	0.018	0.649	0.060

Kleg muscular (dimensionless unit)	4.38 (3.60 to 5.15)	-2.80×10^{-2} (-5.32×10^{-1} to 4.76×10^{-1})	-8.63×10^{-1} (-2.03 to 3.09×10^{-1})	8.60×10^{-2} (-6.51×10^{-1} to 8.23×10^{-1})
t-value	11.20	-1.10×10^{-1}	-1.51	2.32×10^{-1}
P value	0.000	0.912	0.143	0.817
Kleg skeletal (dimensionless unit)	26.76 (24.46 to 29.06)	2.32×10^{-1} (-1.26 to 1.73)	1.98 (-1.14 to 5.10)	-2.35* (-4.76 to 5.70×10^{-2})
t-value	23.11	3.09×10^{-1}	1.30	-1.94
P value	0.000	0.758	0.204	0.056
<p>Abbreviations: M – body mass; g – gravitational constant of 9.81 m/s^2; L – leg length; Kleg – leg stiffness; β = Beta coefficient; CI – confidence interval; DOF – degrees of freedom; DOF for Intercept, Time, Group, Time x Group = 88, 88, 28, 88. ** significant; * non-significant tendency</p> <p>1dimensionless stiffness unit = 828.9 (113.1) N/m 1dimensionless GRF unit = 683.76 (114.35) N 1dimensionless leg length unit = 0.82 (0.05) m</p> <p><u>Example model:</u> Kleg = 31.14 + Time (0.204) + Group (1.12) + Time:Group (-2.27) Baseline Total joint work for Target group = 31.14 + (0) (0.204) + (1)(1.12) + (0) (-2.27) = 32 units = 26.74 kN/m Post intervention Total joint work for Target group = 31.14 + (1) (0.204) + (1) (1.12) + (1) (-2.27) = 30.19 units= 25.03 kN/m</p>				

Table 7-13 Relationship between leg stiffness and impact peak

Group - Time	Intercept (95 % confidence interval)	Leg stiffness (95 % confidence interval)	Adjusted R ²
General-Pre	1.12 (7.82 X 10 ⁻⁰¹ to 1.46)	1.67 X 10^{-02*} (5.99 X 10⁻⁰³ to 2.74 X 10⁻⁰²)	4.22 %
General-Post	8.08 X 10 ⁻⁰¹ (5.41 X 10 ⁻⁰¹ to 1.08)	2.81 X 10^{-02**} (1.95 X 10⁻⁰² to 3.66 X 10⁻⁰²)	16.34 %
Targeted-Pre	1.15 (8.20 X 10 ⁻⁰¹ to 1.48)	1.42 X 10^{-02#} (4.07 X 10⁻⁰³ to 2.43 X 10⁻⁰²)	4.29 %
Targeted-Post	7.88 X 10 ⁻⁰¹ (5.56 X 10 ⁻⁰¹ to 1.02)	2.86 X 10^{-02**} (2.10 X 10⁻⁰² to 3.63 X 10⁻⁰²)	22.00 %
<p>** P < 0.001; * P = 0.002; # P = 0.006</p> <p>1 dimensionless GRF unit = 683.8 (114.4) N 1dimensionless stiffness unit = 828.9 (113.1) N/m</p>			

Supplemental. Leg length change.

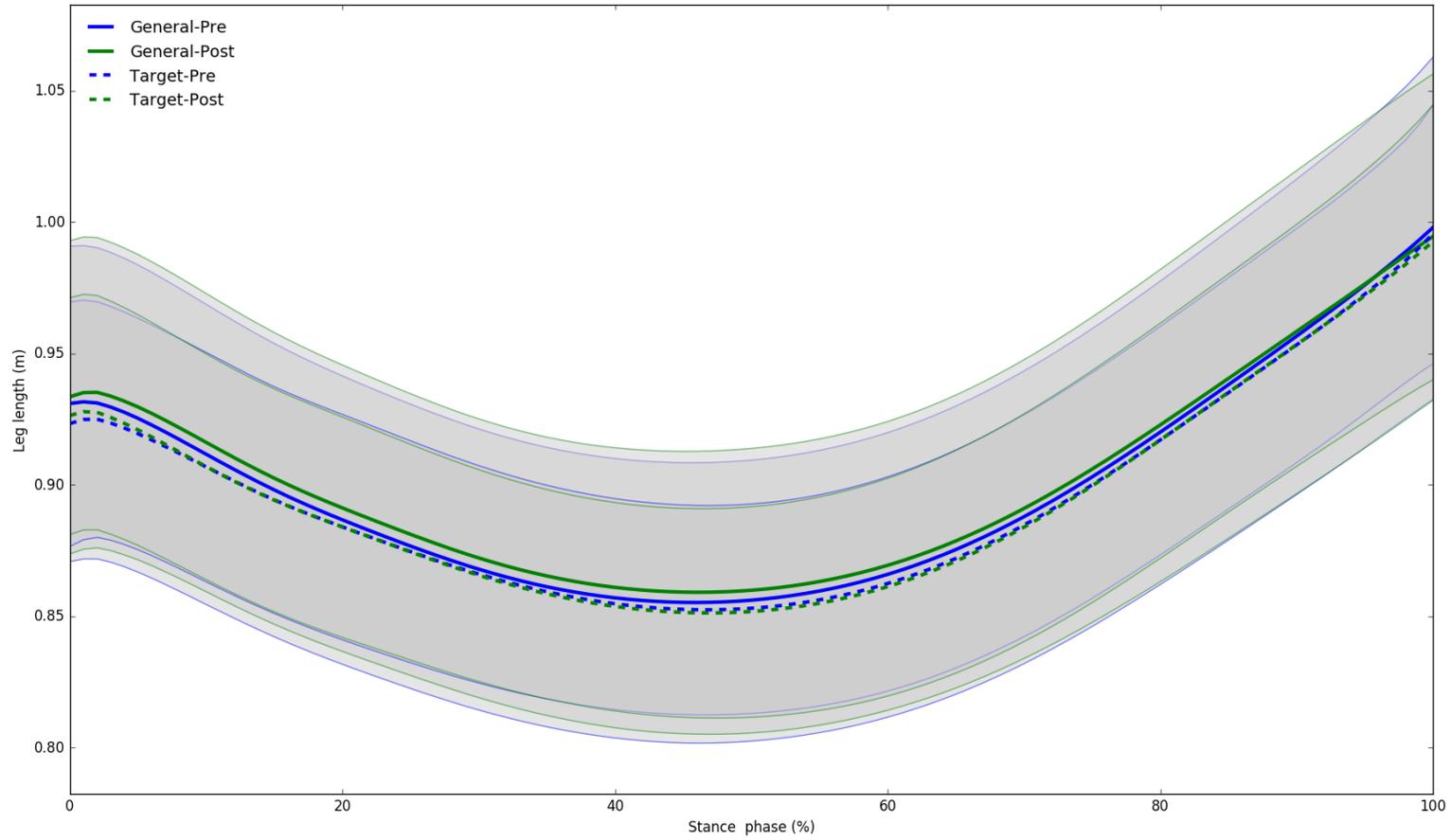


Figure 7-6 Running leg length change (m)

Supplemental. SPM leg length.

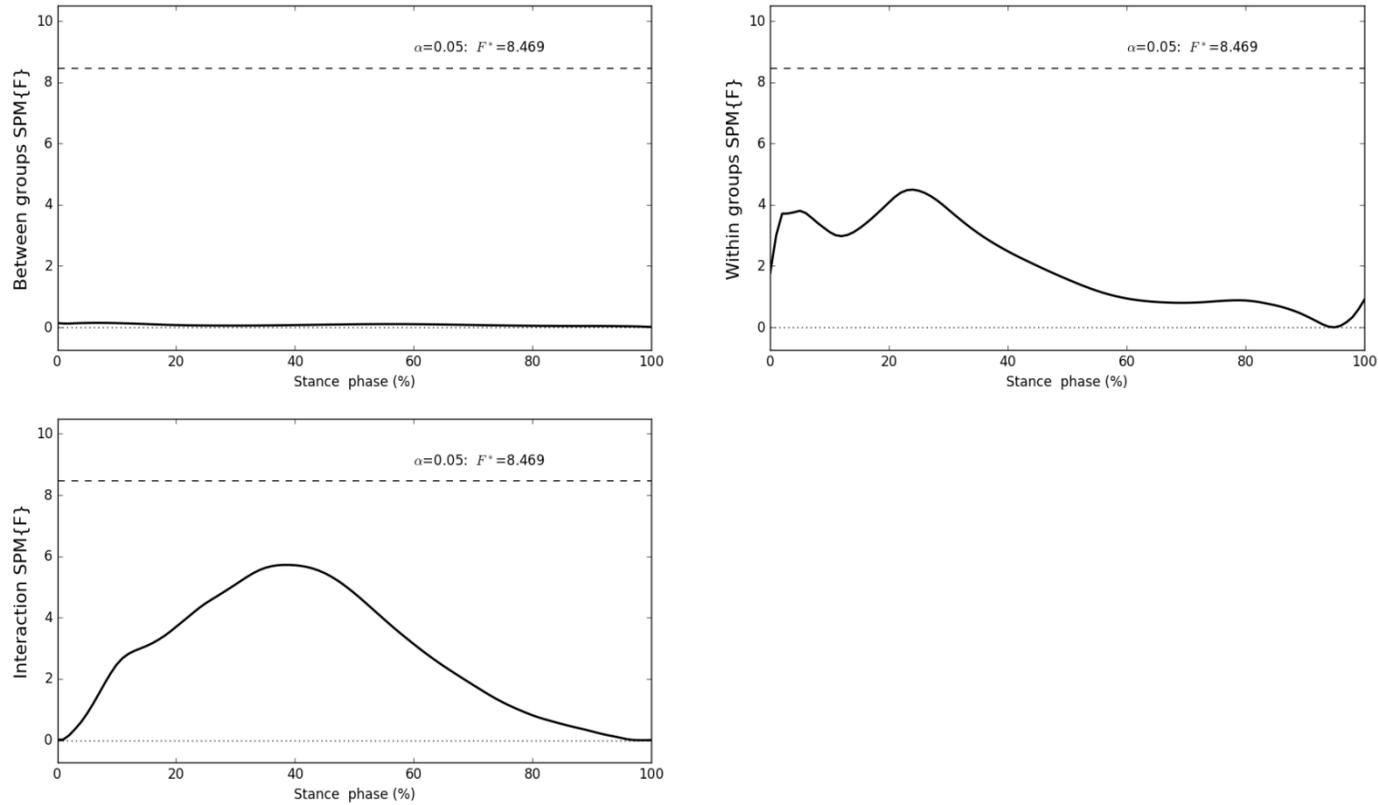


Figure 7-7 SPM results of leg length change. Instantaneous leg length only has a tendency towards significance in the “Targeted” pre-post group. This is evidence from the rising slope of the SPM interaction, and the Mean (SD). The linear mixed model results demonstrate that change in leg length in the “Targeted” group was significantly different from the “General” group.

Supplemental. Ground reaction force change.

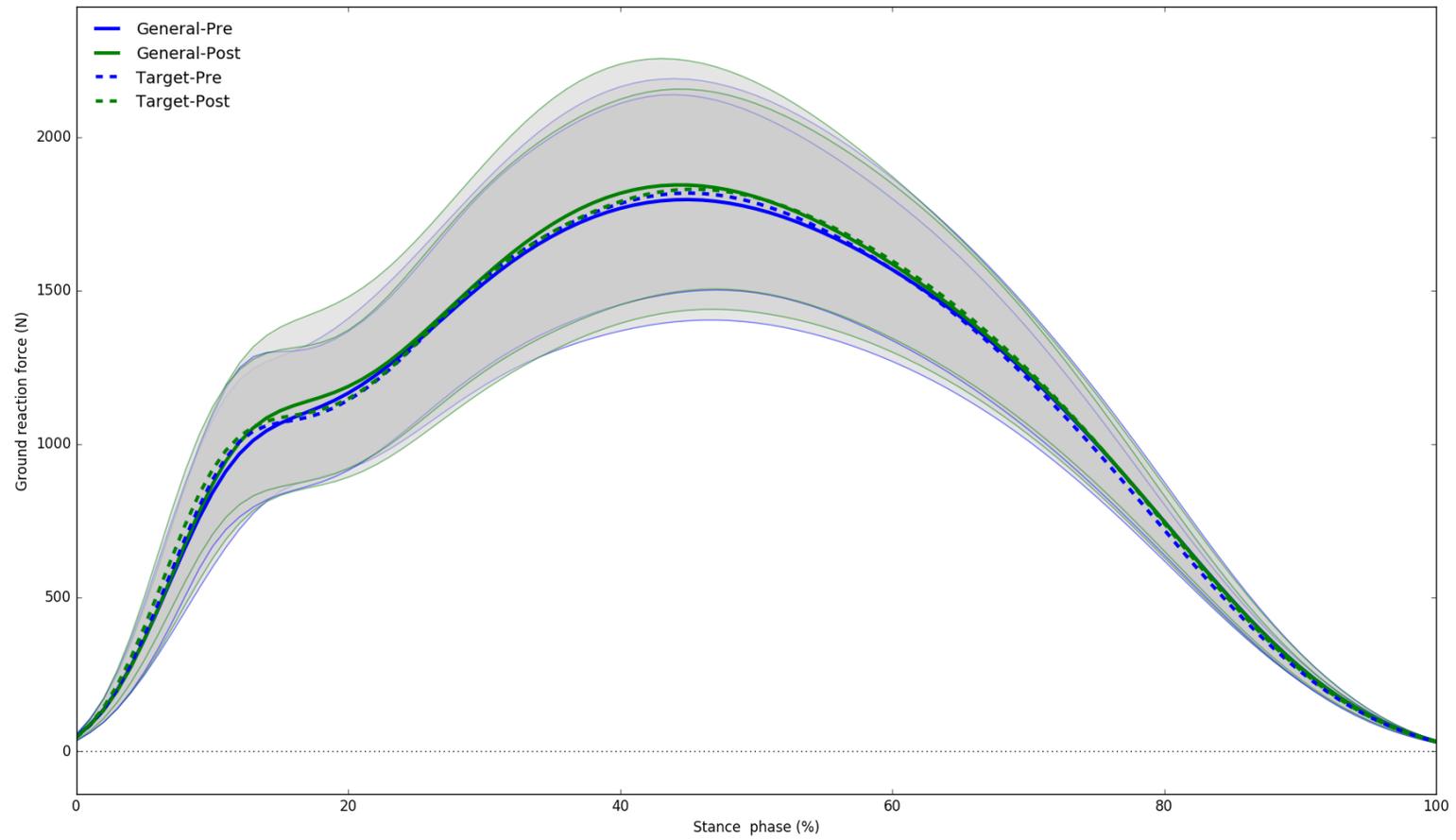


Figure 7-8 Ground reaction force in loaded running

Supplemental. SPM Ground reaction force .

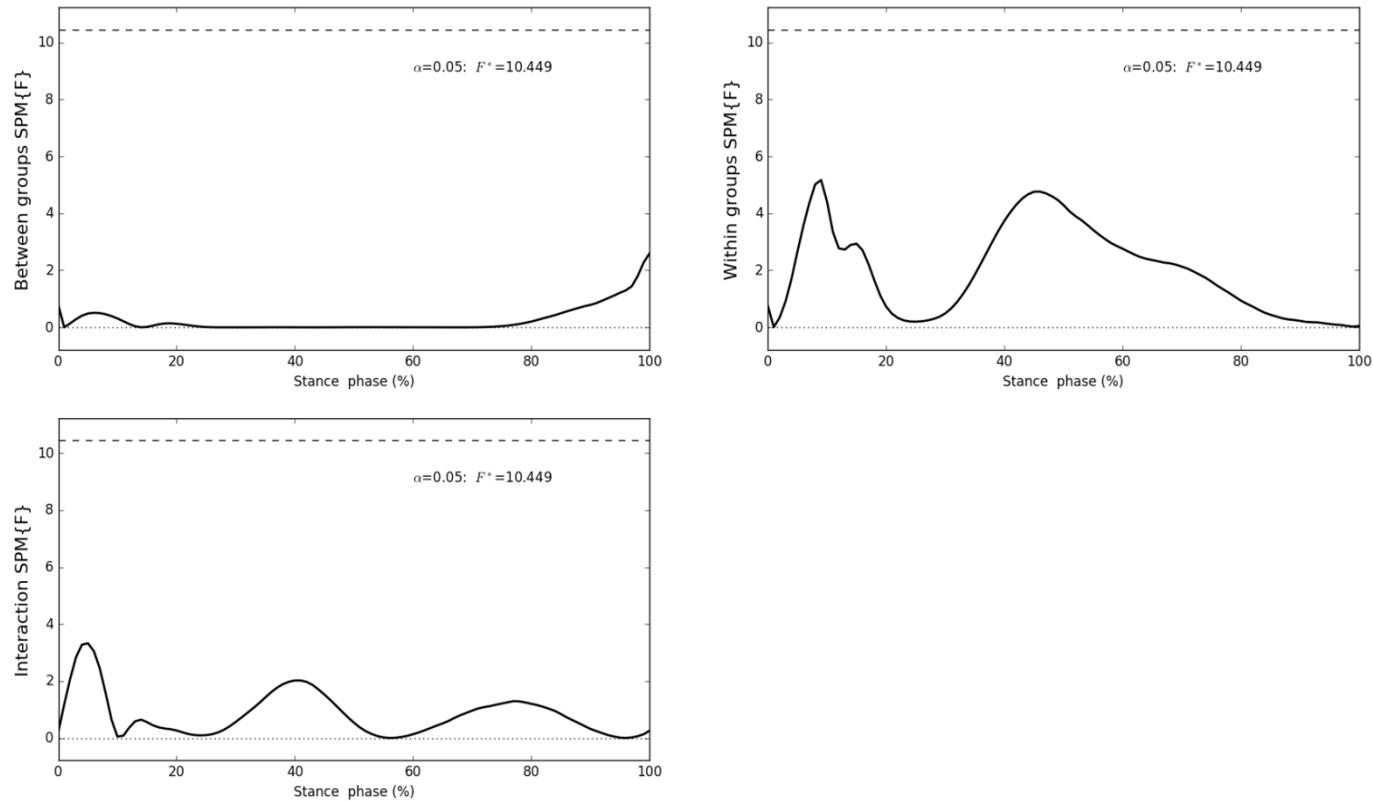


Figure 7-9 SPM results for ground reaction force. * The magnitude of GRF projected in line with the leg non-significantly increased post training in the “General” group. The results here demonstrate change to leg stiffness in the “Targeted” group was not due to changes in projected GRF.

Supplemental

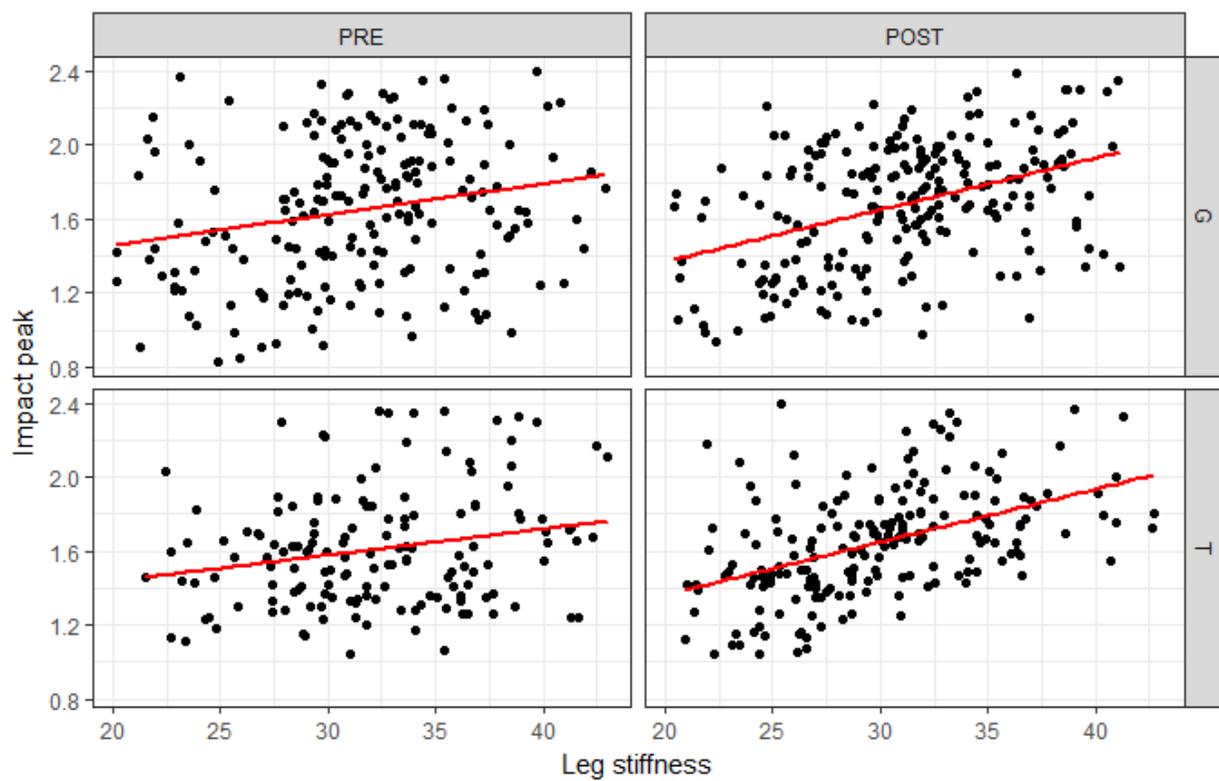


Figure 7-10 Simple linear regression between leg stiffness and impact peak (G = “General”, T = “Targeted”)

7.9 Acknowledgement

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7.10 Conflict of interest and source of funding

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Chapter 8

Loaded running kinematics after resistance training

Synopsis: In Chapter 5, we demonstrated that load carriage increased mid-stance ankle and knee peak flexion angles, and late stance hip adduction angle. These changes could potentially increase the magnitude and asymmetry of tissue loading potential in load carriage running. “Targeted” training was not more effective at reducing the three hypothesized negative kinematic adaptations of loaded running, compared to “General” training. However, only “Targeted” training appeared to increase ankle and knee flexion excursion angles, by adjusting initial contact angles, rather than increasing peak angles. These local kinematic responses could be in response to the high impact force nature of “Targeted” exercises, which in turn drove the reduction in leg stiffness. Exercises designed to optimize loaded running kinematics that may benefit tissue loading potential, may need to be performed without a wearable trunk load, and adopt coaching cues that facilitates an increase in leg stiffness. This study has been prepared as an article for submission.

