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INVESTIGATION OF RUNNING SHOE DESIGN ON THE FOOT KINEMATICS, KINETICS, AND MUSCLE RECRUITMENT PATTERN IN PEOPLE WITH OVERPRONATION PROBLEM

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Ph.D

The Hong Kong Polytechnic University

2010

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ROY TSZ-HEI CHEUNG

A thesis submitted in partial fulfillment of the requirements for the degree of

Doctor of Philosophy

Nov 2009

CERTIFICATION OF ORIGINALITY

I hereby declare that this thesis is my own work and that, to the best of my knowledge and belief, it reproduces no materials previously published or written, nor material that has been accepted for the award of any other degree or diploma, except where due acknowledgement has been made in the text. Any contribution made to the research by others, with whom I have worked at the Hong Kong Polytechnic University or elsewhere, is explicitly acknowledged in the thesis.

I also declare that the intellectual content of this thesis is derived from the discussion between me and my supervisor, Professor Gabriel Ng, and the data presented in this thesis is the product of my own work, except to the extent that assistance from others in the project's design and conception or in style, presentation and linguistic expression is acknowledged.

Roy Tsz-hei CHEUNG

This thesis is dedicated to my wife Shannon and my family.

Your unwavering support continues

my motivation to study and curiosity towards science.

ABSTRACT

Motion control footwear is a common technology in running shoe designs. The function of motion control footwear is mainly to reduce excessive rearfoot pronation in runners for injury prevention. Because erratic rearfoot kinematics may affect proximal joints by the linkage of human kinetic chain, the functions of motion control footwear may not only be confined to kinematic control. The functions of motion control footwear are not well explored and understood. Thus, the overall purpose of this thesis is to explore functions of motion control footwear in runners with excessive rearfoot pronation.

Experiments were conducted to test the efficacy of motion control footwear in terms of 1) rearfoot kinematics control; 2) plantar loading control; and 3) lower extremity muscle activity control. These parameters were also tested after muscle fatigue because runners are more prone to injury in such status.

A total of forty eight female runners were recruited. Three of them were rejected due to their normal rearfoot pronation. All the subjects were provided standard motion control footwear and neutral footwear for treadmill running sessions with 1-week apart.

Motion capture analysis suggested that the maximum rearfoot pronation of runners was reduced in motion control footwear condition, compared with neutral footwear condition ($p < 0.001$; 10.58° with motion control footwear, 13.94° with neutral footwear). After 1500m run, while the rearfoot supinators were fatigued ($p < 0.01$), the maximum rearfoot pronation further increased when runners put on neutral footwear ($p < 0.01$; 6.5° 95% C.I. 4.7-8.2°) but not in motion control footwear condition ($p = 0.06$; 0.7° 95% C.I. -0.3-1.4°).

Subjectively, runners were not able to differentiate the kinematics control function between two test footwear model, indicated by feedback score ($p = 0.711$) in a validated questionnaire. These results highlighted the importance of rearfoot posture assessment by medical professionals, rather than self adjustment by runners.

Plantar loading sensors revealed that the medial foot structures sustained higher loading ($p = 0.001$; 364 to 418 N; 15% increase at medial midfoot) with neutral footwear condition after muscle fatigue caused by running 1500m. This increased loading pattern is highly associated with various running injuries. On the other hand, the plantar loading was similar in motion control footwear condition ($p = 0.572$).

Electromyography indicated that the muscle activity of tibialis anterior (TA) and peroneus longus (PL) increased with mileage during 10km run. The activation of both TA ($p < 0.001$; normalized RMS 10.5% higher) and PL ($p < 0.001$; normalized RMS 9.6% higher) was higher in neutral footwear condition than motion control footwear condition. Also, significantly higher amount of muscle fatigue was noted in PL during

the neutral footwear testing condition ($p < 0.001$; median frequency drop in motion control footwear = 2.10 Hz; median frequency drop in neutral footwear = 11.60 Hz).

In quadriceps muscles, the experiments showed that the amplitude of median frequency (MF) drop in vastus medialis oblique (VMO) was higher in the neutral footwear condition ($p=0.008$; median frequency drop in motion control footwear = 1.23 Hz; median frequency drop in neutral footwear = 9.51 Hz) while the vastus lateralis (VL) had a more significant drop in MF when running with motion control footwear ($p=0.001$; median frequency drop in motion control footwear = 3.80 Hz; median frequency drop in neutral footwear = 1.78 Hz). Early fatigue of the major patella stabilizing muscle VMO, which occurred in neutral footwear condition, may lead to patellofemoral pain syndrome in runners.

The onset timing of VMO and VL was significant different with footwear condition change and mileage ($p=0.001$). Numerically, the VMO of the subjects activated at around 5.3% of a duty cycle earlier than VL when running with motion control footwear; whereas for the neutral footwear running condition, there was a delay in VMO activation by about 4.6% of a duty cycle compared to VL. The onset delay time of VMO was strongly correlated ($r = 0.948$; $p < 0.001$) with the running mileage in neutral footwear condition only. In the motion control footwear condition, this correlation was very weak ($r = 0.258$; $p = 0.472$).

The results of electromyography experiments suggested motion control footwear provided favorable running conditions in terms of higher resistance towards muscle fatigue, more stable activation of shank stabilizing muscles, as well as enhanced temporal activation of patella stabilizers. These footwear functions may be able to reinforce injury prevention in running population.

The findings of this thesis form a basis for the establishment of a running injury prevention program or adjunct therapy intervention by appropriate footwear prescription in runners with excessive rearfoot pronation problem.

PUBLICATIONS & CONFERENCE PRESENTATIONS
ARISING FROM THE THESIS

A. Article published (Appendix 6)

1. Cheung RTH, Ng GYF, Chen, BFC. (2006) Association of footwear with patellofemoral pain syndrome in runners. *Sports Medicine*, 36 (3), 199-205.
2. Cheung RTH, Ng GYF. (2007) A systematic review of running shoes and lower leg biomechanics: a possible link with patellofemoral pain syndrome? *International SportMed Journal*, 8 (3), 107-116.
3. Cheung RTH, Ng GYF. (2007) Efficacy of motion control shoes for reducing excessive rearfoot motion in fatigued runners. *Physical Therapy in Sports*, 8 (2), 75-81.
4. Cheung RTH, Ng GYF. (2008) Influence of different footwear on force of landing during running. *Physical Therapy*, 88 (5), 620-628.
5. Cheung RTH, Ng GYF. (2008) Motion control shoe affects temporal activity of quadriceps in runners. *British Journal of Sports Medicine*, 43 (12): 943-947.
6. Cheung RTH, Ng GYF. (2010) Motion control shoe delays fatigue of shank muscles in runners with overpronating feet. *American Journal of Sports Medicine*, 38 (3), 486-491.

B. Article in preparation

1. Cheung RTH, Ng GYF. (2010) A systemic review on the motion control running footwear functions. *Physical Therapy*. (Submitted)

C. Conference presentations (Appendix 7)

1. Cheung RTH, Ng GYF (2005) Foot motion analysis in runners wearing different footwear before and after fatigue of leg muscles. *The Hong Kong Student Conference of Exercise Science, Health and Rehabilitation, 27 August 2005, Hong Kong*. P. 12 ***(Winner of the Best Paper Award)***

2. Cheung R, Ng G (2005) Efficacy of motion control shoes in exhausted runners: a 3-dimensional kinematics analysis. *Combined Australian Conference of Science and Medicine in Sport, 5th National Physical Activity Conference and 4th National Sports Injury Prevention Conference, 13-16 October 2005, Melbourne, Australia*. In: *Journal of Science and Medicine in Sport* 8 (4) December 2005 Supplement. P. 176

3. Cheung RTH, Ng GYF (2006) Effects of muscle fatigue and footwear on plantar force distribution during running. *The 5th Pan-Pacific Conference on Rehabilitation & The Pre-FIMS World Congress of Sports Medicine 2006*. 9-11 June 2006, Hong Kong. P. 27

4. Cheung RTH, Ng GYF (2008) Footwear affects the temporal activation of the vasti muscles in subjects with excessive rearfoot pronation. *The 6th Pan-Pacific Conference on Rehabilitation, The 2008 Annual Scientific Meeting of the HKARM and 2008 Graduate Student Conference in Rehabilitation Sciences*. 4-5 October 2008, Hong Kong. P. 34

5. Cheung RTH, Ng GYF (2008) Running shoes and sport medicine: analysis of foot motion and plantar force distribution during running with motion control shoes. *Sports Medicine & Rehabilitation Therapy Convention 2008*, 14-16 August 2008, Hong Kong. P. 7

6. Ng GYF, Cheung RTH (2009) Motion control shoe affects the lower leg muscle activities in runners with over-pronation. *4th International State-of-the-Art Congress. Rehabilitation: Mobility, Exercise & Sports*. 7-9 April 2009, Amsterdam, The Netherlands. P. 18-19

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DECLARATION OF INTEREST

No commercial party having a direct financial interest in the results of this series of research studies did provide any kind of support towards the study. No party was conferred or will confer a benefit upon the authors or the organizations with which the authors are associated.

TABLE OF CONTENTS

CERTIFICATION OF ORIGINALITY …… i
DEDICATION …….. ii
ABSTRACT …… iii
PUBLICATIONS & CONFERENCE PRESENTATIONS ARISING FROM THE THESIS …….. v
ACKNOWLEDGEMENTS …… vii
DECLARATION OF INTEREST …… viii
TABLE OF CONTENTS …… ix
LIST OF ABBREVIATIONS …… xv
LIST OF FIGURES …… xvi
LIST OF TABLES …… xviii
LIST OF APPENDICES …… xix

CHAPTER 1 MOTION CONTROL FOOTWEAR
1.1 History and development of motion control footwear …… 1-2
1.2 Rationale of motion control footwear to control motion …… 2-3
1.3 Efficacy in rearfoot kinematics control ……. 3-6
1.4 Loading / kinetics difference in different footwear ……. 6-9
1.5 Functions of motion control footwear …….. 10-16
1.5.1 Relationship between rearfoot pronation and tibial rotation
1.5.2 Relationship of tibial/ femoral rotation on patellofemoral joint mechanics
1.5.3 Other possible functions of motion control footwear
1.5.3.1 Tibial and femoral rotation difference in different footwear
1.5.3.2 Muscle activation difference in different footwear
1.6 Running injury …… 16-27
1.6.1 Risk factors of running injuries
1.6.1.1 Major intrinsic factors
1.6.1.1.1 Unbalanced plantar loading
1.6.1.1.2 Muscle imbalance
1.6.1.1.3 Muscle fatigue
1.6.1.1.4 Excessive rearfoot pronation
1.6.1.2 Major extrinsic factors
1.6.2 Runners at risk
1.7 Objectives of present investigations …….. 27
1.8 Specific aims of the investigations …….. 27-28
1.9 Hypothesis to be tested …….. 28

1.10 Organization of the thesis ……... 28-29

CHAPTER 2 METHODOLOGY FOR MEASURING THE REARFOOT KINEMATICS AND PLANTAR FORCE DISTRIBUTION DURING PROLONGED RUNNING

2.1 Introduction …… 30-33
- 2.1.1 Footwear model
- 2.1.2 Testing protocol
- 2.1.3 Treadmill versus over ground running
- 2.1.4 Running mileage
- 2.1.5 Gender of subjects
- 2.1.6 Skill level of subjects
- 2.1.7 Foot type screening
- 2.1.8 Symptoms

2.2 Choice of subjects …… 34-36
- 2.2.1 Inclusion criteria
- 2.2.2 Exclusion criteria

2.3 Location & duration of the studies …… 36

2.4 Establishment of effective fatigue procedure for rearfoot supinators ……. 37-46
- 2.4.1 Functional definition of muscle fatigue
- 2.4.2 Selection of target muscle group
- 2.4.3 Assessment for muscle fatigue in rearfoot supinators
- 2.4.4 Establishment of fatigue procedure
  - 2.4.4.1 Pilot test A – running on trampoline
  - 2.4.4.2 Pilot test B – running on sand
  - 2.4.4.3 Pilot test C – treadmill running

2.5 Equipment …… 46-61
- 2.5.1 Footwear
  - 2.5.1.1 Footwear size
  - 2.5.1.2 Shoe box tightness standardization
- 2.5.2 Treadmill
- 2.5.3 Motion recording
  - 2.5.3.1 Marker positions
  - 2.5.3.2 Definition of rearfoot pronation
  - 2.5.3.3 Definition of excessive rearfoot pronation
- 2.5.4 Plantar loading measurement
  - 2.5.4.1 Zonal analysis
- 2.5.5 Synchronization between motion capturing system and plantar loading

measurement
2.5.6 Quantification of muscle fatigue
2.5.7 Subjective feedback on footwear function in rearfoot pronation control
2.6 Testing procedure …… 61-62
2.6.1 Running sessions arrangement
2.6.2 Standardized warm up exercise
2.6.3 Kinematics and kinetics data collection and data management
2.7 Statistical analysis of data …… 62-65
2.8 Ethics …… 65

CHAPTER 3 REARFOOT KINEMATICS DURING RUNNING WITH MOTION CONTROL FOOTWEAR
3.1 Introduction ……..…… 66
3.2 Results …… 66-71
3.2.1 Rearfoot kinematics
3.2.2 Level of muscle fatigue
3.2.3 Subjective feedback on rearfoot motion control function of footwear models
3.3 Discussion …… 71-76
3.3.1 Rearfoot kinematics with motion control footwear
3.3.2 Effects of muscle fatigue
3.3.3 Subjective feedback on motion control footwear function
3.3.4 Cut-off amplitude to define overpronator
3.4 Conclusions …… 76-77
3.5 Relevance to the main study …… 77

CHAPTER 4 PLANTAR LOADING DURING RUNNING WITH MOTION CONTROL FOOTWEAR
4.1 Introduction ……..…… 78
4.2 Results …… 78-80
4.2.1 Plantar loading
4.2.2 Level of muscle fatigue
4.3 Discussion …… 80-83
4.3.1 Zonal analysis of plantar loading in different footwear
4.3.2 Increase plantar loading and running injury
4.3.3 Effects of muscle fatigue: from kinematics to kinetics
4.4 Conclusions …… 84
4.5 Relevance to the main study …… 84-85

4.6 Role of muscles and the kinetic chain concept ……... 85

CHAPTER 5 METHODOLOGY FOR MEASURING SHANK AND THIGH MUSCLES ACTIVITIES DURING PROLONGED RUNNING
5.1 Introduction …… 86
5.2 Choice of subjects …… 86-89
5.2.1 Adjustment of inclusion and exclusion criteria
5.2.2 Inclusion criteria
5.2.3 Exclusion criteria
5.3 Location and duration of the investigations …… 89
5.4 Equipment…… 89-101
5.4.1 Footwear
5.4.2 Footwear size
5.4.3 Shoe box tightness standardization
5.4.4 Treadmill
5.4.4.1 Justification for running speed adjustment
5.4.4.2 Step detection by load cell underneath treadmill
5.4.5 Electromyography
5.4.5.1 Selection of target muscles under test
5.4.5.2 Procedure for the muscle activity measurement
5.4.5.2.1 Skin preparation
5.4.5.2.2 Electrode positioning
5.4.5.2.3 Signal amplification, processing and recording
5.4.5.3 Muscle activity measurement
5.4.5.3.1 Definition of muscle onset
5.4.5.3.2 Median frequency
5.4.5.3.3 Amplitude of muscle activity
5.4.5.4 Quantification of impact noise
5.5 Testing procedure ……102-105
5.5.1 Running sessions arrangement
5.5.2 Standardized warm up exercise
5.5.3 Electromyography data collection and data management
5.5.3.1 Division of running session
5.5.3.2 Normalization of electromyographic data
5.6 Statistical analysis of data……105-106
5.7 Ethics…… 106

CHAPTER 6 SHANK MUSCLES ACTIVITIES DURING 10KM RUN WITH MOTION CONTROL FOOTWEAR
6.1 Introduction …… 107
6.2 Results ……107-111
6.2.1 Number of running steps in each checkpoint
6.2.2 Muscle activity of shank muscles
6.2.3 Fatigue of shank muscles
6.3 Discussion ……111-113
6.3.1 Effects of motion control footwear on muscle activity of shank muscles
6.3.2 Effects of motion control footwear on shank muscle fatigue
6.4 Conclusions ……113-114
6.5 Relevance to the main study …… 114

CHAPTER 7 QUADRICEPS ACTIVATIONS DURING 10KM RUN WITH MOTION CONTROL FOOTWEAR
7.1 Introduction …… 115
7.2 Results ……115-117
7.2.1 Muscle onset of vasti muscles
7.2.2 Muscle fatigue of vasti muscles
7.3 Discussion ……117-121
7.3.1 Effects of motion control footwear on vasti muscle onset
7.3.2 Effects of motion control footwear on vasti muscle fatigue
7.4 Conclusions …… 121
7.5 Relevance to the main study …… 121

CHAPTER 8 GRAND DISCUSSIONS AND CONCLUSIONS
8.1 Efficacy of motion control footwear …… 122-138
8.1.1 Rearfoot kinematics control
8.1.2 Plantar loading control
8.1.3 Shank muscle activity control
8.1.4 Quadriceps muscle activity control
8.1.5 Relationship between motion control footwear and running injury
8.2 Limitations…….. 139-142
8.2.1 Accuracy of shoe markers to represent rearfoot movement
8.2.2 Gender of subjects
8.2.3 Asymptomatic subject pool

8.2.4 Skill level of subjects
8.2.5 Treadmill running
8.2.6 Definition of overpronation
8.3 Future study…… 142-145
8.3.1 Research quality
8.3.2 Kinematics of proximal segments
8.3.3 Effectiveness of motion control footwear in running injury prevention
8.3.4 Adjunct therapy for running injury by footwear adjustment
8.4 Conclusion ……..…… 145-146

REFERENCE …… 147-171
APPENDICES…… 172-191

LIST OF ABBREVIATIONS

ANOVA	=	Analysis of variance
ATP	=	Adenosine triphosphate
BMI	=	Body mass index
CI	=	Confidence intervals
Cm	=	Centimeter
CMRR	=	Common mode rejection ratio
CP	=	Check point
dB	=	Decibel
e.g.	=	For example
EMG	=	Electromyography / Electromyographic
Hz	=	Hertz
ICC	=	Intraclass correlation coefficient
i.e.	=	That is
Kgf	=	Kilogram force
Km	=	Kilometer
kΩ	=	Kilo-ohm
M	=	Meter
MF	=	Median frequency
Mm	=	Millimeter
Ms	=	Millisecond
MVC	=	Maximum voluntary contraction
N	=	Newton (force unit)
PFPS	=	Patellofemoral pain syndrome
PL	=	Peroneus longus
RMS	=	Root mean square
SD	=	Standard deviation
TA	=	Tibialis anterior
VL	=	Vastus lateralis
VMO	=	Vastus medialis oblique
α	=	Alpha
°	=	Degree
%	=	Percent
±	=	Plus and minus
>	=	Larger than
<	=	Less than
=	=	Equal to

LIST OF FIGURES

Figure		Page
1	Extrinsic and intrinsic risk factors of running injuries	19
2	Dynamometer setup	39
3	Running on trampoline	41
4	Test shoe models: Adidas Supernova Control (motion control footwear) & Adidas Supernova Cushion (neutral footwear)	49
5	Standardized shoe lace	51
6	Motion capture system setup	52
7	Marker position for motion capturing	54
8	Insole loading sensor setup	58
9	Insole zonal divisions	59
10	Standardized warm-up exercise protocol before running tests	63
11	Flowchart of running sessions in Chapter 2 experiment	64
12	Mean and standard deviations of rearfoot pronation in different footwear conditions before and after muscle fatigue	69
13	Subjective feedbacks on medial/ lateral control of footwear conditions represented by box-plot graph	73
14	When a subject is running on the treadmill, the load cell underneath will receive a higher impact while right running step stance phase, and a lower impact during left running step	93
15	Electrode positioning of (1) Tibialis Anterior, (2) Peroneus Longus and (3) Vastus Medialis Oblique, (4) Vastus Lateralis, and (5) Earth electrode	96
16	An example to show data acquisition for load cell signals (at channel 5) and electromyographic signals from tibialis anterior (TA) (at channel 1), peroneus longus (PL) (at channel 2), vastus medialis obliquus (VMO) (at channel 3) and vastus lateralis (VL) (at channel 4). High amplitude of load cell signals indicated a right stance phase of running step	98
17	Impact noise quantification	101
18	Flowchart of the running sessions in Chapter 5 experiment	103
19	Normalized RMS EMG of tibialis anterior (TA) and peroneus longus (PL) in motion control (MC) and neutral (N) shoe conditions at different check points (CP). The error bars represent standard deviations and ‘#’ indicates significant difference in the neutral shoe condition of that particular CP when compared to the same shoe condition at CP1. The symbol	109

‘*’ indicates significant difference in the motion shoe condition of that particular CP when compared to the same shoe condition at CP1

20 Difference in muscle onset between VMO/ VL throughout the 10km run. The onset of VL relative to VMO was increasing with mileage in the neutral footwear condition (purple line). In the motion control footwear condition, however, VMO was activated earlier than VL (blue line) 116

LIST OF TABLES

Table		Page
1	Demographic data of subjects recruited in Chapter 2	35
2	Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination in pilot test A	42
3	Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination in pilot test B	42
4	Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination in pilot test C	47
5	Change in rearfoot pronation in different footwear before and after muscle fatigue by 1500m run	67
6	Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination before and after 1500m run	70
7	Correlations between change of rearfoot pronation and maximum voluntary contraction production difference. The negative sign means a decrease of the parameter	72
8	Plantar force in neutral footwear condition before and after 1500m run	79
9	Demographic data of subjects recruited in Chapter 5	88
10	Correlation between changes in normalized root mean square of EMG output from shank muscles and running mileage during 10km run	110
11	Drop in median frequency (MF) in shank muscles with different footwear conditions	112
12	Drop in median frequency (MF) in quadriceps muscles with different footwear conditions	118

LIST OF APPENDICES

Appendix	
I	Questionnaire sample to evaluate subjective feedback on pronation control function in test footwear
II	Ethical approval by the Human Ethics Sub-committee of the Hong Kong Polytechnic University in Chapter 2
III	Informed written consent form sample in Chapter 2
IV	Ethical approval by the Human Ethics Sub-committee of the Hong Kong Polytechnic University in Chapter 5
V	Informed written consent form sample in Chapter 5
VI	Publications arising from the thesis
VII	Award
VIII	Raw data

CHAPTER 1 MOTION CONTROL FOOTWEAR

1.1 HISTORY AND DEVELOPMENT OF MOTION CONTROL FOOTWEAR

With the rapid growth of the running population, the running shoe industry is flourishing and the design features of running shoes have evolved in an unparalleled pace. Even though the selection criteria of many people for running shoes may be based on the appearance, brand and price, numerous footwear technologies developed for various purposes in recent decades and have evoked people to choose running shoes according to their specific physical needs. Some technologies are for better sports performance [Stefanyshyn & Nigg, 2000; Roy & Stefanyshyn, 2006] and some are for injury prevention [Curtis, et al., 2008; Wegener, et al, 2008], the concept of motion control footwear that aimed to reduce excessive rearfoot pronation was founded in early 1980s. The mechanism of motion control footwear was initially based on wedging or flares of heel counter [Lafortune, et al., 1994; Nigg & Morlock, 1987; Stacoff et al., 2000; Stacoff et al., 2001] which had later evolved to different deformation rates between lateral and medial midsoles [Clarke, et al., 1983; Nigg, et al., 2003; Stacoff et al, 2000; Stacoff et al., 2001], so as to control the relative rates of midfoot and rearfoot motion.

The current trend of motion control footwear technology utilizes asymmetrical materials in the midsole rather than the flare or wedge design. The main reason for shifting the emphasis from shoe shape to material for the sole is due to biomechanical consideration. As a large shoe wedge would increase the moment arm upon landing, in particular on trailing or outdoor terrain, shoe flare or wedge design may increase the chance of unexpected foot movements and thus ankle sprain [Fordham et al., 2004].

1.2 RATIONALE OF MOTION CONTROL FOOTWEAR TO CONTROL MOTION

Anisotropic deformation of midsole materials in the medial and lateral heel was introduced to control excessive rearfoot pronation [Hamill, et al., 1988; Stacoff, et al., 1996]. At the initial contact phase of a normal running cycle, the shoe lands on the lateral aspect of the heel and maximum rearfoot pronation occurs in this period [McPoil & Cornwall, 1994]. Softer lateral aspect of the midsole would stop the foot from pronating excessively by providing additional cushioning against the impact. A firmer medial midsole could decelerate the movements of the rearfoot by providing supplementary support. In runners who are not heel strikers i.e. runners land on forefoot or midfoot instead of the heel, modification of midsole material distribution is found in motion control footwear designs. Apart from rearfoot pronation control,

decelerated rearfoot movement may also be able to control the plantar loading during landing [Dixon & McNally, 2008; Nigg 2001; Willems et al., 2006].

1.3 EFFICACY IN REARFOOT KINEMATICS CONTROL

Reports on the efficacy of motion control footwear in limiting excessive rearfoot motion are equivocal. Different results [Clarke, et al., 1983; Hamill, et al., 1988; Nigg & Morlock, 1987; Perry & Lafortune, 1995; Stacoff, et al., 1991; 2000; 2001; van Gheluwe, et al.,1995] were reported on rearfoot pronation with motion control footwear.

An early study by Clarke et al. [1983] reported successful control of excessive rearfoot movement by footwear with motion control features. In their study, a total of 36 different test shoe conditions were provided. The test shoe conditions were different in their midsole hardness (3 hardnesses), heel flares (3 types of heel flare direction) and heel height (4 heights). The results suggested that the firmer midsole and lateral (valgus) flare limited the total rearfoot movement (sum of supination and pronation movement) and maximum pronation angle. On the other hand, the heel height of footwear had no effects in pronation control.

A later study by Perry and Lafortune [1995] using two-dimensional kinematics analysis of running with shoes in valgus and varus flare echoed the findings of Clarke

et al. [1983]. They found that a reduction of maximum foot pronation angle by approximately 15^{o} was achieved in valgus-wedged footwear condition when compared with the varus-wedge condition.

The rationale for valgus wedge and firmer midsole to control foot pronation was based on the findings that such deign features could provide stronger deceleration drive on foot pronation [Hamill, et al., 1988; Stacoff, et al., 1996]. However, the above findings by Clarke et al. [1983] and Perry & Lafortune [1995] were obtained by a 2-dimensional foot motion analysis method. When considering the complex spatial orientations of the foot structures, a 2-dimensional method of motion capture might confront projection errors.

Nigg et al. [1988] failed to find pronation control effects in midsole density modified footwear by a small size experimental study. Similar results were also reported by McNair & Marshall [1994] in 10 healthy male subjects. Another small scale study by Kersting & Bruggemann [2006] also failed to recognize any rearfoot kinematics difference between test shoe conditions. Milani et al. [1997] recruited a larger sample size of 27 healthy subjects in their research. However, in this study, the eight test shoe conditions were in a very small range of midsole hardness The hardness of the test shoes were measured by accelerations and the displacements by an impact velocity of 96cm/sec. The peak acceleration ranged 9.6g and 13.7g with the

sole deformation varying between 7.4mm and 11.5mm. Again, the results failed to identify any pronation control effect on runners. In the studies by Nigg et al. [1988], McNair & Marshall [1994] and Milani et al. [1997], all the subjects were not screened if the participants had excessive rearfoot pronation problem.

Concerning heel flare modification, another two small scale studies by Nigg & Morlock [1987] and Stacoff et al. [2001] suggested this modification may not be effective in producing pronation control effects. Nigg & Morlock [1987] examined the anti-pronation effects of 3 types of heel flare modifications namely round/ varus flare, no flare, lateral/ valgus flare, on 14 healthy subjects. The results suggested that only the initial pronation angle at heel strike phase of running was different among the various shoe flares. However, the total pronation angle was similar among the three test conditions. Stacoff et al. [2001] used intra-cortical bone pin to evaluate lower extremity motion in 5 healthy runners and demonstrated that effects of shoe flare modifications on rearfoot motion were highly varied between subjects.

The aforementioned research studies highlighted the importance of subject size estimation and footwear selection. The small sample size would significantly hamper the statistical power. The function of test shoe models should be specifically distinguished between each other so that the effect size of measurement can be enlarged.

Experiments by Butler et al. [2006] assessed the foot posture of subjects and classified them as “high foot arch” and “low foot arch” group before running tests. Two commercially available running shoe models were provided to the subjects. One of the shoe designs was a motion control design with dual-density at midsole and the other model was a neutral cushion model. The results revealed significant pronation difference in different foot arch height and footwear conditions. The results not only illustrated the potential function of footwear on rearfoot kinematics control but also highlighted the importance to screen the subjects before testing. As the motion control footwear is designed for runners with overpronation problem, subjects with normal foot type may contaminate the findings.