Abstract

Purpose: Load carriage has been reported to increase peak flexion angles of the ankle and knee, and late stance hip adduction angle during running. These kinematic changes may represent an increase risk to tissue loading potential. The aim of this study was to investigate if resistance training can reduce these three kinematic changes associated with load carriage running.

Methods: This is a kinematic analysis of a randomized controlled trial on 30 healthy adult runners. Participants received either a load carriage biomechanically informed (“Targeted”) training program (countermovement jumps, single legged hopping, and hip flexor pull) or heavy-resistance (“General”) leg training program (leg press, unilateral calf raises, and lunge). Three-dimensional motion capture was performed on loaded (20 % body weight) running at 3.5 m/s. Statistical parametric mapping mixed ANOVA was used to compare kinematic waveform variables in stance.

Results: For all kinematic waveforms, there was only a time effect. Both training programs did not significantly reduce peak flexion angles of the ankle and knee. In addition, both training programs reduced hip adduction angle at late stance (by $< 1^\circ$). “Targeted” training appeared to increase ankle and knee flexion angular excursion from initial contact to mid-stance, more than “General” training.

Conclusions: Both resistance training programs did not change loaded running kinematics in a manner and magnitude, which reflects better tissue loading management. However, the kinematic adaptations across the ankle and knee after “Targeted” training may reflect a better impact force minimization strategy. The magnitude of impact force experienced during training and the coaching cues used, may be important training parameters that could facilitate changes to loaded running kinematics.

Trial registration (Date of registration): ANZCTR (ACTRN12616000023459) (14 Jan 2016)

Key words: Load carriage; Running; Impact; Kinematics; Resistance Training.

8.1 Introduction

Load carriage has been shown to increase peak flexion angles of the ankle and knee at mid-stance (~ 50 % cycle of stance), and hip adduction angle during late stance (> 90% cycle of stance) of running (Liew et al., 2016b; Silder et al., 2015). These kinematic changes could exacerbate the normal increase in joint contact loads (henceforth termed as maladaptive kinematics) (Willy et al., 2016; Xu et al., 2017) by increasing internal extensor muscle forces to counteract a greater external flexion joint torque, caused by a greater external flexor lever arm of the ground reaction force (GRF) (Perry et al., 1975; Saxby et al., 2016; Steele et al., 2012; Steele et al., 2010). This could explain why load carriage has been shown to increase peak patellofemoral joint (PFJ) contact force (Willy et al., 2016), a joint commonly implicated in overuse running injuries (Lopes et al., 2012). In addition, excessive joint angular kinematics in the frontal and transverse planes have been shown to induce asymmetrical loads to joints such as the PFJ (Powers, 2003; Powers, 2010), and tissues such as the Achilles tendon (Munteanu and Barton, 2011; Wyndow et al., 2010; Wyndow et al., 2013). Overtime, regions within a tissue that have been chronically exposed to elevated loads (Merican and Amis, 2009) or underexposed to normal loads (Ikoma et al., 2013) may be exposed to suboptimal loading environments that result in tissue degradation and pain. This could explain why individuals with knee pain have been shown to run with greater late stance hip adduction angle, compared to healthy controls (Esculier et al., 2015).

The importance of reducing maladaptive kinematics in load carriage running has inspired the design of gait retraining (Willy et al., 2016), and recently resistance training strategies (Liew et al., 2016a). Increasing loaded running cadence by 7.5 % reduces knee joint contact loads (Willy et al., 2016), which could be attributed to a reduced peak knee flexion angle (Heiderscheit et al., 2011). Although studies on resistance training for BW running (running with no load) kinematics exist (Albracht and Arampatzis, 2013; Earl and Hoch, 2011; Fukuchi et al., 2016; Snyder et al., 2009; Willy and Davis, 2011), none have investigated the influence of resistance training on specific maladaptive kinematics in load carriage running (Liew et al., 2016b). In addition, existing resistance training studies have been relatively unsuccessful in remediating gait-phase specific maladaptive BW running kinematics pertinent to the studied population (Albracht and Arampatzis, 2013; Fukuchi et al., 2016; Willy and Davis, 2011). However, there is inconsistent evidence that resistance training may reduce joint excursion angle (Fukuchi et al., 2016; Snyder et al., 2009). Common limitations to current resistance training programs used on BW running have been the use of relatively low training resistive loads (e.g. thera-bands) (Fukuchi et al., 2016; Willy and Davis, 2011), slow muscle contraction velocities and contraction modes (e.g. isometric) (Albracht and Arampatzis, 2013; Willy and Davis, 2011), that may not represent the rapid stretch-shortening muscle dynamics of running.

Training to increase leg stiffness, knee extensor power generation by augmenting muscular strength, coupled with augmented feedback to correct faulty movement patterns, may benefit maladaptive kinematics in load carriage running (Liew et al., 2016a). Training for greater leg stiffness may reduce peak landing flexion angles (Devita and Skelly, 1992; Powell and Williams, 2017; Sinclair et al., 2016). This may be important in load carriage running, given that peak flexion angles are important determinants of tissue loading magnitude during BW running (Lenhart et al., 2014; Whitting et al., 2011). Inadequate knee extensor strength was found to be accompanied by increased frontal and transverse plane hip and knee kinematics and moments (Kuenze et al., 2015; Lephart et al., 2002; Sturnieks et al., 2008; Thomas et al., 2015; Thomas et al., 2010). This may be relevant for overuse injury risk reduction in load carriage running given the frequent association between frontal and transverse plane joint kinematics and the risk of overuse running injuries (Munteanu and Barton, 2011; Powers, 2003; Powers, 2010; Wyndow et al., 2010; Wyndow et al., 2013). Augmented coaching feedback during training has been shown to reduce injurious transverse and frontal plane kinematics during dynamic activities (Greska et al., 2012; Myer et al., 2013), potentially benefiting loaded overuse running injury risk potential.

This study was undertaken to compare the effects of two training programs on loaded running kinematics. One group performed a resistance program focused on improving leg stiffness, knee power generation, and correcting known maladaptive kinematics of loaded running via augmented coaching feedback (termed “Targeted” training) (Liew et al., 2016a). The control group (termed “General” group) performed general machine-based heavy-resistance lower limb exercises (Liew et al., 2016a). The primary hypotheses for this study was that “Targeted” training would reduce mid-stance peak flexion angles of the ankle and knee, and late stance hip adduction angles during load carriage running, more than the “General” training.

8.2 Methods

8.2.1 Design, ethics, blinding

The detailed protocol of the study has been published (Liew et al., 2016a), and the primary analyses on strength, power, and loaded running kinetics have been reported (Chapter 7). This study received ethical approval from the Curtin University Human Research Ethics Committee (RD-41-14). A computer-generated sequence of random numbers was generated prior to participant recruitment (Liew et al., 2016a). Random allocation was concealed using sealed and opaque envelopes by an independent allocator. Only the outcome assessor (B.L.) was blinded to the group assignment. Written informed consent was sought prior to participant enrolment. The report of this trial follows the guidelines of the CONSORT statement.

8.2.2 Sample size

This study was powered on the effects of a core stability program that investigated changes to vertical stiffness (Dupeyron et al., 2013), which was the variable of focus in the primary analysis (Chapter 7). For a desired power of 0.80, and a Type 1 error rate of 0.05, 24 participants were needed to detect a between group mean difference of 3000 N/m (standard deviation of 3600 N/m) (Liew et al., 2016a). To account for a 20 % dropout over the intervention period, 30 participants were recruited.

8.2.3 Participants

Novice recreational runners aged 18 to 60 years old were invited to participate in this study. Individuals were eligible if they participated in running or running related sport for at least 45 minutes per week. Participants were excluded if they had: disorders precluding safe performance of running, load carriage, strength training, plyometric training; running related injuries that stopped their training within the last three months; current running related pain; lower limb surgery in the past year; or females currently pregnant.

8.2.4 Intervention

This was a two-week familiarization, six-week (18 sessions) training program, the details of which can be found in a previously published protocol (Liew et al., 2016a). In the “General” group, participants performed isoinertial training on bilateral leg press, seated unilateral calf raises, and lunges. External resistance progressed from 80 % of estimated one repetition maximum (1RM) during week one, to 88 % of the weekly-adjusted 1 RM during week six. In the “Targeted” training group, participants performed single-leg (SL) hopping to augment leg stiffness, countermovement jump (CMJ) to augment knee power generation, and a standing hip flexor cable isoinertial pull. The inclusion of hip flexor pull in this group was due a previous finding that load carriage increased hip flexor power absorption during terminal stance, to aid in pre-swing phase energetics (Liew et al., 2016b).

For the SL hopping and CMJ, the exercises were performed with a weighted body vest increasing in weight magnitude from 0 % body weight (BW) at week one to 20 % BW at week six. For SL hopping, the frequency was fixed at 2.2 Hz from week one to four, but increased to 3.0 Hz from week five to six. For the hip flexor pull, external load mass was progressed from 80 % of estimated 1RM at week one, to 88 % of the weekly-adjusted 1RM at week six.

Augmented coaching feedback was provided to both groups using the principles of motor learning (Liew et al., 2016a). Feedback was provided by the trainers to correct movement patterns during exercises, which represented an increased tissue loading potential under load (Liew et al., 2016a). Although both groups received feedback to facilitate the learning of optimal exercise movement patterns, only in the “Targeted” group was specific feedback provided with the intention of reducing peak flexion angles, and hip adduction angle. In addition, participants in the “Targeted”, but not the “General” group, received feedback during the performance of load carriage specific exercises.

8.2.5 Assessment

Three-dimensional motion capture was performed pre and post intervention in a biomechanics laboratory. Participants carried an identical backpack (CAMELBAK, H.A.W.G.® NV, 14 litre) filled with sand in sandbags to a weight of 20 % BW. The backpack's chest strap and hip belt were fitted securely to each participant. Participants wore their personnel running attire and shoes. A run up distance of 20 m was given prior to the first force plate, and a tail off distance of 10 m was given prior to the last force plate. Participants performed a minimum of five successful over-ground running trials at 3.5 m/s. Familiarization trials were given to practice the required velocity. Each trial was interspersed with a 30 s rest break. A successful trial was defined when the running velocity was within $\pm 10\%$ of the prescribed velocity, with a clean foot contact within the force plates. Retro-reflective markers were attached to the segments of the thorax, pelvis, bilateral thigh, shank and foot segments (Chapter 7). Static and dynamic modelling was performed in Visual 3D (C-motion, Germantown, MD). An eight-segment model was created with each joint having three rotational degrees of freedom (Liew et al., 2016b).

Marker trajectories were captured with an 18 camera system (Vicon T-series, Oxford Metrics, UK) at 250 Hz, whilst force data was captured with three synchronized in-ground force plates (sampled at 2000 Hz). A threshold of 20 N was used to determine initial contact and toe-off (Liew et al., 2016b). Marker trajectories were filtered at 18 Hz and a Cardan XYZ rotation sequence was used to calculate three-dimensional joint angles (Liew et al., 2016b).

8.2.6 Statistical analysis

Statistical parametric mapping (SPM) was used to perform a univariate between and within groups mixed Analysis of Variance (ANOVA) on nine kinematic waveforms (three lower limb joints with XYZ coordinates) over the stance phase of running. Statistical analysis was performed using spm1d package (v0.4) (www.spm1d.org), implemented in Python 2.7.11 within Enthought Canopy 1.7.4 (Enthought Inc., Austin, USA). The critical threshold for determining statistical significance was calculated using Random Field Theory (Pataky et al., 2016). Where significance was present in a gait cycle, the mean and standard deviation of the change within the significant period was calculated and reported. For example, if significance was found between 5 % and 10 % of the gait cycle, the mean and standard deviation of the six data points was reported. Since there were no drop outs and no cross-over in intervention, the intention-to-treat analysis was equivalent to a per-protocol analysis. All significances were set at $\alpha = 0.05$.

8.3 Results

All 30 participants completed the training program, and were available for final endpoint analysis (Figure 8-1; Table 8-1).

For all nine kinematic waveforms, there were no significant between group or time-by-group interactions. At the ankle joint, both training programs resulted in reduced ankle dorsiflexion in early stance (~ 5 % cycle) and increased plantarflexion in late stance (~85 % cycle) (Table 8-2, Figure 8-2a, d, Figure 8-5). In addition, both training programs increased ankle adduction during early (~ 5 % cycle) and late stance (~ 90 % cycle) (Table 8-2, Figure 8-4a, d, Figure 8-5).

At the knee joint, there was less knee flexion during early stance (~ 2.5 % cycle) and late stance (~ 90 % cycle) after both training programs (Table 8-2, Figure 8-2b, e, Figure 8-6). At the hip joint, there was greater hip flexion during early to mid-stance (~ 10 to 30 % cycle) (Table 8-2, Figure 8-2c, f, Figure 8-7), and less hip adduction during late stance (~ 85 % cycle) post training compared to pre training (Table 8-2, Figure 8-3c, f, Figure 8-7).

8.4 Discussion

Load carriage has been shown to increase peak flexion angles in mid-stance and late stance hip adduction angle in running (Liew et al., 2016b). In this study, we compared the effectiveness of a biomechanically informed, “Targeted” training regime against “General” resistance training at minimising these kinematic adaptations during load carriage running. Contrary to our primary hypotheses, “Targeted” training was not more effective than “General” training at reducing mid-stance peak flexion angles and late stance hip adduction angle. Interestingly, a visual inspection of the post-pre kinematic changes suggested that “Targeted” training increased ankle and knee joint flexion excursion angles from initial contact to mid-stance.

In this study, there were no reductions in both peak ankle and knee flexion angles at mid-stance during loaded running, despite significant strength and power improvements with both training programs (Chapter 7). Our findings generally concur with that of the resistance training literature on BW running kinematics, that phase-specific and axis-specific angular changes may not consistently map with physiological improvements (Albracht and Arampatzis, 2013; Fukuchi et al., 2016; Snyder et al., 2009; Willy and Davis, 2011). A 14-week ankle extensor isometric strengthening program improved ankle extensor isometric strength by 7 %, but did not alter running ankle kinematics (Albracht and Arampatzis, 2013). A six-week hip strengthening program which utilized isometric and slow-isoinertial contractions improved hip isometric strength by 42 %, yet still did not reduce running peak hip kinematics (Willy and Davis, 2011). The lack of transference of strength gains to running kinematics could be due to the utilization of incompatible muscle contraction training modes and contraction velocities, compared to the rapid stretch-shortening requirements of running (Nicol et al., 2006). Yet, despite “Targeted” training incorporating rapid stretch-shortening training modes, this did not translate into running kinematic changes.

The inability to detect changes to local joint angles at specific gait phases after resistance training could imply that instantaneous joint angles are not systematically “controlled” in gait (Auyang et al., 2009; Yen et al., 2009). The mechanical outcomes of running that may be more responsive to resistance training, may lie at global mechanical variables, such as leg stiffness and leg length, rather than local joint angles (Auyang et al., 2009; Yen et al., 2009). From a motor control perspective, individual joint angles can co-vary as long as the global variable is stabilized across repetitions (Auyang et al., 2009; Yen et al., 2009). This suggests that the variability in a global measure between trials and individuals may be less than the variability of local joint angles. This could explain why leg stiffness was observed to reduce after “Targeted”, and not “General” training (Chapter 7), but phase-specific and axis-specific joint angles were not found to be statistically different between groups.

Although not statistically significant and in contrast to this study's hypothesis, there was a tendency for "Targeted" training to increase joint excursion angles of the ankle and knee by mid stance, compared to "General" training. This was brought about by a reduction in flexion angles of the ankle and knee at initial contact after "Targeted", which tapered off by mid-stance (Figure 8-2d, e). In contrast, "General" training reduced ankle and knee flexion angles throughout stance (Figure 8-2d, e). These kinematic adaptations with training may in part be a response to the high impact force exercises in the "Targeted" program. Strategies such as a reduced ankle flexion angle at initial contact, and increased ankle and knee landing excursion angles have been reported to reduce impact force (Begalle et al., 2015; Hunter and Marshall, 2002; Rowley and Richards, 2015; Tam et al., 2016; Zhang et al., 2008). The kinematic adaptations observed after "Targeted" training have also been observed after unshod running training (Tam et al., 2016), and drop jump training from high drop heights (Hunter and Marshall, 2002; Taube et al., 2012), all of which imposes significant impact force.

Impact force magnitude with exercises in "Targeted" training could have been larger than typical plyometric exercises, given that participants could have experienced two collisional forces, instead of one (Figure 8-8). One collision present in all plyometric exercises, happens between the foot and the ground during landing (Zadpoor et al., 2007). However, in the "Targeted" program, hopping and jumping were performed with an additional moveable trunk borne vest (Liew et al., 2016a). When a vest is worn on the trunk during plyometric training, a second potential collision happens during landing between the vest and trunk. This second collision happens because any trunk borne loads cannot be rigidly fixated to the trunk (Ren et al., 2013). Given that participants in the "Targeted" group needed to cope with greater impact force in training, the preferred adaptation could be an increase in joint excursion angles during load carriage activities.

Another factor that may have promoted the increase in joint excursion angles was the coaching cues used in training (Liew et al., 2016a). In the "Targeted" group, participants were verbally instructed to "keep landing soft" during hop and jump training, which could have encouraged an increase in joint excursion angles (Milner et al., 2012; Verniba et al., 2017). Although participants were instructed to "stay tall at the knees" during hopping to increase leg stiffness (Liew et al., 2016a), the cue to "keep landing soft" could have counteracted the exercise's intended goal. Future research investigating the influence of conflicting coaching cues and the sequential nature of cues delivery on motor skill acquisition is warranted.

The current implementation of “Targeted” training may be relevant for individuals who must perform loaded running where an increase in joint angular excursion is beneficial. One example may be the performance of loaded running with combat target shooting (Jaworski et al., 2015), where greater joint angular excursion may reduce impact force. Better impact force management may consequently improve visual-motor stability (Lim et al., 2017). However, to positively alter joint angles which benefits tissue loading potential in loaded running, “Targeted” exercises may need to be modified. First, plyometric exercises may need to be performed either without a trunk load, or using a loaded paradigm that avoids additional collisional forces (e.g. plyometric with elastic band resistance). Without a moveable trunk load, the body may only need to cope with impact forces coming from a foot-ground collision. Second, coaching cues which encourage a “softer” landing should not be used in exercises designed to increase leg stiffness.

One limitation of this study was the absence of a no-intervention group, which means that the reported longitudinal kinematic changes may be a consequence of repeated measurements. However, this is unlikely given that there was a significant time-by-group interaction in the previously reported measure of running leg stiffness and leg length deformation (Chapter 7). A second limitation was the relatively short training period. Training periods for load carriage in the military have spanned between eight to 24 weeks (Knapik et al., 2012). However, the present training duration was long enough to produce significant strength and power improvements (Chapter 7). In addition, longer resistance training periods have not been shown to consistently result in running kinematic changes (Albracht and Arampatzis, 2013; Fukuchi et al., 2016).

A biomechanically informed “Targeted” training does not appear to better optimize loaded running joint angles in a manner that reflects a significantly reduced tissue loading potential. However, the kinematic alterations may occur as a response to the chronic exposure to high impact forces during “Targeted” training, which was absent with “General” machine-based exercises. Training designed to improve loaded running kinematics that benefit tissue loading potential needs to be modified to reduce lower limb joint excursion angles.

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8.6 Figures

Figure 1.

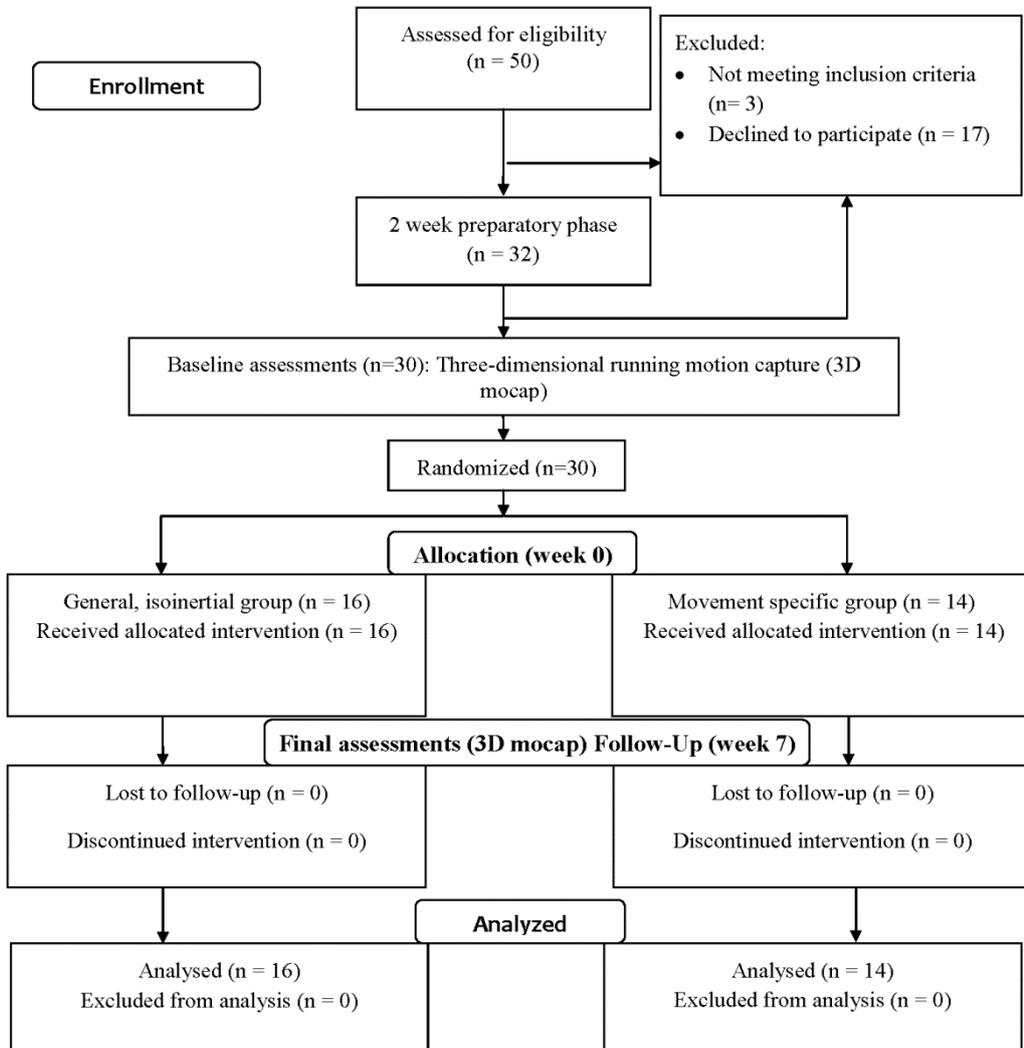


Figure 8-1 CONSORT Flow Diagram

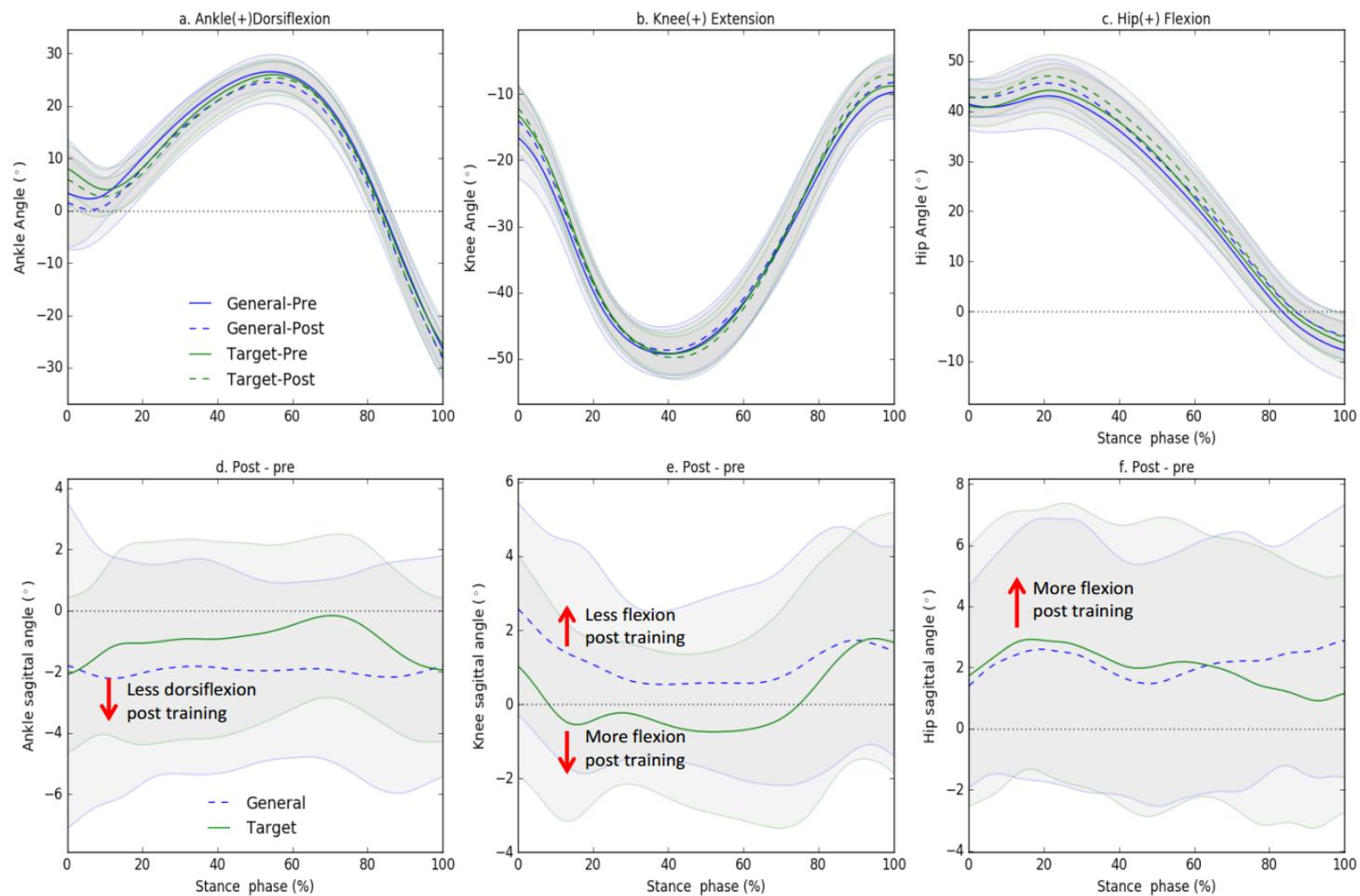


Figure 8-2 (a-c) Mean (standard deviation) waveform of sagittal plane kinematics over the stance phase of running; (d-f) Mean (standard deviation) waveforms of change in kinematics with time (post curves minus pre) for each group.

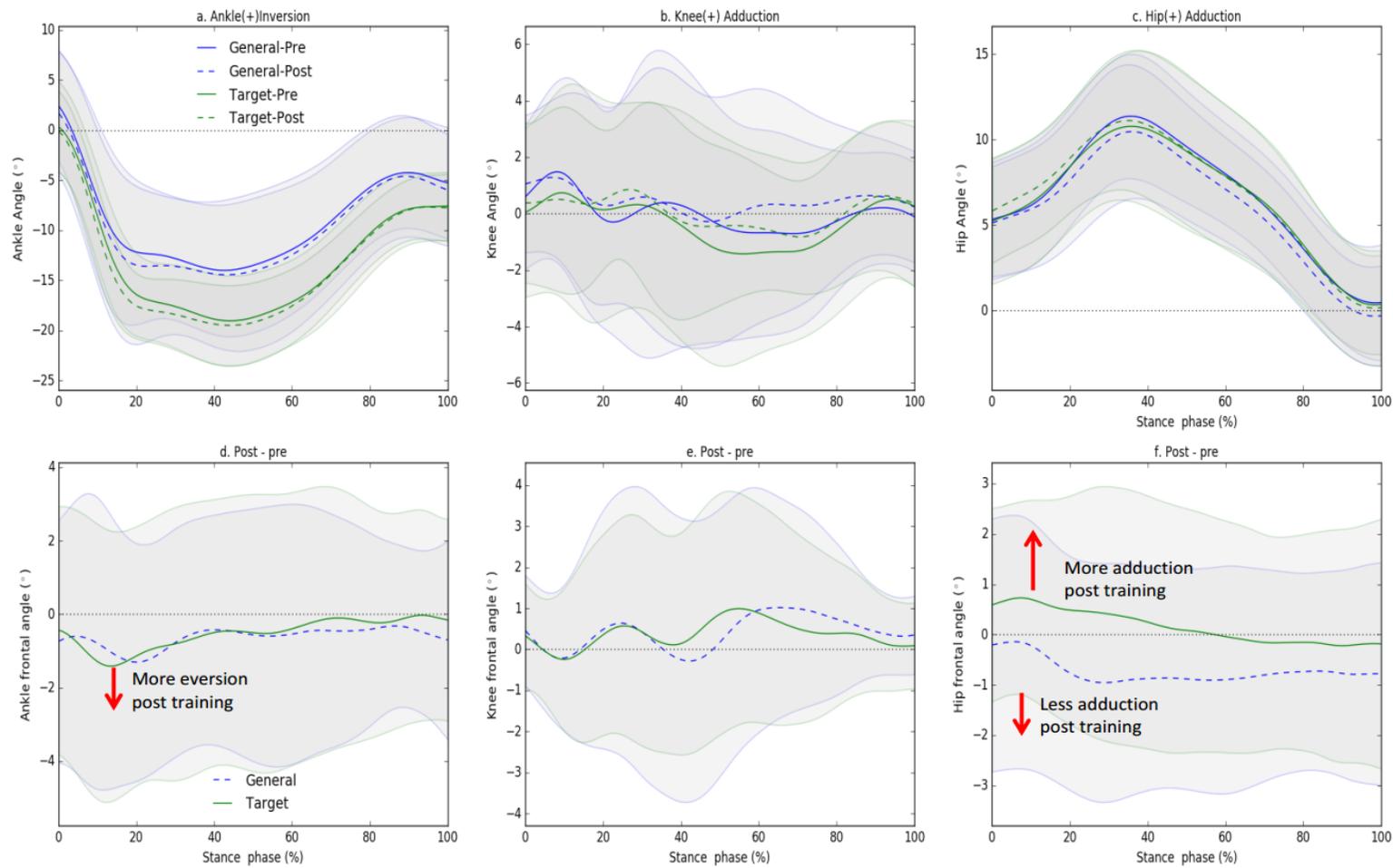


Figure 8-3 (a-c) Mean (standard deviation) waveform of frontal plane kinematics over the stance phase of running. (d-f) Mean (standard deviation) waveforms of change in kinematics with time (post curves minus pre) for each group.

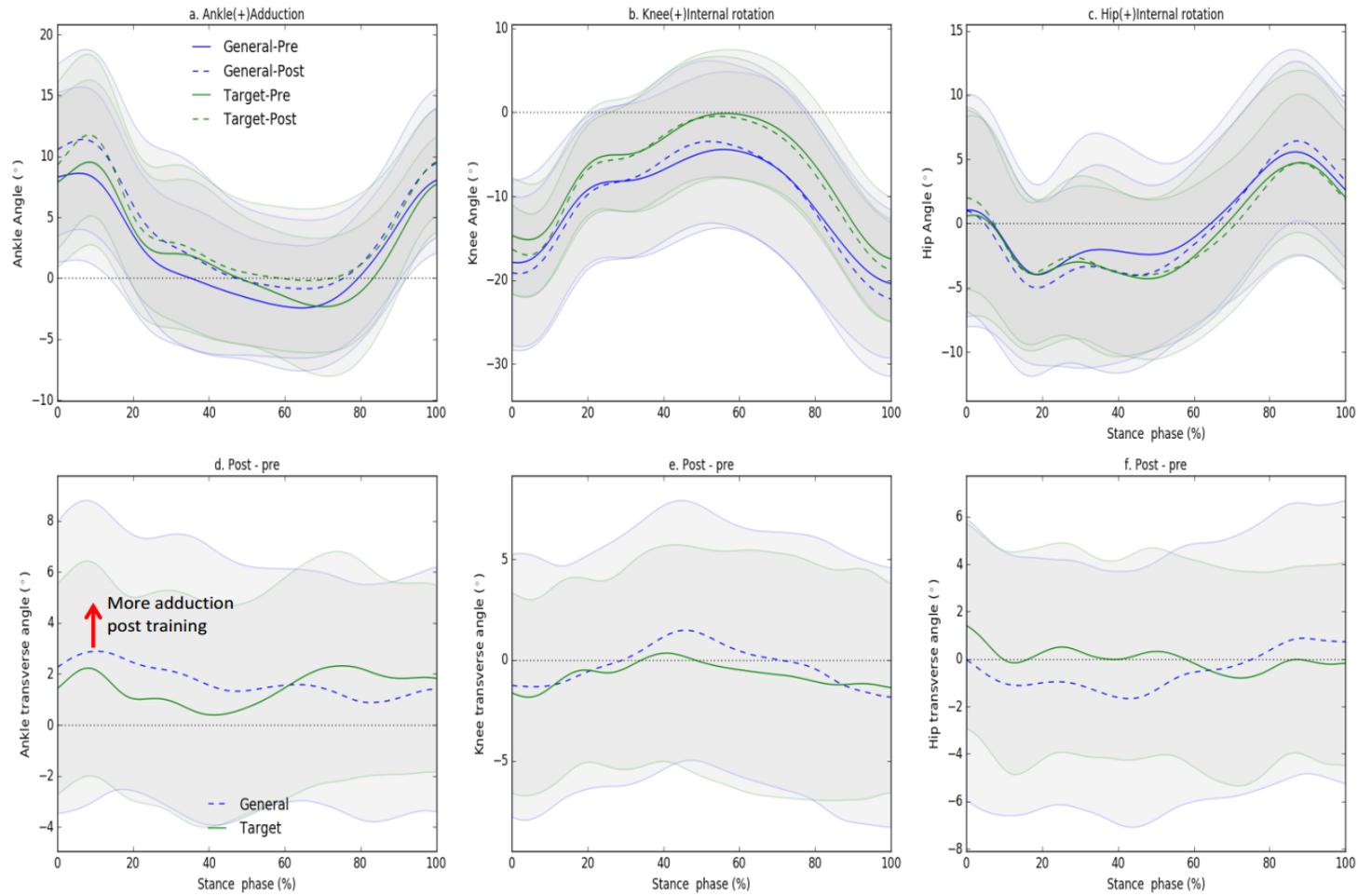


Figure 8-4 (a-c) Mean (standard deviation) waveform of transverse plane kinematics over the stance phase of running; (d-f) Mean (standard deviation) waveforms of change in kinematics with time (post curves minus pre) for each group.

Figure 5.

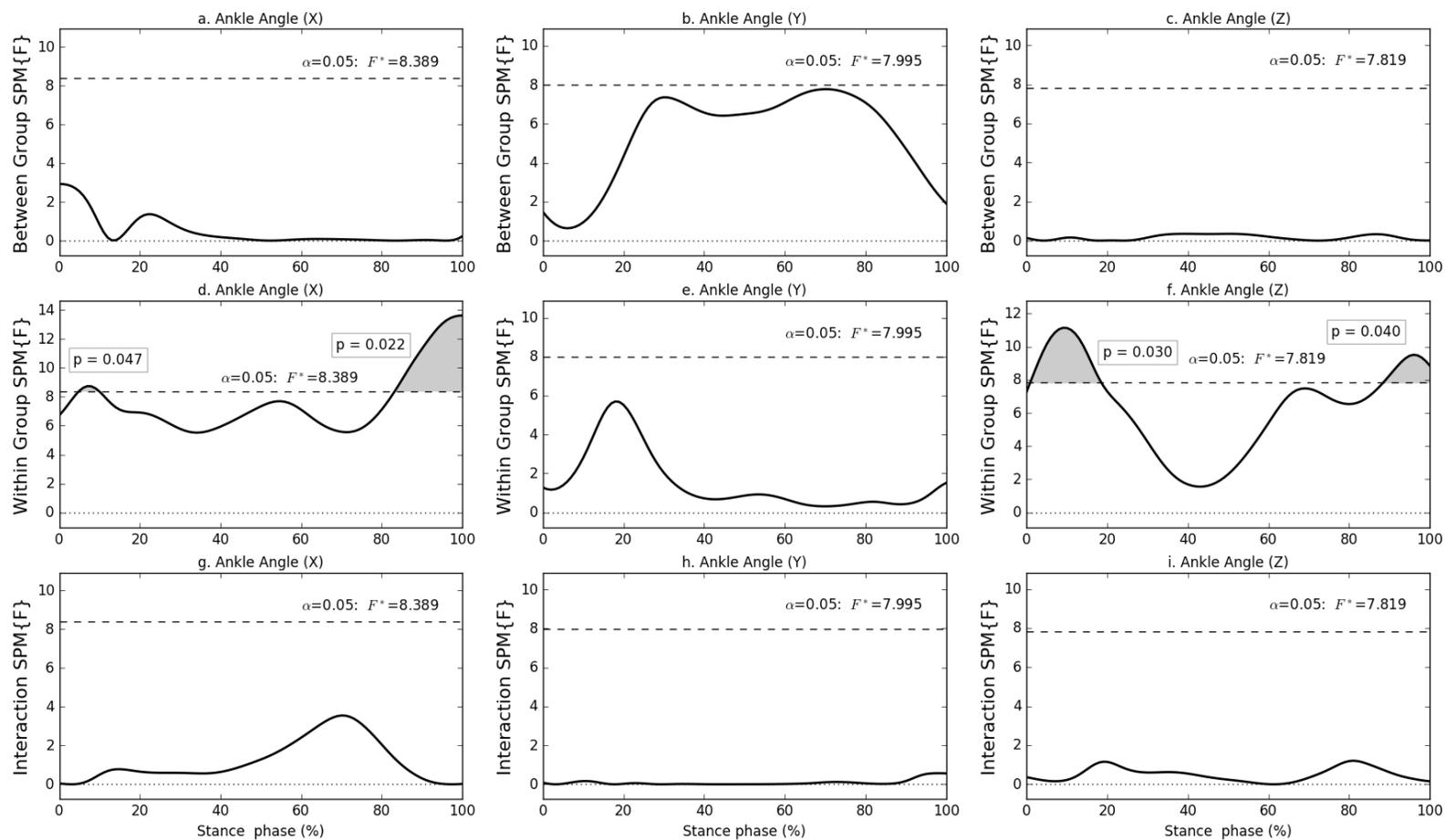


Figure 8-5 Statistical parametric maps of repeated measures anova for ankle X (sagittal plane), Y (frontal plane), Z (transverse plane) angles.

Figure 6.

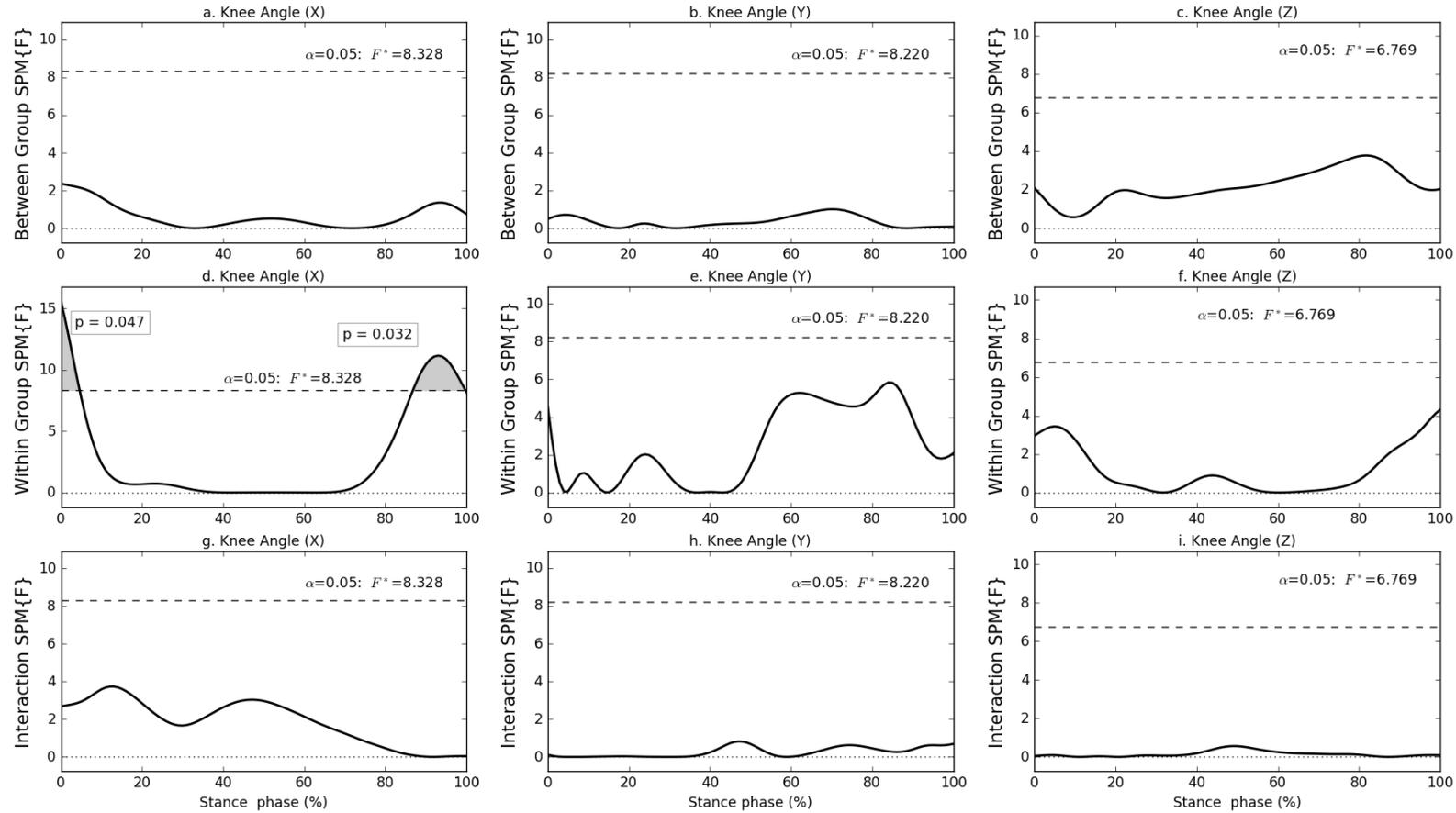


Figure 8-6 Statistical parametric maps of repeated measures anova for knee X (sagittal plane), Y (frontal plane), Z (transverse plane) angles.