1.4 LOADING / KINETICS DIFFERENCE IN DIFFERENT FOOTWEAR

In clinical practice, plantar loading measurement has been used to evaluate foot deformities such as abnormal rearfoot posture, hallux valgus, plantar fasciitis, callus formation and diabetes mellitus related foot ulcers [Orlin & McPoil, 2000]. A typical amount or pattern of plantar loading may reflect lower extremity pathologies. In view that decelerated rearfoot movement may also be able to control the plantar loading during landing [Dixon & McNally, 2008; Nigg 2001; Willems et al., 2006], different loading pattern was investigated among different running footwear models.

Research studies were conducted to test the change in impact force with respect to different hardness of shoe materials [Clarke, et al., 1983]. Nigg & Morlock [1987] was amongst the first research teams to speculate that motion control footwear might alter the joint impact force by correcting the distal joint alignment. However, in their experiment, no significant difference in impact force during running was found with different footwear. Nigg et al. [1988] presented one of the earliest papers about sole hardness on running kinetics and that study involved 16 healthy subjects tested with running on a force plate located in the middle of a runway. Their results did not suggest that change of sole hardness could effectively reduce impact force for runners.

Impact force could be measured by other means than force plate. For example, McNair and Marshall [1994] used accelerometer securing on subjects' tibia to estimate the impact force. In the study by McNair and Marshall [1994], no impact force difference was observed when runners put on footwear in different shock attenuating materials. Milani and colleagues [1997] measured the ground reaction force and overall plantar loading in runners with different footwear and found similar loadings as McNair & Mashall [1994]. Similar results were also reported by Butler [2006] and Roy [2006], who had respectively studied runners' response by changing the footwear and altering the footwear midsole longitudinal stiffness. No significant kinetics difference was found between different shoe conditions.

Two studies [Fong, et al., 2007; Divert, et al., 2005] adopted barefoot running as the control condition to test the footwear function in terms of kinetics parameter. Fong et al. [2007] found that running shoe would provide significantly better cushioning by lowering the normalized peak impact force than barefoot running. This result was contradictory to the findings by Divert et al. [2005], who reported that both passive and active peak of vertical force was lower in barefoot condition. This discrepancy might be explained by the altered running pattern of subjects in the study of Divert et al. [2005], because shorter stance phase time, flight time and stride length were observed in their subjects when running in barefoot.

In a pilot study conducted by Perry and Lafortune [1995], ground reaction force was found to increase with valgus wedge shoes, which was one of the primary designs of motion control footwear. However, the ground reaction force was not reduced in varus shoe condition which might exaggerate the pronation amplitude during running. On the other hand, lateral wedged footwear was suggested to reduce the peak external knee varus moment and peak medial compartment force at the knee during normal walking [Crenshaw, et al., 2000]. It was suggested that lateral wedged footwear could also be an alternative adaptive device in treating people with osteoarthritis of knee.

More recently, a commonly adopted method to quantify kinetics outcome is the plantar loading or what is known as, “pedography”. It is an insole apparatus

consisting of numerous sensors and it measures the direct force acting at the shoe-feet interface. Different from conventional force plate, plantar loading measurement can be divided into different anatomical zones so as to allow more precise analysis of each of the zones under the foot. This is an important advantage over the conventional method because knowing the instantaneous stress acting on a particular body part may better represent the risk of injury during sports. However, this technology is relatively new and there were not many studies that investigated the loading difference between footwear models with this outcome measurement.

Wegener [2008] used plantar loading measurement to verify that the midsole material was able to alter regional peak loading acting on different foot structures in runners. Very recently, Wiegerinck et al. [2009] have reported lower peak plantar loading in 37 runners when running with training shoes than with “racing flats”, which is a light-weight footwear design for competition. The authors suggested prescription of appropriate footwear could help in prevention of metatarsal stress fracture. These results echoed with an earlier study by Kersting & Bruggemann [2006], who also analyzed the kinetics parameter according to different foot regions by plantar loading measurement.

1.5 FUNCTIONS OF MOTION CONTROL FOOTWEAR

Excessive rearfoot pronation was associated with various running injuries [Cook, et al., 1990]. Most injuries are localized in the foot structure e.g. plantar fasciitis [Pascual Huerta, et al. 2008], while some other injuries may affect the proximal body parts e.g. patellofemoral pain syndrome (PFPS) [Eng & Pierrynowski, 1994; Johnston & Gross, 2004; Collins, et al., 2009]. If erratic tibial and compensatory femoral rotations were related with excessive rearfoot pronation, it would be possible to derive an indirect relationship that running injuries in proximal body parts could be prevented by motion control footwear prescription [Cheung & Ng, 2006; 2007]. Also, the functions of motion control footwear may not only confine to kinematics control of rearfoot, but other parameters such as plantar loading and running efficiency, which will be elaborated below.

1.5.1 Relationship between rearfoot pronation and tibial rotation

Tibial rotation is coupled with the supination and pronation movements of the rearfoot throughout a gait cycle at the subtalar joint. This translation of movement has been well studied both *in-vitro* and *in-vivo*.

Theoretical models have postulated that excessive pronation of the rearfoot will delay external rotation of tibia after mid-stance [Tiberio, 1987]. Due to the

architectural design in the body of talus in relation to the distal tibia and fibula, the tibia rotates internally with foot pronation. In the case of excessive pronation, longer duration in foot pronation would detain the supination movement and the resultant tibial external rotation. This change in ankle and foot kinematics may alter the normal mechanics of the tibiofemoral joint by creating a torsional moment at this joint [Tiberio, 1987]. It is speculated that the compensatory internal rotation of femur could balance the alignment during extension of the knee. However, this chained compensation may, in turn, alter the patellofemoral tracking [Tiberio, 1987].

By mounting a foot-leg specimen in a device that allows movements in each anatomical plane, Hintermann et al. [1994] found a positive correlation between foot pronation with tibial internal rotation and foot supination with tibial external rotation. This experiment has provided the evidence of inter-dependent kinematics between foot pronation and tibial internal rotation. A study by Lafortune et al. [1994] confirmed these findings using Steinmann pin markers to track the movements of tibia. Different patterns of tibia movements were investigated when subjects walked with footwear that was modified to induce extreme pronation and supination of the foot. Even though that study only demonstrated a subtle increase of 4^{o} of internal tibial rotation when wearing the pronation-inducing shoes as compared to the supination-inducing shoes, it has supported the notion that foot pronation and tibial

rotation were interrelated.

However, the relationship between tibial and foot movements is not uniform across individuals. High inter-subject variability may result from different integrities of the ankle-foot complex, including the articular surfaces, force of the muscles, and strength of the ligaments [Valderrabano, et al., 2003]. A study by Reischl et al. [1999] using VICON motion analysis had demonstrated that the magnitude and timing of peak pronation was not predictive of the magnitude and timing of tibial and femoral rotation. This study suggested that foot pronation was not the only factor that determined the tibial and femoral rotation, other factors of the lower limb movements, such as muscle activity, should be considered in analyzing the foot and shank kinematics relationship.

1.5.2 Relationship of tibial/ femoral rotation on patellofemoral joint mechanics

The contact pressure of the patellofemoral joint during femoral rotation over a fixed tibia (simulating a closed kinetic chain lower extremity actions), and during tibial rotation with a fixed femur (simulating an open kinetic chain lower extremity actions) has been investigated in some *in-vitro* studies [Lee, et al., 1994; 2001]. Human knee specimens were mounted in a custom designed jig that allowed independent tibial and femoral rotation over each other and contact stress of the

patellofemoral joint was measured by pressure-sensitive films. With a fixed femur, it was revealed that internal tibial rotation did not increase the patellar contact pressure as much as that resulted from external tibial rotation. On the other hand, open chain external tibial rotation resulted in significant increase in contact pressure over the lateral articular facets of patella throughout the whole range of knee flexion/extension movement. Regarding the rotation of femur on a fixed tibia, that study reported a nonlinear increase in patellofemoral joint pressure in either direction. Only a small increase in patellofemoral joint pressure was detected from the initial 20^{o} of rotation in either direction. Beyond that range, the patellofemoral pressure shot up dramatically. In general, external femoral rotation would result in an increase in pressure on the medial facets, whereas internal femoral rotation would add pressure over the lateral patellar facets.

Results of these two studies implied that tibial rotation alone could not totally account for the change in patellofemoral joint pressure. It might be a combined effect of tibial and femoral rotation. In a study analyzing human kinetics in running, Bellchamber & van den Bogert [2000] found that the power flow was mainly from proximal to distal, but there was high between-subject variability. With this individual difference, the pattern of tibial and femoral rotation was believed to be not constant. This variation can also be resulted from different muscle recruitment pattern.

Therefore, the combined tibial and femoral rotation might increase the patellofemoral contact pressure thus contributing to the pathology of PFPS. This model may explain why in some studies that had only examined tibial rotation or femoral anteversion [Milgrom 1991] failed to find any intrinsic risk factor to differentiate PFPS subjects from the asymptomatic subjects.

An early study by Levens et al. [1949] had reported that the femur would internally rotate by approximately 7° during the first half of the stance phase during normal walking. However, more recent studies [Powers, et al., 2002; Reischl, et al., 1999] have reported a wide variability in the femoral rotation angle of around 20°. Therefore, in assessing the relationship between lower extremity rotations and patellofemoral joint kinematics, femoral rotation in excess of 20^{o} may be considered as a potential predisposing factor of PFPS.

1.5.3 Other possible functions of motion control footwear

As stated in sections 1.5.1 & 1.5.2, there is a relationship between rearfoot pronation and tibial rotation and the patellofemoral joint biomechanics can be affected by different combinations of tibial and femoral rotation. It is therefore plausible that motion control footwear can modulate the tibial and femoral rotation by checking excessive rearfoot pronation.

Since the tibial and femoral rotation is controlled by muscle activity, it is also hypothesized that different muscle activities could result with wearing different running footwear models.

1.5.3.1 Tibial and femoral rotation difference in different footwear

The efficacy of motion control footwear in checking proximal bony segment kinematics is still questionable. Lafortune et al. [1994] recruited five healthy subjects and conducted a walking test with pronation induced footwear (valgus wedged), pronation inhibiting footwear (varus wedged), and neutral footwear (no wedge). The 3-dimensional kinematics of tibiofemoral joint and patellofemoral joint were captured by bone pin markers on patella, tibia and femur. The results suggested that the tibial rotation difference between footwear conditions was only 4° and not statistically significant. Besides the small sample size that had compromised the power of the statistics, the use of asymptomatic subjects might also decrease the sensitivity of the test as these subjects may not respond as the same as subjects with foot pronation problem.

Butler et al. [2006; 2007] recruited recreational runners to perform running test with motion control footwear and neutral footwear. The lower extremity kinematics was monitored by skin markers. All the subjects were classified by their foot posture.

Again, a trend of larger internal tibial rotation amplitude was noticed in the neutral footwear condition, but that relationship did not reach a statistically significant level.

1.5.3.2 Muscle activation difference in different footwear

In our recent review papers [Cheung & Ng, 2006; 2007], we have speculated that motion control footwear may be a possible adjunct therapeutic device for people with PFPS. Muscle activation pattern was found to be different in previous research studies that examined orthotics [Murley & Bird, 2006] and footwear with different heel heights [Edwards, et al., 2008]. As muscle activities could have close associations with running injuries due to the changes in force distribution on the joints [Cowan, et al., 2001; Mizrahi, et al., 2000; Sanna & O'Connor, 2008], investigation of lower extremity muscle is warranted for injury prevention and symptomatic control. However, there is no research report published on the effects of motion control footwear on the muscle activation pattern.

1.6 RUNNING INJURY

Running is one of the most popular leisure sports worldwide. Not only is running an individual sport per se, but it is also an important component of many other sports. The running population is growing rapidly with higher recognition for its beneficial

effects on cardio-respiratory, physical as well as psychological health [Lawrence, 1997]. However, adverse effects such as over straining to the muscles [Rolf, 1995], stress fracture to the bones [Wall & Feller, 2006] and other overuse conditions such as patellofemoral pain syndrome (PFPS) [Cook, et al., 1990; Lun, et al., 2004; Thijs, et al., 2008] and posterior tibial syndrome (shin splints) [Thacker, et al., 2002; Craig, 2008] have also been reported in runners. In order to reduce the risk of running injuries, numerous investigations were launched to review different risk factors in this sport. It is important to understand the etiology of running injuries so that preventive measures and treatments can be formulated.

According to an epidemiological research study [van Mechelen et al., 1992], the overall annual incidence rate of running injuries varied between 37% and 56%. If the incidence was calculated according to the exposure of running time, the figures revealed that the incidence of injuries to be 2.5 to 12.1 injuries per 1000 hours of running. Running injuries involve mainly the lower extremities, with the knee joint to be most predominant followed by the leg and foot structures [Cook et al., 1990; van Mechelen et al., 1992; Taunton et al., 2003]. Clement et al. [1981] evaluated the injury data in 1650 runners and found that 87% of all injuries were at the knee and distal to the knee. 41% of which were at the knee and 18% of the total injuries were at the foot.

1.6.1 Risk factors of running injuries

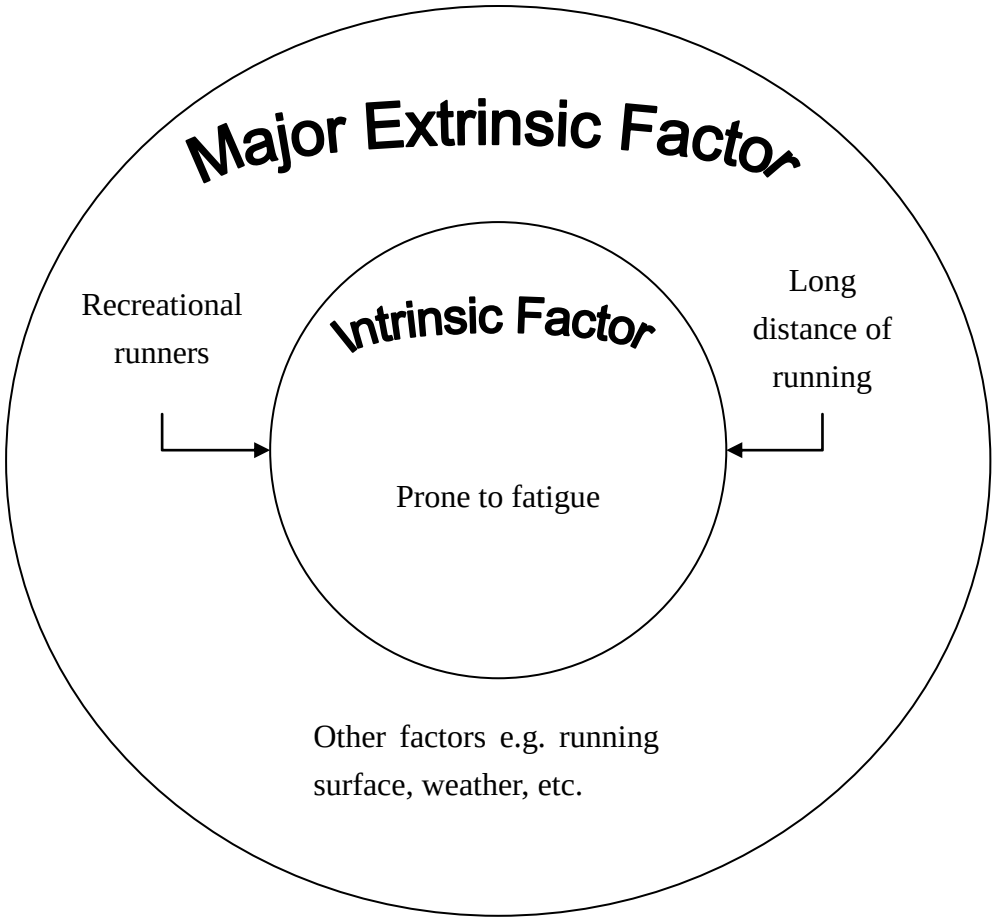
The relationship between lower extremity structure, mechanics and injury is not completely understood. The etiology of lower extremity injuries is uncertain and multi-factorial. This uncertainty may be due to the variability present in the human body and the many degrees of freedom in the whole kinetic chain. Because of this variability, a specific injury may be linked to many possible combinations of structural or mechanical risk factors. These risk factors of running injuries can be classified as extrinsic and intrinsic (FIGURE 1), and they can be inter-related with one another thus making a complex matrix of risk factors.

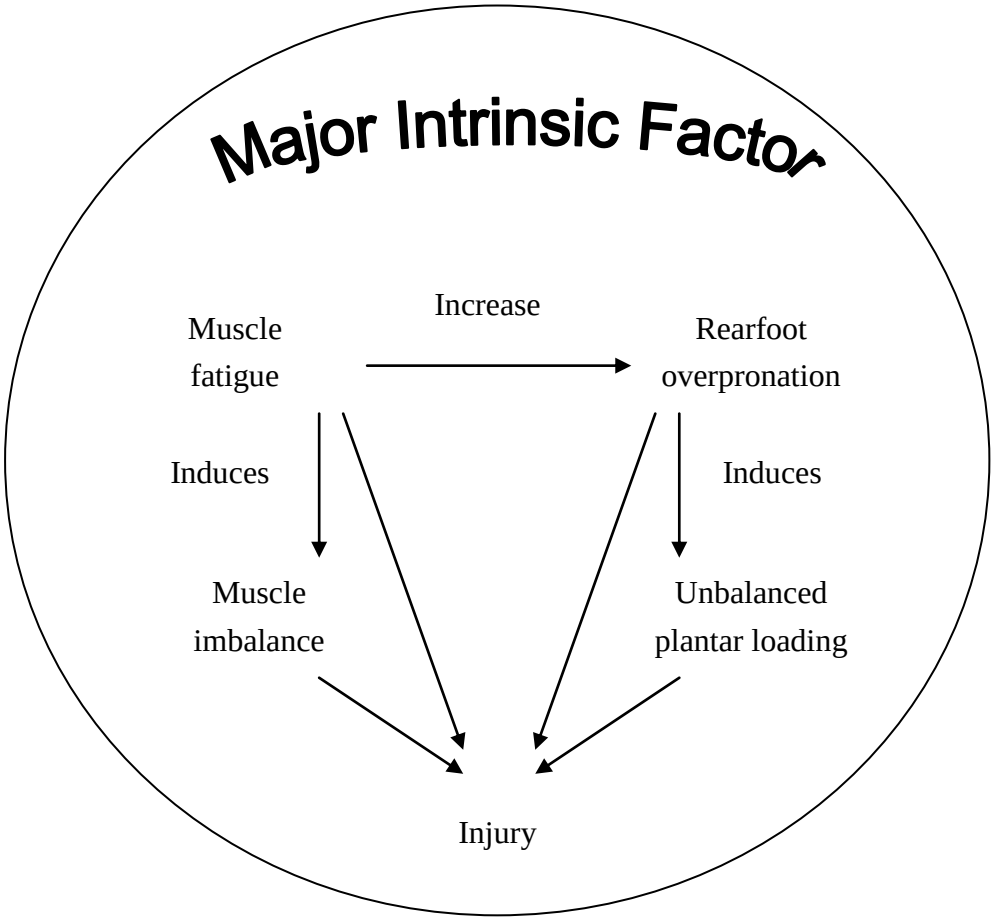
1.6.1.1 Major intrinsic factor

1.6.1.1.1 Unbalanced plantar loading

It has been speculated that biomechanical factors contribute to running injuries. The significance of ground reaction force in injury development is unclear and much of the early work supporting the correlation between impact forces and injury were coming from studies of animal models. For example, studies had found that exposure to impulsive forces and repetitive impact loading would lead to cartilage degeneration in cows [Radin & Paul, 1971] and rabbits [Anderson, et al., 1990; Dekel & Weissman,

Figure 1 Extrinsic and intrinsic risk factors of running injuries





1978; Radin, et al., 1973; Serink, et al., 1977; Swann & Seedholm, 1993; Yang, et al., 1989]. The effect of impact forces on human injuries is still questionable. Several studies have failed to establish a link between repetitive impact loading and osteoarthritis in runners [Burrows & Bird, 2000]. Therefore, another outcome measure known as pedography, was evolved in recent decades.

Pedography, or plantar force evaluation, is used to determine specific loading characteristics at the sole of the foot. It can reflect the amount of stress acting on the foot structures in real-time and this information may provide clues for identification of pathological running patterns which will be useful for injury prevention [Cavanagh, et al., 1987, Teyhen, et al., 2009]. This technique is often applied in orthotics prescription for patients with diabetes mellitus and rheumatoid arthritis [Lobmann et al., 2001; Jackson et al., 2004]. However, pedography is still a relatively new method for evaluating the instant tissue loading in sports science research.

Weist et al. [2004] had reported that the increase plantar loading at forefoot and toe area in 30 fatigued runners might explain the high risk of metatarsal stress fracture in this running population. A prospective study by Willems et al. [2006] which recruited 400 physical education students and correlated their biomechanical factors, including plantar loading, lower limb kinematics and gait patterns, to the incidence of “exercise-related lower leg pain” has shown that runners with

exercise-related lower leg pain had higher plantar loading over the medial foot structures. However, there is no studies have attempted to look at the plantar loading in runners with motion control footwear or in the state of fatigue.

1.6.1.1.2 Muscle imbalance

Muscle activity plays an important role in maintaining the correct kinematics during running. Imbalance or malfunctioning of lower extremity muscles would alter the movement patterns as well as joint loading. Therefore, muscle imbalance has been regarded to be a risk factor for running injuries.

The PFPS is a common knee problem in runners. According to a prospective study [Witvrouw, et al., 2000], 7%–15% of the running population suffered from PFPS and this problem is amongst the most predominant knee ailments in runners [van Mechelen, 1992]. Although the etiology of PFPS is not well understood, most researchers believe that it is a multifactorial problem. Different pathologies and corresponding treatment regimens were proposed and one of the classical speculations is mal-functioning of the active control system [Cowan, et al., 2001; Voight & Weider, 1991]. A manifestation of the active control system mal-functioning is the delayed onset of contraction of vastus medialis obliquus (VMO) relative to vastus lateralis (VL) [Cowan, et al., 2001; Voight & Weider, 1991]. Muscle onset sequence training

with biofeedback and other rehabilitation exercises have been shown to be effective and thus being advocated as appropriate treatment strategies for patients with PFPS [Yip & Ng, 2006; Ng, et al., 2008].

Furthermore, it has been proposed that the extrinsic foot musculature, especially the Tibialis anterior (TA) and peroneus longus (PL) are two major stabilizing muscles for rearfoot control [Murley & Bird, 2006]. It has been hypothesized that these muscles might alter their activations in order to maintain a preferred movement pattern [Nigg, 2001]. Dysfunction of these stabilizing muscles may lead to alternation of the normal joint kinematics and cause various running injuries including metatarsal stress fracture [Mizrahi, et al., 2000]. In order to reduce the abnormal loading on the foot structures, orthotic devices have been used and Nigg & Wakeling [2001] proposed a new paradigm that orthotic devices improve lower extremity symptoms by synchronizing the lower leg muscle activities including TA and PL.

1.6.1.1.3 Muscle fatigue

Muscle fatigue is a natural physiological response to prolonged exercise and it can have either a central or a peripheral cause [Gandevia, et al., 1995]. Central fatigue refers to a decrease in neural drive from the central nervous system, which results in reduction of motor unit discharge rate and failure of excitation of motor neurons

[Asmussen, 1979; Gandevia, et al., 1995]. Peripheral fatigue refers to inefficiency of the excitation-contraction coupling mechanism and (or) neuromuscular junction in reacting to the neural stimulation from the motor nerve [Gandevia, et al., 1995].

According to the theory of metabolic overload, prolonged exercise decreases muscle performance [Salmons, 1997]. The demand for adenosine triphosphate (ATP) exceeds its production, leading to a vicious cycle of calcium ion overloading of the cell and a further decrease in ATP production. Both insufficient supply of ATP and fatigue would induce lactic acid accumulation which would adversely affect the muscle contraction. Fatigue studies have reported decreases in muscle strength [Mercer et al.,2003; Nyland et al., 1997; Rahnama et al., 2003], impaired joint position sense [Rozzi et al., 1999; Skinner et al., 1986] and delayed neuromuscular responses [Gleeson et al., 1998; Mercer et al., 1998; Rozzi et al., 1999]. Muscle fatigue may contribute to running injuries by decreasing the ability of muscles to respond to or withstand the loads placed upon the runners.

Muscle fatigue can also alter the running biomechanics. Fatigued runners were observed to have increased hip internal rotation [Sanna & O'Connor, 2008] and rearfoot pronation [Derrick, et al., 2002] which are associated with various running injuries including PFPS. Change in normal running kinematics resulted from muscle fatigue, is therefore believed to be associated with an increased risk of injury.

1.6.1.1.4 Excessive rearfoot pronation

Excessive rearfoot pronation during touch down phase of a running cycle is typically associated with running overuse injuries [James et al., 1978]. Rearfoot pronation, which is defined as a combined movement of calcaneal eversion, forefoot abduction and dorsiflexion, is hypothesized to play an important role in shock absorption during the heel strike to mid-stance phase of walking or running by providing natural adaptive movement in the subtalar and midtarsal joints of the foot [Buchbinder et al., 1979; Perry & Lafortune, 1995]. Together with the natural knee flexion during this phase of running cycle, rearfoot pronation provides an inherent mechanism to prevent overloading of joints in the lower extremities [Lafortune et al., 1994; Leung et al., 1998]. Also, the rearfoot pronation unlocks the mid-tarsal joints, which enables the forefoot to become more supple and flexible, thus allowing it to adapt to uneven terrain [Hamill et al., 1992].

With the unique anatomical architecture of the subtalar joint, rearfoot pronation is always accompanied by internal tibial rotation [Hintermann & Nigg, 1998]. Therefore, excessive rearfoot pronation may result in erractic internal tibial rotation. To counteract the erratic tibial rotation, femur may also perform compensatory rotation [Levens, et al., 1949; Powers, et al., 2002; Reischl, et al., 1999; Cheung & Ng,

2006]. Thus, rearfoot overpronation not only induces higher local joint loading on the foot structures, but it may also alter the proximal joint kinematics and kinetics.

Excessive rearfoot pronation is associated with Achilles tendonitis and plantar fasciitis [Cook et al., 1990; van Mechelen et al., 1992] in the foot structure and with the kinetic linkage of the entire lower limb, it may also lead to the development of posterior tibial syndrome (shin splints) [Messier & Pittala, 1988; Viitasalo & Kvist, 1983] and patellofemoral pain syndrome (PFPS) higher up in the kinetic chain. [Cheung & Ng, 2007; Ghani Zadeh Hesar et al., 2009].

1.6.1.2 Major extrinsic factors

Long running distance [van Mechelen et al., 1992; Macera, 1992] is considered an extrinsic risk factor of running injuries because longer running distance induces muscle fatigue. In terms of skill level, recreational runners are more prone to injuries than professional runners [Cook et al., 1990] because professional runners usually have an access to running team support, which may include supervision of coaches and other medical support from physiotherapists and physical trainers.

1.6.2 Runners at risk

Running injury is very common among runners. Different risk factors may be

interrelated (FIGURE 1). Long running distance causes muscle fatigue whereas recreational runners are less fatigue resistant than professional runners due to their difference in level of training. Therefore, muscle fatigue problem may be more prone to affect long distance recreational runners. Also, as recreational runners are not supported by medical professionals, they may have accumulated minor injuries not properly managed or excessive rearfoot pronation problem which are not even identified.

Fatigued runners were found to have increased rearfoot pronation [Derrick, et al., 2002] and delayed neuromuscular responses [Gleeson et al., 1998; Mercer et al., 1998; Rozzi et al., 1999] which may induce muscle imbalance. As rearfoot pronation serves the purpose of shock absorption, erratic pronation may also induce unbalanced plantar loading.

According to the flowchart in FIGURE 1, it is speculated that the group of runners who are most vulnerable are:

1. Long distance recreational runners
2. With excessive rearfoot pronation
3. Under muscle fatigue

In order to develop the best strategy to prevent running injury, it is most appropriate to study the group that is most vulnerable. Leg muscle activation pattern and plantar

loading should be recorded to determine if these outcome measures are changed with the different testing conditions.

1.7 OBJECTIVES OF PRESENT INVESTIGATIONS

The functions of motion control footwear have not been well studied. There were controversies in previous findings on different aspects of motion control footwear and their functions and these functions may be associated with various running injuries. Thus, the main objectives of the present investigations are to examine the motion control footwear functions on kinematics control, kinetics change and relevant muscle responses in runners with different footwear.

1.8 SPECIFIC AIMS OF THE INVESTIGATIONS

There are several experiments in this study. Specifically, the aims of the investigations are to study:

1) the kinematics response in runners with overpronating feet with different running footwear in prolonged running;
2) the runners' feedback on the perception of running footwear functions;
3) the plantar force distribution with different running footwear during prolonged running;

4) the lower extremity muscle activities in runners with different running footwear in prolonged running.

1.9 HYPOTHESIS TO BE TESTED

1) Excessive rearfoot pronation can be reduced by motion control footwear.
2) Runners are able to differentiate the footwear function by their subjective sensory feedback.
3) The plantar force distribution is different when runners put on motion control footwear as compared to neutral cushioned footwear.
4) The muscle activity of lower extremity is different when runners put on motion control footwear as compared to neutral cushioned footwear.
5) The aforementioned functions of motion control footwear are maintained after muscles have fatigued.