Figure 7.

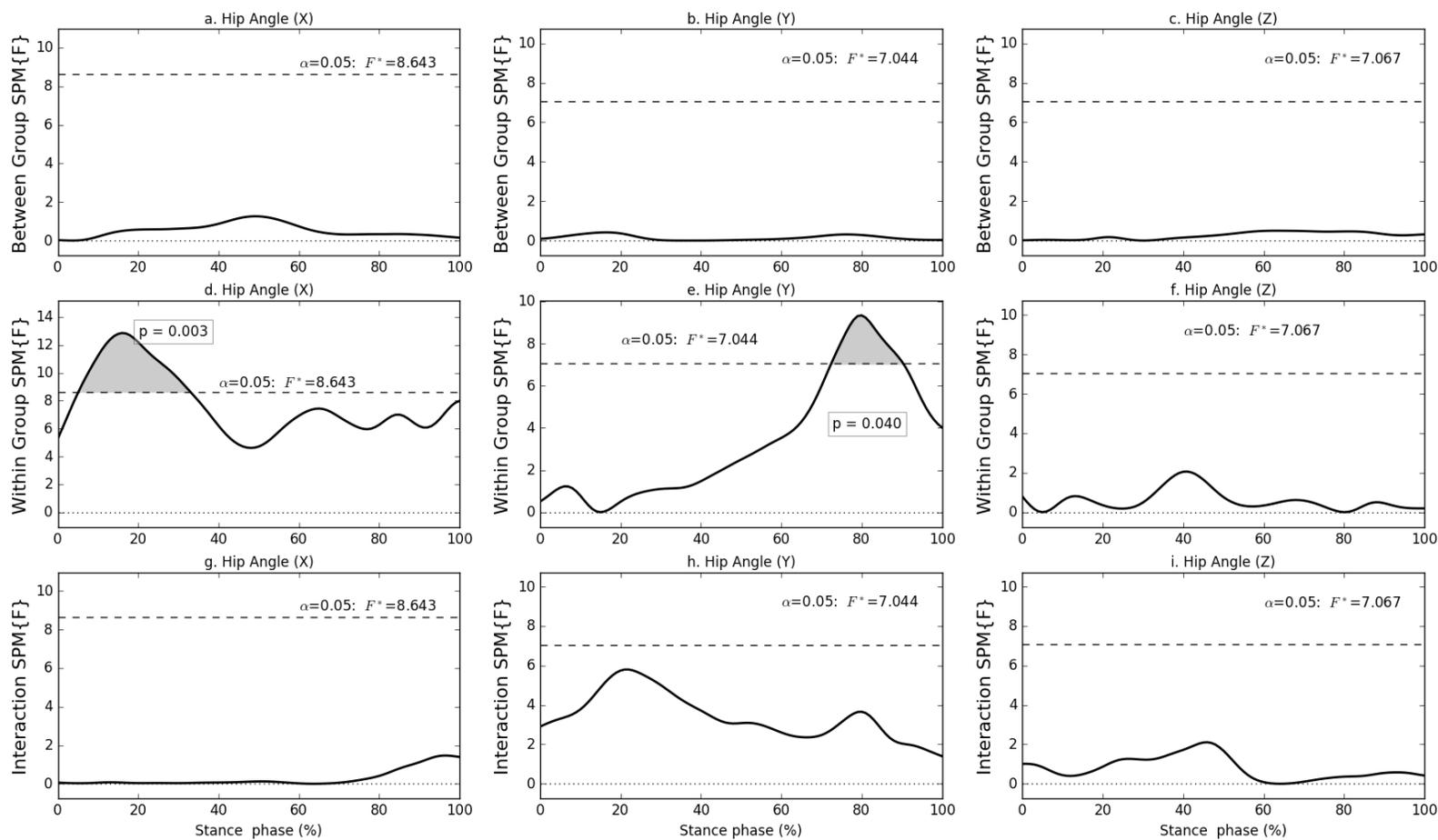


Figure 8-7 Statistical parametric maps of repeated measures anova for hip X (sagittal plane), Y (frontal plane), Z (transverse plane) angles.

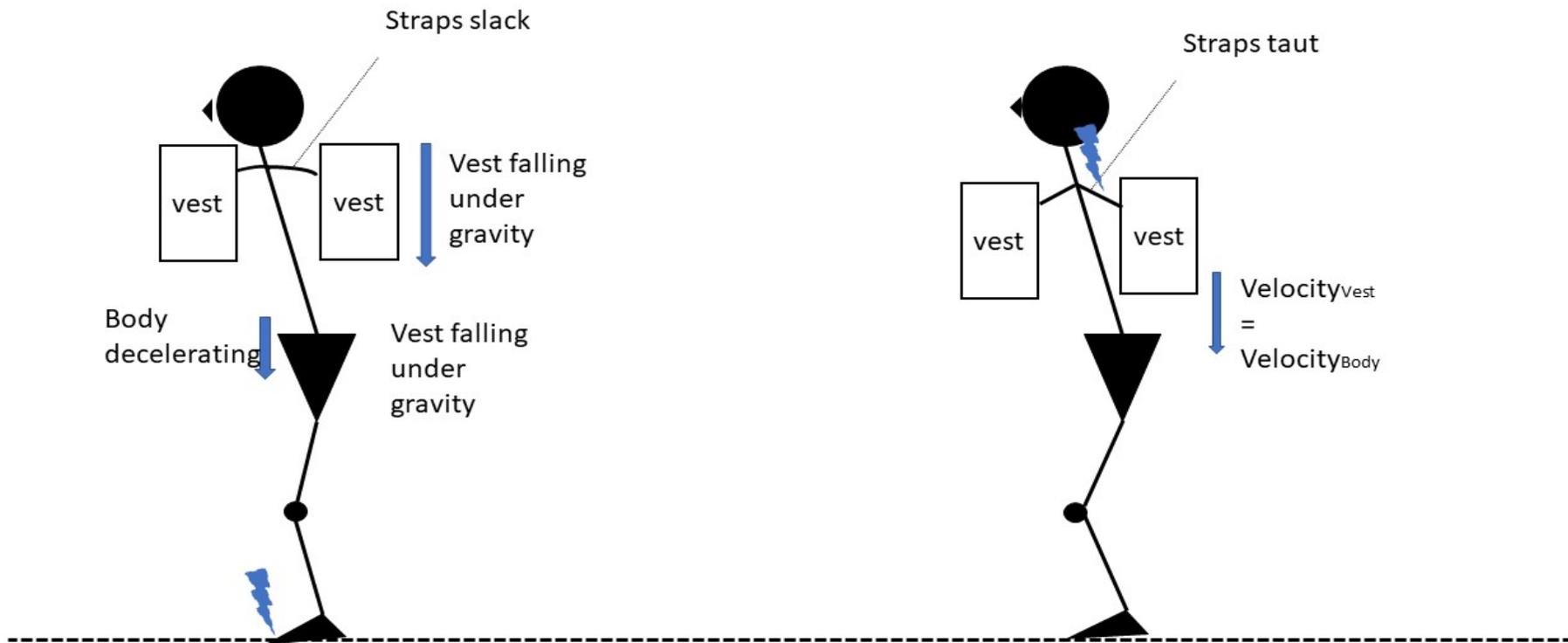


Figure 8-8 Illustration of foot-ground and backpack-trunk collisions during loaded hopping training

8.7 Tables

Table 8-1 Baseline participant characteristics

	“General” (n = 16)	“Targeted” (n = 14)	p value
Age (years)	30.8 (8.8)	29.4 (9.9)	0.691
Body mass (kg)	67.8 (14.0)	69.2 (10.3)	0.750
Height (cm)	172.4 (8.8)	171.7 (6.4)	0.811
Sex (Male:Female)	8:8	7:7	1.00
Resistance training frequency over past 6 weeks (times/week)	1.0 (0.8)	1.7 (1.6)	0.146
Resistance training frequency over past 12 months (times/week)	1.2 (1.2)	3.0 (3.3)	0.067
Resistance training experience (B:I:A)#	12:4:0	6:5:3	0.085
Running experience (years)	9.6 (5.9)	9.3 (6.3)	0.903
Running frequency over past 6 weeks (times/week)	2.4 (1.3)	3.0 (1.5)	0.235
Running frequency over past 12 months (times/week)	2.1 (1.4)	3.4 (3.4)	0.007
Running distance over past 6 weeks (km/week)	12.0 (13.7)	21.8 (20.2)	0.138
Running distance over past 12 months (km/week)	12.4 (18.6)	18.6 (13.3)	0.301
Foot contact pattern baseline (RFS:MFS:FFS)	9:5:2	13:1:0	0.064
Foot contact pattern post training (RFS:MFS:FFS)	10:3:3	13:1:0	0.140
Load running experience* (Y:N)	13:3	12:2	1.00

Attendance (maximum of 18)	16.7 (1.8)	16.9 (1.8)	0.801
<p>Abbreviations: DF – degree of freedom; B – beginner; I – intermediate; A- advanced; Y – Yes; N – No; RFS – rear foot strike; MFS – mid-foot strike; FFS – forefoot strike</p> <p>#Beginner = Currently not resistance training or started but ≤ 2 months with a frequency of ≤ 1 session/week; Intermediate = Currently doing resistance training and started within the last 2 to 6 months with a frequency of 2 to 3 sessions/week; Advanced = Currently doing resistance training with ≥ 1 years' experience with a frequency of ≥ 4 sessions/week</p> <p>*Experience (Y) = running for a minimum of 30 minutes at least once per month with a load of ≥ 10 % of your weight, for at least 6 times per year</p>			

Table 8-2 Mean (standard deviation) of joint angle change with training.

Joint angle	% Gait cycle with significance	Targeted angle (°) change (post – pre)	General angle (°) change (post – pre)
↓ Ankle DF	4.9 to 9.9	-1.55 (2.50)	-2.11 (4.24)
↑ Ankle PF	83.3 to 100	-1.54 (2.49)	-2.05 (3.66)
↑ Ankle Add	1.0 to 19.0	1.79 (4.04)	2.71 (5.56)
↑ Ankle Add	88.4 to 100	1.85 (3.66)	1.22 (4.60)
↓ Knee Flex	0.0 to 4.6	0.75 (2.56)	2.31 (2.81)
↓ Knee Flex	86.7 to 99.7	1.67 (3.20)	1.63 (2.79)
↑ Hip Flex	5.2 to 33.0	2.72 (4.34)	2.40 (4.06)
↓ Hip Add	72.4 to 90.4	-0.17 (2.15)	-0.75 (1.97)
↑ = Increase; ↓ Decrease; DF = Dorsiflexion; PF = Plantarflexion; Add = Adduction; Flex = Flexion; Ext = Extension			

8.8 Acknowledgement

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8.9 Conflict of interest and source of funding

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Load carriage poses significant problems to the physical performance and injury risk potential of individuals who must run with load (Carlton and Orr, 2014; Orr et al., 2015; Orr et al., 2014). Presently, physical conditioning is the most immediate and practically feasible solution to the physical burden of load carriage (Knapik et al., 2012). A significant gap in the literature has been the absence of a biomechanically informed training program for load carriage, especially towards the optimization of running mechanics. This gap exists predominantly because of a dearth in background knowledge of how running mechanics change with load carriage. A second reason is due to the absence of a translational framework in load carriage research, which integrates background knowledge into foreground intervention design. Hence, the aim of this thesis was to address two significant thesis questions. First, how does the biomechanics of running change under load? Second, what is the efficacy of a biomechanically informed resistance training program on loaded running mechanics?

9.1 Lessons learnt on load carriage running

In the pursuit to address the two thesis questions, four lessons were learnt during the conduct of three experimental studies.

9.1.1 Lesson one: Accurate experimentation requires valid task-specific experimental methods

Prior to being able to perform any experiments involving carrying a backpack during running, I needed to design and validate a pelvic marker set that would not be visually occluded or physically displaced by a backpack (Chapter 3: Pilot study). The pelvic segment forms an important “root” within a chain of linked segments that make up a kinetic model. This study confirmed that the new lateral pelvic cluster marker set produced good to excellent coefficient of multiple correlation magnitudes, and similar sensitivity in detecting three dimensional pelvic and hip joint angles during the stance phase. compared to the traditional International Society of Biomechanics pelvic marker set. The results of this study allowed me the confidence to pursue valid experimentations of load carriage running mechanics.

9.1.2 Lesson two: Knowing “what” to train in loaded running

The design of a biomechanically informed resistance program requires understanding what muscle groups, muscle variables, and muscle contraction modes to train in load carriage running (section 2.7). To answer these questions, I conducted a kinetic and kinematic analysis of running with load carriage involving three different load magnitudes (0 %, 10 %, 20 % BW), over three different velocities (3, 4, 5 m/s). From the analysis in Chapter 4 (Study one), all joint-level muscle groups were important to allow the body to sustain constant velocity running during load carriage. A smaller contribution to the adaptation of loaded running was a shift in work from the hip to the ankle. This first analysis suggested that resistance training for load carriage running needs to focus on all joint-level muscle groups, rather than a single joint’s muscle group.

Chapter 5 (Study one) provided an added layer of information to aid in the programming of resistance training parameters. Rather than a constant increase in power with load, running with load demanded transient increase in muscle power at specific periods of the gait cycle. Load carriage influenced this increase in muscle power at mid-stance for the ankle, push-off for the knee, and late-stance for the hip. Based on this information, resistance training designed for loaded running needed to augment force generation/absorption capacity within a short period of time, by incorporating plyometric exercises. These results also informed the pertinent muscle contraction modes during training for each joint-level muscle group, from a rapid eccentric-concentric transition mode for the ankle, concentric mode for the knee, and eccentric mode for the hip. The gait phase-specific and joint-axis specific changes to running kinematics under load assisted in the development of coaching cues needed to ensure optimal movement patterns when training under load.

9.1.3 Lesson three: Optimization of different cost functions in loaded running

The role of resistance training in altering running mechanics may be different from interventions such as gait retraining, and both should be viewed as complementary. Resistance training plays a role in augmenting physiological muscular (and tendon) mechanical properties, and generic movement skills. Given that movement pattern is not instructed during running, mechanical changes after resistance training may depend on an implicit trade off analysis by the motor control system, on which cost functions are to be optimized. In contrast, gait retraining explicitly instructs on the desired movement pattern during running (Davis and Futrell, 2016; Willy et al., 2016), and movement pattern is determined less by an internal trade off analysis. This could account for the lesser consistency in running mechanical changes with resistance training, as compared to gait retraining (Davis and Futrell, 2016; Willy et al., 2016).

A yet unconsidered role of resistance training is in “tuning” the muscular system to the input ground reaction force (GRF) stimulus (Nigg and Wakeling, 2001). Resistance training in principle focuses on augmenting muscular force-velocity deficits (Cormie et al., 2011). Yet, any land based exercises will provide an input GRF stimulus, which causes the motor control system to react and alter the control of muscles and movement (Nigg and Wakeling, 2001). Even without explicit kinematic instructions, it would appear that the motor control system intrinsically tunes the control of muscles and movement to manage input forces (Tam et al., 2016; Zelik and Kuo, 2012). Evidently, any selected running pattern cannot optimize all possible cost functions (Miller and Hamill, 2015). When the goal of training is to augment impact force management, a prolonged period of exposure to high impact force exercises may be beneficial. However, high impact force exercises may place the optimization of mechanical work and tissue loading management at a disadvantage during load carriage running. This could occur by high impact force training inducing lower limb running kinematic and kinetic alterations that increases the lever arm of the GRF to individual joint’s centres, thereby requiring greater counteracting internal joint torques and work.

9.1.4 Lesson four: Leg stiffness may underlie work and tissue loading optimization

When the goal of training is on the reduction of loaded running mechanical work and the negative kinematic changes associated with tissue loading potential, muscular strength gains must be coupled with an increase to leg stiffness. Based on the results of Study two, “Targeted” training parameters need to be modified to augment loaded running leg stiffness. First, heavy-resistance exercises should be performed by novice strength trainees to enable a more rapid strength improvement. Second, plyometric exercise components should mimic the contact time of loaded running to augment rapid eccentric and concentric force generation capacity. Third, plyometric training needs to be performed without a moveable trunk load, to avoid additional collisional impact force. Fourth, only coaching cues which facilitates an increase in leg stiffness should be used. With appropriate levels of leg stiffness, the motor control system would have the appropriate mechanical leverage to respond to gait-coordination modifying interventions, such as gait retraining.

9.2 Limitations of research

The body of research conducted for this thesis has several limitations. First, the biomechanical outcome measures used in the studies are surrogate outcome measures. Actual outcome measures of physical performance such as a loaded running time trial, actual running related injury cases, and in-vivo tissue loading were not measured. Generalisation of our findings of load carriage and physical training from the surrogate outcomes, to clinically relevant outcomes should be made with caution. In addition, given that this thesis focused on changes to running mechanics during acute loaded conditions, extrapolation to chronic loaded conditions, where variables such as fatigue are present, should also be made with caution.

The second limitation of this study is a lack of long term follow-up of participants in Study two. This precludes us from making long term inferences of the retention effects of the training programs to running biomechanics. Moreover, the associative relationship between running biomechanics with strength and power alterations during the detraining phase cannot be established. However, the “permanence” of strength and power gains with resistance training is contingent on the persistence of resistance training in daily life (Rønnestad et al., 2010), and the former should not be expected to be retained when resistance training is discontinued (Kubo et al., 2010).

The third limitation of this thesis is that inferences about the effects of load carriage and physical training on running beyond the study population should be made with caution. This is because the effects of load carriage on gait mechanics have been shown to be related to the skill expertise of load carriage (Liew et al., 2016), and the effects of resistance training can be influenced by different athletic populations (de Villarreal et al., 2009).

Lastly, the focus of this thesis was to determine the optimal type of exercises needed to improve load carriage running mechanics. In so doing, the thesis did not explore the effect of periodised resistance training on loaded running mechanics (Hartmann et al., 2015).

Periodised training aims to incorporate a systematic plan of training to achieve peak physical performance at the end of a sports cycle. It involves breaking up a training period into smaller sub-periods, where different performance variables are trained within each sub-period. When examining the interventions used in this thesis within the periodization landscape, different training modalities may need to be incorporated at different periods to optimize load carriage running mechanics. It could be that for novice strength trainees wanting to improve loaded running mechanics, “General” isoinertial training needs to occur first to provide a better muscular strength foundation. Next, modified “Targeted” training can be added to emphasize technique and rapid stretch shortening contractions. Lastly, a period of gait retraining would be needed to alter running coordination patterns.

9.3 Future research directions

Clearly, there has been a growing body of research into load carriage over the last decade, especially into more dynamic tasks, such as running. I believe that the excitement surrounding load carriage is not simply that it is an important task unto itself (which it is), but that load carriage research ultimately aims to uncover mechanisms that aid in human performance optimization. In Figure 9-1, optimal individualized load carriage training can be developed through answering four questions. First, there is a need to determine what tasks are required of an individual carrying load. Second, there is a need to understand the cost functions required to successfully accomplish the required tasks. Third is the determination of modifiable performance variables (Latash et al., 2007), that map to the desired cost functions needed for improvement. Fourth is the determination of the individual muscular and joint requirements (elemental variables) that influence the desired performance variables.

To answer these four questions, I believe there needs to be greater involvement of multiple research disciplines in load carriage research. Disciplines such as biomechanics, motor control, physiology, and strength and conditioning, all have important contributions to make to assist in the optimization of load carriage training methods.

From a biomechanics perspective, greater knowledge fidelity can be gleaned using musculoskeletal modelling. Musculoskeletal modelling not only allows the quantification of in-vivo tissue loading and muscle-tendon energetics, it is also capable of predicting optimal movement patterns under a variety of candidate neuromuscular control schemes (Miller and Hamill, 2015). A motor control perspective enables scientists to understand performance variables that the central nervous system uses to coordinate a redundant system of muscles during load carriage. Understanding the performance variables involved in load carriage, allows the development of training methods to positively augment loaded running coordination. Gait mechanics are often driven by a need to reduce energy cost. From a physiological perspective, future studies should investigate how different running coordination strategies (e.g. increasing cadence) influence loaded running energy cost, under varying fatiguing conditions. Lastly from the strength and conditioning perspective, future studies should investigate the effects of periodised resistance training on load carriage running mechanics and performance. This means that optimal physical training for load carriage may be subject-specific, tailored to baseline neuromuscular status, and progressively targets different “layers” of neuromuscular control.