1.10 ORGANIZATION OF THE THESIS

The methodology of the experiments is introduced in Chapter 2 and Chapter 3. In Chapter 4, 5, 6, & 7, results of four experiments are presented, aiming at testing the motion control footwear functions which include rearfoot kinematics control, kinetics

(plantar force distribution), and lower extremity muscle control, accordingly. The final chapter is the grand discussion and conclusion.

CHAPTER 2 METHODOLOGY FOR MEASURING THE REARFOOT KINEMATICS AND PLANTAR FORCE DISTRIBUTION DURING PROLONGED RUNNING

2.1 INTRODUCTION

Due to the scarcity of reports on the efficacy of motion control footwear in rearfoot pronation control and the load distribution under the foot, this experiment was conducted to examine the motion control footwear function in terms of rearfoot kinematics and plantar loading. Furthermore, in view of the fact that fatigue is a natural sequel of running and its detrimental effects on the running performance, thus study also aimed to examine the effects of fatigue on the outcome parameters to further enhance the quality of this experiment, several issues about the methodology were addressed.

2.1.1 Footwear model

The rationale of motion control footwear technology is originated by heel wedge (or flare). However, the application of heel wedge has become less common due to the fact that shoe wedge and flare induce higher risk of ankle sprain [Fordham et al., 2004]. Currently, motion control footwear is usually composed of duo-materials in the

midsole which allow anisotropic deformation. Also, the selection of motion control footwear with midsole material modification is simply because this kind of motion control footwear is more easily available in the market.

2.1.2 Testing protocol

The mainstream of motion control footwear is running shoe design. Running test was preferred as it would have higher clinical implications than walking. Also, running involves higher loading to the joints and larger joint movements compared to walking test [Ounpuu, 1994]. It can therefore enlarge the potentially small effect size of the outcome measures in the investigations.

2.1.3 Treadmill versus over ground running

In previous research, most of the running trials involved 10-20m [Bates, 1989]. Even though the data were collected in the middle phase of the run to eliminate acceleration and deceleration effects, this protocol did not simulate the real situation in runners' usual practice. To control the speed throughout the study, treadmill running is a better choice. Moreover, treadmill running in the gymnasium or physical fitness club is now a modern trend of exercise in most developed countries. Selection of treadmill running has therefore a higher clinical relevance to the study.

2.1.4 Running mileage

As stated in section 2.1.3, short running mileage trial has been commonly adopted in previous studies. Very few studies had looked at the effects of longer mileage on running performance e.g. middle to long distance running. Mileage of 1500m is a standard event in many international and local competitions. Even regular recreational runners are able to complete this mileage, thus suitable subjects would be more easily available.

2.1.5 Gender of subjects

As different running patterns were found between male and female runners [Decker, et al., 2003; Ferber, et al., 2003; Lephart, et al., 2002], a subject pool with mixed gender may contaminate the results. The present study only examined females because females are clinically more prone to have excessive rearfoot pronation problem due to higher joint flexibility [Riegger-Krugh & LeVeau, 2002].

2.1.6 Skill level of subjects

Only recreational runners were tested in this study. The reasons being were firstly, recreational runners are more prone to running injuries [Cook, et al., 1990]. Secondly, they might be innocent about their excessive rearfoot problem thus the recruitment of

re recreational runners could have a higher clinical impact. Thirdly, the fatigue resistance in these runners is lower. To study the effects of muscle fatigue on motion control footwear function, it may involve much longer distance of running test to induce muscle fatigue if professional or elite runners were tested. Therefore, selection of recreational runner is suggested.

2.1.7 Foot type screening

In the attempt to ensure the subjects were overpronators i.e. subjects with excessive rearfoot pronation, all the subjects had to go through a screening of foot type. The reason for such screening was because the motion control footwear was specifically designed for overpronators, subjects with normal foot type may not benefit from the motion control footwear design and this could lead to a statistical type II error.

2.1.8 Symptoms

The running pattern can be altered when runners were in pain. Subject standardization may be a problem with a mixed subject pool of different levels of pathology, location of problem and stage of problem. Therefore, this study had only focused on testing subjects who were asymptomatic.

2.2 CHOICE OF SUBJECTS

Initially, twenty-eight female recreational runners were recruited, but after the first assessment, 3 subjects were excluded because their rearfoot pronation had fallen short of the required range (For details of rearfoot pronation screening, please refer to section 2.5.3.3). For the remaining 25 subjects, their mean age was 23.5 ± 6.8 years. Other demographic data about the subjects are presented in TABLE 1.

2.2.1 Inclusion criteria

1 Female subjects

2 Recreational runners

- With regular practice (at least once per week) for six months or more
- Non-professional runners and not supported by any running club

3 Healthy in terms of musculoskeletal and cardiopulmonary functions

4 Overpronators

- The functional definition of overpronator will be described in details at section 2.5.3.3.

Table 1 Demographic data of subjects recruited in Chapter 2

Age	23.6 ± 6.8
Running Experience (Yrs)	3.6 ± 3.2
Average Mileage / Week (km)	2.1 ± 1.2
Average Frequency of Run/ Week	2.6 ± 1.4
Height (m)	1.55 ± 0.07
Weight (kg)	46.3 ± 4.0
Body Mass Index (kgm^{-2})	19.3 ± 1.4

2.2.2 Exclusion criteria

1 Male subjects

2 Professional runners

- Member of any running clubs
- Athlete of any district or Hong Kong (or any other country) running team

3 Subjects with any known active musculoskeletal and cardiopulmonary problems that required medical consultation within one year

4 Subjects with normal foot type.

2.3 LOCATION AND DURATION OF THE STUDIES

All experiments were performed in the Motion Analysis Laboratory of Department of Rehabilitation Sciences at The Hong Kong Polytechnic University. There were 2 running sessions with 1-week apart. Each session consisted of the same testing procedures except that the footwear condition was different. The test procedure in each session involved pre-fatigue and post-fatigue outcome measurement, with fatigue procedure in between the two measurements.

2.4 ESTABLISMENT OF EFFECTIVE FATIGUE PROCEDURE FOR REARFOOT SUPINATORS

2.4.1 Functional definition of muscle fatigue

In this study, “muscle fatigue” is functionally defined as the drop of force produced by a maximum voluntary contraction (MVC) of a single physiological movement after exercise. This method of defining muscle fatigue simulates the clinical situation in measuring muscle strength, and should be more clinically relevant.

2.4.2 Selection of target muscle group

Because the major counter force against rearfoot pronation is the force produced by the rearfoot supinators, in this study, the rearfoot supinators were therefore examined.

2.4.3 Assessment for muscle fatigue in rearfoot supinators

Hand held dynamometer (MicroFet2 Force Gauge Testing Device, Hoggan Health Industries, Utah, United States of America) was used to measure the isometric torque produced by the rearfoot supinators. Good test-retest reliability of this device

was shown in previous studies [Kwoh, et al., 1997; Malliopoulos & Thevenon, 2002]. A preceding pilot study involved 10 healthy subjects had also revealed excellent intra-rater reliability (ICC (3,1) =0.9695) of the measurement. To further promote the accuracy of the measurement, the dynamometer was fixed on a stool so as to minimize the dynamometer from shifting as a result of the force difference.

The subjects were asked to produce an MVC of isometric rearfoot supination on the dynamometer fixing on a stool against the wall (FIGURE 2) for 3 times. Mean value derived from 3 repetitions of MVC would be regarded as the ability of muscle to counteract rearfoot pronation.

2.4.4 Establishment of fatigue procedure

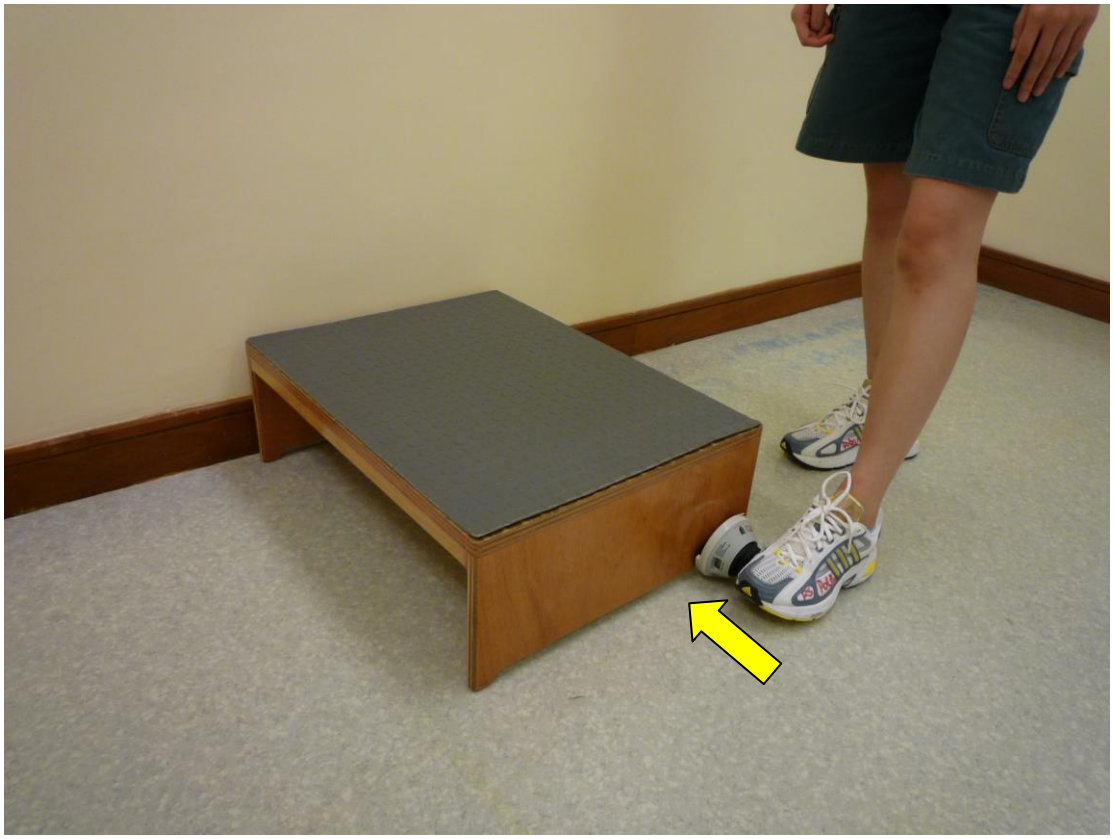
In order to examine the duration and the format of exercise to effectively induce muscle fatigue in rearfoot supinators, a total of 3 pilot studies had been conducted

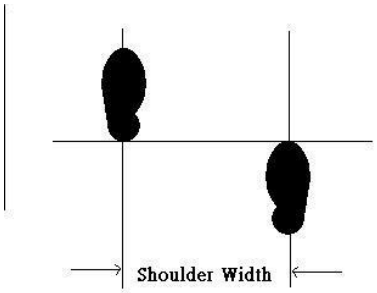
2.4.4.1 Pilot test A – Running on trampoline

To induce fatigue of rearfoot supinators, running on a trampoline was one of the options because it required the use of the calf and lower leg muscle group extensively during landing.

Two healthy recreational runners were recruited in this pilot study. They were

Figure 2 Dynamometer setup. The subject was asked to produce an isometric rearfoot supination against the dynamometer which fixed at the stool. The stool was placed next to the wall to provide a firm platform for force measurement. The foot placement should be in shoulder width apart with tested leg one step length in front. Yellow arrow indicates the direction of action.





free of any known musculoskeletal problems in their lower extremities including past history of ankle or foot injuries that required medical treatments, and they have normal foot arch as judged by physical examination.

Prior to the running test, a baseline MVC was taken according to the methodology described in section 2.4.3. After the initial MVC measurement, subjects were instructed to mimic the action of running by stationary stepping barefoot on a trampoline at a frequency of 3Hz by following the pace of a metronome (FIGURE 3). The MVC of rearfoot supinators was retested in every 3 minutes. The whole measurement only took 10-20 seconds so as not to allow the fatigued muscle to recover. The MVC level was monitored for 10 intervals (i.e. half hour running).

Results of the two subjects' MVC were shown in TABLE 2. Both subjects did not show any trend of decrease in the rearfoot supinators performance after the 30-minute step running on a trampoline.

There were two possible reasons for the failure of this fatigue procedure not being effective:

1) The running pattern on the trampoline was not standardized. The subjects were only asked to mimic the action of running but they might not have used the muscles in their usual running pattern due to the difference in kinematics as compared with actual running.

Figure 3 Running on trampoline. Subjects were instructed to mimic the running action on trampoline by stationary stepping barefoot.

Table 2 Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination in pilot test A

	0'	**3'**	**6'**	**9'**	**12'**	**15'**	**18'**	**21'**	**24'**	**27'**	**30'**
Subject A	47	47	48	42	44	48	46	44	47	45	46
Subject B	32	32	30	33	31	30	31	31	31	32	31

Number represents the torque produced by the rearfoot supinators (kgf)

Table 3 Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination in pilot test B

	0'	**5'**	**10'**	**15'**	**20'**	**25'**	**30'**
Subject A	23	13	10.5	(18)	(22)	(22)	(22)
Subject B	58	39	34	33	18	(30)	(53)
Subject C	38	34	32	36	24	28	27
Subject D	27	28	20	26	24	26	25
Subject E	27	19	19	18	20	21	19

Number represents the torque produced by the rearfoot supinators (kgf)

Number in blanket indicates the torque production during rest

2) The subjects produced strong intrinsic muscle power and thus their foot arches were supported. Their rearfoot supinators might have higher fatigue resistance. The situation could be different if the subjects were having excessive rearfoot pronation problem.

3) Subjective report from the participants suggested that the muscle recruitment in trampoline running task was mainly on the thigh muscles e.g. quadriceps and hamstrings instead of calf muscles.

Owing to the above reasons, other studies were conducted to develop a better method to induce muscle fatigue in the rearfoot supinators.

2.4.4.2 Pilot test B – Running on sand

The results of pilot test A did not suggest that running on a trampoline was an effective method to induce significant muscle fatigue in the rearfoot supinators. It could be due to the subjects' normal foot arch which had higher resistance against fatigue. Therefore, only subjects with pronated foot were recruited in pilot study B.

A total of 5 healthy subjects (3 females and 2 males) with clinically overpronating foot were recruited in this pilot study. The foot posture measurement was referred to the procedures reported by Keenan & Bach [2006]. The subjects were

otherwise free of any known musculoskeletal problems in their lower extremities, including past history of ankle or foot injury. Three of them did not have any exercise habit before the test. One subject was a recreational runner and the other was an amateur dancer.

The procedure to measure MVC was the same as in Pilot study A. Prior to the fatigue exercise, a baseline MVC was taken in the standard position mentioned.

This pilot test was conducted on a sandy beach. The reason for running on sand was because the sandy surface would not produce a firm reaction force thus the foot structures might be deformed more and the foot muscles would be working in a less physiologically favorable length. After initial MVC measurement, the subjects were instructed to run barefoot on sand with stepping frequency at 3Hz, which was controlled by a metronome. The MVC of the rearfoot supinators was retested in every 5 minutes. Once the measured MVC was less than 50% of the baseline MVC, the subjects were asked to rest in a sitting position.

The MVC of the subjects is illustrated in TABLE 3. According to the results, only 2 subjects had developed muscle fatigue (more than 50% drop of initial MVC) with this protocol of running on sand. Also, the fatigue was only temporary as recovery occurred within 10 minutes in both subjects. The other 3 subjects only developed mild fatigue with 10 – 40% of MVC drop. Among those 3 subjects who did

not develop muscle fatigue with this method, one of them had exercise habit and the others had not.

Running on sand aimed to accelerate the fatigue rate of the rearfoot supinator muscles in a functional manner. In this pilot study, it showed that the supinator muscles were not fatigued after running on sand. Moreover, the recovery rate was also very fast with MVC returning to almost 90% of its initial MVC after a 10-minute rest. Also, the efficacy of this fatigue procedure seemed not to relate with the exercise habit of the subjects.

All the 5 subjects complained of shortness of breath rather than foot muscle fatigue during the testing procedure. The limitation of the exercise method is that it may overload the cardiovascular system more than the muscular system.

2.4.4.3 Pilot test C – Treadmill running

In view of the previous two pilot studies had failed to induce localized muscle fatigue on foot supinator muscles, a third protocol was therefore developed in the attempt to effectively fatigue the supinator muscles. Five recreational runners with overpronating feet were recruited.

The MVC measurement procedures were the same as the previous 2 pilot studies. After the baseline MVC measurement, the subjects were asked to run on a treadmill at

a speed of 10 kilometres per hour i.e. 2.78 meters per second without any inclination for 1500m. The selection of running speed and mileage were based on the average training dosage of the runners.

After 1500m run, the subjects' MVC were retested. The MVC values were reduced by around 50% of their original value in all the subjects (TABLE 4). The runners subjectively reported not much shortness of breath and no other discomfort during and after the test. The success of this fatigue procedure may be due to the following:

1) The fatigue procedure was not interrupted by MVC measurement.
2) The exercise selected was functionally specific and able to simulate normal running.
3) The subjects were overpronators thus there would be higher demands on their supinators in foot kinematics control.

Therefore, the aforementioned procedure was selected as the fatigue procedure in the main study.

2.5 EQUIPMENT

2.5.1 Footwear

Two highly comparable shoe models were used in this study. For motion control

Table 4 Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination in pilot test C

	MVC before run	MVC after 1500 m run
Subject A	34	12
Subject B	22	10
Subject C	23	10
Subject D	49	23
Subject E	40	18

Number represents the torque produced by the rearfoot supinators (kgf)

footwear, Adidas “Supernova control” was selected whereas Adidas “Supernova cushion” was selected as the control footwear condition (FIGURE 4). The construct of both shoe models were similar except for the midsole material. The “Supernova cushion” composes of a single midsole material which does not have pronation control function. The shoe was designed to reduce the impact rate but it has little effect on foot motion control and there is no reported difference in the kinematics profile between barefoot running and running with neutral footwear [Stacoff, et al., 1996; Stacoff, et al., 2001]. The “Supernova control”, however, comprises two materials with different hardness in the lateral and medial midsole and it was designed to check excessive foot pronation [Olympic Series Sports Medicine Workshop II, 2003].

2.5.1.1 Footwear size

The subjects were allowed to try different sizes to identify the one that fitted their feet best. The size was recorded so that the subjects put on the same size of shoes in different test models. The corresponding size of insole sensor was matched to the runners’ shoe size.

Figure 4 Test shoe models: Adidas Supernova Control (motion control footwear) & Adidas Supernova Cushion (neutral footwear)

Supernova control, Adidas

Supernova cushion, Adidas

2.5.1.2 Shoe box tightness standardization

To ensure that the data were collected under similar shoe fitting conditions, the same pair of laces was used in the two shoe models for each subject. This pair of shoe laces had color markers on both ends which could guide the positions of knots thus standardizing the tightness of the shoe fitting for the subjects (FIGURE 5).

2.5.2 Treadmill

Treadmill running was adopted in the test because it was more stable and easier to control than running on a track around a field. Also, it was shown to be an effective method to produce muscle fatigue in rearfoot supinator in the previous pilot test of section 2.4.4.3. Even though minor differences such as shorter stride length, higher cadence and a shorter non-support phase had been reported to exist in treadmill running, these differences could be minimized with practice [Nigg, et al., 1995]. A running speed at 10 km per hour videlicet 2.78 meters per second was used.

2.5.3 Motion recording

A Vicon 3-dimensional motion analysis system (Hardware Model: V-370; Software Model: Workstation 4.0, Oxford, UK) with 3 cameras was used to capture the left lower extremity movements during the run (FIGURE 6). After a set of

Figure 5 Standardized shoe lace: 4 red markers (indicated by arrows) on the lace to standardize the tightness of shoe box. After subject adjusted their usual tightness of shoe box by adjusting the tension of shoe lace, 4 markers were highlighted on the lace for that particular subject so that tightness of shoe box was standardized in the next running session.

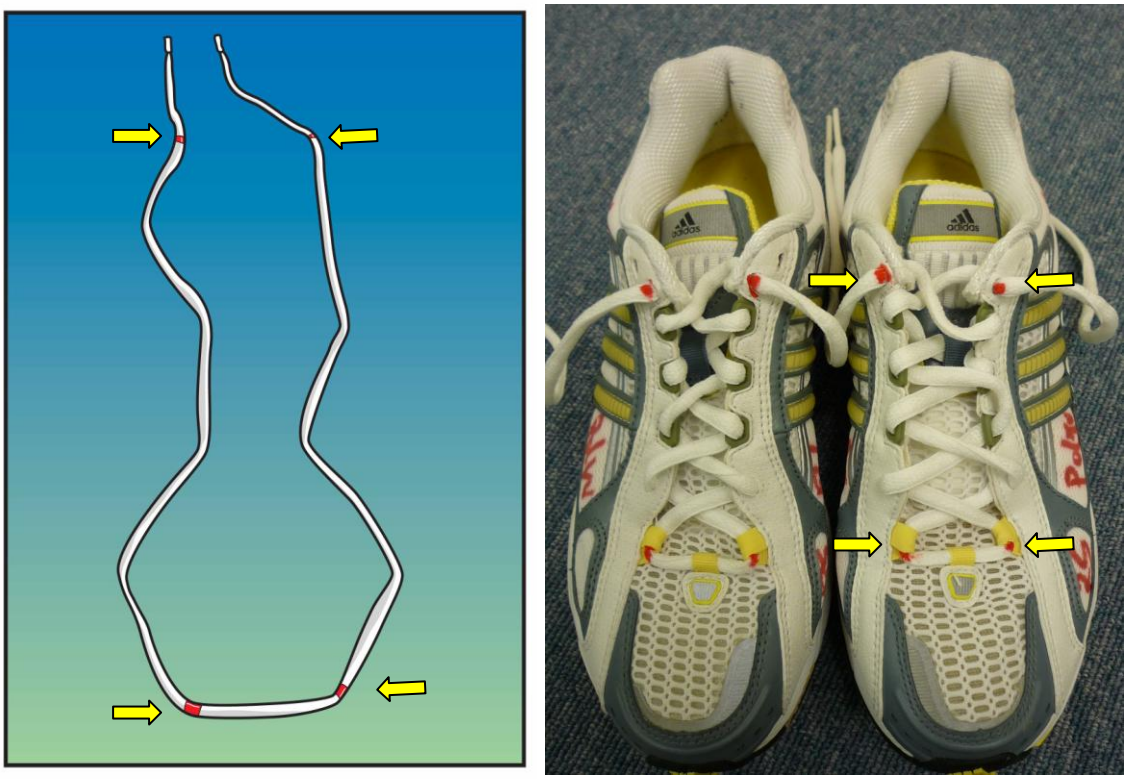


Figure 6 Motion capture system setup. Three VICON cameras were positioned at left rear corner of the treadmill to capture the markers on the left leg of runners.

standard calibration procedures, the left lower extremity was filmed at 60 Hz. Although higher sampling frequency was preferred, this rate was regarded as acceptable in slow to moderate running studies [van Gheluwe & Madsen, 1997].

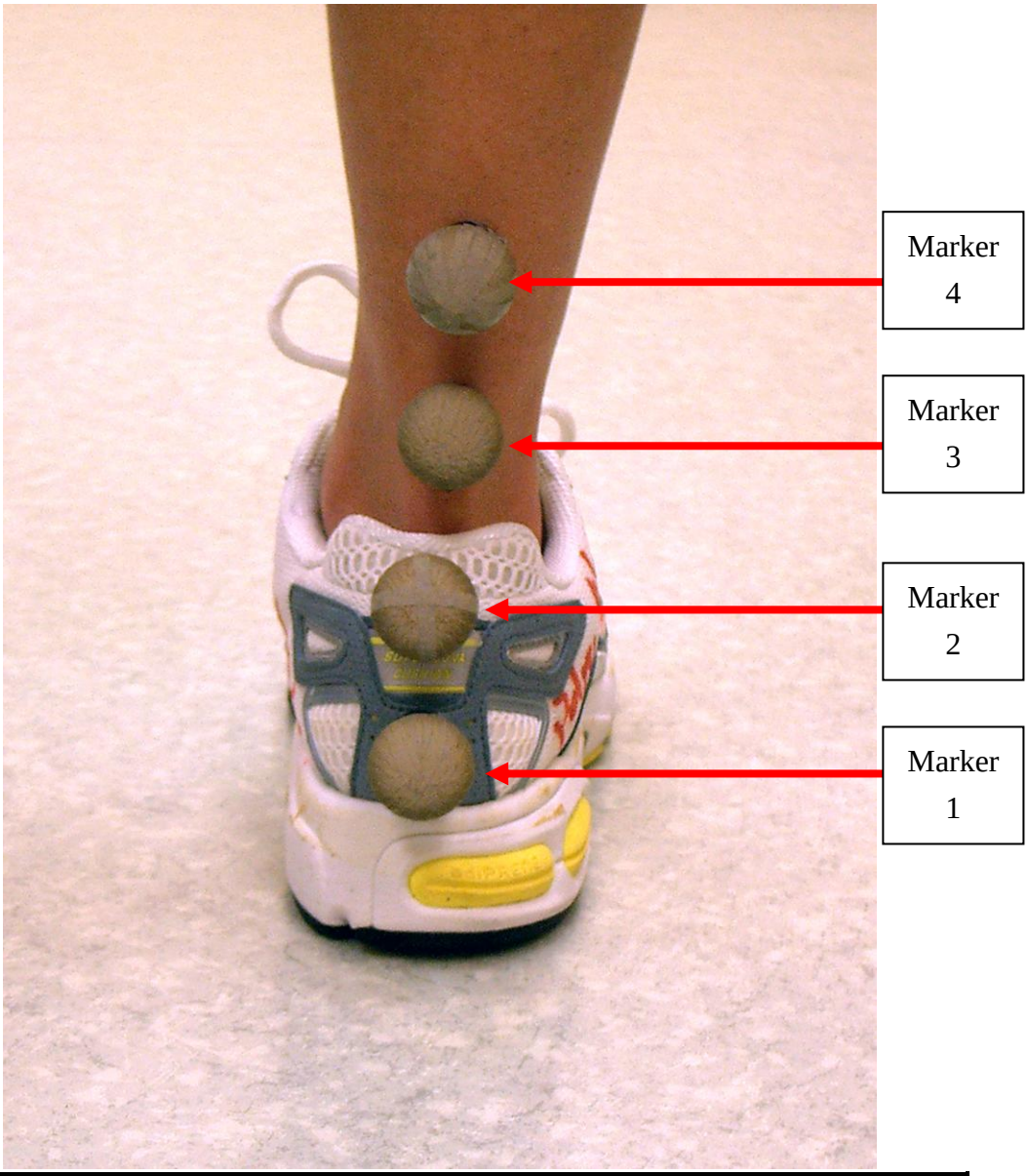
2.5.3.1 Marker positions

Four light reflective spherical markers were attached on the subjects (FIGURE 7) with reference from the methods of previous studies [Nigg & Morlock, 1987; Perry & Lafortune, 1995; Stacoff, et al., 1991]. The line between Markers 1 and 2 formed vector a (Va) and that between markers 3 and 4 formed vector b (Vb). The rearfoot angle was defined as the acute intercept angle between these two vectors using the formula:

$$\text{Rearfoot angle} = \text{Arcos}\,(Va \cdot Vb \,/\, |Va| \times |Vb|)$$

The above formula for calculating rearfoot eversion-inversion has been previously reported in the literature and the emphasis was on the maximum angle [Nigg & Morlock, 1987; Perry & Lafortune, 1995; Stacoff, et al., 1991; van Gheluwe & Madsen, 1997] which relates with various overuse injuries. Despite the fact that the aforementioned measurement might include an error of translation movement

Figure 7 Marker position for motion capturing



Marker	Location of Markers
1	Center of heel cap just above the shoe sole
2	Center of heel cap at the insertion of the Achilles tendon
3	Center of the Achilles tendon at the height of medial malleolus
4	15 cm above marker 3 at the center of the leg

between the shoe and the foot, this measurement demonstrated a similar motion pattern compared with skeletal motion measured by bone markers [Reinschmidt, et al., 1997]. To ensure repeatability of the study, the positions of the reflective markers were recorded as offset in a static manner before every running trial. The cameras were within 2m from the subject so as to maximize the resolution of the recorded images. The test-retest reliability of this setup was found to be very strong with ICC (3,25) of 0.89.

2.5.3.2 Definition of rearfoot pronation

The primary kinematics outcome was the maximum rearfoot pronation angle, which was defined as the pronation range of motion travelled from the time point of heel strike till the maximum amplitude of angle. However, due to the limitation of the VICON data capturing frequency, other important rearfoot motion parameters such as rearfoot angular velocity and acceleration were not measured in this study.

2.5.3.3 Definition of excessive rearfoot pronation

The screening consisted of both physical examination and objective examination. Physical examination involved a static measurement of subtalar neutral position according to two previous studies [Root, et al., 1971; Keenan & Bach, 2006].

After the physical examination, subjects with clinical signs of overpronating feet were invited to proceed to the testing. The first testing session involved running on a treadmill with neutral footwear which aimed to identify the amount of foot pronation and running pattern of the subjects. Subjects with maximum rearfoot angle higher than 6° compared with the offset angle measured in standing would proceed to the next testing session of treadmill running with motion control footwear at one week later. Amplitude of 6° was selected because it was around the average cut-off rearfoot motion range to be regarded as being excessive, which is derived from the mean cut-off value obtained from the studies of Eng & Pierrynowski [1994], Hintermann & Nigg [1998], Johnson et al. [1994] and LeLievre [1970].

2.5.4 Plantar loading measurement

Plantar force measurement and detection of heel strike were performed with an insole sensor (Pedar Novel GMBH, Ismaninger Strasse 51, Munich, Germany) in the left shoe. Each insole comprised 99 sensors covering the whole plantar surface with each sensor having a sampling frequency of 99 Hz. The thickness of the insole was less than 1 mm so that it has minimal disturbance to the normal gait pattern. To further reduce the risk of normal running pattern disturbance, another insole sensor

was placed in the right shoe as well but data from that insole sensor were not recorded.

This equipment has been reported to have good to excellent test-retest reliability (ICC ranged 0.84-0.99) in both walking and running and it was regarded as one of the most accurate instruments for in-shoe kinetic measurements highly comparable to the force plate and F-scan measurements [Barnett, et al., 2001; Quesada, et al., 1997; McPoil, et al., 1995; McPoil & Cornwall, 1997; Kernozek & Zimmer, 2000].

In each testing session, the system was calibrated before use. The left insole sensor was connected by leads to an 8-bit analog-to-digital converter box, which was suspended on the railing of the treadmill. The converter box was connected to a laptop computer for data collection (FIGURE 8). Another insole sensor in the right shoe was not connected so that the running pattern was not further disturbed by extra wirings.

2.5.4.1 Zonal analysis

To quantify the loading in different areas of the foot, the insole was subdivided into 10 anatomical regions (FIGURE 9) according to the suggestion by Cavanagh [1992]. These different zones corresponded to specific foot structures (FIGURE 9), including medial heel, lateral heel, medial midfoot, lateral midfoot, 1st metatarsal head, 2nd & 3rd metatarsal head, 4th & 5th metatarsal head, hallux, second toe and lateral

Figure 8 Insole loading sensor setup

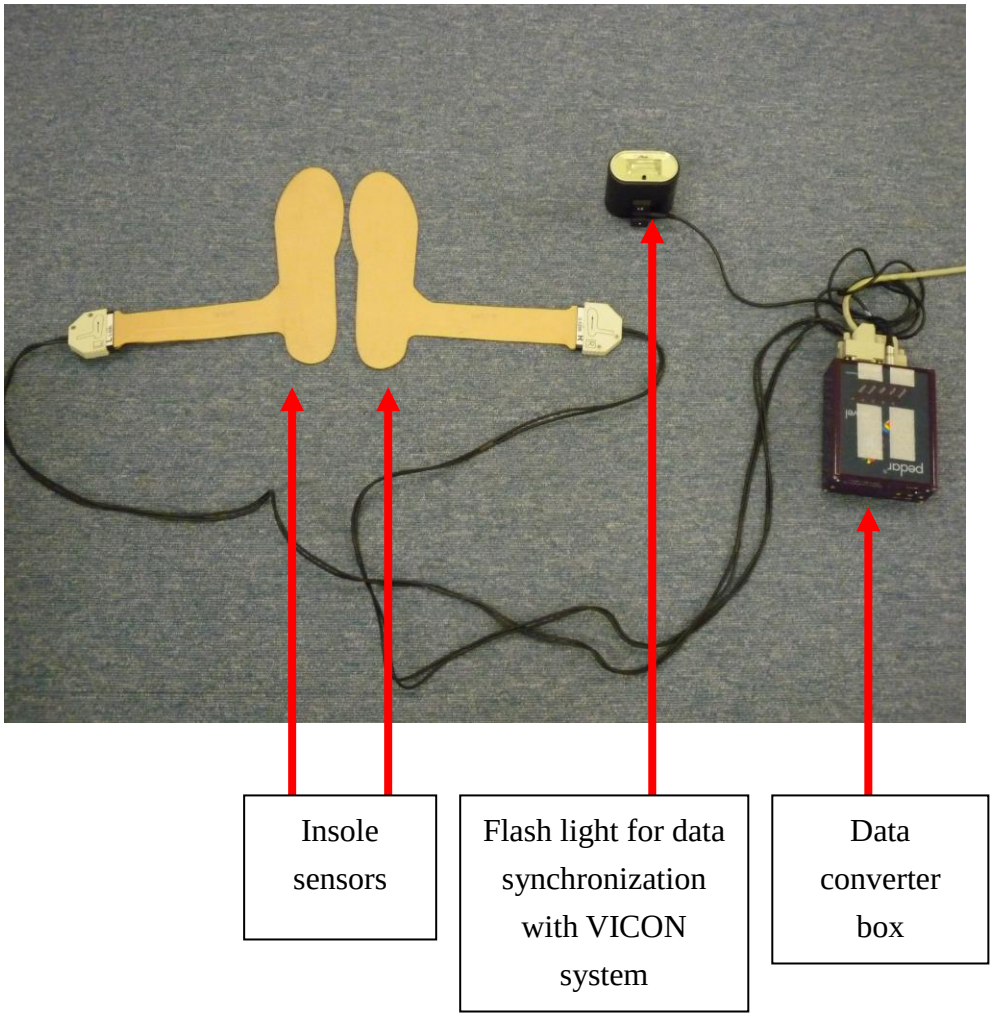
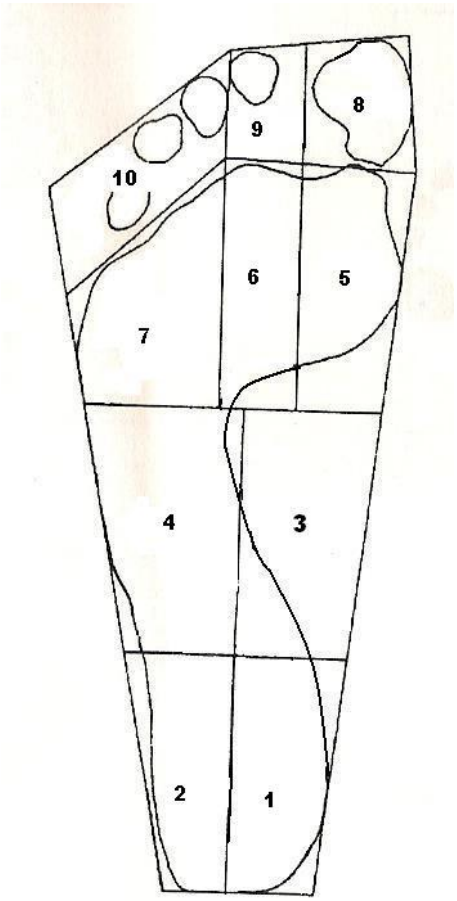


Figure 9 Insole zonal divisions. Ten anatomical related areas were divided to represent different loading corresponding to different foot structures.



Area Number	Anatomy-related Area	Area Number	Anatomy-related Area
1	Medial Heel	6	2nd & 3rd Metatarsal Head
2	Lateral Heel	7	4th & 5th Metatarsal Head
3	Medial Midfoot	8	Hallux
4	Lateral Midfoot	9	Second Toe
5	1st Metatarsal Head	10	Lateral Toes

toes.

2.5.5 Synchronization between motion capturing system and plantar loading measurement

A photocell was used to detect a flashlight (FIGURE 8) produced by the sensor insole system (Pedar Novel GMBH, Ismaninger Strasse 51, Munich, Germany). The light signal would be converted into an analog signal, which was fed into the Vicon 3D motion analysis system for synchronization.

2.5.6 Quantification of muscle fatigue

Hand held dynamometer (MicroFet2 Force Gauge Testing Device, Hoggan Health Industries, Utah, United States of America) was used to measure the isometric torque produced by the foot supination immediate before and after the running session. The procedures of measurement were described in section 2.4.3.

2.5.7 Subjective feedback on footwear function in rearfoot pronation control

A validated questionnaire from Mundermann, et al. [2002] was used to assess the subjects' subjective feeling on the comfort level of the footwear. There were nine domains covered by the questionnaire (APPENDIX I). The analysis was focused on

the domain "medio-lateral control" of shoes only. Subjects were asked to mark on a 15-cm long visual analog scale after running with each footwear condition. A score was calculated in terms of a "0 to 1" scale. Zero score indicates poor medial lateral control, and 1 indicates good control in this aspect. This outcome measurement was to examine if the runners were able to differentiate the functional difference between the two tested footwear models.

2.6 TESTING PROCEDURE

2.6.1 Running sessions arrangement

This study involved 2 test sessions at 1-week apart. Before each session, a standardized warm-up stretching exercise program was prescribed. Ample time was given for the subjects to adapt to the treadmill running in each session. In the first session, subjects were asked to complete a 1500m run on the treadmill in neutral footwear condition. Subjects wore motion control footwear in the second running session one week later and the running speed and distance were identical to the first session.

2.6.2 Standardized warm up exercise

Prior to each test session, subjects were asked to complete a standardized warm

up exercise protocol (FIGURE 10). The exercise protocol involved stretching exercise of lower extremities and low resistance cycling exercise. The warm up exercises would last for about 10 minutes and then the test would start within 1 minute after the warm up exercises.

2.6.3 Kinematics and kinetics data collection and data management

To standardize the acquisition of data, kinematics and kinetics data were collected in 25 running steps before and after the 1500m running bout which were regarded to be representative of the initial (fresh) and end (fatigue) phase of the test. The flow chart of the investigation was illustrated in FIGURE 11.

2.7 STATISTICAL ANALYSIS OF DATA

Repeated measures ANOVA (general linear model), with a Bonferroni adjusted significance level at 0.025 ($\alpha = 0.05 \div 2$), was used to test the maximum rearfoot pronation on the effect of fatigue and shoe type. Pearson correlation was used to establish association between the level of muscle fatigue and kinematics variable. Paired t-test was used to detect difference between subjects' feedback on the level of comfort for each shoe model.

Another general linear model analysis was used to test the effects of timing and

Figure 10 Standardized warm-up exercise protocol before running tests

The standardized warm-up exercise comprised:

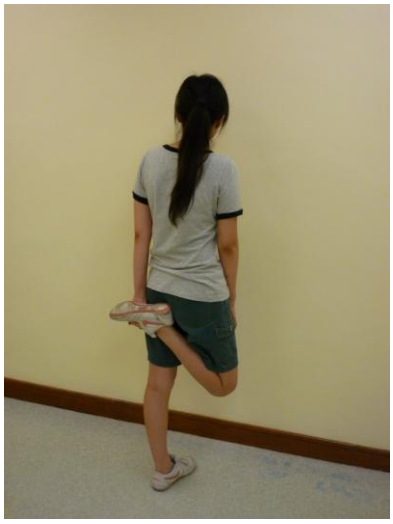
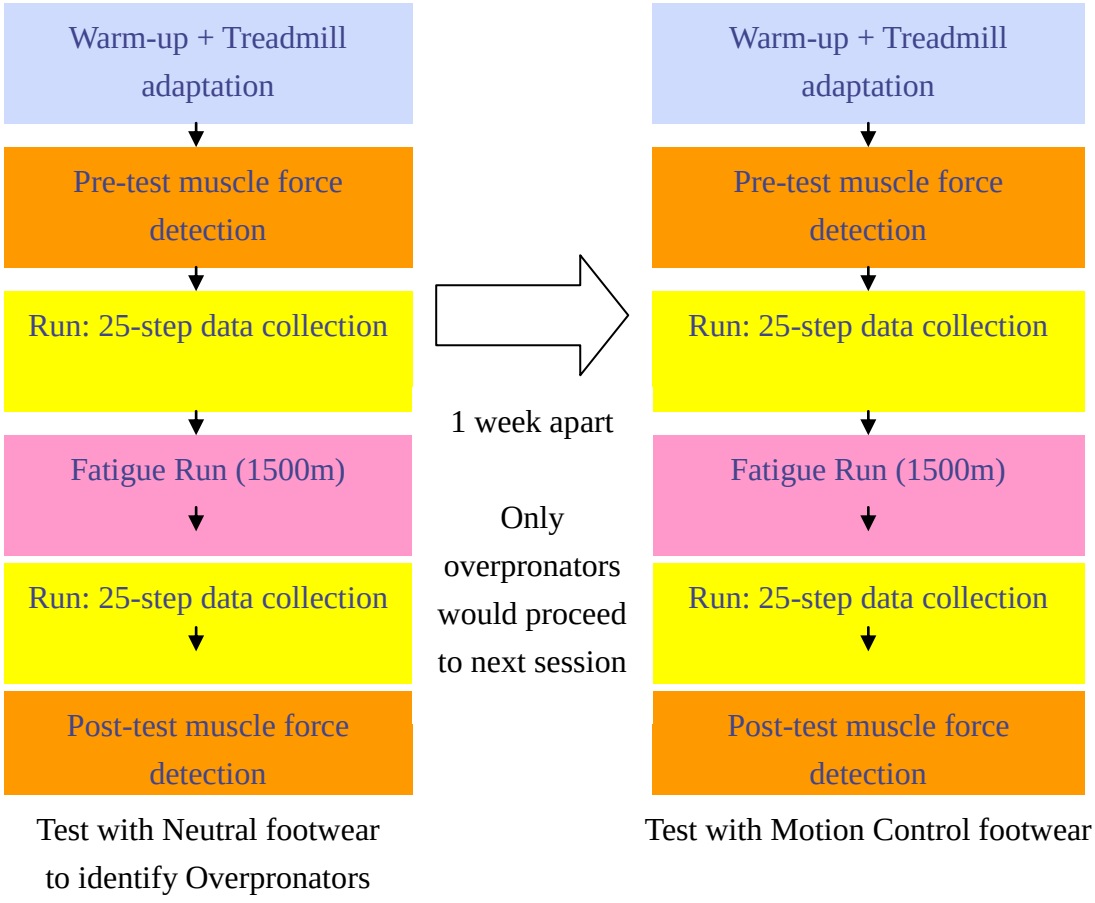
	
1) Stretching of quadriceps muscle in standing (15 seconds hold x 10 repetitions for each leg)	2) Stretching of calves muscle in lunging position (15 seconds hold x 10 repetitions for each leg)
3) Stretching of hamstrings muscle in lunging position (15 seconds hold x 10 repetitions for each leg)	4) Cycling of low-resistance static bike x 5 minutes

Figure 11 Flowchart of running sessions in Chapter 2 experiment



shoe type on the regional mean peak force. Both footwear model and timing were treated as the within factors. Since the tests were done repeatedly for the 10 anatomical regions, Bonferroni adjustment was applied and α was adjusted to 0.005 ($\alpha = 0.05 \div 10$). Simple paired t-test was used to detect the differences between subjects' MVC of foot supination before and after the 1500m running bout.

2.8 ETHICS

The present investigation was reviewed and approved by the Ethics Review Committee of Department of Rehabilitation Sciences at the Hong Kong Polytechnic University (APPENDIX II) prior to the commencement of the study. Written informed consent (APPENDIX III) was obtained from each subject prior to testing.

CHAPTER 3 REARFOOT KINEMATICS DURING RUNNING WITH MOTION CONTROL FOOTWEAR

3.1 INTRODUCTION

This chapter describes the results (kinematics part) of the experiment described in Chapter 2. In this experiment, 25 female recreational runners with excessive rearfoot pronation, were asked to run on a treadmill for 1500m in 2 sessions. In each session, subjects put on either motion control footwear or neutral footwear. Rearfoot motion was captured and level of muscle fatigue was monitored by dynamometer. Also, subjects were asked to rate the subjective feedback on the footwear function on the rearfoot control through a validated questionnaire.

3.2 RESULTS

3.2.1 Rearfoot kinematics

Mean rearfoot pronation in different footwear conditions before and after muscle fatigue were illustrated in TABLE 5. Before 1500m running bout, mean rearfoot pronation in motion control footwear and neutral footwear conditions were 10.6 ± 3.53° and 13.9 ± 3.25° respectively. After the running bout, the rearfoot pronation in

Table 5 Change in rearfoot pronation in different footwear before and after muscle fatigue by 1500m run

Rearfoot Angle (Mean ± SD)	**Neutral Shoe**	**Motion Control Shoe**
Before 1500m run	13.9 ± 3.25	10.6 ± 3.53
After 1500m run	17.7 ± 3.14	11.2 ± 3.86

neutral footwear condition increased to 17.7 ± 3.14°. In motion control footwear condition, the rearfoot pronation was 11.2 ± 3.86°.

Results of the ANOVA revealed that the rearfoot pronation angles were different (FIGURE 12) with the footwear and states of fatigue ($F=17.126$; $p<0.01$). Pair-wise comparison revealed marginally insignificant difference in rearfoot pronation ($p=0.06$) when running with motion control footwear before and after muscle fatigue.

The change in rearfoot pronation in motion control footwear condition after fatigue was 0.7° (95% C.I. -0.3° to 1.4°). However, when running with the neutral footwear, the rearfoot pronation increased by 6.5° (95% C.I. 4.7° to 8.2°) ($p<0.01$) when the subjects were exhausted. Furthermore, rearfoot pronation with the neutral footwear was higher than that of motion control footwear in both the pre- ($p<0.01$) and post-fatigue test ($p<0.01$).

3.2.2 Level of muscle fatigue

The force generated from maximum voluntary contraction (MVC) of rearfoot supinators in different footwear conditions before and after exercise was shown in TABLE 6.

Numerically, the MVC produced by rearfoot supinators was reduced by 30% - 40% in both footwear conditions ($p<0.01$) after running for 1500m. However, there

Figure 12 Mean and standard deviations of rearfoot pronation in different footwear conditions before and after muscle fatigue

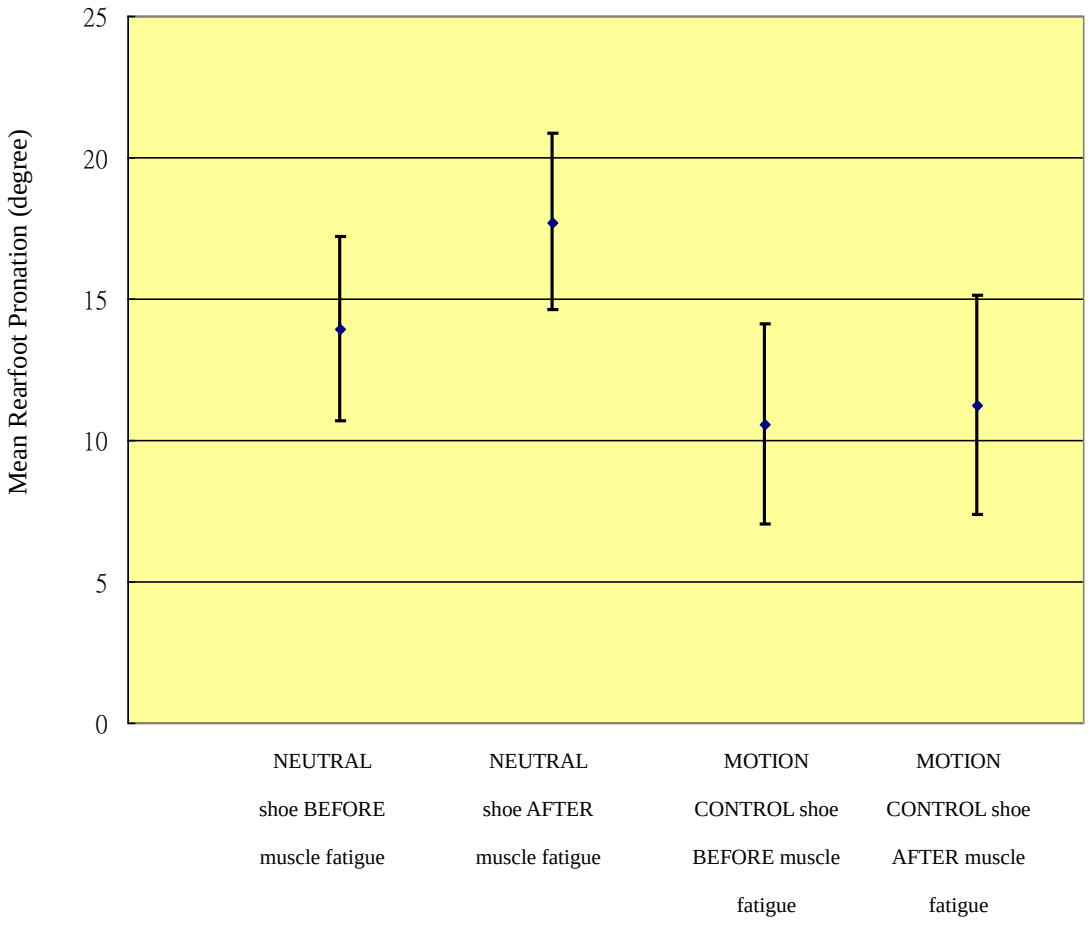


Table 6 Change of subjects' maximum voluntary contraction (MVC) of rearfoot supination before and after 1500m run

MVC Output (Mean ± SD)	**Neutral Shoe**	**Motion Control Shoe**
Before 1500m Run	16.5 ± 4.7	16.2 ± 4.0
After 1500m Run	10.7 ± 3.9	10.2 ± 3.3

was no significant correlation between the level of muscle fatigue and any kinematics variables (TABLE 7).

3.2.3 Subjective feedback on rearfoot motion control function of footwear models

Representing by a score from 0 (worst rearfoot control performance) to 1 (best rearfoot control performance), the average rate of motion control footwear and neutral footwear were 0.54 ± 0.17 and 0.55 ± 0.16 respectively (FIGURE 13). There was also no statistically significant difference between motion control footwear and neutral footwear (p=0.711) in the subjective feedback on medio-lateral control.

3.3 DISCUSSION

The present experiment aimed to study the rearfoot kinematics when running with different footwear before and after fatigue of the lower leg muscles in recreational female runners who had increased rearfoot pronation. Results demonstrated significant difference in rearfoot kinematics when subjects were running with different footwear.

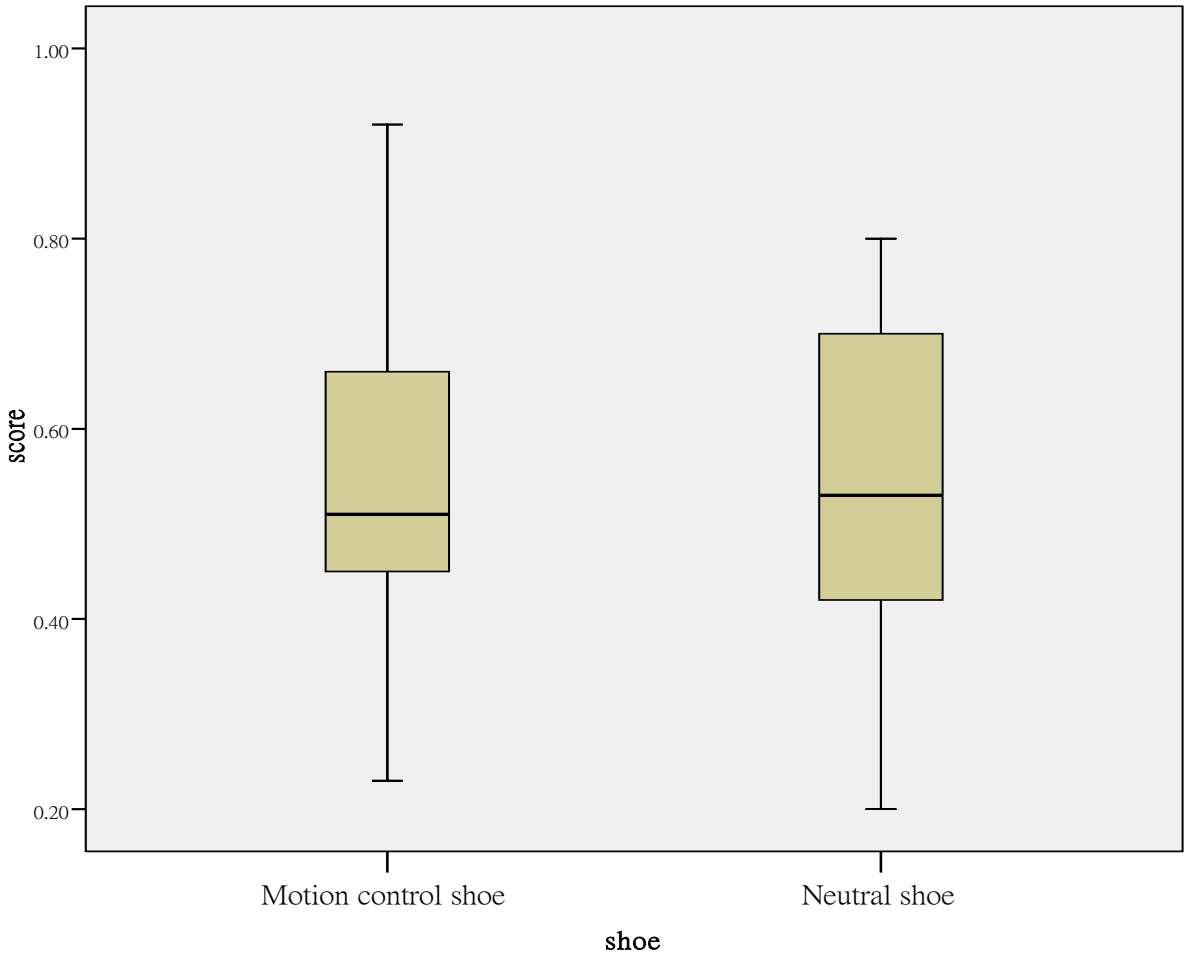
3.3.1 Rearfoot kinematics with motion control footwear

All the subjects had clinical evidence of rearfoot overpronation and the results

Table 7 Correlations between change of rearfoot pronation and maximum voluntary contraction production difference. The negative sign means a decrease of the parameter

	Change in Rearfoot Pronation (degree)	**Change in Maximum Voluntary Contraction (kgf)**	**Correlations**	**P**
Neutral Footwear	3.78 ± 4.07	- 3.00 ± 2.78	0.016	0.940
Motion Control Footwear	0.67 ± 1.69	- 2.57 ± 1.78	0.036	0.865

Figure 13 Subjective feedbacks on medial/ lateral control of footwear conditions represented by box-plot graph. Score of 1.0 and 0.0 indicated the best and the worst performance of the running footwear respectively.



revealed that they had larger rearfoot pronation when wearing neutral footwear than motion control footwear. With the motion control footwear, rearfoot pronation was reduced by around 4°. Results of this study are in line with some previous studies [Clarke et al., 1983; Perry & Lafortune, 1995; Nigg & Morlock, 1987; Stacoff et al., 2001; Butler, et al., 2006] despite the fact that the methodology in terms of footwear designs, running speed and experimental set-up were different amongst these past studies.

After the rearfoot supinators, which is major antagonist of rearfoot pronation, was fatigued, rearfoot pronation difference between motion control footwear and neutral footwear increased from 4° to 6.5°.

3.3.2 Effects of muscle fatigue

Increased rearfoot pronation in the neutral footwear condition was found to be more severe after muscle fatigued. With fatigue of the rearfoot supinators, the rearfoot pronation difference between the two tested shoe models was increased to 6.5°. This increase of rearfoot pronation after muscle fatigue has not been reported in any previous studies [Clarke et al., 1983; Perry & Lafortune, 1995; Nigg & Morlock, 1987; Stacoff et al., 2001; Butler, et al., 2006] with different footwear conditions. With the

motion control footwear, however, the rearfoot pronation did not change significantly even after the muscles had developed fatigue.

Originally, a correlation between the decrease in rearfoot supination force and increase in rearfoot pronation was expected. However, the present results did not reveal a strong correlation. This could be explained by the difference in mode of muscle contractions between the MVC testing and running. During running, rearfoot pronation is mainly checked by the eccentric actions of rearfoot supinators, but in MVC measurement, such muscle group is under isometric contraction. These two modes of muscle contraction may not be strictly comparable.

3.3.3 Subjective feedback on motion control footwear function

Subjectively, all the subjects in this study were not able to differentiate the functional difference between the two tested footwear models in controlling their rearfoot motion. Almost all the subjects reported that they were not aware of their rearfoot pronation problem and they never knew the “motion control” footwear technology before this study. Their main footwear selection criteria were based on the price, brand and appearance of the footwear. This would highlight the importance of professional input from appropriate experts, such as physiotherapists and podiatrists, in assessing the individual’s foot type and educating them to make the best selection

of footwear that suits their needs.

3.3.4 Cut-off amplitude to define overpronator

Subjects involved in this study were regarded as overpronators with more than 6^{o} of rearfoot pronation. However, there is no consensus in the literature for classifying overpronators. The cut-off pronation angle in this study was taken as the average value of the previous studies [Hintermann & Nigg, 1998; Eng & Pierrynowski, 1994; Johnson, et al., 1994; LeLievre, 1970]. The clinical applicability of this cut off value has not been tested and the results may not be generalized to people with different amplitude of rearfoot pronation.