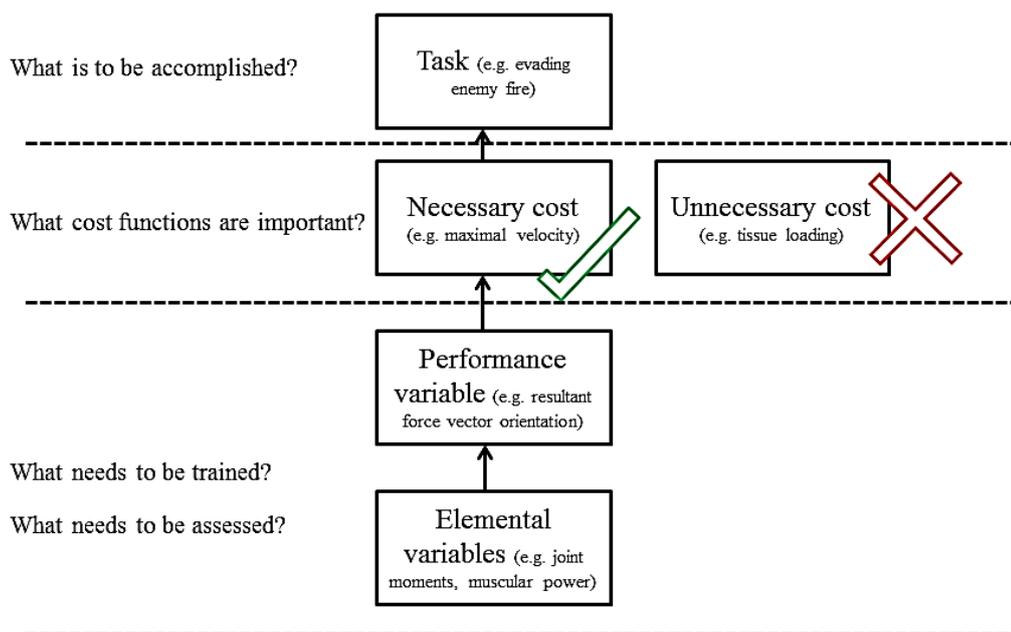


Figure 9-1 Decision tree for designing load carriage training

9.4 Conclusion

Investigations on how load carriage influences physical performance and injury risk in running are important from an occupational and sporting, health and safety perspective. The new biomechanically informed training program appears to be more effective at managing some cost function (e.g. impact force), than others (e.g. mechanical work and kinematic exacerbations of tissue loading potential). Significant advancement in load carriage training design requires the understanding of the task requirements during load carriage, the cost functions which are pertinent, and the modifiable mechanical and physiological variables. These questions can benefit from a multi-disciplinary approach to load carriage research to generate further candidate training variables for investigation. To facilitate background knowledge translation into foreground intervention design, I have shown that an “adaptive-maladaptive” translational framework is a useful tool for future research in load carriage.

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Appendix A Short commentary on swing mechanics in loaded running

A.1 Introduction

The focus of Study one was to describe the biomechanics of loaded running that would aid in the design of a biomechanically informed resistance training program. The purpose of the training program was aimed at improving mechanics that would drive improvements in energy cost and running injury risk. Most biomechanical variables that are related to energy cost and injury mechanisms are associated with stance phase mechanics (Arellano and Kram, 2014; Neal et al., 2016). Indeed, the energy cost needed to swing the legs represents less than 10 % of running's total energy cost (Arellano and Kram, 2014). In addition, very limited studies have identified swing phase biomechanical variables that could contribute to running overuse injuries (Schmitz et al., 2014). Hence, the focus of Study one was on the report of stance phase loaded running mechanics. The purpose of this commentary is to report how swing phase mechanics in running change with load, to provide a comprehensive descriptive analysis of load carriage running biomechanics. Again, the focus of the report of this short commentary is restricted to gait phases and variables which have been linked to running energy cost and injury mechanisms.

A.2 Methods

The methods follow those reported in Chapter 4 and Chapter 5. Swing phase was defined between toe-off and initial contact of the right limb. The following swing phase biomechanical variables were reported: positive and negative joint work of the ankle, knee, and hip, joint power and kinematic waveforms of the ankle, knee, and hip. Joint work and joint power were presented as dimensionless units. One unit of joint work represents a mean of 532 J, whilst one unit of joint power represents a mean of 1845 W. The report of kinematics followed the right-hand rule. Positive angles for the X axis were flexion at the hip, extension at the knee, and dorsiflexion at the ankle. Positive angles for the Y axis were adduction at the hip and knee, and inversion at the ankle. Positive angles for the Z axis were internal rotation at the hip and knee, and adduction at the ankle. The statistical analysis of discrete variables followed that of Chapter 4 and waveform variables followed that of Chapter 5.

A.3 Results

Load carriage significantly increased hip positive work and its effect was greater at faster velocities (Table 9-1). For example, a 20 % body weight (BW) load increased hip positive work by 0.01 units from BW running at 3 m/s, but by 0.02 units at 5 m/s. Load carriage reduced hip and knee negative work and the reductions were smaller at faster velocities (Table 9-1). For example, a 20 % BW load reduced hip negative work by 0.004 units from BW running at 3 m/s, but by 0.001 units at 5 m/s. Load carriage increased early swing hip power generation, and this was greater as velocity increased (Figure 9-2, Figure 9-3). Load carriage increased early swing knee power absorption, and this was greater as velocity increased (Figure 9-2, Figure 9-3).

For swing phase kinematics, load carriage increased hip flexion angle during early to mid-swing, but reduced knee flexion angle in mid-swing (Figure 9-4, Figure 9-5). Load carriage increased terminal swing knee flexion angle. However, similar knee flexion angle was achieved just prior to initial contact with and without load (Figure 9-5).

A.4 Discussion

In BW running, early-swing phase hip joint power and work has been shown to contribute to increased running velocity, particularly at velocities greater than 5 m/s (Schache et al., 2015). Our results concurred with that finding, seeing that increasing velocity had a significant effect on hip positive work. The effect of load carriage on swing phase hip positive work is very much smaller than stance phase mechanical work. For example, carrying a 20 % BW load increased stance phase ankle positive work by 0.3 units (Table 4-1), which is very much larger than the 0.01 units increase in swing phase hip positive work. The extra mechanical work needed to sustain constant velocity in load carriage running is thus driven largely by stance phase mechanical work, at least up to a running velocity of 5 m/s.

In BW running, an increased thigh segment angle at mid-swing was correlated to a reduced impact peak of the vertical ground reaction force (Schmitz et al., 2014). This could mean that the observed hip flexion angle at mid-stance with load was an adaptive mechanism in reducing impact force during loading response of running. However, a limitation of the analysis was that we measured hip angle and not segment angle. An increased hip flexion angle can be driven proximally by an increased trunk-pelvic segment angle or distally by increased thigh segment flexion angle.

A consistent phenomenon in load carriage is an increase in trunk segment flexion angle during walking and running when carrying a backpack (Brown et al., 2014; Liew et al., 2016). It is very likely that a portion of the increased hip flexion angle with load was driven by proximal mechanics. However, distal thigh mechanics cannot be excluded. A previous study on soldiers performing a vertical step obstacle clearance task was the finding that minimum foot clearance increased with load carriage (Brown et al., 2016). To increase minimum foot clearance, one strategy is to increase thigh segment flexion angle. Thus, it is still possible that an increase in thigh segment flexion may contribute to the increased hip flexion angle with load. Even though an increased swing phase hip flexion angle could reduce the impact force during loading response of load carriage running, there has been no cross-sectional or prospective evidence to link swing phase mechanics to overuse BW running injury risk. This short commentary justifies the appropriateness of our exercise selection in Chapter 6 to target stance phase muscular strength and power variables.

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A.6 Tables

Table 9-1 Effect of speed and load on running swing phase joint work

	Intercept	Speed (m/s)	Load (% body weight)	Speed:Load
Dependent variables	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)	β (95 % CI: lower – upper)
Hip negative work	2.59×10^{-02} (2.18×10^{-02} to 2.99×10^{-02})	$-1.32 \times 10^{-02**}$ (-1.48×10^{-02} to -1.16×10^{-02})	4.03×10^{-04} (1.70×10^{-04} to 6.36×10^{-04})	-6.75×10^{-05} (-1.34×10^{-04} to -1.43×10^{-06})
t-value	12.40	-16.30	3.39	-2.00
Hip positive work	-4.69×10^{-02} (-6.23×10^{-02} to -3.14×10^{-02})	$3.05 \times 10^{-02**}$ (2.46×10^{-02} to 3.65×10^{-02})	$-9.11 \times 10^{-04**}$ (-1.60×10^{-03} to -2.21×10^{-04})	$5.02 \times 10^{-04**}$ (2.96×10^{-04} to 7.08×10^{-04})
t-value	-5.96	10.10	-2.59	4.77
Knee negative work	3.08×10^{-02} (2.17×10^{-02} to 4.00×10^{-02})	$-2.79 \times 10^{-02**}$ (-3.16×10^{-02} to -2.43×10^{-02})	$5.20 \times 10^{-04**}$ (9.27×10^{-05} to 9.47×10^{-04})	$-1.53 \times 10^{-04**}$ (-2.79×10^{-04} to -2.77×10^{-05})
t-value	6.59	-15.10	2.39	-2.39
Knee positive work	-9.20×10^{-04} (-2.65×10^{-03} to 8.11×10^{-04})	$2.48 \times 10^{-03**}$ (1.98×10^{-03} to 2.98×10^{-03})	-2.64×10^{-05} (-1.23×10^{-04} to 6.98×10^{-05})	$-3.41 \times 10^{-05**}$ (-6.05×10^{-05} to -7.65×10^{-06})

t-value	-1.04	9.70	-5.39×10^{-01}	-2.53
Ankle negative work	1.11×10^{-04} (-3.64×10^{-05} to 2.58×10^{-04})	-1.91×10^{-04}** (-2.42×10^{-04} to -1.41×10^{-04})	2.83×10^{-06} (-4.59×10^{-06} to 1.03×10^{-05})	-4.69×10^{-07} (-2.56×10^{-06} to 1.62×10^{-06})
t-value	1.48	-7.45	7.48×10^{-01}	-4.40×10^{-01}
Ankle positive work	-3.88×10^{-04} (-7.73×10^{-04} to -2.13×10^{-06})	6.94×10^{-04}** (5.58×10^{-04} to 8.31×10^{-04})	-5.45×10^{-06} (-1.87×10^{-05} to 7.79×10^{-06})	4.30×10^{-06}** (4.94×10^{-07} to 8.11×10^{-06})
t-value	-1.97	9.98	-8.08×10^{-01}	2.22
Abbreviations: β = Beta coefficient; CI – confidence interval; **P < 0.05				
Example: Hip positive work at 3.0 m/s carrying 20 % BW equates to: $-4.69 \times 10^{-02} - 9.11 \times 10^{-04} (20) + 3.05 \times 10^{-02} (3) + 5.02 \times 10^{-04} (20)(3) = 0.06$ units.				

A.7 Figures

Figure 1. Joint power

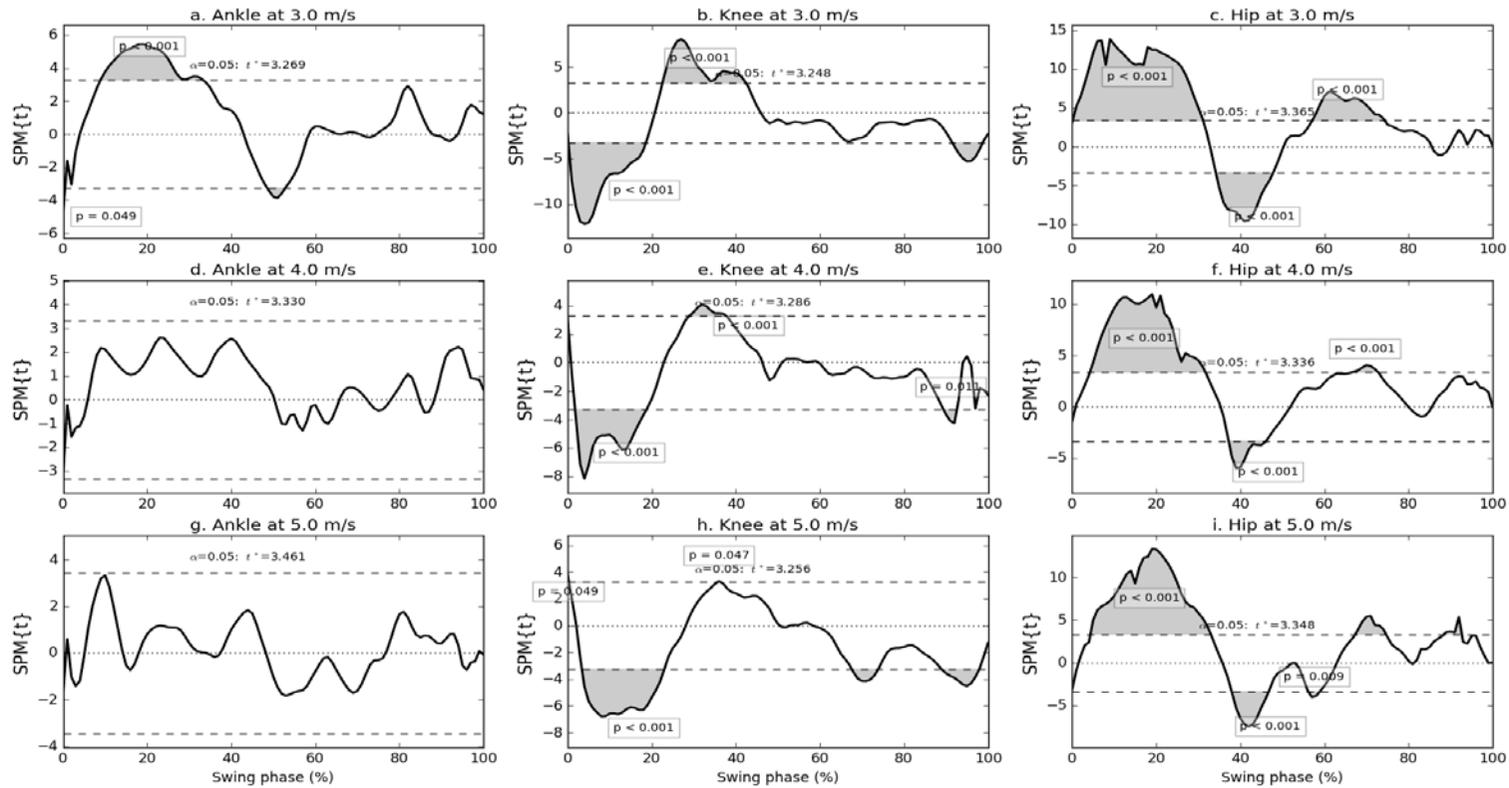


Figure 9-2 Swing phase joint power SPM results

Figure 2. Joint power

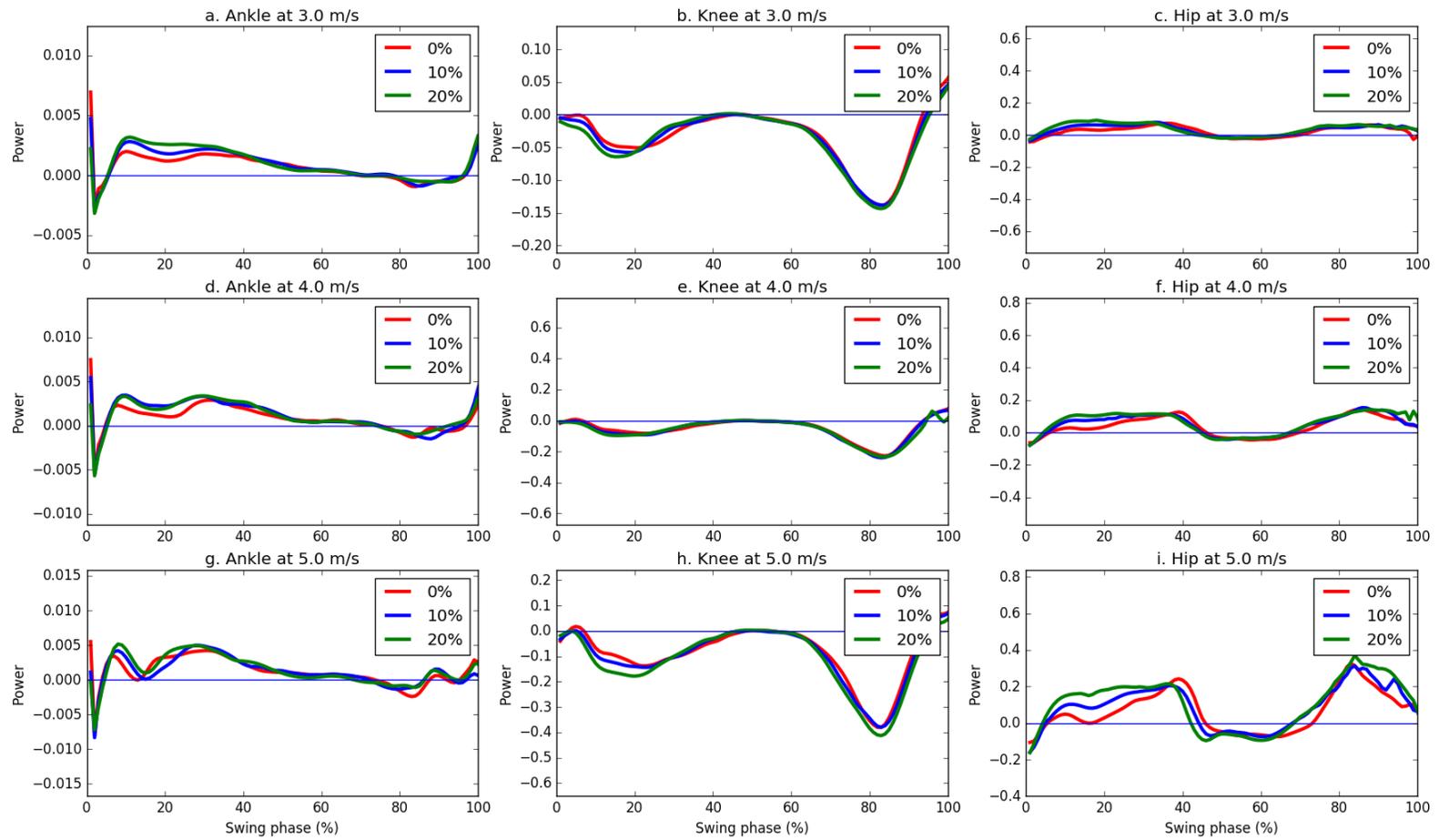


Figure 9-3 Swing phase joint power mean waveforms

Figure 3. Joint angle (medio-lateral axis)

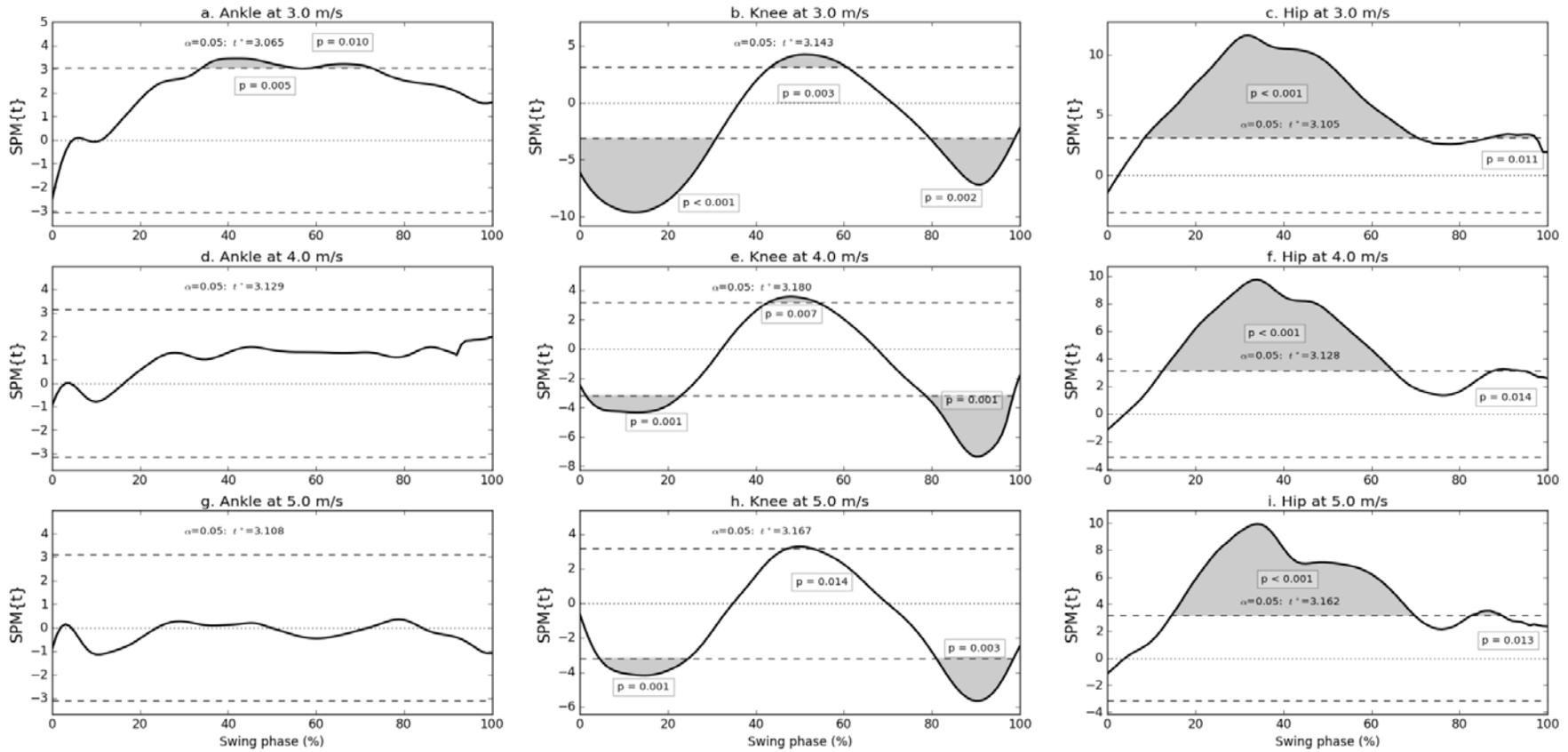


Figure 9-4 Swing phase joint angle (X axis) SPM results

Figure 4. Joint angle (medio-lateral axis)