3.4 CONCLUSIONS

1 Excessive rearfoot pronation movement is reduced when runners with overpronation put on motion control footwear, as compared to neutral footwear.

2 The effects of kinematics control with motion control footwear are maintained after the rearfoot supinators are fatigued.

3 If the subjects with rearfoot overpronation put on inappropriate footwear e.g. neutral footwear, the excessive rearfoot pronation is further increased after prolonged running.

4 The subjective judgments from recreational runners could not differentiate the footwear designs that had different functional values.

3.5 Relevance to the main study

1. Hypothesis “excessive rearfoot pronation can be reduced by motion control footwear” (see section 1.9) was true for both before and after muscle fatigue.
2. The running injury risk factor “incresed rearfoot pronation” can be controlled by footwear adjustment.
3. The increased rearfoot pronation, which is related to “muscle fatigue”, can also be reduced by appropriate footwear prescription.
4. If the kinematics risk factor is not controlled, the rearfoot pronation would increase with mileage which could add to the risk of injury
5. Hypothesis “runners are able to differentiate the footwear function by their subjective feedback” (section 1.9) is not substantiated in this study.
6. Runners are not able to “sense” the footwear function on rearfoot motion control. Examination or expert advice from professionals may be crucial in injury prevention.

CHAPTER 4 PLANTAR LOADING DURING RUNNING WITH MOTION CONTROL FOOTWEAR

4.1 INTRODUCTION

This chapter describes the results (kinetics part) of the experiment in Chapter 2. In this experiment, 25 female recreational runners with excessive rearfoot pronation, were asked to run on treadmill for 1500m in 2 sessions. In each session, subjects put on either motion control footwear or neutral footwear. Plantar loading of the left foot was monitored by an insole sensor and level of muscle fatigue was measured by a dynamometer.

4.2 RESULTS

4.2.1 Plantar loading

Repeated measures ANOVA revealed significant interaction of the two factors of time and footwear at the area under the medial midfoot ($p=0.001$). Therefore, the two factors were analyzed separately. Zonal pedographic analysis (TABLE 8) revealed that in the neutral footwear condition, there were increases in peak force in the medial midfoot (364 to 418 N; 15% increase, $p = 0.001$) region after the 1500m run. There was a pattern of increase of peak force over the first metatarsal head region (524 to

Table 8 Plantar force in neutral footwear condition before and after 1500m run

	Mean Peak Force (N)		
	Before	**After**	**P**
Medial Heel	387	389	NS
Lateral Heel	407	405	NS
Medial Midfoot	364	418	**0.001
Lateral Midfoot	419	412	NS
First Metatarsal	524	565	*0.021
Second & Third Metatarsal	503	511	NS
Fourth & Fifth Metatarsal	491	505	NS
Hallux	340	353	NS
Second & Third Toes	99	104	NS
Fourth & Fifth Toes	76	80	NS

NS = non-significant

* = Only significant before Bonferroni adjustment

** = Significant after Bonferroni adjustment

565 N; 8% increase, p=0.021). In the motion control footwear testing condition, the plantar force remained similar (p=0.572) among the different anatomical regions before and after the running bout.

4.2.2 Level of muscle fatigue

The MVC produced by the rearfoot supinators was reduced in both footwear conditions ($p<0.01$) after the 1500m run. There was an average drop of 30% - 40% MVC in both footwear conditions (TABLE 6).

4.3 DISCUSSION

The aim of this study was to compare the pedographic measurements before and after a moderate running distance of 1500m in two different footwear conditions. The findings indicated that the peak plantar force varied according to different footwear and running mileage.

4.3.1 Zonal analysis of plantar loading in different footwear

If the pedographic data was analyzed as a whole i.e. not zonal analysis, no significant difference in plantar force measurement was detected before and after the running sessions. Similar findings had also been reported by others [Nigg & Morlock,

1987; Perry & Lafortune, 1995]. This highlights the sensitivity of the zonal analysis for evaluation of plantar force distribution with different footwear.

When the data were analyzed on an anatomical zonal basis, results of the neutral footwear condition suggested that the medial foot structure sustained a higher loading at the end of the running bout, despite the increase in loading under the first metatarsal area was only significant before Bonferroni adjustment.

As shown in the results of Chapter 4, neutral footwear, which comprised a single material in the midsole, was not able to control excessive rearfoot motion before and after fatigued running, it was therefore speculated that the increased plantar force over the medial foot structures was not resulted by increase in rearfoot pronation moment only. A possible explanation for the kinetics difference may be related with the fatigued rearfoot supinator muscle at the end of the running bout. The reduced ability of the muscles in counteracting the rearfoot pronation may contribute to resist the pronation moment, which induced a higher medial plantar force increase. Since the amount of rearfoot pronation was maintained before and after prolonged running, the plantar loading was not increased even though the rearfoot supinator muscles were fatigued.

It is hypothesized that after 1500m of running, the rearfoot supinator muscle group had become less efficient in controlling rearfoot pronation, and the increased

pronation movement could have led to an increase in plantar force on the medial structures. Such an increase in plantar force on the medial side of the foot could result in an extra moment to the foot structure so that additional muscle work was required. This phenomenon could be explained by the muscle tuning with respect to different surfaces and environments. Muscle tuning was proposed by Nigg [2001] that the effect of the impact force at heel-strike can be regarded as an input signal to the soft-tissue vibrations. These vibrations are heavily damped and the paradigm of muscle tuning suggests that the muscle activity would adapt to different input signals to minimize vibrations. When the muscles are fatigued, they would be less efficient, thus lowering the responsiveness of the muscle tuning system to the impact force which may alter the movement pattern [Nigg, 2001].

The pattern of plantar force was not different in the motion control footwear testing condition before and after the 1500m running bout. This finding suggested that the motion control footwear was not only able to control the rearfoot motion after a moderate running distance, but it also checked the potential increase of the plantar force.

4.3.2 Increase plantar loading and running injury

According to two recent prospective cohort studies, increased plantar force pattern

at medial midfoot and first metatarsal head region could be associated with metatarsal stress fracture [Weist, et al., 2004] and exercise related lower leg pain in runners [Lobmann, et al., 2001]. The potential cause of metatarsal stress fracture and exercise related lower leg pain may be due to an increase in repetitive force and demand on shank muscle activity to counteract excessive pronation.

4.3.3 Effects of muscle fatigue: from kinematics to kinetics

According to the findings of Chapter 3, muscle fatigue led to increase rearfoot pronation which has also been reported by Derrick et al. [2002]. In muscle fatigue condition, such increase of rearfoot pronation may influence shock wave attenuation during heel strike [Lafortune, et al., 1994; Perry & Lafortune, 1995] due to reduction of force attenuating ability of the musculoskeletal system [Mizrahi, et al., 2000; Voloshin, et al., 1998]. A recent study by Bixiaux & Moretto [2008] had investigated change of plantar loading before and after muscle fatigue. In their study, the peak plantar loading in medial midfoot and forefoot area were raised after prolonged exercise. In our study, such an increased plantar loading could be balanced by motion control footwear, which was effective in rearfoot pronation control. In other words, apart from the kinematics control, motion control footwear did not avoid muscle fatigue, but control part of the consequences after muscle fatigue.

4.4 CONCLUSIONS

1 When runners with overpronation ran on motion control footwear, their corresponding plantar loading remained stable throughout prolonged running.

2 If the overpronators ran on neutral footwear, the plantar loading at medial foot areas would increase with running mileage, while the rearfoot supinator muscles would have less resistance to fatigue.

3 As increase of medial midfoot loading is associated with running injuries, motion control footwear may prevent injuries in runners by checking excessive force at this area.

4 Function of motion control footwear is not confined to kinematics control; but kinetics control as well.

4.5 RELEVANCE TO THE MAIN STUDY

1. Hypothesis “the plantar force distribution is different when runners put on motion control footwear as compared to neutral footwear” (section 1.9) is valid only when muscle is fatigued.

2. The running injury risk factor “unbalanced plantar loading” can be avoided by footwear adjustment.

3. The change of plantar loading in different footwear models was related to leg muscle activities instead of change in kinematics alone.

4.6 ROLE OF MUSCLES & THE KINETIC CHAIN CONCEPT

Despite the presence of kinematics and kinetics data, electromyography provides unique information about muscle activity which control joint kinematics and consequent plantar loading. Future study on leg muscle activity is warranted. The next section of this thesis describes studies on assessing the role of muscles in the higher segments of the kinetic chain; so that the role of foot kinematics and kinetics may be linked with the major control of the more proximal segments in explaining the overall functioning of each of the footwear.

CHAPTER 5 METHODOLOGY FOR MEASURING SHANK AND THIGH MUSCLES ACTIVITIES DURING PROLONGED RUNNING

5.1 INTRODUCTION

Muscle activities of lower extremity are highly associated with running injuries because their activities may alter the force distribution on the joints [Cowan, et al., 2001; Mizrahi, et al., 2000; Sanna & O'Connor, 2008]. Muscle activity also plays an important role in maintaining correct kinematics during running. Imbalance or malfunctioning of lower extremity muscles would alter the movement patterns as well as joint loading. Muscle imbalance has therefore been regarded to be a risk factor for running injuries. However, there are very few reports on runners' muscle activity in different footwear conditions, in particular, the motion control footwear. This part of study aimed to fill this knowledge gap.

5.2 CHOICE OF SUBJECTS

Twenty female recreational runners with excessive rearfoot pronation (mean rearfoot pronation during running > 6° than normal standing posture) as determined by the VICON motion analysis system (screening procedures can be referred to section 2.5.3.3) were recruited by convenience sampling. Their mean age was 25.8 ±

3.7 years. Other demographic data about the subjects are presented in TABLE 9.

5.2.1 Adjustment of inclusion and exclusion criteria

Generally, the characteristics of subjects involved in this investigation were similar to those described in Chapter 2. However, in order to differentiate the potentially small effect sizes in this investigation, a longer running mileage of 10km was adopted. Therefore, all the subjects in this study were required to be able to accomplish 10km run in usual training speed.

5.2.2 Inclusion criteria

5 Female subjects

6 Recreational runners

- With regular practice (at least once per week) for six months or more
- Non-professional runners and not supported by any running club
- Able to complete 10km run

7 Healthy in terms of musculoskeletal and cardiopulmonary functions

8 With excessive rearfoot pronation

- The functional definition of excessive pronation has been described in detail in section 2.5.3.3.

Table 9 Demographic data of subjects recruited in Chapter 5

Age	25.8 ± 3.7
Running Experience (Yrs)	4.6 ± 2.4
Average Mileage / Week (km)	33.1 ± 12.2
Average Frequency of Run/ Week	3.75 ± 1.2
Height (m)	1.72 ± 0.06
Weight (kg)	60.84 ± 6.7
Body Mass Index (kgm^{-2})	20.54 ± 1.27

5.2.3 Exclusion criteria

5 Male subjects

6 Professional runners

- Member of any running clubs
- Athlete of any district or Hong Kong (or any other country) running team

7 Subjects with any known active musculoskeletal and cardiopulmonary problems that required medical consultation in the last one year of the study

8 Subjects with normal foot type.

5.3 LOCATION AND DURATION OF THE INVESTIGATIONS

The whole experimental procedures were performed in the Sports Training and Rehabilitation Laboratory at The Hong Kong Polytechnic University. There were 2 running sessions with 1-week apart. Each session consisted of the same testing procedures except that the footwear condition was different.

5.4 EQUIPMENT

5.4.1 Footwear

Same test footwear models, Adidas Supernova control and Adidas Supernova

cushion (FIGURE 4), were used in the study.

5.4.2 Footwear size

The subjects were allowed to try different sizes to identify the one that fitted their feet best. The size was recorded so that the subjects put on the same size of shoes in different test models.

5.4.3 Shoe box tightness standardization

To ensure that the data was collected under similar shoe fitting conditions, the same pair of laces was used in the two shoe models for each subject. This pair of shoe laces had color markers on both ends which could guide the positions of knots thus standardizing the tightness of the shoe fitting for the subjects (FIGURE 5).

5.4.4 Treadmill

The treadmill model and setup were identical with that described in section 2.5.2. However the running session was conducted at a speed of 8km/ hour for a total of 10km.

5.4.4.1 Justification for running speed adjustment

The speed selection of the 10km treadmill run was based on a preceding pilot test which involved 3 recreational female runners who ran regularly. They all had experience to complete 10km race before. They were asked to run on a treadmill for 10km in a self-adjusted constant speed. The testing speed of 8km/ hour was determined by averaging the speed among those 10 km runners recruited in this pilot test.

5.4.4.2 Step detection by load cell underneath treadmill

It is important to define the instance of touch down loading in determining the onset time relative to the running duty cycle. Young [1993] had used F-scan foot switch to identify running step and loading response during the whole stance phase. However, in consideration that the present study involved long distance running and the foot switch may not withstand the repeated loading, the detection of running steps was not suggested by the insole sensor e.g. Pedar (Pedar Novel GMBH, Ismaninger Strasse 51, Munich, Germany) or F-scan (Tekscan Inc., Boston, United States of America).

A new method was therefore introduced in this investigation to determine the instance of loading. A load cell (Vishay Electronic GmbH, Geheimrat-Rosenthal-Str.

100, Selb, D-95100, Germany) was positioned at the right rear base of a leveled treadmill (FIGURE 14). The treadmill was maintained at zero inclination by 3 blocks of the same height with the load cell. The zero inclination was confirmed with a level bar. An analog level of alternatively high and low waves would indicate the right and left steps respectively, interlaced by a signal-free air borne phase.

A pilot test involving 6 subjects running for 2 minutes was conducted. Identification of the last 10 steps was measured by both F-scan (Tekscan Inc., Boston, United States of America) and the load cell setup. The 95% limits of agreement [Bland & Altmand, 1986] between the two measurements in step determination was 0.03 sec ± 0.00345.

5.4.5 Electromyography

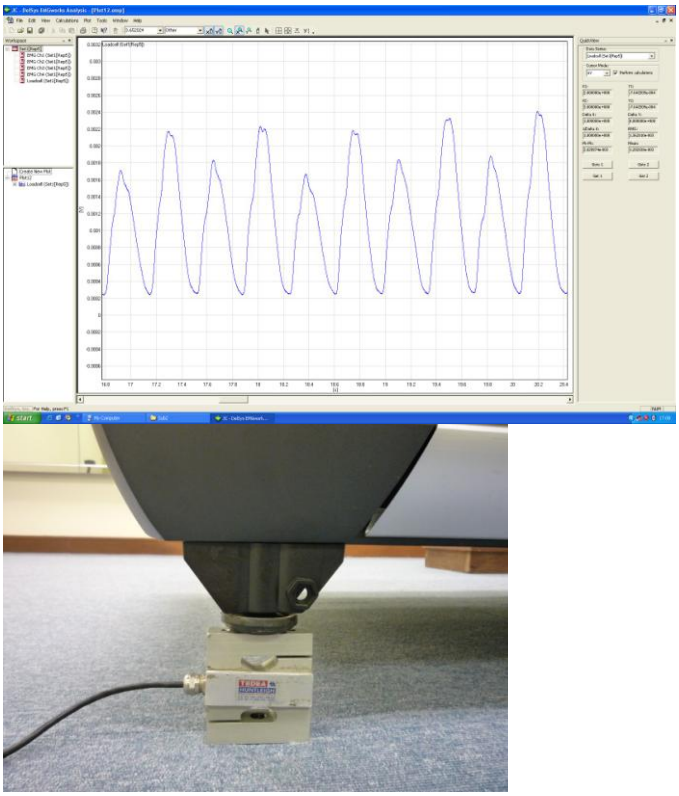
5.4.5.1 Selection of target muscles under test

Malfunctioning of the active system [Cowan, et al., 2001; Voight & Weider, 1991] is believed to be a risk factor of patellofemoral pain syndrome (PFPS), which is one of the most common knee problems in runners. A manifestation of active system malfunctioning is onset delay of vastus medialis obliquus (VMO) relative to vastus lateralis (VL) [Cowan, et al., 2001; Voight & Weider, 1991].

Also, it has been proposed that the extrinsic foot musculature, especially the

Figure 14 When a subject is running on the treadmill, the load cell underneath will receive a higher impact while right running step stance phase, and a lower impact during left running step.

Three other blocks of same height with load cell to balance the treadmill landing



Load cell to detect right foot landing by identifying the higher analog output signal

tibialis anterior (TA) and peroneus longus (PL) are two major stabilizing muscles for rearfoot control [Murley & Bird, 2006]. It has been hypothesized that these muscles alter their activations in order to maintain a preferred movement pattern in different running environment [Nigg, 2001].

Dysfunction of these stabilizing muscles may lead to alternation of normal joint kinematics and cause various running injuries including metatarsal stress fracture [Mizrahi, et al., 2000]. In order to reduce the abnormal loading on the foot structures, orthotic devices have been used and Nigg & Wakeling [2001] proposed a paradigm that orthotic devices improve lower extremity symptoms by synchronizing the lower leg muscle activities including TA and PL. Therefore, VMO, VL, TA and PL were examined in this study.

5.4.5.2 Procedures for the muscle activity measurement

The EMG activity from the right lower extremity was recorded using DelSys (DelSys Inc., Boston, United States of America) double differential Ag-Ag Cl surface electrodes with an inter-electrode distance of 10mm. The common mode rejection ratio (CMRR) of the current system was 92dB.

5.4.5.2.1 Skin preparation

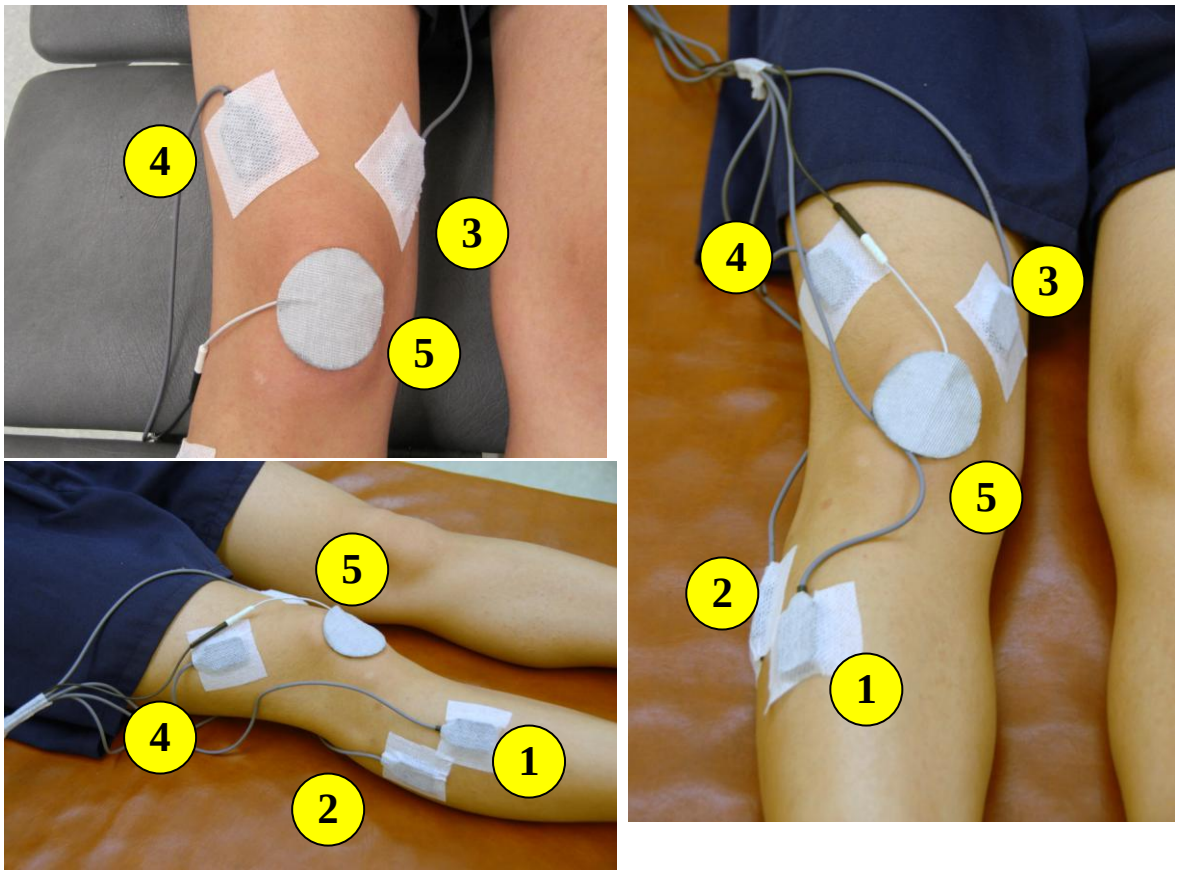
In order to reduce skin impedance, the skin was prepared by shaving, lightly abrading with fine-grade sand paper to remove the dead cell layer and cleansing with alcohol to reduce the skin-electrode interface impedance to below 5kΩ [Basmajian & De Luca, 1985; Gilmore & Meyers, 1983].

5.4.5.2.2 Electrode positioning

The electrode on TA was applied at 1/4 to 1/3 the distance between the distal patella and lateral malleolus and lateral to tibia (FIGURE 15), whereas the PL electrode was applied at 5-7 cm distal to the lateral head of fibula along the lateral shin (Figure 15) [Leis & Trapani, 2000].

Electrodes on VMO and VL were fixed, respectively, at 55° medially and 15° laterally to the longitudinal axis of femur and 1/10 of the femur length above the supero-medial corner of patella (FIGURE 15) [Wong & Ng, 2005]. The location of VMO electrode was confirmed by palpation during quadriceps contraction with and without hip adduction to distinguish between VMO and vastus medialis longus [Bose, et al., 1980]. Correct location of VL electrode was confirmed by quadriceps contraction with and without hip flexion to differentiate rectus femoris action [Cerny, 1995].

Figure 15 Electrode positioning of (1) Tibialis Anterior, (2) Peroneus Longus and (3) Vastus Medialis Oblique, (4) Vastus Lateralis, and (5) Earth electrode



To prevent movement artifact, wires between the electrodes and the computer were secured to the skin with adhesive tape and leads braided to minimize any stray electromagnetic interference [Wong & Ng, 2006]. The earth electrode was applied at the patella (FIGURE 15).

5.4.5.2.3 Signal amplification, processing and recording

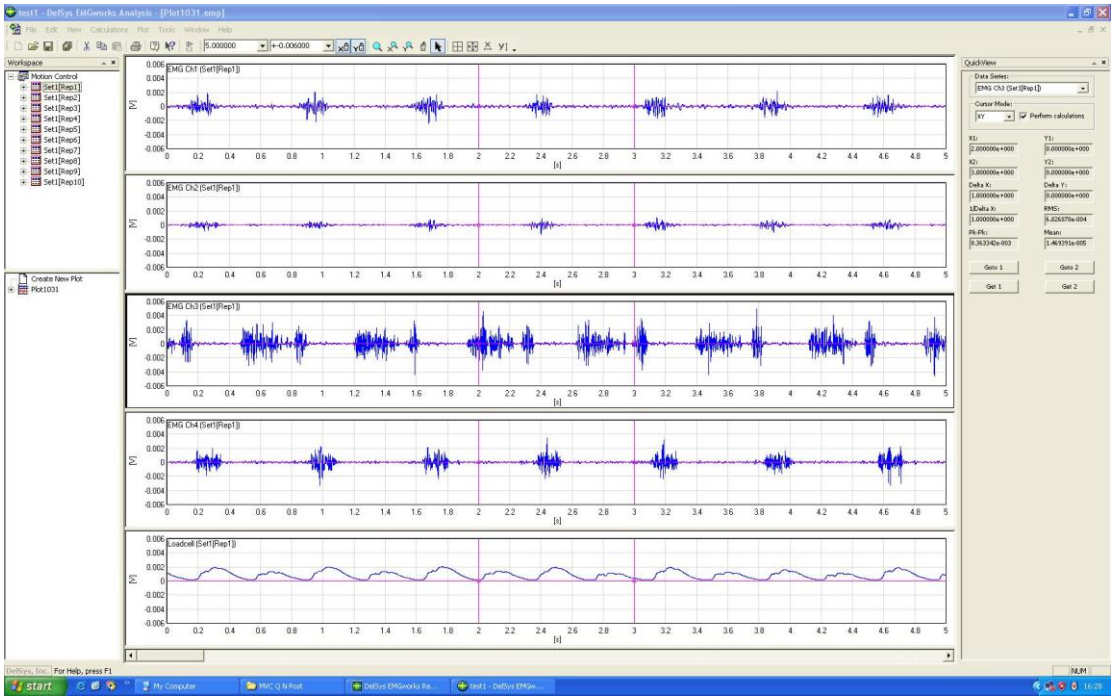
The EMG signal was 1K amplified (Bagnoli-4 Main Amplifier) and sampled at a rate of 1000 Hz using a data-acquisition program (DelSys EMGworks Acquisition). Before sampling, the raw EMG signals were analog high-pass filtered with 50 Hz and full wave rectified (FIGURE 16). The processed signal was saved and then input to the Excel software (Office 2003, Microsoft, Redmond, United States of America) for the muscle onset analysis. With the fast Fourier transform (FFT) function of the data acquisition program (DelSys EMGworks Acquisition), the raw EMG data was transferred from time domain into frequency domain for MF calculation.

5.4.5.3 Muscle activity measurement

5.4.5.3.1 Definition of muscle onset

The determination of EMG onset was based on the instance that EMG signal was at three standard deviations above the mean (mean + 3 SD) of the resting baseline

Figure 16 An example to show data acquisition for load cell signals (at channel 5) and electromyographic signals from tibialis anterior (TA) (at channel 1), peroneus longus (PL) (at channel 2), vastus medialis obliquus (VMO) (at channel 3) and vastus lateralis (VL) (at channel 4). High amplitude of load cell signals indicated a right stance phase of running step.



value and such activity would last for 25ms or more [Hodges & Bui, 1996]. The onset time was presented in terms of percentage of the whole running cycle, starting from the instance of heel strike. As previous studies [Cowan, et al., 2001; Voight & Weider, 1991] had reported that delay onset of VMO to be a risk factor for PFPS, onset of the VL was subtracted from that of VMO so that a negative value would indicate VMO onset before VL, whereas a positive value would indicate the opposite.

5.4.5.3.2 Median frequency

Assessment of the frequency domain has been used in determining muscle fatigue responses. As there is a concomitant change in the power spectrum derived from surface electrodes where there is an increase in the amplitude of the low frequency band and a relative decrease in the higher frequencies, therefore, shifting to lower median frequency (MF) is a common indication of muscle fatigue [Basmajian & De Luca, 1985; Solomonow, et al., 1990; Kupa, et al., 1995; Roy, et al., 1997; Wretling, et al., 1997; Rainoldi, et al., 1999; Gerdle, et al., 2000].

5.4.5.3.3 Amplitude of muscle activity

The root mean square (RMS) voltage is an indicator of the amplitude of the EMG signals. It is the summation of the square root values of the mean raw EMG

signals [DeLuca, 1997]. This outcome measure is commonly selected to represent the electrical power of the EMG signals.

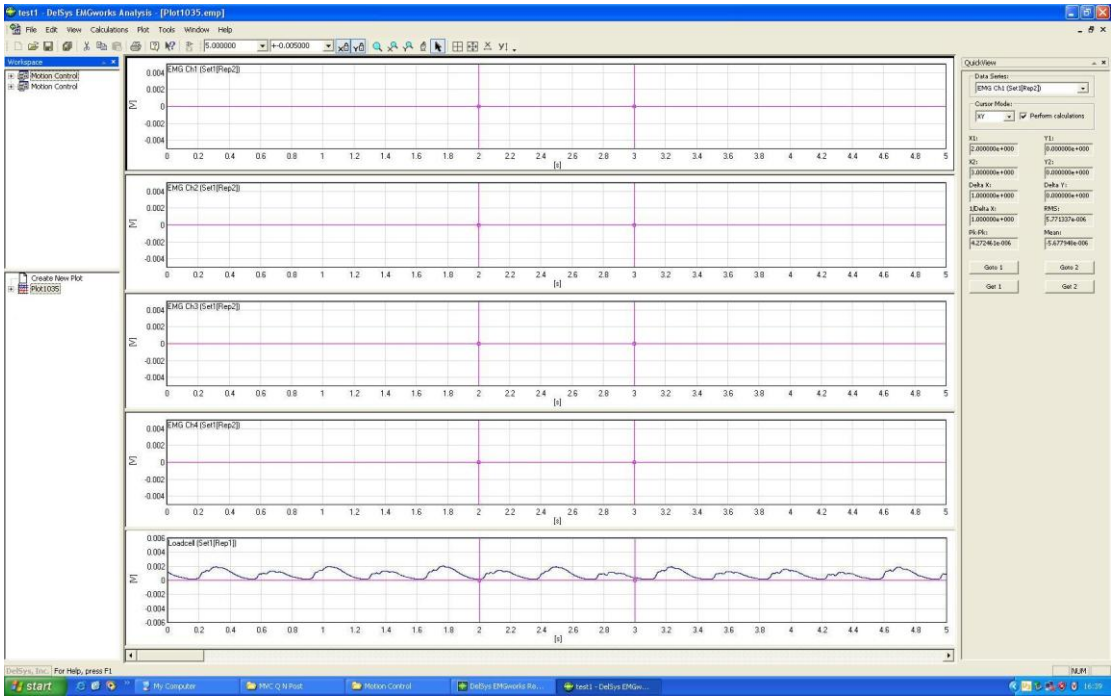
5.4.5.4 Quantification of impact noise

To quantify the amount of impact noise recorded by the electrodes during treadmill running and to obtain the best filtering of the impact noise, two healthy regular runners were tested in a pilot study. They were free of any known musculoskeletal problems in their lower extremities.