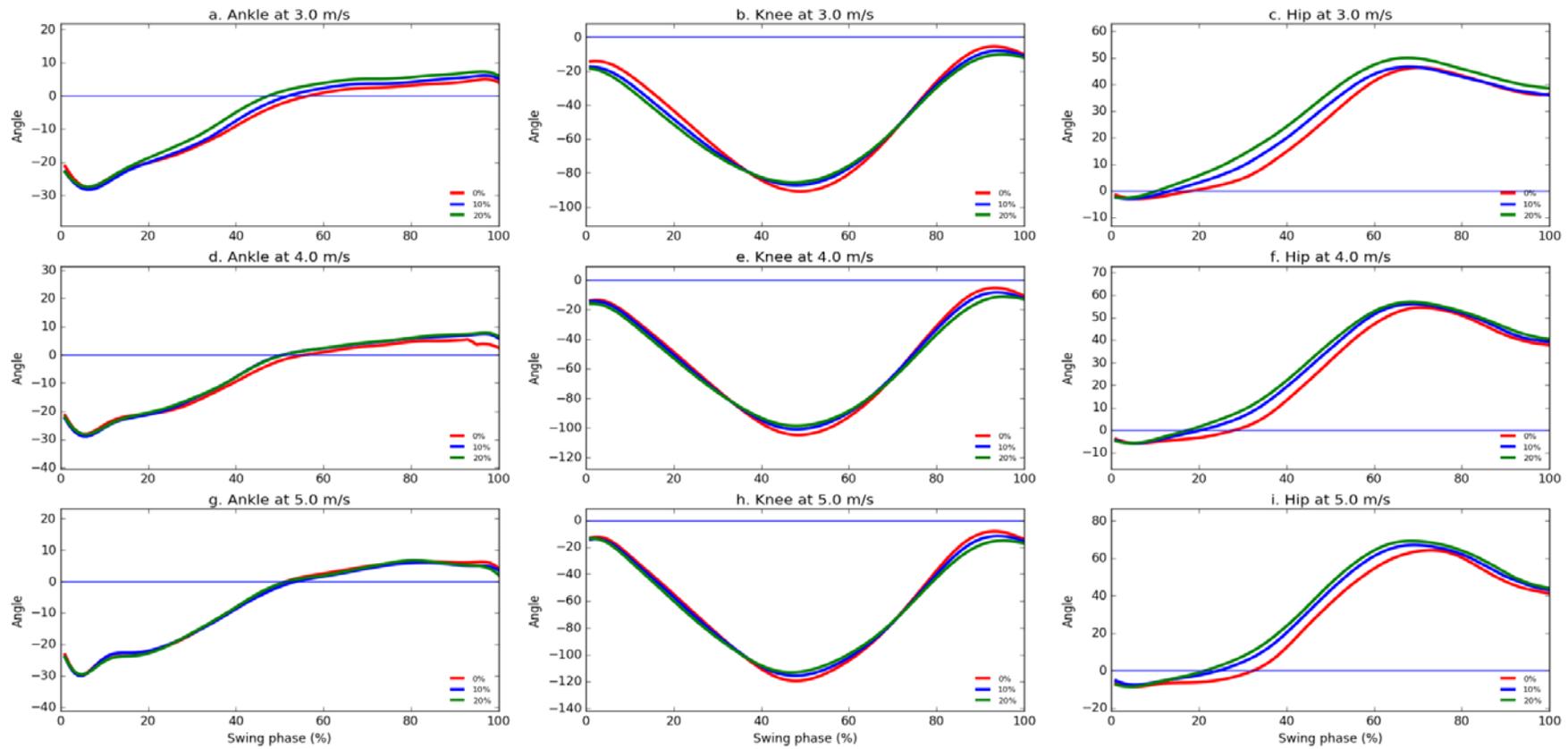


Figure 9-5 Swing phase joint angle (X axis) mean waveforms

Figure 5. Joint angle (Anterior-posterior axis)

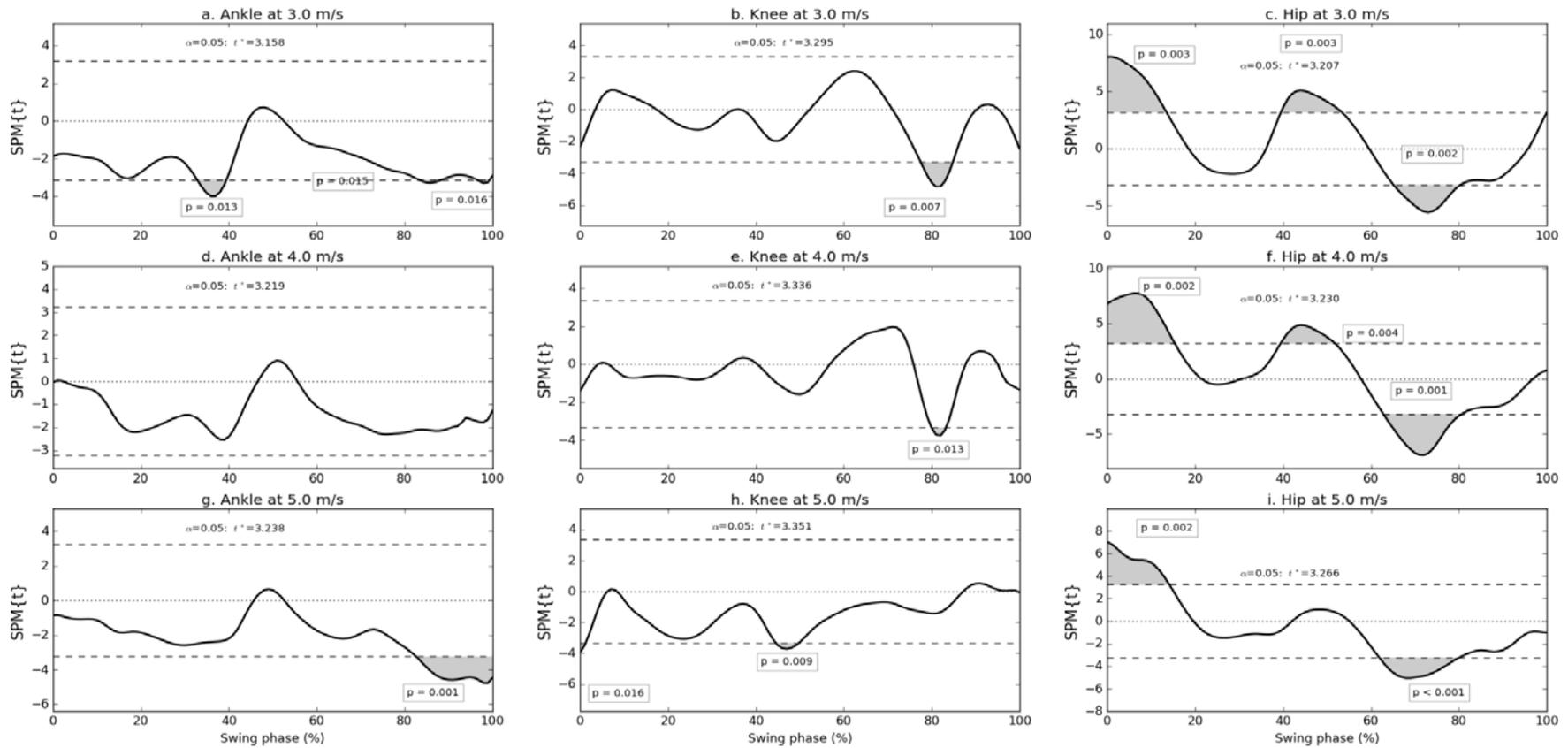


Figure 9-6 Swing phase joint angle (Y axis) SPM results

Figure 6. Joint angle (Anterior-posterior axis)

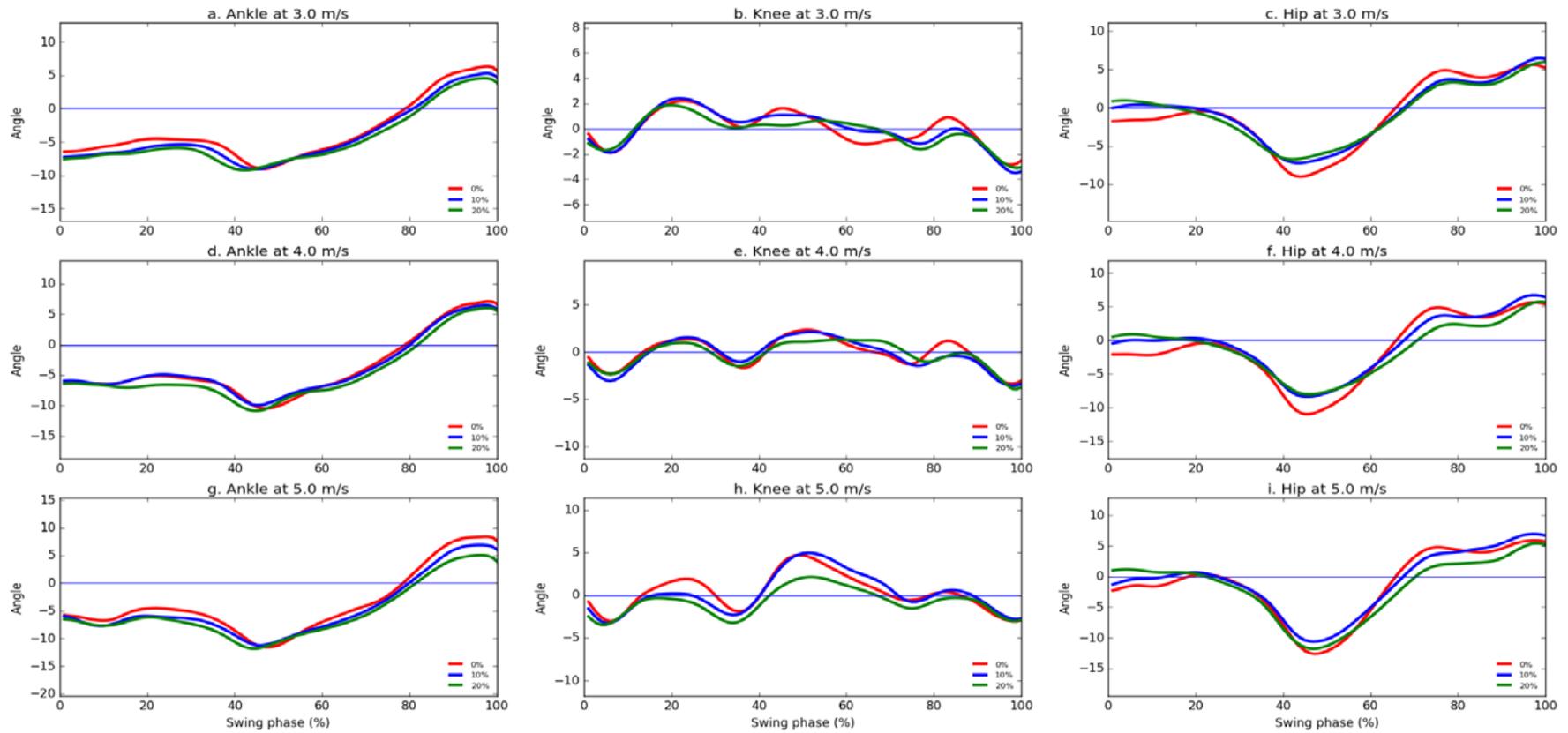


Figure 9-7 Swing phase joint angle (Y axis) mean waveforms

Figure 7. Joint angle (Vertical axis)

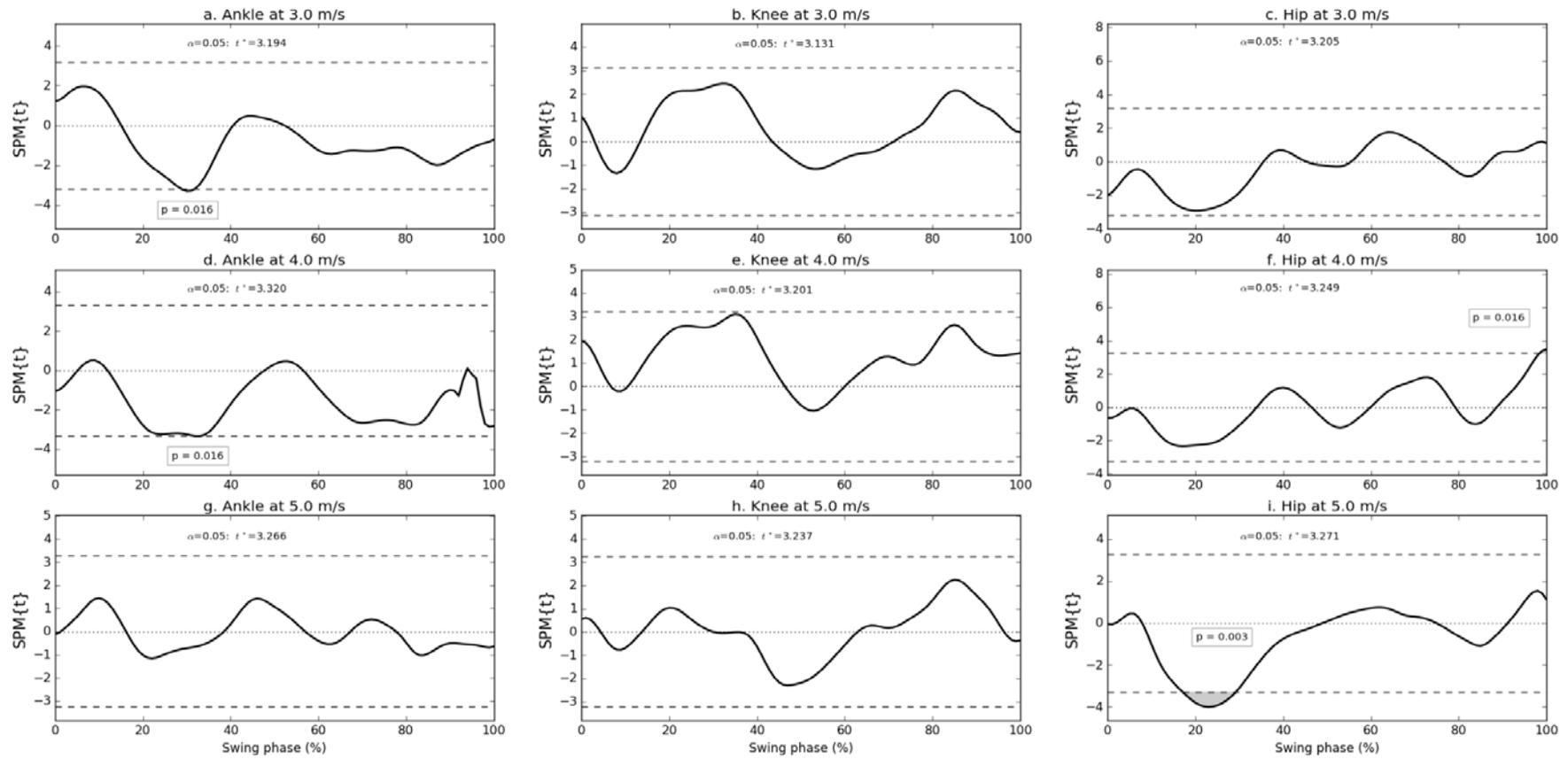


Figure 9-8 Swing phase joint angle (Z axis) SPM results

Figure 8. Joint angle (Vertical axis)

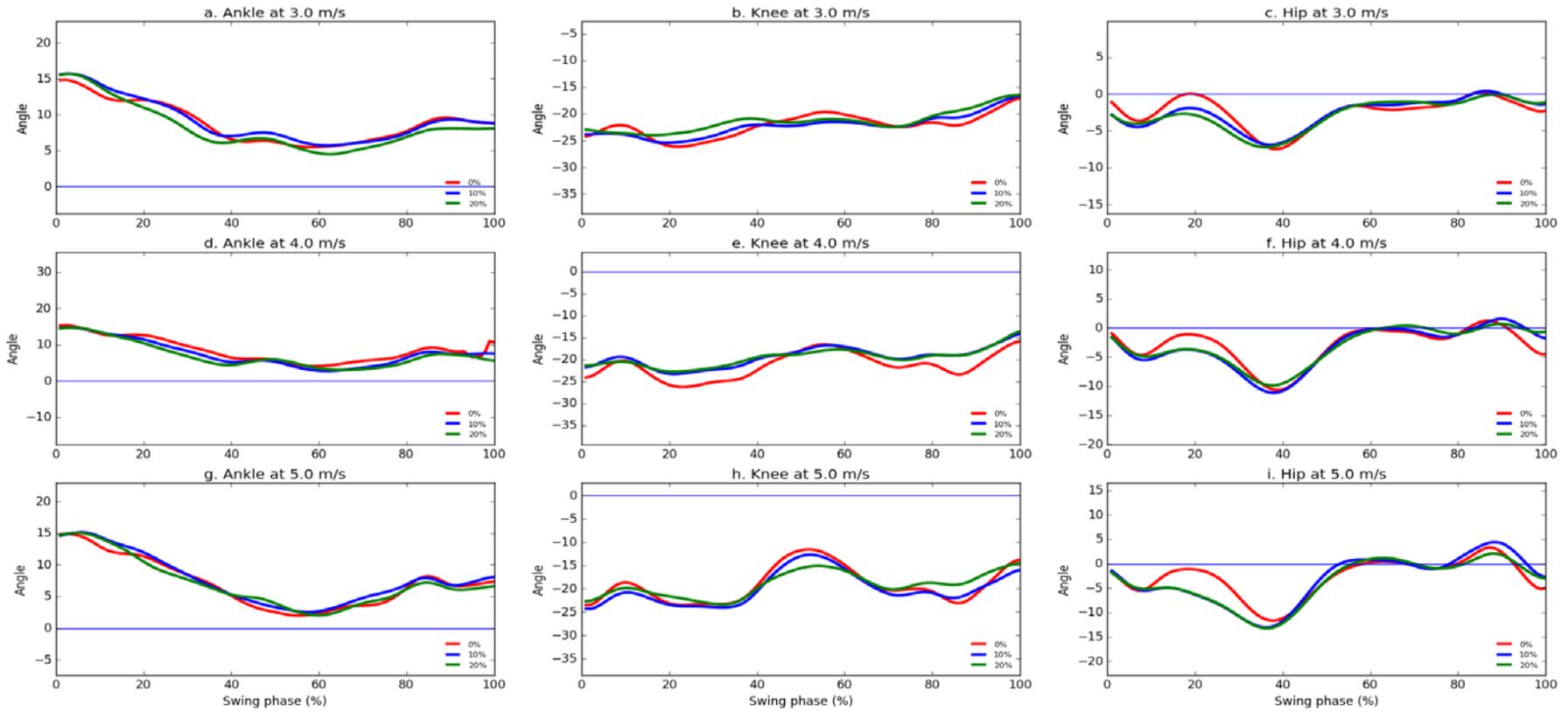


Figure 9-9 Swing phase joint angle (Z axis) mean waveforms

Appendix B

Statement of contribution from co-authors



To Whom It May Concern

I, Liew Xian Wei Bernard, am the primary contributor to the conception, design, execution, processing, analysis, and manuscript writing of the following publications:

1. Performance of a lateral pelvic cluster technical system in evaluating running kinematics. *J Biomech.* 2016 Jun 14;49(9):1989-93. doi: 10.1016/j.jbiomech.2016.05.010.
2. The effects of load carriage on joint work at different running velocities. *J Biomech.* 2016 Oct 3; 49(14):3275-3280. doi: 10.1016/j.jbiomech.2016.08.012
3. Joint power and kinematics coordination in load carriage running: Implications for performance and injury. *Gait Posture.* 2016 Jun;47:74-9. doi: 10.1016/j.gaitpost.2016.04.014.
4. Effects of two neuromuscular training programs on running biomechanics with load carriage: a study protocol for a randomised controlled trial. *BMC Musculoskelet Disord.* 2016 Oct 22;17(1):445.
5. Resistance training on load carriage mechanical work (Prepared for submission)
6. Loaded running kinematics after resistance training (Prepared for submission).

Bernard Liew

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To Whom It May Concern

I, Liew Xian Wei Bernard, contributed to the conception, design, execution, processing, analysis, and manuscript writing to the publication entitled "Performance of a lateral pelvic cluster technical system in evaluating running kinematics. J Biomech. 2016 Jun 14;49(9):1989-93. doi: 10.1016/j.jbiomech.2016.05.010."

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To Whom It May Concern

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To Whom It May Concern

I, Liew Xian Wei Bernard, am the primary contributor to the conception, design, execution, processing, analysis, and manuscript writing of the publication entitled "Loaded running kinematics after resistance training (Prepared for submission)."



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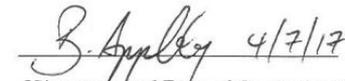


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I, Liew Xian Wei Bernard, am the primary contributor to the conception, design, execution, processing, analysis, and manuscript writing of the publication entitled "Loaded running kinematics after resistance training (Prepared for submission)."



Bernard Liew

I, as Co-Author, endorse that this level of contribution by the candidate indicated above is appropriate.

Justin Keogh

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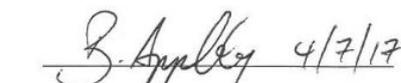


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 - b. There is no mention within the correspondence regarding your ability to publish your thesis in an institutional repository. However referring to SHERPA/RoMEO below it is permissible to post your final manuscript "On the author's ... institutional repository".Provided your comply with the Publisher's requirement to include the final manuscript, as accepted for publication version and not the final published article on espace, further permission is not necessary.

The screenshot shows the SHERPA/RoMEO website interface. At the top, there is a green header with the SHERPA/RoMEO logo and navigation links for 'Home' and 'Search'. Below the header, the main heading is 'Search - Publisher copyright policies & self-archiving'. A search result is displayed for the 'Journal of Applied Biomechanics' (ISSN: 1065-8483, EISSN: 1543-2688). The result is categorized as a 'RoMEO blue journal' and indicates that a 'Paid OA' option is available. The 'Author's Pre-print' status is 'archiving status unclear', while the 'Author's Post-print' status is 'author can archive post-print (ie final draft post-refereeing)'. The 'Publisher's Version/PDF' status is 'author cannot archive publisher's version/PDF'. The 'General Conditions' section lists several restrictions: the article cannot be on a personal website or institutional repository, the publisher's version/PDF cannot be used, publisher copyright and source must be acknowledged, a DOI link to the publisher version is required, a statement to accompany the deposit 'as accepted for publication' is needed, NIH authors may deposit in PubMed Central, and the publisher retains copyright. The 'Mandated OA' status is 'Awaiting information'. The 'Paid Open Access' status is 'Open Access'. The 'Notes' section indicates the publisher was last reviewed on 02/02/2017 and last contacted on 05/12/2013. The 'Copyright' status is 'Policy'. The 'Updated' date is 02-Feb-2017, with a link to 'Suggest an update for this record'. The 'Link to this page' is <http://sherpa.ac.uk/romeo/issn/1065-8483/>. The result is published by 'Human Kinetics - Blue Policies in RoMEO'.

3. BioMed Central
(I am assuming that your article was not published in one of the following journals: Alzheimer's Research & Therapy, Arthritis Research & Therapy, Breast Cancer Research, Critical Care, Genome Biology, Genome Medicine, Stem Cell Research & Therapy).
 - a. You retain the right to reuse all or part of the article without any restriction, (subject only to proper attribution) as your journal has been published under the Creative Commons Attribution (CC-BY) license.
 - b. Similarly you retain the right to include the article in your thesis when depositing into an institutional repository.Provided you comply with the stipulated conditions <https://creativecommons.org/licenses/by/3.0/us/>, no further permissions are required.

I hope this has clarified your queries in relation to copyright permissions. I would recommend you attach copies of your correspondence with regard to copyright permissions as an appendix to your thesis.

Please don't hesitate to contact me if you require any further information.

Kind regards

Sian

Sian Dewar
BA GradDipInfoLibStds(Curtin) AALIA
Copyright Support Officer | University Library
Please note my office hours are Tuesday, Wednesday and Thursdays.

Curtin University
Tel | +61 8 9266 3852
Fax | +61 8 9266 3213

Email | sian-louise.dewar@curtin.edu.au
Web | <http://curtin.edu.au>
<http://copyright.curtin.edu.au>



CRIICOS F Enslin Cook 00011

From: Xian Wei Bernard Liew [<mailto:liew@postgrad.curtin.edu.au>]
Sent: Thursday, 2 March 2017 7:42 AM
To: Amanda Bellenger <A.Bellenger@curtin.edu.au>
Cc: Kevin Netto <kevin.netto@curtin.edu.au>; Susan Morris <S.Morris@curtin.edu.au>
Subject: Permission to reproduce articles as thesis chapters

Dear Amanda,

I am Bernard a PhD student who is nearing submission. I have previously asked you on how to get permission to reproduce some articles I have published as chapters in my thesis. I know of the template and whilst I have submitted to the relevant bodies, every publisher I have asked mentioned that as an author, I can reproduce the articles as a thesis chapter, with no need for any signatures. How should I proceed? Can I put this email as a word document in my thesis appendix for justification?

PS: I have cc'ed my supervisors into this mail so they can have a record of this

Regards,
Bernard

From: Amanda Bellenger [<mailto:A.Bellenger@curtin.edu.au>]
Sent: Thursday, 1 December 2016 8:44 AM
To: Xian Wei Bernard Liew
Subject: Link to Copyright Permission template

Hi Bernard,

Information for HDR students about copyright is available from <http://copyright.curtin.edu.au/research/higher-degree-theses/>

The permission template in Word format is at http://copyright.curtin.edu.au/wp-content/uploads/2016/10/HDR_Students_Permission_Request1.docx and in pdf format at http://copyright.curtin.edu.au/wp-content/uploads/2016/09/student_permission_form_rev_141205.pdf

If you can please reply back with the link that isn't working, I will make sure the linking is fixed ASAP.

Regards,
Amanda.

Amanda Bellenger
BA GradDipEd MAppSc
Manager | Strategy, Planning & Copyright
University Library

Curtin University
Tel | +61 8 9266 7494
Fax | +61 8 9266 3213

Email | a.bellenger@curtin.edu.au
Web | <http://curtin.edu.au>



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CRIICOS F Enslin Cook 00011

C.2 Correspondence with Elsevier publisher

From: [Shridhar, Lakshmi Priya \(ELS-CHN\)](#)
To: [Xian Wei Bernard Liew](#)
Subject: RE: Seeking copyright permissions for articles for PhD thesis
Date: Wednesday, 1 March 2017 7:53:40 PM
Attachments: [CopyrightPermission\[Elsevier\].docx](#)



ELSEVIER

Dear Bernard Liew,

Thank you for your email.

Please note that as one of the Authors of this article, you retain the right to include the journal article, in full or in part, in a thesis or dissertation. You do not require permission to do so.

For full details of your rights as a Journal Author, please visit:

<http://www.elsevier.com/wps/find/authorsview.authors/copyright#whattrights>

Your retained rights allow you to submit your article in electronic format and to post this Elsevier article online if it is embedded within your thesis. You are also permitted to post your Author Accepted Manuscript online; however posting of the final published article is prohibited.

Please refer to Elsevier's Posting Policy for further information:

<http://www.elsevier.com/wps/find/authors.authors/postingpolicy>

Feel free to contact me if you have any queries.

Regards,

Priya

Lakshmi Priya
Sr Copyrights Coordinator
Global Rights Department

ELSEVIER

International Tech Park | Crest - 12th Floor | Taramani Road | Taramani | Chennai 600 113 | India
Tel: +91 44 42994660 | Fax: +91 44 42994701
E-mail: L.shridhar@elsevier.com | url: www.elsevier.com

From: Xian Wei Bernard Liew [<mailto:b.liew@postgrad.curtin.edu.au>]
Sent: Thursday, February 23, 2017 9:51 AM
To: Rights and Permissions (ELS)
Subject: FW: Seeking copyright permissions for articles for PhD thesis

Dear Sir/Mdm,

I like to follow up on my prior request to reproduce the pre-proof version of my manuscripts in my PhD thesis chapters.

Bernard

From: Xian Wei Bernard Liew
Sent: Friday, 3 February 2017 4:42 PM
To: permissions@elsevier.com

C.3 Correspondence with Biomedical Central publisher

From: "Jess Ramos Jr" <jesus.ramosjr@springernature.com>
To: Xian Wei Bernard Liew
Subject: 00663798 RE: Request for copyright to use article for PhD thesis
Date: Friday, 3 February 2017 4:55:58 PM



Dear Dr Liew,

Thank you for contacting BioMed Central.

The open access articles published in BioMed Central's journals are made available under the Creative Commons Attribution (CC-BY) license, which means they are accessible online without any restrictions and can be re-used in any way, subject only to proper attribution (which, in an academic context, usually means citation).

The re-use rights enshrined in our license agreement (<http://www.biomedcentral.com/about/policies/license-agreement>) include the right for anyone to produce printed copies themselves, without formal permission or payment of permission fees. As a courtesy, however, anyone wishing to reproduce large quantities of an open access article (250+) should inform the copyright holder and we suggest a contribution in support of open access publication (see suggested contributions at <http://www.biomedcentral.com/about/policies/reprints-and-permissions/suggested-contributions>).

Please note that the following journals have published a small number of articles that, while freely accessible, are not open access as outlined above: Alzheimer's Research & Therapy, Arthritis Research & Therapy, Breast Cancer Research, Critical Care, Genome Biology, Genome Medicine, Stem Cell Research & Therapy.

You will be able to find details about these articles at <http://www.biomedcentral.com/about/policies/reprints-and-permissions>

If you have any questions, please do not hesitate to contact me.

Best wishes,

Jess Ramos Jr.
Global Open Research Support Executive
Open Research Group

Springer Nature
236 Gray's Inn Road, London WC1X 8HB, UK
T +44 (0)203 192 2009
F +44 (0)203 192 2010
Jess.RamosJr@springernature.com
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No. 1738860 / VAT Registration No. GB 823826326

-----Your Question/Comment -----

From: b.liew@postgrad.curtin.edu.au

To: info@springeropen.com
Dear Person in charge,

I am writing in request to gain copyright permissions for the above articles, to be included as chapters in my thesis. Could you please assist?

Kind regards,

Appendix D Study 1

D.1 Ethics and amendment



Memorandum

To	Dr Kevin Netto, Physiotherapy and Exercise Science Mr Bernard Liew, Physiotherapy and Exercise Science Dr Susan Morris, Physiotherapy and Exercise Science
From	Professor Peter O'Leary, Chair Human Research Ethics Committee
Subject	Protocol Approval RD-41-14
Date	19 November 2014

Office of Research and Development
Human Research Ethics Committee

TELEPHONE 9266 2784
FACSIMILE 9266 3793
EMAIL hrec@curtin.edu.au

Thank you for your "Form C Application for Approval of Research with Low Risk (Ethical Requirements)" for the project titled "*The biomechanics of running with load*". On behalf of the Human Research Ethics Committee I am authorised to inform you that the project is approved.

Approval of this project is for a period of four years **20-11-14 to 20-11-18**.

The approval number for your project is **RD-41-14**. Please quote this number in any future correspondence.

Your approval has the following conditions:

- i) Annual progress reports on the project must be submitted to the Ethics Office.
- ii) **It is your responsibility, as the researcher, to meet the conditions outlined above and to retain the necessary records demonstrating that these have been completed. See: Western Australian University Sector Disposal Authority (WAUSDA).**

Applicants should note the following:

It is the policy of the HREC to conduct random audits on a percentage of approved projects. These audits may be conducted at any time after the project starts. In cases where the HREC considers that there may be a risk of adverse events, or where participants may be especially vulnerable, the HREC may request the chief investigator to provide an outcomes report, including information on follow-up of participants.

The attached **Progress Report** should be completed and returned to the Secretary, HREC, C/- Office of Research & Development annually.

Our website https://research.curtin.edu.au/guides/ethics/low_risk_hrec_forms.cfm contains all other relevant forms including:

- Completion Report (to be completed when a project has ceased)
- Amendment Request (to be completed at any time changes/amendments occur)
- Adverse Event Notification Form (If a serious or unexpected adverse event occurs)
- Western Australian University Sector Disposal Authority (WAUSDA)

Yours sincerely

Professor Peter O'Leary
Chair Human Research Ethics Committee

Please Note: The following standard statement must be included in the information sheet to participants:
This study has been approved under Curtin University's process for low-risk Studies (Approval Number RD-41-14). This process complies with the National Statement on Ethical Conduct in Human Research (paragraph 5.1.7 and paragraphs 5.1.18-5.1.21). For further information on this study contact the researchers named above or the Curtin University Human Research Ethics Committee. c/- Office of Research and Development, Curtin University, GPO Box U1987, Perth 6845 or by telephoning 9266 9223 or by emailing hrec@curtin.edu.au.

D.2 Patient information sheet

Project Title: The biomechanics of running with loads.

Principal Supervisor: Dr Kevin Netto

Co-supervisor: Dr Susan Morris

Principal researcher: Bernard Liew (PhD candidate)

Why is this study important?

Carrying a backpack while running is a common activity involved in various occupations, such as the military. It is also a common activity required in certain sports. However, little is known behind how carrying a influence running behaviour, and if certain individuals are more tolerant than others in carrying heavy loads during running. This study will help us understand how an individual's physical condition influences load carriage running ability.

How is this study being done?

If you agree to be a participant of this study, we will arrange a convenient time for you to come to the Curtin University motion analysis laboratory. A questionnaire will be given to ensure you are suitable to participate in this study, as well as collect basic information of age, weight, height, and sporting activity level. If you are suitable for participation, reflective markers and adhesive electrodes will be attached to the skin of your hips and legs. These markers and electrodes will be used to collect information of body movements, and muscle activity. You will be required to perform repeated running trials within the laboratory at three speeds and carrying four different loads in the backpack. Speeds selected will be "slow", "intermediate", and "moderate". Loads carried will be set at 0 %, 10 %, 20 %, and 30 % of your body weight. You will run an approximate total of 100 trials of a 20 metre walkway. Each trial is interspersed with a minimum rest period of 30s. A minimum rest period of 5 min will be given after every condition. You should expect to spend a total of 3 hours in the laboratory.

What are the risks of the research?

There is minimal risk in this study as the load carried is tailored to your body weight, and adequate rest is provided. You may experience muscle soreness in your shoulders and legs, appearing within 24-48hrs after the experiment. This is a normal sign of muscle fatigue and will be expected to recover within 2-3days from onset. Another possible risk associated with the study is an allergic reaction to the tape used to apply the reflective markers. To keep this risk minimal, we will be using low allergy tape and if a reaction does occur, the tape will be removed immediately.

What are the benefits of the research?

The information gained will help us understand if load carriage running ability can be predicted. Understanding the predictors of load carriage running ability would provide a platform in designing an exercise program to improve an individual's running capacity when carrying a load. By extension, this would enable us to improve sporting performance and occupational performance during running.

What will happen to the information you provide?

Any information that you give will remain completely confidential. During the analysis of the results you will be given a number to ensure complete anonymity. Your screening questionnaire and any results from the study will be kept for a period of 7 years with the project supervisors in a secure place. After this period, all records will be destroyed. This is a requirement of Curtin University.

Upon your agreement, all data collected will be used for future research in this field.

You may refuse to participate in the study. If you change your mind once you have agreed to participate, you are free to withdraw at any time and without fear of prejudice. If you decide to withdraw please contact the principal researcher (Bernard Liew) at the earliest opportunity. In the event of this occurrence, all the data related to you will be destroyed.

The project researchers are Mr. Bernard Liew (0435223717; b.liew@postgrad.curtin.edu.au), Dr. Kevin Netto (9266 3689), and Dr. Susan Morris (9266 4644). If you wish to make a complaint on ethical grounds, then you can contact the Secretary of the Human Research Ethics Committee (phone: 9266 2784 or hrec@curtin.edu.au).

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number). The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its key role is to protect participants. If needed, verification of approval can be obtained either by writing to Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University, GPO Box U1987, Perth WA 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au.

Thank you for considering participation in this study.

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number RD-41-14).

D.3 Informed consent

Project Title: The biomechanics of running with load

PLEASE NOTE THAT PARTICIPATION IN RESEARCH STUDIES IS VOLUNTARY AND YOU CAN CHOOSE TO WITHDRAW AT ANY TIME.

- I understand the purpose and procedures of the study.
- I have been provided with the participant information sheet.
- I understand that the procedure itself may not have any benefits.
- I understand that no personal identifying information like names and addresses will be used and that all information will be securely stored for 7 years before being destroyed.
- I understand that the data collected will be used for future research.
- I have been given the opportunity to ask questions.
- Any questions I have asked, have been answered to my satisfaction.

I

(Given name of participant)

(Surname)

of

(Address)

(Postcode)

.....

(Home phone)

(Mobile)

(Email)

have read the information provided and agree to participate in this study.

Participant's Signature

Date

I have explained to the participant the procedures of the study to which they have consented their involvement and have answered all questions. In my appraisal, the participant has voluntarily and intentionally given informed consent and possesses the legal capacity to give informed consent to participate in this research study.

Principal Investigator: _____ Date: _____

This study has been approved by the Curtin University Human Research Ethics Committee
(Approval Number: RD-41-14)

D.4 Video informed consent

Project Title: The biomechanics of running with loads

Principal Supervisor: Dr Kevin Netto

Co-supervisor: Dr Susan Morris

Principal researcher: Bernard Liew (PhD candidate)

The investigator has explained to me that visual images (video and still images) are important to verify marker position in the gait laboratory and can be useful to illustrate aspects of the study in reports describing the research. These images may also be converted to electronic formats for use in multimedia presentations and documents accessible to others by computer for teaching purposes and the purpose of sharing results of the study. By giving consent I authorize the researchers to use any of the visual images taken of me in printed format, in slides for presentations and in electronic format.

I, (Name).....

consent to allow visual images of me to be

used for the following:

Verification purposes to assess the data collected

Research purposes for use in conferences and journals

Educational purposes

Signature:

Date:

I have explained to the participant how the visual images will be used, and have answered all questions.

Investigator Signature:

Date:

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number RD-41-14).

D.5 Gait collection sheet

Baseline assessment form

Participant's name: _____

Participant's identification number: _____

Date of testing: _____

Height (m): _____

Weight (kg): _____

Age: _____

Sex: Male/Female

Volume of running, jumping, hopping-based sports per week (hours): _____

Task	Trial no./name	Description	Processed
Countermovement jump			
Countermovement jump			
Countermovement jump			
Squat Jump			
Squat Jump			
Squat Jump			
Hopping			

D.6 Advertisement

HEALTHY VOLUNTEERS NEEDED FOR RUNNING BIOMECHANICS PHD STUDY



Carrying a load during running is common in sports, athletic training, military, and even in people running to work/school.

We often overlook the impact load carrying has on running behavior. Depending on various factors, it may be beneficial or detrimental.

We are looking to determine under what conditions do load carrying become detrimental to running behavior.

Participation includes jump assessments, running across a motion analysis lab with backpack loads of up to 30% of your body weight, at slow to moderate speeds.

Contact Bernard (PhD student):

Mobile: 0435223717

Email: b.liew@postgrad.curtin.au



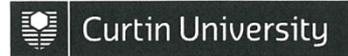
Curtin University

SCHOOL OF PHYSIOTHERAPY AND EXERCISE SCIENCE,
CURTIN UNIVERSITY
Kent St, Bentley WA 6102

Appendix E Study 2

E.1 Ethics and amendment

MEMORANDUM



To:	Dr Kevin Netto Physiotherapy and Exercise Science
CC:	Mr Bernard Liew
From:	Dr Catherine Gangell, Manager Research Integrity
Subject	Amendment approval Approval number: RD-41-14
Date:	29-Jul-15

Office of Research and
Development
Human Research Ethics Office
TELEPHONE 9266 2784
FACSIMILE 9266 3793
EMAIL hrec@curtin.edu.au

Thank you for submitting an amendment to the Human Research Ethics Office for the project:

RD-41-14	The biomechanics of running with load
----------	---------------------------------------

The Human Research Ethics Office approves the amendment to the project.

Amendment number: RD-41-14/AR03

Approval date: 29-Jul-15

The following amendments were approved:

Various changes to the protocol to facilitate the '2nd Study' as outlined in the original project.

Please ensure that all data are stored in accordance with WAUSDA and Curtin University Policy.

Yours sincerely

Dr Catherine Gangell
Manager, Research Integrity

E.2 Patient information sheet

Instructions

1. Read-only pages 1-4 (Participant information sheet)
2. Read-only pages 5-6 (Do not need to fill up yet)
3. FILL UP pages 7 – 15 (Medical, training, injuries)

PARTICIPANT INFORMATION SHEET (For your keep)

Project Title: The effect of two different resistance training programs on the biomechanics of running with load: a randomized controlled trial

Principal Supervisor: Dr Kevin Netto

Co-supervisor: Dr Susan Morris

Principal researcher: Bernard Liew (PhD candidate)

Invitation to participate:

You are being invited to participate in a research study. Before you decide, it is important for you to understand the background of this study, and what it entails. This short information sheet will help you decide whether you would like to join. Please take your time to read the information carefully, and discuss with your friends and relatives. Please ask us if there is anything not presented clearly or if you would like more details.