The setup was the same as in sections 5.4.5.2 – 5.4.5.3 except that the contact surface between the electrodes and skin was blocked by an insulating plastic sheet. The runners were asked to run on the treadmill with the same as detailed in section 3.4.4 but the mileage was only 1 km. The EMG signal processing was the same as protocol as stated in the section 5.4.5.2.3 and 5.4.5.3.

The frequency spectrums of the collected EMG signals from both subjects were low frequency in nature. After adjustment of high pass filter to 50Hz, the noises from the foot impact were negligible (FIGURE 17). This filtering indicated that the impact noise from this setup was minimal and would not contaminate the muscle activity signal during data collection.

Figure 17 Impact noise quantification: A signal sample to show filtered (low-pass of 50Hz) EMG data acquisition of muscle activities from 4 channels (VMO, VL, TA and PL) were negligible when the electrodes were insulated with plastic sheet.



5.5 TESTING PROCEDURE

5.5.1 Running sessions arrangement

Each subject was asked to attend three testing sessions in this investigation. In the first session, it was for foot posture screening. Subjects ran on treadmill in 10km per hour with neutral footwear condition for 1500m. Subjects with maximum rearfoot angle higher than 6° compared with the offset angle measured in standing were classified as overpronators and they would proceed to the next testing session.

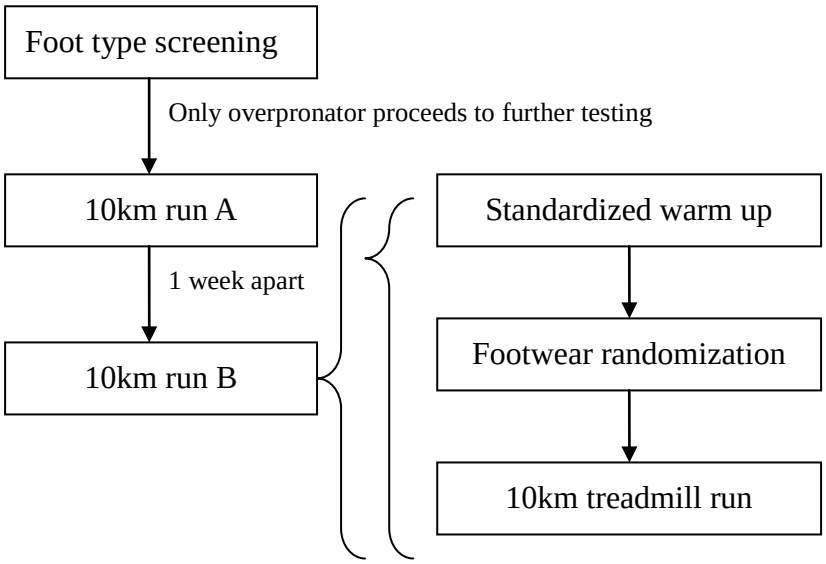
The following two running sessions, which involved a 10km treadmill run, were identical except the sequence of footwear condition was randomized by throwing a dice. Odd number indicated the use of motion control shoe in the first session and vice versa.

The flow chart of running sessions in this investigation was illustrated in FIGURE 18.

5.5.2 Standardized warm up exercise

Prior to each running session, the subjects were asked to perform standardized warm up exercise as shown in FIGURE 10.

Figure 18 Flowchart of the running sessions in Chapter 5 experiment



5.5.3 Electromyography data collection and data management

5.5.3.1 Division of running session

In order to compare different stages of running, the total 10km running distance was equally divided into 10 recording checkpoints. To reduce the data acquisition and storage problem, the EMG data was only collected during the last minute of each recording checkpoint. All the recording checkpoints were included in this study for analysis.

5.5.3.2 Normalization of electromyographic data

Because the absolute value of EMG recordings cannot be directly compared between subjects or between muscles, or even within the same subject [Gilmore & Meyers, 1983], EMG normalization was therefore performed.

Before each running session, every subject performed a maximal contraction for both TA and PL according to the method of Yang and Winter [1984]. Each muscle contraction was held for 8 seconds and the actual window for measurement was confined to the last 4 seconds. This procedure was repeated for 3 times and the averaged root mean square (RMS) EMG value of the 3 repetitions would represent the MVC EMG value. In each recording checkpoint of the running tests, the RMS values

of EMG from all the running steps were averaged and normalized against the MVC EMG value.

5.6 STATISTICAL ANALYSIS OF DATA

Repeated measures ANOVA was used to test the effects of footwear and mileage on the normalized RMS EMG values of both muscles. Since 20 (10 recording checkpoints and 2 footwear models) pairwise comparisons would be performed, Bonferroni adjustment was applied by adjusting α to 0.0025 ($\alpha = 0.05 \div 20$) so as to control the family type I error. For significant ANOVA results, post hoc Wilks' Lambda test was used to identify the data pairs that were significantly different from one another. Furthermore, the relationships between mileage and EMG were examined with Pearson's correlation. For the fatigue profile, the corresponding MF difference between the 1^{st} and the 10^{th} checkpoint was analyzed using paired t-tests. The alpha value for paired t-test was set at 0.05.

Another repeated measures ANOVA was used to test the effects of footwear and mileage on vasti muscle onset. The amplitude of onset delay was defined as the onset difference between VMO and VL in terms of duty cycle. Pearson's correlation was used to indicate the relationship between mileage and the delay in EMG onset time. For the fatigue profile, the corresponding MF difference between the 1^{st} and the 10^{th}

checkpoint was analyzed using independent t-test.

5.7 ETHICS

The present investigation was approved by the Ethics Review Committee of the Department of Rehabilitation Sciences of The Hong Kong Polytechnic University (Appendix IV). Written informed consent (Appendix V) was obtained from each subject prior to testing. All the procedures in the experiment were non-invasive.

CHAPTER 6 SHANK MUSCLES ACTIVITIES DURING 10KM RUN WITH MOTION CONTROL FOOTWEAR

6.1 INTRODUCTION

This chapter describes the shank muscles activities in the experiment mentioned in Chapter 5. In this experiment, 20 female recreational runners with excessive rearfoot pronation were tested with running on a treadmill for 10km in 2 sessions. In each session, subjects put on either motion control footwear or neutral footwear in randomized order. Activities of tibialis anterior (TA) and peroneus longus (PL) of the right leg were monitored by surface electromyography (EMG).

6.2 RESULTS

6.2.1 Number of running steps in each checkpoint

The number of running steps in each recording session was highly comparable between footwear conditions. The mean step numbers in motion control footwear and neutral footwear testing condition in each session were 85.6 ± 1.2 and 84.9 ± 0.7, respectively.

6.2.2 Muscle activity of shank muscles

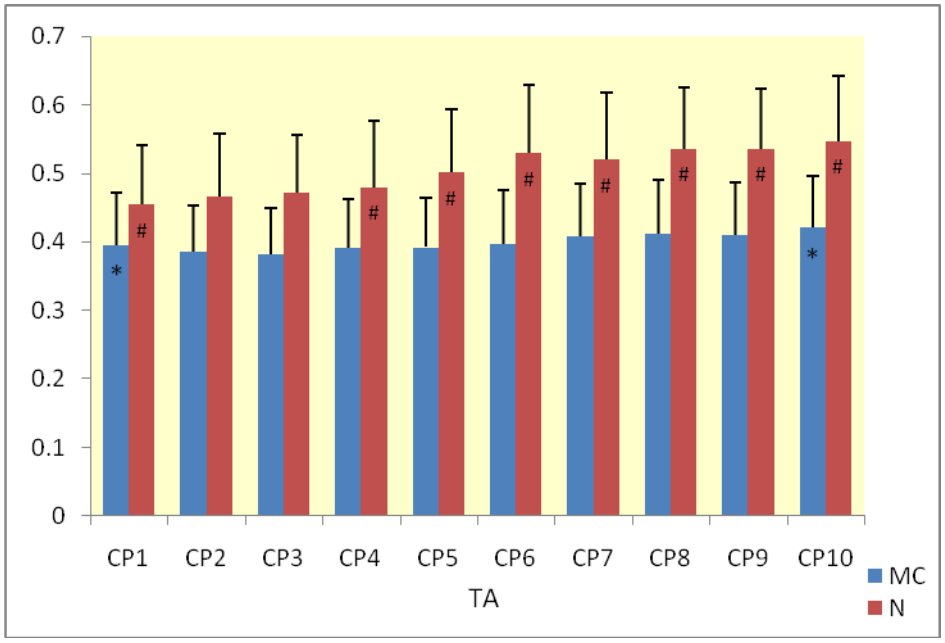
The normalized root mean square (RMS) EMG of TA and PL revealed significant statistical difference with different footwear and mileage ($p<0.001$). The TA activation with the neutral footwear testing condition was around 10.5% higher than the motion control footwear condition. For the PL activation, the result revealed a 9.6% higher activity in the neutral footwear than the motion control footwear condition (FIGURE 19). Regardless of the footwear condition, EMG activities in both muscles increased with running mileage. When comparing to the baseline value at the 1st check point, the TA and PL recruitment increased since the 5th and 2nd check-point, respectively.

Significant correlations were found between the change in TA RMS EMG of both motion control footwear and neutral footwear conditions with mileage of running. For the PL muscle, such a correlation was only revealed in the neutral footwear condition. As a whole, the correlations are more evident in the neutral footwear than the motion control footwear condition (TABLE 10).

6.2.3 Fatigue of shank muscles

There was a significant downward shift ($p<0.001$) in median frequency (MF) after the 10km running bout. Comparing the two footwear conditions, significantly larger shift in MF was found in PL during the neutral footwear testing condition ($p<0.001$).

Figure 19 Normalized RMS EMG of tibialis anterior (TA) and peroneus longus (PL) in motion control (MC) and neutral (N) shoe conditions at different check points (CP). The error bars represent standard deviations and '#' indicates significant difference in the neutral shoe condition of that particular CP when compared to the same shoe condition at CP1. The symbol '*' indicates significant difference in the motion shoe condition of that particular CP when compared to the same shoe condition at CP1



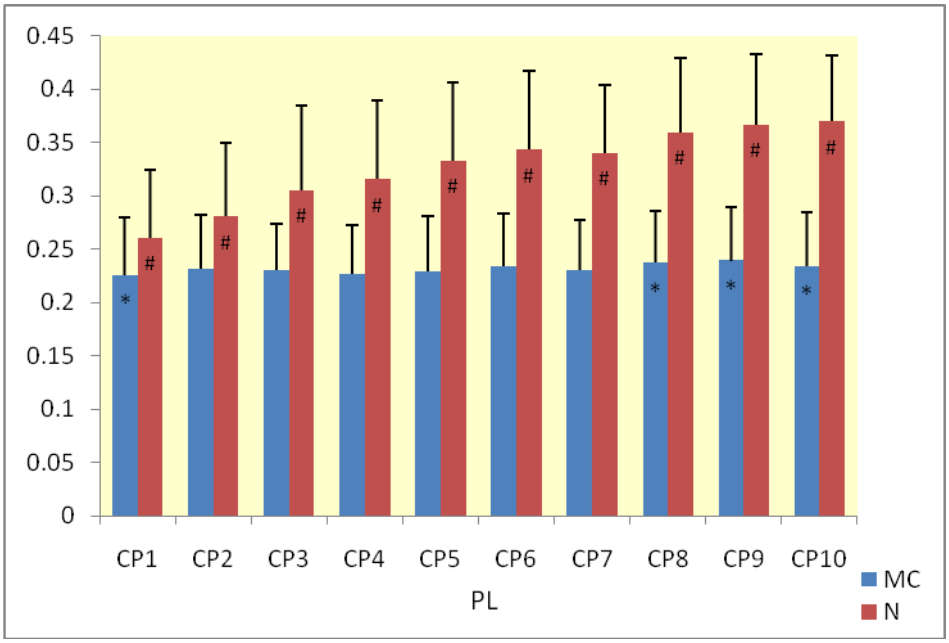


Table 10 Correlation between changes in normalized root mean square of EMG output from shank muscles and running mileage during 10km run

	Motion Control	**Neutral**
Tibialis Anterior	0.148 (p=0.037)	0.329* (p<0.001)
Peroneus Longus	0.066 (p=0.35)	0.444* (p<0.001)

For the TA however, the difference between footwear conditions was insignificant (p=0.074), though a larger amplitude of MF decrease was observed in the neutral footwear condition (TABLE 11).

6.3 DISCUSSION

This study compared the lower leg muscles recruitment with and without fatigue in recreational runners who have excessive rearfoot pronation when running with different footwear. The results revealed that motion control footwear was able to maintain more stable muscle activity in TA and PL muscles and delay fatigue of these muscles with prolonged running better than neutral cushioned footwear.

6.3.1 Effects of motion control footwear on muscle activity of shank muscles

The RMS EMG findings revealed an increased activation level in TA and PL in the neutral footwear condition. Furthermore, there were also significant correlations between the change in RMS EMG of both muscles and mileage of running with this type of footwear. For the motion control footwear however, the EMG activation levels of both muscles were more stable than the neutral footwear condition throughout the running bout.

Table 11 Drop in median frequency (MF) in shank muscles with different footwear conditions

	Motion Control	**Neutral**	**P**
Tibialis Anterior	5.51 Hz ± 6.59 1st checkpoint: 75.59 Hz 10th checkpoint: 70.08 Hz	12.94 Hz ± 13.93 1st checkpoint: 74.66 Hz 10th checkpoint: 61.72 Hz	0.074
Peroneus Longus	2.10 Hz ± 4.44 1st checkpoint: 69.18 Hz 10th checkpoint: 67.08 Hz	11.60 Hz ± 5.96 1st checkpoint: 77.76 Hz 10th checkpoint: 66.17 Hz	<0.001

6.3.2 Effects of motion control footwear on shank muscles fatigue

The present findings revealed that MF had dropped in both muscles after running with motion control or neutral footwear conditions (TABLE 11), but the drop was significantly more in the neutral footwear condition than the motion control footwear condition for PL ($p<0.001$) and a similar trend was also observed for TA ($p=0.074$). These suggested that PL and possibly, TA, had developed more fatigue in the neutral footwear testing condition throughout the 10km of running. As both TA and PL are important dynamic stabilizers [Kirby, 1989; Payne & Danaberg, 1997] that control rearfoot pronation, fatigue of these muscles might lead to further increase in pronation with running mileage as demonstrated in the previous experiment of Chapter 4. Therefore, motion control footwear which helps delay muscle fatigue in PL and TA could be beneficial for long distance running in maintaining stability of the foot. These results suggest motion control footwear may promote performance of the runners by increasing the endurance and delaying the onset of leg muscle fatigue.

6.4 CONCLUSIONS

1 Increased shank muscles activity is noted when subjects with rearfoot overpronation ran with neutral footwear.

2 Motion control footwear is demonstrated to have more stable shank muscle activation pattern for these subjects.

3 Shank stabilizing muscles are more preferentially fatigued when the subjects run with neutral footwear.

6.5 RELEVANCE TO THE MAIN STUDY

1 The hypothesis “the muscle activity of shank is different when runners put on motion control footwear as compared to neutral footwear” was proved in this study both before and after muscle fatigue.

2 Motion control footwear provides a more stable shank activation pattern in people with rearfoot overpronation during prolonged running. Thus the risk factor of “muscle imbalance” in running can be reduced by footwear adjustment.

3 The preferential fatigue of shank muscles is evident in neutral footwear model in runners with overpronation problem.

4 Because the most dominant body part in running injury is the knee joint, electromyography of thigh muscle in long distance running with different footwear may provide further information in running injury prevention.

CHAPTER 7 QUADRICEPS ACTIVATIONS DURING 10KM RUN WITH MOTION CONTROL FOOTWEAR

7.1 INTRODUCTION

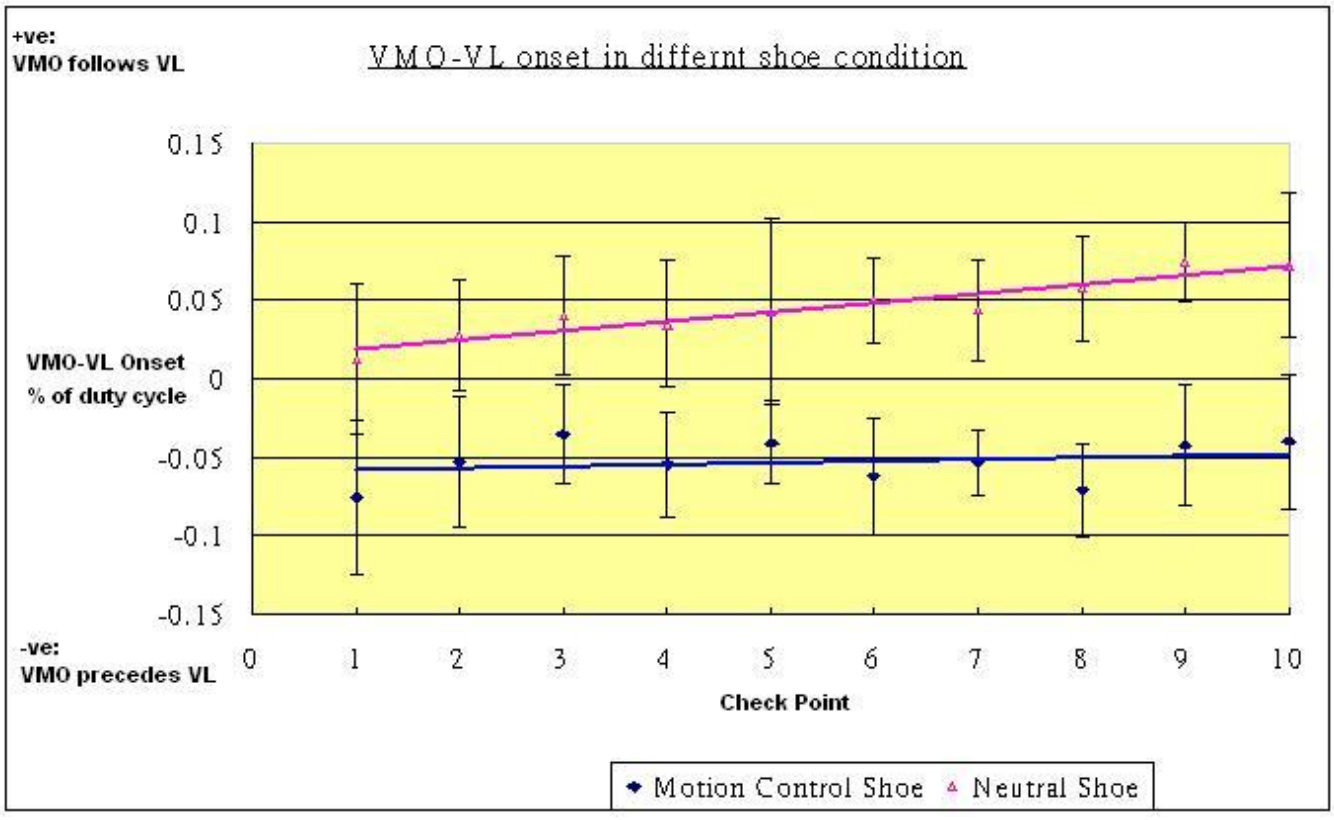
This chapter describes the quadriceps activities in the experiment mentioned in Chapter 5. In this experiment, 20 female recreational runners with excessive rearfoot pronation were asked to run on a treadmill for 10km in 2 sessions. In each session, subjects put on either motion control footwear or neutral footwear in randomized order. Activities of vastus medialis obliquus (VMO) and vastus lateralis (VL) of the right leg were monitored by surface electromyography (EMG).

7.2 RESULTS

7.2.1 Muscle onset of vasti muscles

The EMG onset of VMO and VL revealed significant difference with different footwear ($p=0.001$) and mileage ($p=0.001$). Numerically, the VMO was activated at around 5.3% of a duty cycle earlier than VL when running with motion control footwear; whereas for the neutral footwear running condition, there was a delay in VMO activation by about 4.6% of a duty cycle compared to VL (FIGURE 20). The delay in onset time of VMO was strongly correlated ($r=0.948$; $p<0.001$) with the

Figure 20 Difference in muscle onset between VMO/ VL throughout the 10km run. The onset of VL relative to VMO was increasing with mileage in the neutral footwear condition (purple line). In the motion control footwear condition, however, VMO was activated earlier than VL (blue line).



running mileage in the neutral footwear condition only. In the motion control footwear condition, the correlation was weak (r=0.258; p=0.472).

7.2.2 Muscle fatigue of vasti muscles

Significant drops in median frequency (MF) in VMO and VL were noted toward the end of the run in both motion control footwear ($p = 0.008$) and neutral footwear conditions ($p < 0.001$), which was suggestive of muscle fatigue in both muscles regardless of the type of footwear. The pattern of drop in MF was presented in TABLE 12. The figure revealed that the amplitude of drop in MF in VMO was higher ($p < 0.001$) in the neutral footwear condition while the VL had a more significant drop in MF ($p = 0.008$) when running with motion control footwear.

7.3 DISCUSSION

The purpose of this study was to compare the onset of vasti muscle and pattern of fatigue in recreational runners with excessive rearfoot pronation when running with different footwear for 10km. Motion control footwear was shown to be able to maintain a more stable temporal activity of VMO in these runners. Also, a larger shift in MF was observed in VMO when the subjects ran with neutral footwear. It is believed that both the delay in VMO onset [Cowan, et al., 2001; Neptune, et al., 2000]

Table 12 Drop in median frequency (MF) in quadriceps muscles with different footwear conditions

	Motion Control	**Neutral**	**P**
Vastus Medialis Oblique	1.23 Hz ± 1.42	9.51 Hz ± 5.37**	0.008
Vastus Lateralis	3.80 Hz ± 2.74*	1.78 Hz ± 2.68	0.001

and fatigue of this muscle [Callaghan, et al., 2001] were two common risk factors for patellofemoral pain.

7.3.1 Effects of motion control footwear on vasti muscle onset

A delayed onset of VMO activity was observed when the subjects with excessive rearfoot pronation were running with the neutral footwear and the difference in onset timing between the two muscles was positively correlated with running mileage. It was noted that most of the subjects were running with 80 steps per minute during the test, thus the time for completing each step was about 0.75 seconds. The VMO was activated at about 4.0 milliseconds earlier than VL when the subjects were running with motion control footwear, whereas the same muscle would activate at about 3.4 milliseconds later than VL when running with neutral footwear condition. Cowan, et al. [2001] found that patients with PFPS demonstrated around 10 milliseconds of delay in VMO onset when performing stair climbing activity than normal healthy subjects.

Neptune, et al. [2000] tested nine healthy subjects with a 3-dimensional forward dynamic simulation modeling technique and reported that a delay in VMO activation by as little as 5 milliseconds than VL would affect the patellar tracking and medio-lateral patellofemoral contact pressure. In a fast movement such as running, the

timing of activities would be shorter than that of stair climbing thus the difference in onset activation of VMO and VL observed in this study could be a risk factor of PFPS development.

With a longer running mileage, the delay in VMO onset timing was found to increase in the neutral footwear condition but not so for the motion control footwear condition. In Chapter 3 of the thesis, the experiment suggested that excessive rearfoot pronation was controlled when subjects were running with motion control footwear even when the muscles were fatigued. These findings were suggestive that the motion control footwear could enhance a stable temporal activity of VMO on top of providing control over the pronation problem.

7.3.2 Effects of motion control footwear on muscle fatigue

Reduction in MF of quadriceps was observed during the running bout in both footwear conditions. Such a shift in MF was found to have preferentially occurred in VMO (9.51Hz) than VL (1.78Hz) in the neutral footwear condition. This suggests that VMO could have developed more fatigue than VL in the neutral footwear running condition. As VMO is the most important dynamic stabilizer [Lieb & Perry, 1968] that controls the movements of patella during extension of the knee and maintains normal patellar tracking, fatigue of VMO might lead to unbalanced lateral shift of patella,

which could be another causative factor of PFP [McConnell, 1986].

7.4 CONCLUSIONS

1 When runners with excessive rearfoot pronation put on neutral footwear, their VMO muscle activation was delayed as compared to VL muscle, whereas the VMO muscle would activate before VL when runners put on motion control footwear.

2 The muscle delay in VMO is enlarged in the neutral footwear condition.

3 Muscle fatigue is found to have preferentially occurred in VMO than VL in the neutral footwear running condition

7.5 RELEVANCE TO THE MAIN STUDY

1 The hypothesis that “the muscle activity of quadriceps is different when runners put on motion control footwear as compared to neutral footwear” is substantiated in this study for both before and after muscle fatigue.

2 The risk factor of “muscle imbalance” in running related injuries can be reduced by proper selection of footwear.

CHAPTER 8 GRAND DISCUSSIONS AND CONCLUSIONS

8.1 EFFICACY OF MOTION CONTROL FOOTWEAR

Motion control footwear was a technology in running shoe design for people with excessive rearfoot pronation for almost three decades. This technology, evolved from heel wedge to different midsole composition, aimed to control excessive rearfoot pronation so as to prevent running injuries. In this thesis, a series of investigations were launched to examine the functions of motion control footwear in different aspects, including not only rearfoot kinematics, but also plantar loading pattern (kinetics) and lower extremity muscle activities.

The justification of investigations on non-localized outcome measures based on a paradigm that the foot structure may influence the more proximal anatomical structures. Change of proximal body alignment may alter the muscle activity and recruitment pattern.

The information of the investigations was believed to provide important data for prevention and treatment of various running injuries by appropriate footwear selection.

8.1.1 Rearfoot kinematics control

The findings of previous research studies investigating the motion control footwear function in rearfoot kinematics control were not consistent. The reasons were associated with their difference in motion capture method, subject population, and lack of foot type screening.

With possible projection error, two-dimensional motion analysis system was weeded in kinematics analysis. This kind of system was mainly adopted in some early studies. Among those 2 studies involving two-dimensional motion analysis, the results were conflicting. Perry & Lafortune [1990] found significant control of rearfoot pronation by wedged footwear in ten male runners. Approximately 15^{o} of maximum rearfoot pronation reduction was achieved in valgus-wedged (pronation inhibiting) footwear condition compared with the varus-wedge (pronation inducing) condition. McNair & Marshall [1994] tested four different running shoe models with different midsole density in ten male subjects. They failed to find any difference in the kinematics between shoe models in treadmill running.

Another factor may lead to inconsistent findings was the heterogeneity of the recruited subject pool. Because previous researchers suggested the running pattern and corresponding joint kinematics of male and female runners were different [Decker, et al., 2003; Ferber, et al., 2003; Lephart, et al., 2002], direct comparison for a mixed

gender subject pool may be problematic.

Clarke et al. [1983] reported successful control of excessive rearfoot movement by footwear with firmer midsole and lateral (valgus) flare in ten runners. Nigg & Morlock [1987] studied fourteen healthy subjects and showed reduction of initial pronation angle, but not the total pronation angle, by lateral heel flare. In these two studies, the gender of the subject pool was not mentioned.

Moreover, only very few studies investigating rearfoot kinematics had involved subjects with foot type screening. This may explain the conflicting results because of the possibility of a type II statistical error if subjects with normal foot type were tested by motion control footwear.

In the studies by Nigg et al. [1988], McNair & Marshall [1994] and Milani et al. [1997], which failed to find pronation control effects in the motion control footwear, all their subjects were not screened whether they had excessive rearfoot pronation problem. On the contrary, the study by Butler et al. [2006] assessed that the rearfoot posture of subjects by classifying them as “high foot arch” and “low foot arch” group before running tests, had revealed a systemic trend of pronation difference in motion control footwear condition in low foot arch group.

The results of Butler et al. [2006] not only illustrated the potential function of footwear on rearfoot kinematics control but also revealed the importance to screen the

subjects before testing. As the motion control footwear is designed for runners with rearfoot overpronation problem, subjects with normal foot type may not benefit from this footwear thus the results could be confounded.

As stated in Chapter 2 in the kinematics part of investigation, 3-dimensional motion capturing technique was adopted with single gender (female) subjects who were all screened for foot type. It was believed that the results from this investigation provided stronger evidence for the motion control footwear efficacy.

The results of the present series of studies showed that the maximum pronation was statistically different in motion control and neutral footwear. Before the 1500m run, the mean difference of maximum pronation was 3.3° (motion control footwear = 10.6 ± 3.53°; neutral footwear conditions = 13.9 ± 3.25°). In an unpublished prospective clinical study cited by Reinschmidt et al. [1992], the amount of pronation would correlate with the occurrence of pain and injuries in tennis players. However, the clinical significance of a 3.3° reduction in rearfoot pronation seems to be rather small thus the implication on pain will need further investigation.