Why is this study important?

Carrying a backpack while running is a common activity involved in running-related sports and occupation. Our previous study identified that carrying a moderately heavy load increases leg muscle work to maintain constant running speed. We also identified several important muscle groups that are important in load carriage running. It is hypothesised that an exercise program targeted to these muscle groups could modify an individual's running speed and running pattern. However, it is unknown what specific exercise parameters are best suited to positively influence load carriage running. The aim of this intervention is to compare two exercise programs (6 weeks plus 2 weeks pre-conditioning) on its purported effects on self-determined running speed and biomechanics.

Can anyone participate?

We are looking for a broad group of novice and experienced runners residing in Western Australia. If you are between 18 to 60 years of age, in good general systemic health, and doing some form of running or running-related sports (e.g. football), for a total of at least one hour per week, you are suitable to join. The following criteria preclude us from accepting you as participants to ensure your safety:

1. Presence of any disorders that could affect your running and load carrying ability.
2. Medical conditions that preclude heavy resistance training and sprinting.
3. Suffered a running related injury or pain within the last three months, which resulted in complete absence of all planned running sessions for at least a week.
4. Current pain that affects your running performance or exercise performance.
5. Any lower limb surgery within the past 12 months.
6. Females who are pregnant.

Do I have to take part?

The decision to participate is made entirely by you, it is voluntary. We will explain the details of this study and go through the information sheet. We will then ask you to sign a consent form as evidence that you have agreed to participate. You are free to withdraw at any time, without prejudice or disadvantages to you, without giving a reason.

How is this study being done?

Attached is a detailed list of what is involved. In brief, this study will span a total duration of 10 weeks, where participants have to undergo a 2 weeks pre-conditioning session, a first laboratory test, 6 weeks of training, and a final week of laboratory test. All testing and training will take place in the School of Physiotherapy and Exercise Science (Building 408 and 400) at Curtin University. For each testing, you will be required to come down once for 3.5h testing session. Two global tests will be performed: 1) a biomechanics running, jumping and hopping assessment performed at the motion analysis laboratory (Build 408, level 1, room 1510), and 2) a strength assessment performed at the strength and conditioning lab (Building 400, level3, room 363). For the biomechanics assessment, reflective markers and adhesive electrodes will be attached to the skin of the lower limb and trunk. You will be required to perform repeated running trials at two speed conditions (your own comfortable speed and 3.5 m/s) while carrying two different loads (no load and a 20 % body weight load). For the strength assessments, maximum strength testing of the legs will be performed.

For the exercise program, regardless of the group you are allocated to, it is conducted within Building 400, in the strength and conditioning gym. It will be a 6 week exercise program, three sessions per week, and each session will last between 45-60min. Each session will involve 3 resistance exercises that will be strenuous. The intensity of the exercises will be gradually built up, and you will be closely supervised by two strength and conditioning personnel.

What happens next?

Taking part in this study involves being randomly allocated to one of two groups. The groups will be randomly selected by a computer (like a coin toss), so you cannot select the group you will be joining. Randomization will only happen after you perform the two week pre-conditioning session and baseline assessments.

Group 1: If you are in this group, you will be performing a series of 3 exercises, leg press, calf raisers, and lunges

Group 2: If you are in this group, you will be performing a series of 3 exercises, single leg hopping, maximal jumps and hip flexor cable pull.

What do I have to do to be involved in this study?

You will first be asked to sign an inform consent sheet. You will be given a personal copy of this information sheet.

What are the benefits of participating in the research?

By taking part in this study, you will find out information about your running biomechanics, lower limb muscle strength and power. At the end of the study, you will also receive information on changes in running biomechanics, strength and power. You will also be participating in a 6 week supervised resistance training program in a secured training environment at no financial cost to you. During these training sessions, you will be taught and demonstrated the optimal execution of the above exercises. The information gained will help us understand the optimal exercise parameters needed to design an exercise program to improve an individual's capacity to run with load. Previous studies have found that resistance training programs in general benefit running economy and performance for runners of all expertise level.

What are the risks of the research?

The risk associated with assessment and training is likely to be minimal. The pre-conditioning sessions are there to familiarize you with the exercises and get accustomed to the training load. There is a minimal risk of sustaining an acute musculoskeletal injury during testing. This risk will be further reduced by ensuring adequate warm-up and rest between running repetitions. A previous study has demonstrated that our testing protocol is feasible and safe. For the training program, the intensity of the resistance program will be gradually built up over the 6 weeks. Correct exercise technique will be ensured by the trainers which will increase the safety of the program. You may experience muscle soreness in the exercised muscles post training, but this is a normal sign of fatigue and will be expected to recover within 1-2 days. In addition, exercise volume will vary within the week to account for potential muscle soreness and fatigue. All participants must be attired appropriately during testing and exercise (i.e. covered shoes, and gym attire) to minimize the risk of accidental knocks. Another possible risk is an allergic reaction to the tape used to apply the reflective markers. To keep this risk minimal, we will be using low allergy tape and if a reaction does occur, the tape will be removed immediately.

What if something goes wrong?

There is a very low chance that harm may come to you during this study. However, if you do come to harm through your direct participation in this research project, there are no special compensation arrangements. Appropriate first-aid will be rendered by the testers and trainers on the spot should an injury or medical event occur. The principal researcher (Bernard Liew) is a registered physiotherapist in Western Australia (Reg No PHY0001910104). However, expenses for further treatments may have to be paid by you.

Parking and accessibility

During the periods of testing or exercising, there are designated car parking lots for research participants. You will need to provide your car registration number to the investigator 24 hours before you arrive. Parking requirements are only applicable from school hours of Monday to Friday, 8am to 4.30pm.

What will happen to the information you provide?

Any information that you give will remain completely confidential. During the analysis of the results you will be given a number to ensure complete anonymity. Your screening questionnaire and any results from the study will be kept for a period of 7 years with the project supervisors in a secure place. After this period, all records will be destroyed. This is a requirement of Curtin University. Upon your agreement, all data collected will be used for future research in this field.

You may refuse to participate in the study. If you change your mind once you have agreed to participate, you are free to withdraw at any time and without fear of prejudice. If you decide to withdraw please contact the principal researcher (Bernard Liew) at the earliest opportunity. In the event of this occurrence, all the data related to you will be destroyed.

The project researchers are Mr. Bernard Liew (0435223717; b.liew@postgrad.curtin.edu.au), Dr. Kevin Netto (9266 3689), and Dr. Susan Morris (9266 4644). If you wish to make a complaint on ethical grounds, then you can contact the Secretary of the Human Research Ethics Committee (phone: 9266 2784 or hrec@curtin.edu.au).

This study has been approved by the Curtin University Human Research Ethics Committee (RD-41-14). The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its key role is to protect participants. If needed, verification of approval can be obtained either by writing to Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University, GPO Box U1987, Perth WA 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au.

Thank you for considering participation in this study.

This study has been approved by the Curtin University Human Research Ethics Committee (RD-41-14).

E.3 Informed consent

Project Title: The effects of two resisted exercise programs on the biomechanics of load carriage running: a randomized controlled trial

PLEASE NOTE THAT PARTICIPATION IN RESEARCH STUDIES IS VOLUNTARY AND YOU CAN CHOOSE TO WITHDRAW AT ANY TIME.

- I understand the purpose and procedures of the study.
- I have been provided with the participant information sheet.
- I understand that the procedure itself may not have any benefits.
- I understand that no personal identifying information like names and addresses will be used and that all information will be securely stored for 7 years before being destroyed.
- I understand that the data collected will be used for future research.
- I have been given the opportunity to ask questions.
- Any questions I have asked, have been answered to my satisfaction.

I

(Given name of participant)

(Surname)

of

(Address)

(Postcode)

.....

(Home phone)

(Mobile)

(Email)

have read the information provided and agree to participate in this study.

Participant's Signature

Date

I have explained to the participant the procedures of the study to which they have consented their involvement and have answered all questions. In my appraisal, the participant has voluntarily and intentionally given informed consent and possesses the legal capacity to give informed consent to participate in this research study.

Principal Investigator: _____ Date: _____

This study has been approved by the Curtin University Human Research Ethics Committee (RD-41-14)

E.4 Video informed consent

Project Title: The effects of two resisted exercise programs on the biomechanics of load carriage running: a randomized controlled trial

Principal Supervisor: Dr Kevin Netto

Co-supervisor: Dr Susan Morris

Principal researcher: Bernard Liew (PhD candidate)

The investigator has explained to me that visual images (video and still images) are important to verify marker position in the gait laboratory and can be useful to illustrate aspects of the study in reports describing the research. These images may also be converted to electronic formats for use in multimedia presentations and documents accessible to others by computer for teaching purposes and the purpose of sharing results of the study. By giving consent I authorize the researchers to use any of the visual images taken of me in printed format, in slides for presentations and in electronic format.

I, (Name).....

consent to allow visual images of me to be

used for the following:

Verification purposes to assess the data collected

Research purposes for use in conferences and journals

Educational purposes

Signature:

Date:

I have explained to the participant how the visual images will be used, and have answered all questions.

Investigator Signature:

Date:

This study has been approved by the Curtin University Human Research Ethics Committee (RD-41-14).

E.5 Gait collection sheet

Biomechanics trials

Subject name: _____

Testing date: _____

What went wrong?

Electrode number	Muscle attached
	VL
	RF
	SOL
	GL

Tasks	Performed	Reasons
cmj_bw (3max)		
cmj_back(3max)		
sj_bw(3max)		
sj_back(3max)		
hop_fix_bw_r (10s)		
hop_fix_bw_l (10s)		
hop_self_bw_r (10s)		
hop_self_bw_l (10s)		
hop_fix_back_r (10s)		
hop_fix_back_l (10s)		
hop_self_back_r (10s)		
hop_self_back_l (10s)		
run_fix_bw (5-10)		
run_self_bw (5-10)		
run_fix_back (5-10)		
run_self_back (5-10)		

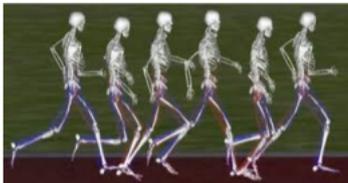
E.6 Advertisement

Novice/Experienced runners wanted – Training study to modify running mechanics

- Are you between 18 to 60 years old?
- Do you participate in running or running-related activities and sports?
- Are you in good general medical health, without conditions that preclude you from safely performing resistance exercise and running?



If you answered 'YES' to the above questions, you could be eligible for:



Scan the QR code to access my word-press link for more details,

or visit it at <https://liewxwb.wordpress.com/>



**FREE RESISTANCE TRAINING PROGRAM
(6WEEKS) + 2 WEEKS PREPARATION (Feb to
May 2016).**

and

FREE RUNNING AND STRENGTH ASSESSMENT

Curtin's School of Physiotherapy and Exercise Science PhD student (Bernard) is investigating the effects of different resistance training programs on running biomechanics while carrying heavy load. Ethics approval (RD-41-14).

For further information, please contact:

PhD student: Bernard Liew

Phone: 0435223717

Email: b.liew@postgrad.curtin.edu.au

E.7 Pre-test questionnaire

NAMESex: F/M

ID No.

Date of Birth

Height

(m):.....

Baseline test date

Weight

(kg):.....

As you are to be a subject in this laboratory/project, would you please complete the following questionnaire. Your cooperation in this is greatly appreciated.

Please tick appropriate box

NO

YES

Has the test procedure been fully explained to you?

Any information contained herein will be treated as confidential

1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?
2. Do you feel pain in your chest when you do physical activity?
3. In the past month, have you had chest pain when you were not doing physical activity?
4. Do you lose your balance because of dizziness or do you ever lose consciousness?
5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?
6. Is your doctor currently prescribing drugs for your blood pressure or heart condition?
7. Do you know of any other reasons why you should not undergo physical activity? This might include severe asthma, diabetes, a recent sports injury, or serious illness.

8. Have you any blood disorders or infectious diseases that may prevent you from providing blood for experimental procedures?

- If you have answered NO to all questions then you can be reasonably sure that you can take part in the physical activity requirement of the test procedure

I declare that the above information is correct at the time of completing this questionnaire Date/...../.....

Please Note: If your health changes so that you can then answer YES to any of the above questions, tell the experimenter/laboratory supervisor. Consult with your doctor regarding the level of physical activity you can conduct.

- If you have answered YES to one or more questions:

Talk with your doctor in person discussing with him/her those questions you answered yes. Ask your doctor if you are able to conduct the physical activity requirements.

Doctor's signature Date/...../.....

Signature of Experimenter..... Date
...../...../.....

E.8 Baseline health questionnaire

Medical condition

1. Indicate below ALL current medical conditions you have (even conditions which are well controlled (e.g. high blood pressure)), and any treatments you are receiving for them.

Medical condition	Hospital Treatment	Medication

2. Do you have any medical condition or disability that precludes you from performing HIGH INTENSIT Y running and resistance training? _____ (yes/no)

3. Have you visited the hospital seeking any treatment in the past 12 months?
_____ (yes/no)

Medical condition	Hospital Treatment	Approximate date

5. Have you had any surgery within the past 12 months? _____ (yes/no)

Medical condition	Surgery type	Approximate date

6. Are you currently pregnant (if female)? _____ (yes/no)

Running training

1. How long have you been running? _____ (years)

(One running year = running at least 15mins per session, at least four times a month, for most month of the year)

2. On average, how many times do you run per week? Over past 1 year: _____ Last 6 weeks: _____

3. On average, how many kilometres do you run in total per week? Over past 1 year: _____ Last 6 weeks: _____

4. Do you have CURRENT experience in carrying load while running? _____ (yes/no)

(Definition of experience = is running for a minimum of 30 minutes at least once per month with a load of $\geq 10\%$ of your weight, for at least 6 times per year)

If you answered YES to Q4, continue to Q5, if you answered NO, go to Q9

5. On average, how heavy do you carry on your runs, if known? _____ (kg)

6. Do you have PAST experience walking or running with heavy loads? _____ (yes/no). If YES: how many years' experience: _____

[Heavy load = $\geq 20\%$ weight. 1 year experience = walking or running with heavy load on at least 12 separate occasions/year]

7. Indicate below the race distances you currently compete in/ train for (put NA if not applicable)?

a) _____

b) _____

c) _____

8. What are your personal best timings in recent races (put NA if not applicable)?

Distance	Time	Date of race

--	--	--

9. How many times per week do you perform resistance training/ plyometric training/ endurance strength training?

Over past 1 year: _____ Last 6 weeks: _____

10. Based on the table below, what is your CURRENT resistance training status***?

	Resistance training				
Resistance training status***	Current program	Training age	Frequency (per week)	Training stress*	Technique experience and skill
Beginner (untrained)	Not training or just begun	≤ 2 months	≤ 1	None or low	None or minimal
Intermediate (moderately resistance trained)	Currently training	2-6 months	2-3	Medium	Basic
Advanced (moderately resistance trained)	Currently training	≥ 1 years	≥ 4	High	High
*In this example, “training stress” refers to the degree of physical demand or stimulus of the resistance training program.					

Running injuries

1. Over the past year, have you suffered a time-loss running related pain?

_____ (yes/no).

(Time-loss running related pain = pain that resulted in complete absence from all planned running sessions for at least 1 week)

Pain location (if self-diagnosed)	Injury (if diagnosed by healthcare practitioner)	Number of weeks off running	Approximate time when it occurred

2. Do you have current running related pain? _____ (yes/no).

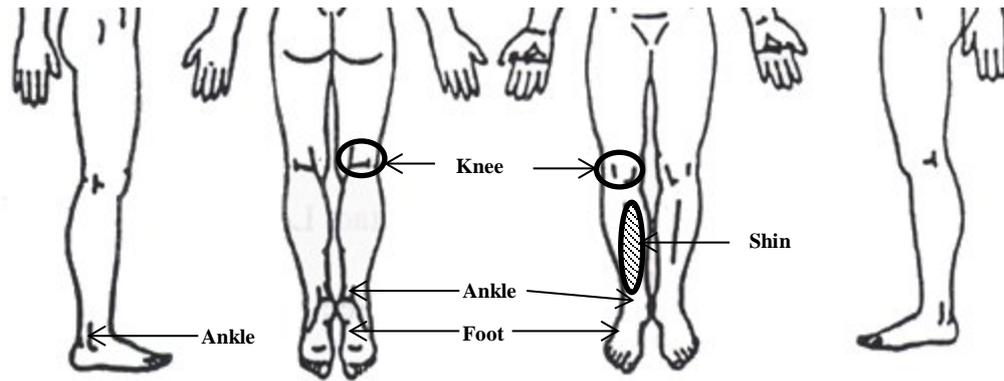
(Running related pain = pain (except blisters or muscle soreness) that is experienced during all planned running sessions for one week or more, regardless on the effects of your performance)

Pain location (if self-diagnosed)	Injury (if diagnosed by healthcare practitioner)	Duration of symptoms (weeks/months)

3. Do you currently use any form of orthotics in your running shoes? _____

(yes/no).

Overuse injury form (adapted from OSTRC Overuse Injury Questionnaire)



Part 1: Knee Problems

Please answer all questions regardless of whether or not you have problems with your knees. Select the alternative that is most appropriate for you, and in the case that you are unsure, try to give an answer as best you can anyway.

The term "knee problems" refers to pain, ache, stiffness, swelling, instability/giving way, locking or other complaints related to one or both knees.

Question 1

Have you had any difficulties participating in normal training and competition due to knee problems during the past week?

- Full participation without knee problems
- Full participation, but with knee problems
- Reduced participation due to knee problems
- Cannot participate due to knee problems

Question 2

To what extent have you reduced your training volume due to knee problems during the past week?

- No reduction
- To a minor extent
- To a moderate extent

- To a major extent
- Cannot participate at all

Question 3

To what extent have knee problems affected your performance during the past week?

- No effect
- To a minor extent
- To a moderate extent
- To a major extent
- Cannot participate at all

Question 4

To what extent have you experienced knee pain related to your sport during the past week?

- No pain
- Mild pain
- Moderate pain
- Severe pain

Part 2: Shin (only front of leg (*region between knee to ankle*) problems)

Please answer all questions regardless of whether or not you have problems in your lower back. Select the alternative that is most appropriate for you, and in the case that you are unsure, try to give an answer as best you can anyway.

The term "shin problems" refers to pain, aching, stiffness or other problems in your shin.

Question 1

Have you had any difficulties participating in normal training and competition due to Shin problems during the past week?

- Full participation without Shin problems

- Full participation, but with Shin problems
- Reduced participation due to Shin problems
- Cannot participate due to Shin problems

Question 2

To what extent have you reduced your training volume due to Shin problems during the past week?

- No reduction
- To a minor extent
- To a moderate extent
- To a major extent
- Cannot participate at all

Question 3

To what extent have Shin problems affected your performance during the past week?

- No effect
- To a minor extent
- To a moderate extent
- To a major extent
- Cannot participate at all

Question 4

To what extent have you experienced Shin pain related to your sport during the past week?

- No pain
- Mild pain
- Moderate pain

- Severe pain

Part 3: Ankle Problems

Please answer all questions regardless of whether or not you have problems in your ankle. Select the alternative that is most appropriate for you, and in the case that you are unsure, try to give an answer as best you can anyway. The term "Ankle problems" refers to pain, aching, stiffness, looseness or other complaints in one or both of your ankles.

Question 1

Have you had any difficulties participating in normal training and competition due to ankle problems during the past week?

- Full participation without ankle problems
- Full participation, but with ankle problems
- Reduced participation due to ankle problems
- Cannot participate due to ankle problems

Question 2

To what extent have you reduced your training volume due to ankle problems during the past week?

- No reduction
- To a minor extent
- To a moderate extent
- To a major extent
- Cannot participate at all

Question 3

To what extent have ankle problems affected your performance during the past week?

- No effect
- To a minor extent

- To a moderate extent
- To a major extent
- Cannot participate at all

Question 4

To what extent have you experienced ankle pain related to your sport during the past week?

- No pain
- Mild pain
- Moderate pain
- Severe pain

Part 4: Foot Problems

Please answer all questions regardless of whether or not you have problems in your foot. Select the alternative that is most appropriate for you, and in the case that you are unsure, try to give an answer as best you can anyway. The term "foot problems" refers to pain, aching, stiffness, looseness or other complaints in one or both of your feet.

Question 1

Have you had any difficulties participating in normal training and competition due to foot problems during the past week?

- Full participation without foot problems
- Full participation, but with foot problems
- Reduced participation due to foot problems
- Cannot participate due to foot problems

Question 2

To what extent have you reduced your training volume due to foot problems during the past week?

- No reduction

- To a minor extent
- To a moderate extent
- To a major extent
- Cannot participate at all

Question 3

To what extent have foot problems affected your performance during the past week?

- No effect
- To a minor extent
- To a moderate extent
- To a major extent
- Cannot participate at all

Question 4

To what extent have you experienced foot pain related to your sport during the past week?

- No pain
- Mild pain
- Moderate pain
- Severe pain