After 1500m of running bout, when the rearfoot supinators muscle had developed state of fatigue, the maximum rearfoot pronation in neutral footwear condition increased significantly from 13.9 ± 3.25° to 17.7 ± 3.14°. In contrast, when the runners put on motion control footwear, no statistically significant difference was

shown in rearfoot pronation (before fatigue = 10.6 ± 3.53°; after fatigue = 11.2 ± 3.86°). In terms of percentage increase in maximum pronation angle, the subjects had demonstrated a 31% increase of rearfoot pronation if they were equipped with inappropriate footwear and a further 27% of pronation increase was evident after they were fatigued.

The figures revealed the function of motion control footwear in rearfoot kinematics control was maintained even after the runners were in the state of muscle fatigue. On the other hand, in the population of overpronators, the pronation angle was further increased when running with neutral footwear, which may confront a higher risk of running injury.

8.1.2 Plantar loading control

Kinetics evaluation of footwear performance was common because other than motion control footwear, the cushioning design is another very popular footwear technology. Although the relationship between impact force and injury was not strongly correlated in human study [Burrows & Bird, 2000], loading of running steps in different footwear was frequently investigated.

One early report by Nigg & Morlock [1987] speculated that motion control footwear may alter the joint impact force by correcting rearfoot joint alignment.

However, in their experiment, no significant difference in impact force was found in runners with different footwear. Nigg et al. [1988] also investigated the effects of sole hardness on running kinetics. Again, the results did not suggest that change of sole hardness could effectively alter impact force for runners. Later studies [McNair & Marshall, 1994; Milani, et al., 1997] echoed the results of Nigg [Nigg & Morlock, 1987; Nigg, et al., 1988] that no difference in impact force was observed between footwear comprising different shock attenuating materials in the midsole.

In a small scale study by Perry and Lafortune [1995], ground reaction force was found to be increased in ten healthy runners with motion control footwear. However, the ground reaction force was not reduced in pronation-inducing footwear. Lateral wedged footwear was suggested to reduce the peak external knee varus moment and peak medial compartment force at the knee during normal walking [Crenshaw, et al., 2000]. It was suggested that lateral wedged footwear can also be an alternative in treating people with osteoarthritis of knee.

The conflicting results in the investigations of footwear kinetics may be due to the methodological discrepancies between studies. In kinetics analysis, subjects were usually asked to produce a heel strike running pattern at the middle of the hallway. The procedure was repeated so that an average value was obtained. This testing protocol might alter the normal running pattern of the subjects and the results may not

be generated by same running condition, as the running speed, cadence, and step length might not be comparable in different trials.

Moreover, same as kinematics analysis, absence of foot type screening was still common in kinetics anlysis. Kersting & Bruggemann [2006] tried to analyze the kinetics parameter according to different foot regions by plantar loading measurement. In their test with eight runners, different neuromuscular adaptations were noticed towards different footwear conditions. Such difference response could be due to the heterogeneity among the subjects' foot posture.

Apart from methodological difference, the clinical relevance of direct interpretation of ground reaction force was doubtful, as the linkage between ground reaction force measurement and running injury might not be directly related. Therefore, another methodology entitled pedography, or plantar force measurement was developed, which is a relatively new technology in running footwear research. The measurement of pedography is directly taken at the shoe-foot interfaces. Force is usually represented with respect to different anatomical areas of the foot. The relative ease of data collection from a number of running steps is another advantage of the pedographic technology. Thus, this outcome measure has bcome more common and it can provide more clinically relevant information.

Since the pedography method is relatively new, only few research papers using

this method for evaluation of the kinetics profile of running were found. Weist et al. [2004] stated that the increase in plantar loading at forefoot and toe area among 30 fatigued runners may explain the high risk of metatarsal stress fracture in this running population. In a large scale prospective study of 400 physical education students by Willems et al. [2006] to correlate plantar loading to the incidence of “exercise-related lower leg pain”, the results showed that runners with exercise-related lower leg pain had higher plantar loading over the medial foot structures.

Currently, concerning the change in plantar loading with different footwear, only two studies were found. Wegener [2008] used plantar loading measurement to verify that the midsole material was able to reduce regional peak loading acting on different foot structures in runners. Wiegerinck et al. [2009] demonstrated a lowering in peak loading of the sole in 37 runners when running with training shoes than racing flats. The authors suggested proper running shoe selection could help in prevention of metatarsal stress fracture.

No previous studies have attempted to look at the plantar loading in runners with motion control footwear and in the state of fatigue. The current investigation provided new information on the kinetics in runners with excessive rearfoot pronation under the motion control footwear condition in long distance running.

The present results suggested that after a 1500m running bout, the runners with

excessive rearfoot pronation had sustained a higher plantar loading in the medial foot structures when they were equipped with neutral footwear. Despite the increase in loading under the first metatarsal area was only significant at the 0.05 level but not after Bonferroni adjustment, the pattern suggested that the first metatarsal, which stabilized the midtarsal and subtalar joint in propulsive phase of the gait cycle [Bennet & Duplock, 1993], may be under extra loading in prolonged running.

Such an increase in plantar loading of the medial foot structures can be induced by simply increasing the rearfoot pronation moment. However, based on the fact that the rearfoot pronation had increased in neutral footwear at the early phase of running and the plantar loading was similar between the motion control and neutral footwear at the beginning of the running bout, the elevated plantar force at the end of the running bout should be caused by factors other than the change in kinematics alone.

A possible explanation for the difference in kinetics may be related with the fatigued rearfoot supinator muscle at the end of the running bout. The reduced ability of the muscles in counteracting the rearfoot pronation may contribute to resist the pronation moment, which induced an increase in plantar force on the medial side. This explanation is also valid for the runners in motion control footwear condition. Since the amount of rearfoot pronation was maintained before and after prolonged running, the plantar loading was not increased even though the rearfoot supinator

muscles were fatigued.

This has brought about the possibility that the lower extremity muscle activity could be different with change in footwear condition. Therefore, lower extremity EMG investigations in runners with different footwear were launched.

8.1.3 Shank muscle activity control

Apart from the speculation in section 8.1.2, the shank muscle activation pattern was found to be different in runners with or without orthotics [Murley & Bird, 2006], altered muscle recruitment in lower extremity was possible in runners with motion control footwear.

Because muscle activities could have close associations with running injuries due to the changes in force distribution on the joints [Cowan, et al., 2001; Mizrahi, et al., 2000; Sanna & O'Connor, 2008], investigation of lower extremity muscle would provide useful information for injury prevention and symptomatic control. Mizrahi et al. [2000] suggested that dysfunction the major stabilizing muscles of ankle joint, namely tibialis anterior (TA) and peroneus longus (PL), may lead to alteration of normal joint kinematics and cause various running injuries including metatarsal stress fracture. Nigg & Wakeling [2001] also proposed a paradigm that orthotic devices would improve lower extremity symptoms by synchronizing the lower leg muscle

activities including TA and PL.

Surprisingly, there were limited studies that had investigated the muscle activity change with different footwear conditions. A newly designed footwear model called "Masai Barefoot Technique", which comprised an unstable outsole shape, was examined [Nigg, et al., 2006; Romkes, et al., 2006], and it was found that the muscle activity of TA was increased in the tested footwear in a static standing position [Nigg, et al., 2006] and during walking [Romkes, et al., 2006]. Edwards et al. [2008] noticed an increase of muscle activation in quadriceps muscles of subjects when wearing footwear with higher heel height.

The present investigation is original in examining the shank muscle activity in subjects with excessive rearfoot pronation and comparing between motion control footwear and neutral footwear. In this part of the study, the subjects were asked to finish a 10km run on a treadmill. The increased mileage would have magnified the potentially minute difference in the change in muscle activity in each 1-km interval.

The results indicated that the level of fatigue in PL was significantly higher when running with neutral footwear. A similar trend was noticed in TA but the difference has just fell short of statistical significance. The major ankle stabilizing muscles were less prone to muscle fatigue in motion control footwear condition. This higher fatigue resistance in motion control footwear can be due to the lowering in muscle

recruitment required for controlling rearfoot kinematics. Such reduction in muscle recruitment could therefore lead to higher resistance against muscle fatigue.

This postulation was confirmed with higher normalized RMS of TA and PL in runners with neutral footwear. This greater demand of ankle stabilizing muscles was expected to counteract the increasing rearfoot pronation with mileage.

One of the assumptions in measuring the MF of selected muscles was the change in muscle length in different footwear condition was comparable. In a controlled running speed (8km/ hour) provided by treadmill, the number of steps per checkpoint were found to be similar in both motion control (85.6 ± 1.2) and neutral footwear conditions (84.9 ± 0.7). It indicated that the step frequency and step length were similar in the subjects in different test trials. With the fact that the body height of the subjects recruited in this study was highly comparable (1.72m ± 0.06), similarity in the change of muscle length was thus assumed.

8.1.4 Quadriceps muscle activity control

As stated in section 1.5, excessive rearfoot pronation may affect proximal joint biomechanics through erratic internal tibial rotation and compensatory femoral rotation and cause symptoms in the proximal body parts. Among the different running injuries, the knee is the most commonly injured body part and patellofemoral pain

syndrome (PFPS) is one of the most prevalent knee conditions.

Although there is ample of research in patellofemoral pain syndrome since the 1970s, its pathology is still not well understood [Outerbridge, 2001]. One of the classical speculations on the cause of PFPS is malfunctioning of the active muscular system [Cowan, et al., 2001; Voight & Weider, 1991]. A manifestation of active system malfunctioning is the delay onset of vastus medialis obliquus (VMO) relative to vastus lateralis (VL) [Cowan, et al., 2001; Voight & Weider, 1991]. Therefore, muscle onset sequence training with biofeedback and other rehabilitation exercises have been advocated as appropriate treatment strategies for PFPS [Yip & Ng, 2006; Ng, et al., 2008].

The current investigation is the first research to examine the quadriceps activity in recreational runners with motion control footwear. The results suggested that, if subjects with excessive rearfoot pronation were equipped with motion control footwear, the temporal activation of vastus medialis obliquus (VMO) would precede than that of vastus lateralis (VL). Whereas when the subjects put on neutral footwear, later onset of VMO than VL was observed. Moreover, such delay in onset of VMO was increasing with running mileage.

The delayed onset of VMO activity was shown when the subjects were running with neutral shoe and the difference in onset timing between the two muscles was

positively correlated with running mileage. It was noted that most of the subjects were running with 80 steps per minute during the test, thus the time for completing each step was about 0.75 seconds. The VMO was activated at about 4.0 milliseconds earlier than VL when the subjects were running with motion control shoe, whereas the same muscle would activate at about 3.4 milliseconds later than VL when running with neutral shoe condition. Cowan et al. [2001] found that patients with PFPS demonstrated around 10 milliseconds of delay in VMO onset when performing stair climbing activity than normal healthy subject. Neptune et al. [2000] reported that a delay in VMO activation by as little as 5 milliseconds than VL would affect the patellar tracking, measured by medio-lateral patellofemoral contact pressure. In a fast movement such as running, the timing of activities would be shorter than that of stair climbing thus the difference in onset timing of activation of VMO and VL observed in this study could be a risk factor of PFP development.

With a longer running mileage, the delay in VMO onset timing was found to increase in the neutral shoe condition but not so for the motion control shoe condition. In Chapter 3 the results showed that the excessive rearfoot pronation was controlled when subjects were running with motion control shoe even under the muscle fatigue condition. These findings are suggestive that the motion control shoe could facilitate a stable temporal activity of VMO on top of providing overpronation control.

Reduction in MF of quadriceps was observed during the running bout in both shoe conditions. Such a shift in MF was found to have preferentially occurred in VMO (9.51 Hz) than VL (1.78 Hz) in the neutral shoe condition. This suggests that VMO could have developed more fatigue than VL in the neutral shoe running condition. As VMO is the most important dynamic stabilizer that realigns the patella during extension of the knee and maintains the patellar tracking during knee movements [Lieb & Perry, 1968], fatigue of VMO could lead to unbalanced lateral shifting of patella, which was also a potential risk factor of PFP [McConnell, 1986].

It is believed that both delay onset of VMO and fatigue of this muscle were two common risk factors for patellofemoral pain. The findings suggested that the motion control footwear has provided a more favorable condition, in terms of quadriceps muscle activity, for subjects with excessive rearfoot pronation who engage in running to avoid the most common running related injuries such as PFPS.

8.1.5 Relationship between motion control footwear and running injury

In light of the lower limbs resembling a close kinetic chain arrangement during running, as well as the unique architecture of the subtalar joint, rearfoot pronation is always accompanied by internal tibial rotation [Hintermann, et al., 1994]. Therefore, erractic internal tibial rotation can be resulted by overpronation [Cook, et al., 1990].

To counteract the tibial rotation, femur may also perform compensatory rotation [Levens, et al., 1949; Powers, et al., 2002; Reischl, et al., 1999; Cheung & Ng, 2006]. Thus, excessive rearfoot pronation not only induces higher local joint loading on the foot structures, but it may also alter the proximal joint kinematics and kinetics. Locally, excessive rearfoot pronation was associated with Achilles tendonitis and plantar fasciitis. Indirectly, posterior tibial syndrome (shin splints) and patellofemoral pain syndrome are suggested to be related to high pronation amplitude during running [Cook, et al., 1990; van Mechelen, 1992].

The efficacy of motion control footwear in injury prevention and symptom control was not well established. According to previous epidemiological studies, excessive rearfoot pronation was associated with Achilles tendonitis and plantar fasciitis [Cook et al., 1990; van Mechelen et al., 1992] in the foot structure and with the kinetic linkage of the entire lower limb, it may also lead to the development of posterior tibial syndrome (shin splints) [Messier & Pittala, 1988; Viitasalo & Kvist, 1983] and patellofemoral pain syndrome (PFPS) higher up in the kinetic chain. [Cheung & Ng, 2007; Ghani Zadeh Hesar, et al., 2009].

Changes in symptoms in runners that run with different rearfoot pronation controlling devices were studied [Gross, et al., 1991; Donoghue, et al., 2008; Vicenzion, et al., 1997; 2000; Smith, et al., 2004]. In a large sample research by Gross

et al. [1991], a very high proportion of around 90% of long distance runners reported that the use of orthosis was effective in symptoms control of their various running injuries. Donoghue et al. [2008] tested the effects of anti-pronation orthotic device in runners with Achilles tendon injury, but interestingly, the kinematics evaluation did not reveal a successful rearfoot pronation control by orthosis. However, around 90% of the recruited subjects of their studies reported improvement in terms of reduction in severity of the symptoms. Anti-pronation taping was another commonly adopted method to reduce rearfoot pronation in clinical setting and this technique was suggested to be effective in controlling excessive rearfoot pronation [Vicenzino, et al., 1997; 2000]. Clinically, Smith et al. [2004] reported a single case study that the anti-pronation taping technique to have promoted a 10-fold increase of pain-free jogging distance in an active athlete. Besides that, there is no higher level of evidence available to examine the clinical efficacy of such taping technique.

Compared to the quantity of reports on orthosis or taping technique, clinical footwear research is still in its germinating stage. No experimental study to test the efficacy of motion control footwear on prevention or treating running injuries was available. Because the rationale of motion control footwear was completely different with that of orthosis or taping technique, direct application of the data from these research studies should be treated with caution.

8.2 LIMITATIONS

8.2.1 Accuracy of shoe markers to represent rearfoot movement

The use of high speed camera has been used in assessing the foot motion since the early 1980's [Frederick, 1986]. Markers were attached on the footwear to track the rearfoot motion. However, if the movements of shoe markers were not identical with movements of the skin markers, measurements obtained from shoe markers may be problematic. By cutting a small window at the heel counter area of the footwear, both Clarke et al. [1984] and Nigg [1986] had found the difference between the shoe markers and the skin markers, which were located within the heel counter windows, to be very small (less than 3 degrees) and the error was systemic.

Due to the fact that there was no research study to show if there was any kinematics difference between footwear with and without heel counter window cutting, in the present investigations, the overall footwear construct was therefore kept intact.

8.2.2 Gender of subjects

In this study, only female subjects were recruited. This sampling method would minimize the variance due to gender difference despite the limitation on the

generalizability of this study to both genders. The major reason for not including male subjects in this study was because of the potential difference in running biomechanics between male and female [Decker, et al., 2003; Ferber, et al., 2003; Lephart, et al., 2002]. Whether the present findings of motion control footwear could be applied to male subjects is still unknown.

8.2.3 Asymptomatic subject pool

The subject pool in this study comprised only healthy individuals without any active signs or symptoms. The effect of motion control footwear on subjects with running injuries or symptomatic patients is still questionable. Future study is warranted to test the efficacy of motion control footwear on symptom control in subjects with different running injury conditions.

8.2.4 Skill level of the subjects

Exclusion of professional runners in this study was based on the fact that this population was under expert care of running team. Coaches or team physiotherapists may spot out the runners' potential foot problem and deliver treatment. This kind of support may bias the runners' subjective feedback on the motion control footwear function. Also, intensively trained professional runners would have a much higher

resistance towards muscle fatigue. Even in the state of fatigue, these runners may have some unique strategy to cope with further mileage of running.

In this study, the skill level of the subjects was defined only in terms of their training habit, but not other physiological measures such as cardiovascular function or track record. This method may subject to a mixture of runners in different performance and skill level.

8.2.5 Treadmill running

Treadmill running was used in this series of studies in consideration that the set up has the advantages of: 1) providing a speed controlled environment for long distance running; and 2) being able to simulate the general trend of the new generation of exercising in a fitness club.

However, treadmill running may not simulate the actual over-ground running. The fact that the current experiments being conducted with treadmill running may limit the generalization of the results to outdoor running because there may be potential difference in the kinematics and EMG activities of the key muscles as compared with outdoor running [Wank, et al., 1998], caution is therefore needed when applying the results of this study to other conditions.

8.2.6 Definition of overpronation

The operational definition of overpronators in this study was a maximum rearfoot pronation of more than 6^{o} during running than in normal standing position. However, there is no consensus in the literature for the cut off pronation angle. The cut-off pronation angle taken in this study was the mean value of the previous studies that assessed overpronators [Hintermann & Nigg, 1998; Eng & Pierrynowski, 1994; Johnson, et al., 1994; LeLievre, 1970]. The clinical applicability of this cut off value has not been tested and the results may not be generalized to people with different amount of rearfoot pronation.

8.3 FUTURE STUDY

8.3.1 Research quality

Footwear research is still under development with studies from single or very small population to more structured clinical trials as well as detail mechanical testing reports. However, to provide more solid ground work for evidence based practice, randomized controlled trials are highly advocated. A common problem in current clinical trials is the lack of subject randomization and blinding method. This resulted in poor rating in the Jadad scale, thus rendering it difficult to conduct high level meta analysis.

8.3.2 Kinematics of proximal segments

The current studies provide new information on the kinematics, kinetics and neuromuscular control of running with different footwear. In light of the lower limbs resembling a close kinetic chain arrangement during running, the effects of footwear may not be limited to the foot structures but also affecting the more proximal muscle controls of the lower limb thus having a potential influence on the control of the proximal joints. However, change of kinematics of tibial rotation, femoral rotation and patella motion with respect to different footwear are still unanswered by the investigations in this thesis. The authors had tried using the method of 3-dimensional tracking device suggested by Laprade and Lee [2005] to quantify the patella kinematics during running. However, compared to a simple knee flexion extension in unloaded manner, running with the setup involving magnetic tracking sensors required non-metallic environment. Also in the pilot testing, the thermoplastic cap for the patellar fixation was often detached during vigorous movements in running.

There were only 2 studies that investigated the proximal bony segment kinematics with respect to footwear modification. Crenshaw et al. [2000] tested wedged footwear and normal footwear on the kinematics of hip and knee by a motion analysis system at 60 Hz. The subjects did not demonstrate any kinematics difference

in both joints. However, the subjects in this experiment were asked to walk in normal comfortable pace instead of running, the mechanical effects of the footwear might not be totally revealed. A more recent study [Eslami, et al., 2007] that recruited 16 healthy subjects to run in a laboratory runway with either barefoot or in shoe condition was not able to find any significant difference of tibial rotation between the two conditions. However, the function of the type of footwear being tested was not mentioned and therefore, simple conclusion could not be drawn.

8.3.3 Effectiveness of motion control footwear in running injury prevention

The present studies provided information about the basic functional performance of motion control footwear in healthy subjects. However, further study involving clinical outcomes e.g. prospective epidemiological study is recommended. One of the suggested clinical research designs could be recruitment of runners with excessive rearfoot pronation problem. Those subjects could be assigned into a treatment group (motion control footwear) and a control group (neutral footwear). Symptoms of various running injuries could be traced in a longitudinal research manner. Results from this suggested research design will provide the evidence of motion control footwear in running injury prevention.

8.3.4 Adjunct therapy for running injury by footwear adjustment

Apart from running injury prevention, motion control footwear could be tested in injured runners with excessive rearfoot pronation. Eng & Pierrynowski [1994] examined the extra effects of orthosis on top of regular physiotherapy in subjects with patellofemoral pain syndrome. The patients receiving physiotherapy treatment and foot orthosis demonstrated better outcome than the patient group that received physiotherapy treatment only. The results of that study [Eng & Pierrynowski, 1994] had shed light on the clinical benefits from rearfoot posture control in treating proximal joint problems.

8.4 CONCLUSION

Experiments in this series of studies suggested motion control footwear provides potential benefits to runners with excessive rearfoot pronation. This technology is able to control rearfoot overpronation, check excessive plantar loading on the medial foot structures, as well as provide a stable activation pattern in shank stabilizing muscles and better VMO/ VL firing pattern during prolonged running. These factors might be associated with different running injuries.

As the subjects were not able to differentiate the footwear function by their subjective feeling, subjects with excessive rearfoot pronation are recommended to

have foot posture assessment by medical professional such as physical therapist and podiatrist, and run with appropriate footwear.

Future studies are recommended to further substantiate the clinical efficacy of motion control footwear in running injury prevention and treatment.

< End >

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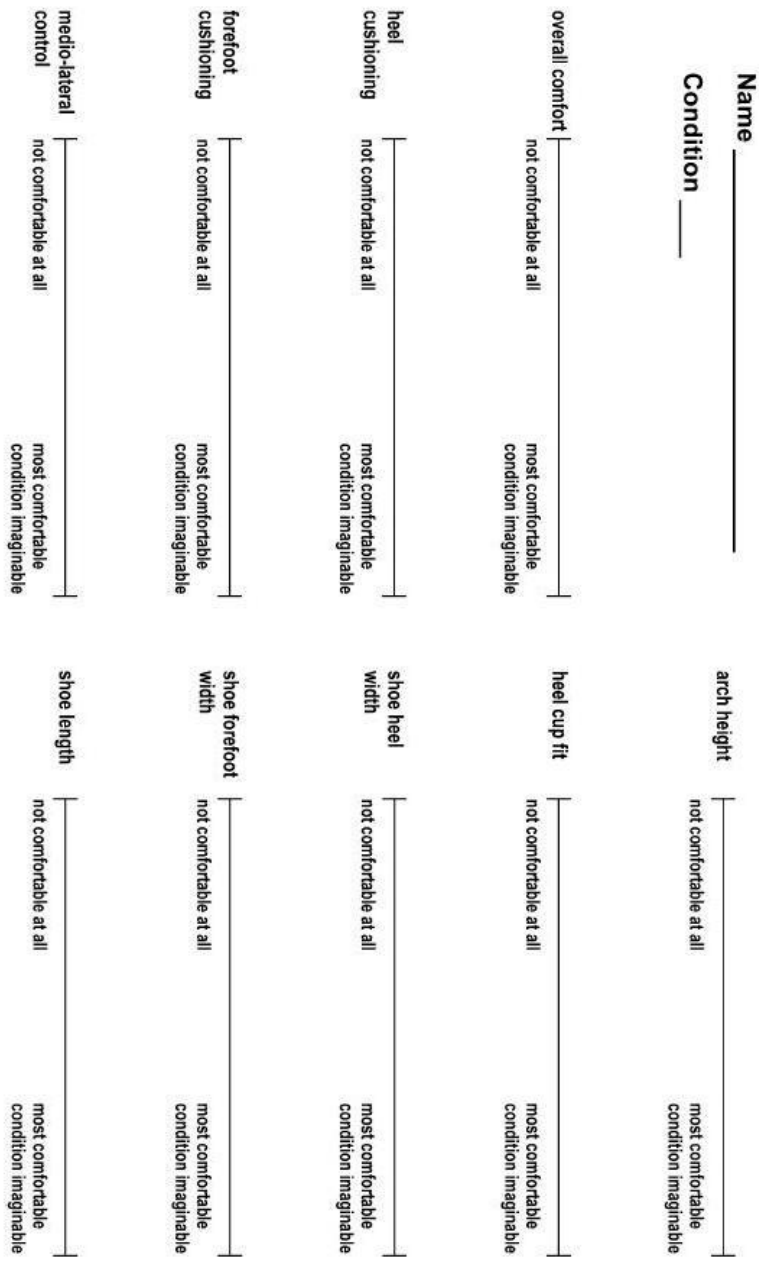
APPENDIX I

Questionnaire sample to evaluate subjective feedback on pronation control function in test footwear



APPENDIX II

Ethical approval by the Human Ethics Sub-committee of the HKPolyU in Chapter 2

THE HONG KONG POLYTECHNIC UNIVERSITY
香港理工大學

MEMO

To : NG Yin Fat, Department of Rehabilitation Sciences

From :KWONG Shek Chuen, For Chairman, Faculty Research Committee, Faculty of Health & Social Sciences

Ethical Review of Research Project Involving Human Subjects

I write to inform you that approval has been given to your application for human subjects ethics review of the following research project for a period from 02/03/2004 to 01/01/2010:

Project Title : Efficacy of motion control shoes in exhausted runners: a 3-dimensional kinematics & kinetic analysis

Department : Department of Rehabilitation Sciences

Please note that you will be held responsible for the ethical approval granted for the project and the ethical conduct of the research personnel involved in the project. In the case the Co-PI has also obtained ethical approval for the project, the Co-PI will also assume the responsibility in respect of the ethical approval (in relation to the areas of expertise of respective Co-PI in accordance with the stipulations given by the approving authority).

You are responsible for informing the Faculty Research Committee Faculty of Health & Social Sciences in advance of any changes in the research proposal or procedures which may affect the validity of this ethical approval.

You will receive separate notification should you be required to obtain fresh approval.

KWONG Shek Chuen
For Chairman
Faculty Research Committee
Faculty of Health & Social Sciences

APPENDIX III

Informed written consent form sample in Chapter 2

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences
Research Project Informed Consent Form

Project title: Efficacy of motion control shoes in exhausted runners: a 3-dimensional kinematics and kinetic analysis

Investigator: CHEUNG Roy, MPhil student, Department of RS, PolyU
Supervisors: Prof. NG Gabriel, Professor, Department of RS, PolyU

Project information:
Running injury is commonly associated with uncontrolled rearfoot motion. This study aims to investigate the effectiveness of motion control shoes, in particular, during muscle fatigue, which is vital for injury prevention.

This study will take place at the gait laboratory in PolyU. You will need to come for 2 sessions of testing and each session will last for about an hour and they will be at one week apart. During the tests, you will wear different foot-wears and run on a treadmill at speed 10 kilometers per hour for 1500 meters. Reflective markers will be attached to your leg and an insole will be put into the shoe so as to record your lower limb movements and forces of landing. Also the ankle force will be measured before and after the run.

The above procedures do not have any known risk. You shall not have any pain or discomfort sensation except for the normal feeling of tiredness after running. This sensation should disappear within 1 or 2 days after the exercise.

Your involvement will not be of direct benefit to you. However, findings of this study will be useful in determining the effectiveness of different shoe types for controlling lower limb movements and muscle activities during running.

Consent:
I, _____, have been explained the details of this study. I voluntarily consent to participate in this study. I understand that I can withdraw from this study at any time without giving reasons, and my withdrawal will not lead to any punishment or prejudice against me. I am aware of any potential risk in joining this study. I also understand that my personal information will not be disclosed to people who are not related to this study and my name or photograph will not appear on any publications resulted from this study.

I can contact the investigator, Roy Cheung, at telephone 9670 for any questions about this study. If I have complaints related to the investigator(s), I can contact Mrs. Michelle Leung, secretary of Departmental Research Committee, at 27665397. I know I will be given a signed copy of this consent form.

Signature (subject): Date:

Signature (witness): Date:

APPENDIX IV

Ethical approval by the Human Ethics Sub-committee of the HKPolyU in Chapter 5

THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

MEMO

To : NG Yin Fat, Department of Rehabilitation Sciences
From : KWONG Shek Chuen, Chairman, Faculty Research Committee, Faculty of Health & Social Sciences

For

Ethical Review of Research Project Involving Human Subjects

I write to inform you that approval has been given to your application for human subjects ethics review of the following research project for a period from 23/04/2007 to 01/01/2010:

Project Title : Effects of footwear on leg muscle activities during a 10-km run

Department : Department of Rehabilitation Sciences

Principal Investigator : NG Yin Fat

Please note that you will be held responsible for the ethical approval granted for the project and the ethical conduct of the research personnel involved in the project. In the case the Co-PI has also obtained ethical approval for the project, the Co-PI will also assume the responsibility in respect of the ethical approval (in relation to the areas of expertise of respective Co-PI in accordance with the stipulations given by the approving authority).

You are responsible for informing the Faculty Research Committee Faculty of Health & Social Sciences in advance of any changes in the research proposal or procedures which may affect the validity of this ethical approval.

You will receive separate notification should you be required to obtain fresh approval.

For KWONG Shek Chuen
Chairman
Faculty Research Committee
Faculty of Health & Social Sciences

APPENDIX V

Informed written consent form sample in Chapter 5

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences
Research Project Informed Consent Form

Project title: Efficacy of motion control shoes on leg muscle activity

Investigator: CHEUNG Roy, MPhil student, Department of RS, PolyU
Supervisors: Prof. NG Gabriel, Professor, Department of RS, PolyU

Project information:
Running injury is commonly associated with muscle imbalance of lower extremity. This study aims to investigate the leg muscle activity when runners in different shoe conditions. Such information is vital for injury prevention.

This study will take place at the Exercise laboratory in PolyU. You will need to come for 2 sessions of testing and each session will last for about an hour and they will be at one week apart. During the tests, you will wear different footwear and run on a treadmill at speed 8 kilometers per hour for 10 km. Electrodes will be attached to your leg. Also force production by ankle and knee will be measured before the run.

The above procedures do not have any known risk. You shall not have any pain or discomfort sensation except for the normal feeling of tiredness after running. This sensation should disappear within 1 or 2 days after the exercise.

Your involvement will not be of direct benefit to you. However, findings of this study will be useful in determining the effectiveness of different shoe types for controlling lower limb movements and muscle activities during running.

Consent:
I, _____, have been explained the details of this study. I voluntarily consent to participate in this study. I understand that I can withdraw from this study at any time without giving reasons, and my withdrawal will not lead to any punishment or prejudice against me. I am aware of any potential risk in joining this study. I also understand that my personal information will not be disclosed to people who are not related to this study and my name or photograph will not appear on any publications resulted from this study.

I can contact the investigator, Roy Cheung, at telephone 9670 for any questions about this study. If I have complaints related to the investigator(s), I can contact Mrs. Michelle Leung, secretary of Departmental Research Committee, at 27665397. I know I will be given a signed copy of this consent form.

Signature (subject): Date:

Signature (witness): Date:

APPENDIX VI

Publications arising from the thesis

1. Cheung RTH, Ng GYF, Chen, BFC. (2006) Association of footwear with patellofemoral pain syndrome in runners. *Sports Medicine*, 36 (3), 199-205.
2. Cheung RTH, Ng GYF. (2007) A systematic review of running shoes and lower leg biomechanics: a possible link with patellofemoral pain syndrome? *International SportMed Journal*, 8 (3), 107-116.
3. Cheung RTH, Ng GYF. (2007) Efficacy of motion control shoes for reducing excessive rearfoot motion in fatigued runners. *Physical Therapy in Sports*, 8 (2), 75-81.
4. Cheung RTH, Ng GYF. (2008) Influence of different footwear on force of landing during running. *Physical Therapy*, 88 (5), 620-628.
5. Cheung RTH, Ng GYF. (2008) Motion control shoe affects temporal activity of quadriceps in runners. *British Journal of Sports Medicine*, 43 (12): 943-947.
6. Cheung RTH, Ng GYF. (2010) Motion control shoe delays fatigue of shank muscles in runners with overpronating feet. *American Journal of Sports Medicine*, (In-press)

APPENDIX VII

Award

The Hong Kong Student Conference of Exercise Science,
Health & Rehabilitation 2005

CHEUNG Tsz Hei Roy

Department of Rehabilitation Sciences
The Hong Kong Polytechnic University

has won the Best Presentation Award at the Hong Kong Student Conference of Exercise Science, Health & Rehabilitation on August 27, 2005

Dr. Richard Masters
Chair, Organizing Committee

Professor Bruce Abernethy
Director, Institute of Human Performance

APPENDIX VIII

Raw data

Rearfoot kinematics

Rearfoot Pronation (degree)	Motion control footwear BEFORE 1500m run	Motion control footwear AFTER 1500m run	Neutral footwear BEFORE 1500m run	Neutral footwear AFTER 1500m run
Subject 1	12.2	11.3	17.7	18.7
Subject 2	13.6	12.3	16.4	18.1
Subject 3	11.4	16.9	17.7	25.2
Subject 4	15.8	16.6	18.9	23.4
Subject 5	14.3	18.4	14.2	16.8
Subject 6	14.3	18.4	14.7	15.5
Subject 7	7.6	8.52	10.16	15.4
Subject 8	5.9	7.07	7.2	19.29
Subject 9	4.45	6.2	11.2	17.6
Subject 10	7.38	9.55	9.52	24.41
Subject 11	6.61	7.57	9.61	17.98
Subject 12	6.91	6.95	9.77	16.13
Subject 13	7.25	7.27	11.02	18.93
Subject 14	7.3	7.4	12.3	16.31
Subject 15	8.2	8.42	13.1	17.8
Subject 16	11.2	11.4	15.3	18.5
Subject 17	9.77	9.53	12.77	13.77
Subject 18	12.4	12	16.98	17
Subject 19	10.75	9.89	13.74	14.92
Subject 20	12.35	12.01	15.92	16.43
Subject 21	14.21	14.22	19.11	20.35
Subject 22	12.3	11.9	16.23	16.42
Subject 23	8.93	9	13.5	13.2
Subject 24	10.36	10.11	14.46	12.53
Subject 25	18.99	18.24	17.09	18.43

Plantar loading at the medial midfoot area

Plantar loading (N)	Motion control footwear BEFORE 1500m run	Motion control footwear AFTER 1500m run	Neutral footwear BEFORE 1500m run	Neutral footwear AFTER 1500m run
Subject 1	208	222	617	734
Subject 2	184	170	600	710
Subject 3	230	209	585	699
Subject 4	168	260	519	677
Subject 5	192	335	445	626
Subject 6	230	422	410	627
Subject 7	300	508	372	590
Subject 8	379	578	339	548
Subject 9	474	623	277	486
Subject 10	546	610	280	448
Subject 11	579	542	245	383
Subject 12	551	511	302	346
Subject 13	521	497	282	283
Subject 14	509	493	257	292
Subject 15	508	477	244	283
Subject 16	510	452	256	265
Subject 17	487	416	264	323
Subject 18	465	365	267	312
Subject 19	435	347	268	281
Subject 20	391	329	262	278
Subject 21	349	292	304	274
Subject 22	317	261	411	261
Subject 23	289	247	452	251
Subject 24	262	261	447	246
Subject 25	215	221	396	227

Plantar loading at the first metatarsal head area

Plantar loading (N)	Motion control footwear BEFORE 1500m run	Motion control footwear AFTER 1500m run	Neutral footwear BEFORE 1500m run	Neutral footwear AFTER 1500m run
Subject 1	684	674	636	668
Subject 2	811	746	738	732
Subject 3	533	497	571	596
Subject 4	483	499	209	459
Subject 5	502	568	540	595
Subject 6	661	632	567	585
Subject 7	582	656	531	759
Subject 8	537	550	547	568
Subject 9	464	442	914	940
Subject 10	517	448	371	354
Subject 11	564	568	519	534
Subject 12	384	423	516	612
Subject 13	462	445	418	430
Subject 14	417	508	438	457
Subject 15	513	517	438	451
Subject 16	472	480	450	479
Subject 17	547	563	560	581
Subject 18	474	454	926	952
Subject 19	527	459	382	365
Subject 20	574	578	529	544
Subject 21	394	441	534	630
Subject 22	472	452	425	437
Subject 23	427	517	447	466
Subject 24	523	522	443	456
Subject 25	482	495	465	494

Maximum voluntary contraction (MVC) force production

MVC force production (kgf)	Motion control footwear BEFORE 1500m run	Motion control footwear AFTER 1500m run	Neutral footwear BEFORE 1500m run	Neutral footwear AFTER 1500m run
Subject 1	19.27	10.27	18.6	10.07
Subject 2	20.43	8.37	21.67	11.2
Subject 3	26.67	9.6	30.34	17.2
Subject 4	18.4	11.2	18.8	10.9
Subject 5	14.8	8.1	15.1	9.1
Subject 6	20.7	17.4	20.7	16.7
Subject 7	16.8	9.33	14.97	9.73
Subject 8	14.6	8.87	12.67	8.33
Subject 9	10.4	6.37	12.07	6.6
Subject 10	13.4	9.63	17.27	11.8
Subject 11	9.53	5.2	8.2	5.53
Subject 12	13	9.33	12.53	8.07
Subject 13	17.1	12.1	17.4	10.2
Subject 14	14	9.67	15.27	8.67
Subject 15	16.33	11.2	17.87	10.57
Subject 16	14.2	8.2	13.8	10.1
Subject 17	16.8	8.2	17.09	11.25
Subject 18	15	9.45	15.23	10.56
Subject 19	22.3	16.3	23.88	17.98
Subject 20	19.08	15.3	20.12	15.54
Subject 21	13.1	9.27	11.21	8.78
Subject 22	15.32	12.4	14.26	9.48
Subject 23	9.79	6.23	8.99	6.94
Subject 24	18.77	13.13	18.24	12.79
Subject 25	15.98	10.77	16.65	9.96

Subjective score for footwear medial-lateral control

VAS (0-1)	Motion Control Footwear	Neutral Footwear
Subject 1	.30	.56
Subject 2	.70	.34
Subject 3	.50	.80
Subject 4	.92	.52
Subject 5	.66	.53
Subject 6	.28	.80
Subject 7	.38	.42
Subject 8	.61	.52
Subject 9	.66	.38
Subject 10	.47	.41
Subject 11	.48	.39
Subject 12	.75	.74
Subject 13	.69	.72
Subject 14	.45	.50
Subject 15	.30	.54
Subject 16	.54	.71
Subject 17	.53	.61
Subject 18	.43	.40
Subject 19	.70	.70
Subject 20	.50	.80
Subject 21	.46	.50
Subject 22	.51	.49
Subject 23	.71	.67
Subject 24	.23	.20
Subject 25	.65	.55

Normalized root mean square electromyographic signal from the tibialis anterior in Neutral Footwear condition

Normalized RMS EMG	CP1	CP2	CP3	CP4	CP5	CP6	CP7	CP8	CP9	CP10
Subject 1	0.64	0.63	0.65	0.67	0.67	0.71	0.7	0.72	0.73	0.71
Subject 2	0.42	0.41	0.43	0.42	0.49	0.47	0.51	0.48	0.47	0.5
Subject 3	0.33	0.33	0.35	0.35	0.34	0.61	0.4	0.41	0.42	0.33
Subject 4	0.36	0.4	0.41	0.41	0.4	0.39	0.42	0.44	0.47	0.46
Subject 5	0.53	0.55	0.54	0.56	0.58	0.61	0.61	0.6	0.59	0.63
Subject 6	0.33	0.34	0.35	0.36	0.37	0.36	0.37	0.41	0.39	0.41
Subject 7	0.35	0.33	0.36	0.34	0.38	0.39	0.33	0.41	0.45	0.47
Subject 8	0.4	0.49	0.44	0.45	0.46	0.45	0.48	0.51	0.44	0.52
Subject 9	0.51	0.5	0.49	0.53	0.55	0.54	0.5	0.56	0.57	0.56
Subject 10	0.42	0.44	0.46	0.51	0.52	0.5	0.53	0.49	0.54	0.54
Subject 11	0.39	0.4	0.42	0.41	0.49	0.46	0.47	0.47	0.48	0.49
Subject 12	0.47	0.51	0.53	0.51	0.53	0.57	0.58	0.59	0.59	0.59
Subject 13	0.46	0.47	0.48	0.44	0.49	0.51	0.53	0.55	0.54	0.54
Subject 14	0.61	0.68	0.67	0.67	0.63	0.68	0.63	0.62	0.67	0.68
Subject 15	0.55	0.54	0.5	0.6	0.61	0.66	0.6	0.67	0.61	0.68
Subject 16	0.51	0.48	0.46	0.49	0.53	0.55	0.57	0.56	0.56	0.58
Subject 17	0.41	0.42	0.48	0.43	0.42	0.44	0.48	0.49	0.53	0.5
Subject 18	0.43	0.43	0.43	0.44	0.42	0.49	0.46	0.51	0.48	0.53
Subject 19	0.49	0.47	0.49	0.44	0.53	0.57	0.58	0.61	0.6	0.62
Subject 20	0.47	0.49	0.49	0.56	0.61	0.62	0.65	0.62	0.59	0.6

Normalized root mean square electromyographic signal from the tibialis anterior in Motion Control Footwear condition

Normalized RMS EMG	CP1	CP2	CP3	CP4	CP5	CP6	CP7	CP8	CP9	CP10
Subject 1	0.54	0.51	0.55	0.58	0.5	0.52	0.56	0.58	0.61	0.59
Subject 2	0.34	0.33	0.34	0.3	0.29	0.33	0.34	0.36	0.37	0.39
Subject 3	0.27	0.27	0.28	0.3	0.26	0.29	0.27	0.32	0.3	0.31
Subject 4	0.31	0.3	0.35	0.33	0.33	0.32	0.34	0.29	0.3	0.34
Subject 5	0.48	0.45	0.43	0.45	0.5	0.5	0.49	0.51	0.48	0.51
Subject 6	0.32	0.35	0.33	0.33	0.34	0.37	0.39	0.41	0.36	0.37
Subject 7	0.36	0.37	0.34	0.39	0.4	0.4	0.42	0.41	0.41	0.41
Subject 8	0.37	0.39	0.38	0.39	0.4	0.41	0.42	0.41	0.43	0.44
Subject 9	0.46	0.44	0.43	0.44	0.43	0.47	0.44	0.45	0.41	0.44
Subject 10	0.31	0.33	0.34	0.32	0.31	0.3	0.36	0.34	0.32	0.31
Subject 11	0.41	0.39	0.38	0.39	0.41	0.4	0.37	0.38	0.4	0.43
Subject 12	0.39	0.38	0.4	0.42	0.43	0.4	0.39	0.38	0.41	0.43
Subject 13	0.39	0.38	0.33	0.37	0.41	0.34	0.44	0.38	0.41	0.4
Subject 14	0.51	0.49	0.48	0.48	0.5	0.52	0.54	0.53	0.53	0.52
Subject 15	0.5	0.51	0.48	0.49	0.46	0.51	0.5	0.52	0.5	0.51
Subject 16	0.48	0.39	0.41	0.42	0.45	0.46	0.44	0.48	0.42	0.5
Subject 17	0.37	0.36	0.32	0.34	0.4	0.41	0.42	0.4	0.42	0.39
Subject 18	0.36	0.37	0.33	0.35	0.34	0.36	0.36	0.34	0.37	0.36
Subject 19	0.41	0.39	0.4	0.38	0.37	0.33	0.39	0.42	0.42	0.41
Subject 20	0.32	0.3	0.33	0.34	0.31	0.3	0.29	0.33	0.34	0.35

Normalized root mean square electromyographic signal from the peroneus longus in Neutral Footwear condition

Normalized RMS EMG	CP1	CP2	CP3	CP4	CP5	CP6	CP7	CP8	CP9	CP10
Subject 1	0.27	0.3	0.31	0.33	0.34	0.37	0.35	0.34	0.37	0.38
Subject 2	0.3	0.33	0.34	0.31	0.37	0.38	0.35	0.39	0.41	0.4
Subject 3	0.21	0.25	0.26	0.27	0.3	0.31	0.33	0.3	0.34	0.33
Subject 4	0.33	0.35	0.4	0.41	0.45	0.41	0.42	0.46	0.44	0.45
Subject 5	0.18	0.2	0.22	0.23	0.23	0.27	0.25	0.26	0.28	0.3
Subject 6	0.27	0.3	0.31	0.33	0.33	0.34	0.32	0.38	0.37	0.36
Subject 7	0.24	0.24	0.23	0.27	0.29	0.29	0.3	0.33	0.34	0.33
Subject 8	0.2	0.19	0.21	0.23	0.27	0.26	0.26	0.29	0.3	0.33
Subject 9	0.3	0.33	0.37	0.36	0.39	0.41	0.34	0.39	0.39	0.42
Subject 10	0.34	0.37	0.39	0.41	0.42	0.41	0.47	0.46	0.44	0.43
Subject 11	0.36	0.39	0.41	0.42	0.43	0.49	0.48	0.48	0.49	0.48
Subject 12	0.4	0.42	0.49	0.49	0.48	0.49	0.42	0.49	0.51	0.5
Subject 13	0.24	0.28	0.29	0.33	0.34	0.36	0.32	0.35	0.36	0.36
Subject 14	0.18	0.2	0.19	0.23	0.27	0.29	0.31	0.33	0.34	0.36
Subject 15	0.26	0.28	0.29	0.28	0.3	0.33	0.34	0.32	0.35	0.37
Subject 16	0.21	0.22	0.25	0.26	0.25	0.28	0.3	0.33	0.31	0.31
Subject 17	0.18	0.2	0.21	0.23	0.22	0.21	0.25	0.26	0.27	0.26
Subject 18	0.29	0.31	0.34	0.35	0.36	0.34	0.33	0.37	0.38	0.37
Subject 19	0.22	0.25	0.3	0.26	0.31	0.33	0.34	0.38	0.36	0.33
Subject 20	0.24	0.2	0.3	0.31	0.31	0.3	0.33	0.28	0.29	0.33

Normalized root mean square electromyographic signal from the peroneus longus in Motion Control Footwear condition

Normalized RMS EMG	CP1	CP2	CP3	CP4	CP5	CP6	CP7	CP8	CP9	CP10
Subject 1	0.21	0.22	0.2	0.19	0.2	0.22	0.23	0.21	0.22	0.22
Subject 2	0.28	0.26	0.24	0.25	0.27	0.27	0.28	0.29	0.28	0.3
Subject 3	0.18	0.19	0.2	0.19	0.18	0.21	0.2	0.21	0.2	0.19
Subject 4	0.27	0.27	0.26	0.28	0.29	0.28	0.25	0.29	0.3	0.28
Subject 5	0.17	0.18	0.19	0.18	0.18	0.19	0.2	0.19	0.19	0.18
Subject 6	0.24	0.23	0.22	0.22	0.2	0.22	0.21	0.23	0.23	0.27
Subject 7	0.25	0.25	0.26	0.24	0.26	0.27	0.24	0.25	0.26	0.25
Subject 8	0.14	0.16	0.15	0.15	0.16	0.14	0.13	0.15	0.16	0.14
Subject 9	0.22	0.23	0.22	0.24	0.23	0.22	0.21	0.22	0.2	0.21
Subject 10	0.29	0.3	0.28	0.27	0.27	0.28	0.29	0.3	0.3	0.31
Subject 11	0.33	0.33	0.29	0.3	0.33	0.31	0.31	0.33	0.34	0.31
Subject 12	0.31	0.32	0.31	0.3	0.33	0.32	0.3	0.31	0.3	0.29
Subject 13	0.23	0.22	0.23	0.24	0.22	0.2	0.21	0.24	0.25	0.24
Subject 14	0.13	0.14	0.15	0.14	0.16	0.15	0.15	0.15	0.16	0.14
Subject 15	0.24	0.24	0.25	0.26	0.25	0.26	0.28	0.24	0.26	0.23
Subject 16	0.19	0.2	0.21	0.2	0.2	0.21	0.23	0.21	0.21	0.22
Subject 17	0.17	0.18	0.2	0.18	0.17	0.18	0.18	0.21	0.19	0.2
Subject 18	0.23	0.25	0.27	0.23	0.26	0.26	0.24	0.26	0.27	0.25
Subject 19	0.21	0.22	0.2	0.21	0.2	0.23	0.22	0.23	0.24	0.22
Subject 20	0.22	0.25	0.27	0.26	0.23	0.27	0.24	0.23	0.23	0.24

Shift of median frequency of electromyographic signal from the tibialis anterior (TA) and peroneus longus (PL) after 10km run

Median frequency (Hz)	TA with Motion control footwear	TA with Neutral footwear	PL with Motion control footwear	PL with Neutral footwear
Subject 1	9.32	13.74	-3.69	12.21
Subject 2	-10.33	65.04	-4.04	8.65
Subject 3	9.04	18.86	-10.73	4.15
Subject 4	4.85	13.66	3.11	12.77
Subject 5	5.61	10.03	5.71	10.31
Subject 6	3.84	4.8	-0.25	11.77
Subject 7	10.87	8.56	3.98	19.11
Subject 8	-7.7	-2.03	-0.83	17.05
Subject 9	9.11	10.98	3.82	6.89
Subject 10	11.32	9.02	3.87	23.87
Subject 11	-2.36	10.01	4.03	6.32
Subject 12	5.56	7.08	8.52	3.32
Subject 13	4.35	23.5	4.08	12.2
Subject 14	13.17	3.77	2.31	22.51
Subject 15	11.63	12.31	-1.02	14.26
Subject 16	9.84	3.32	4.02	8.49
Subject 17	-1.25	-1.34	7.4	0.99
Subject 18	12.31	13.24	4.47	11.18
Subject 19	9.01	13.42	3.67	12.69
Subject 20	1.97	20.79	3.53	13.21

* Positive value indicates a downward shift of median frequency

Median frequency of electromyographic signal from vasti muscles

Median frequency (Hz)	VMO with Motion control footwear BEFORE 10km run	VMO with Motion control footwear AFTER 10km run	VMO with Neutral footwear BEFORE 10km run	VMO with Neutral footwear BEFORE 10km run	VL with Motion control footwear BEFORE 10km run	VL with Motion control footwear AFTER 10km run	VL with Neutral footwear BEFORE 10km run	VMO with Neutral footwear BEFORE 10km run
Subject 1	87.96	84.12	72.62	56.13	80.87	75.62	89.67	83.24
Subject 2	87.56	85.33	98.99	83.41	92.29	85.48	104.15	102.53
Subject 3	67.82	68.28	72.88	63.00	53.43	48.37	73.33	78.30
Subject 4	74.91	76.68	105.19	85.88	64.48	66.21	82.68	85.09
Subject 5	92.55	91.76	72.99	62.77	76.02	74.74	89.16	85.98
Subject 6	79.26	78.99	82.25	72.16	62.48	56.06	103.64	100.22
Subject 7	64.85	64.85	88.98	86.93	83.81	81.55	73.28	68.82
Subject 8	101.52	99.10	103.57	96.47	78.05	75.67	82.37	83.17
Subject 9	96.62	95.46	73.21	72.55	77.29	71.69	88.68	86.58
Subject 10	77.17	74.49	82.27	80.15	99.84	93.16	103.61	100.45
Subject 11	60.46	60.92	88.67	79.33	96.99	88.57	72.68	69.40
Subject 12	84.66	83.21	104.06	100.35	73.47	68.88	82.53	78.32
Subject 13	80.65	77.12	72.51	66.22	60.85	58.76	89.58	87.45
Subject 14	79.52	76.64	82.31	68.95	85.13	80.30	103.29	101.33
Subject 15	84.66	83.08	88.90	74.07	76.90	71.32	72.78	74.85
Subject 16	106.44	104.95	103.58	92.48	99.06	96.62	81.78	80.07
Subject 17	102.01	101.94	72.94	62.55	95.12	91.14	89.00	89.09
Subject 18	87.33	86.37	81.69	79.01	75.90	71.53	103.69	99.85
Subject 19	82.55	81.49	89.41	74.14	60.47	62.53	73.25	71.51
Subject 20	64.48	63.48	82.16	72.37	85.26	83.58	83.51	80.84

Onset time differences between VMO and VL during 10km run in motion control footwear

% of duty cycle	CP1	CP2	CP3	CP4	CP5	CP6	CP7	CP8	CP9	CP10
Subject 1	-0.12	-0.03	0.01	0.00	-0.01	-0.03	-0.06	-0.09	-0.01	0.00
Subject 2	0.00	-0.11	-0.04	-0.10	-0.07	-0.12	-0.05	-0.08	-0.09	-0.04
Subject 3	-0.11	-0.03	-0.05	-0.06	-0.01	-0.02	-0.02	-0.04	0.00	-0.05
Subject 4	-0.06	-0.02	-0.01	-0.09	-0.06	-0.08	-0.09	-0.11	-0.07	-0.12
Subject 5	-0.12	-0.03	0.01	-0.01	-0.02	-0.06	-0.05	-0.10	-0.01	0.00
Subject 6	-0.01	-0.12	-0.05	-0.08	-0.07	-0.12	-0.04	-0.08	-0.08	-0.04
Subject 7	-0.11	-0.03	-0.04	-0.03	-0.01	-0.03	-0.03	-0.03	0.00	-0.05
Subject 8	-0.05	-0.02	-0.09	-0.05	-0.04	-0.09	-0.07	-0.08	-0.07	-0.13
Subject 9	-0.13	-0.03	0.01	-0.01	-0.02	-0.04	-0.07	-0.10	-0.01	0.00
Subject 10	-0.01	-0.12	-0.05	-0.07	-0.07	-0.10	-0.05	-0.07	-0.09	-0.04
Subject 11	-0.11	-0.03	-0.04	-0.07	-0.02	-0.02	-0.03	-0.06	0.00	-0.05
Subject 12	-0.06	-0.02	-0.01	-0.09	-0.07	-0.08	-0.07	-0.10	-0.07	-0.12
Subject 13	-0.13	-0.03	0.01	-0.01	-0.02	-0.03	-0.06	-0.07	-0.01	0.00
Subject 14	0.00	-0.14	-0.05	-0.10	-0.07	-0.12	-0.05	-0.09	-0.09	-0.05
Subject 15	-0.11	-0.05	-0.04	-0.03	-0.03	-0.03	-0.03	-0.04	0.00	-0.05
Subject 16	-0.06	-0.03	-0.07	-0.09	-0.04	-0.07	-0.09	-0.09	-0.07	-0.06
Subject 17	-0.13	-0.04	0.00	-0.01	-0.02	-0.01	-0.06	-0.01	-0.01	0.00
Subject 18	-0.01	-0.12	-0.07	-0.08	-0.09	-0.09	-0.06	-0.09	-0.09	-0.01
Subject 19	-0.11	-0.04	-0.07	-0.05	-0.02	-0.03	-0.02	-0.01	0.00	-0.01
Subject 20	-0.07	-0.02	-0.06	-0.07	-0.06	-0.08	-0.07	-0.09	-0.07	0.01

* Positive values indicate that VL activated before VMO and negative values indicate that VMO activation precedes VL

Onset time differences between VMO and VL during 10km run in neutral footwear

% of duty cycle	CP1	CP2	CP3	CP4	CP5	CP6	CP7	CP8	CP9	CP10
Subject 1	0.1	0.03	0.02	0.04	0.02	0.03	0.05	0.03	0.05	0.02
Subject 2	-0.01	0.06	0.07	0.06	0.05	0.07	0.04	0.06	0.08	0.08
Subject 3	-0.02	0.05	0.08	0.07	0.13	0.08	0.04	0.03	0.07	0.15
Subject 4	-0.01	-0.03	-0.01	-0.03	-0.03	0.02	0.06	0.11	0.05	0.11
Subject 5	0.13	0.03	0.02	0.04	0.02	0.03	0.03	0.03	0.04	0.01
Subject 6	-0.01	0.06	0.07	0.06	0.05	0.07	0.04	0.06	0.07	0.08
Subject 7	-0.02	0.05	0.08	0.07	0.13	0.08	0.03	0.03	0.06	0.19
Subject 8	-0.01	-0.03	-0.01	-0.03	-0.03	0.02	0.01	0.11	0.11	0.13
Subject 9	0.1	0.03	0.02	0.04	0.02	0.03	0.03	0.03	0.05	0.02
Subject 10	-0.01	0.06	0.07	0.06	0.05	0.07	0.05	0.06	0.06	0.08
Subject 11	-0.02	0.05	0.08	0.07	0.13	0.08	0.02	0.03	0.04	0.06
Subject 12	-0.01	-0.03	-0.01	-0.03	-0.03	0.02	0.11	0.11	0.07	0.08
Subject 13	0.08	0.03	0.02	0.04	0.02	0.03	0.11	0.03	0.06	0.03
Subject 14	0.01	0.06	0.07	0.06	0.05	0.07	0.05	0.06	0.09	0.04
Subject 15	-0.02	0.05	0.08	0.07	0.13	0.08	0.01	0.03	0.09	0.03
Subject 16	-0.02	-0.03	-0.01	-0.03	-0.03	0.02	0.11	0.11	0.14	0.05
Subject 17	0.03	0.03	0.02	0.04	0.02	0.03	0.03	0.03	0.08	0.05
Subject 18	-0.01	0.06	0.07	0.06	0.05	0.07	0.02	0.06	0.11	0.08
Subject 19	-0.02	0.05	0.08	0.07	0.13	0.08	0.02	0.03	0.1	0.08
Subject 20	-0.01	-0.03	-0.01	-0.03	-0.03	0.02	0.01	0.11	0.07	0.08

* Positive values indicate that VL activated before VMO and negative values indicate that VMO activation precedes VL