

2000

Force And Electromyographic Responses To Ergometer Rowing

Darryl A. Turner
Edith Cowan University

Follow this and additional works at: <https://ro.ecu.edu.au/theses>



Part of the [Sports Sciences Commons](#)

Recommended Citation

Turner, D. A. (2000). *Force And Electromyographic Responses To Ergometer Rowing*. Edith Cowan University. Retrieved from <https://ro.ecu.edu.au/theses/1544>

This Thesis is posted at Research Online.
<https://ro.ecu.edu.au/theses/1544>

Theses

Theses: Doctorates and Masters

Edith Cowan University

Year 2000

Force And Electromyographic Responses
To Ergometer Rowing

Darryl A. Turner
Edith Cowan University

This paper is posted at Research Online.
<http://ro.ecu.edu.au/theses/1544>

Edith Cowan University

Copyright Warning

You may print or download ONE copy of this document for the purpose of your own research or study.

The University does not authorize you to copy, communicate or otherwise make available electronically to any other person any copyright material contained on this site.

You are reminded of the following:

- Copyright owners are entitled to take legal action against persons who infringe their copyright.
- A reproduction of material that is protected by copyright may be a copyright infringement. Where the reproduction of such material is done without attribution of authorship, with false attribution of authorship or the authorship is treated in a derogatory manner, this may be a breach of the author's moral rights contained in Part IX of the Copyright Act 1968 (Cth).
- Courts have the power to impose a wide range of civil and criminal sanctions for infringement of copyright, infringement of moral rights and other offences under the Copyright Act 1968 (Cth). Higher penalties may apply, and higher damages may be awarded, for offences and infringements involving the conversion of material into digital or electronic form.

USE OF THESIS

The Use of Thesis statement is not included in this version of the thesis.

**FORCE AND ELECTROMYOGRAPHIC RESPONSES TO
ERGOMETER ROWING**

By

Darryl Alan Turner B.Sc. Hons. (Sports Science)

This thesis is presented for the award of Doctor of Philosophy (Sports Science) at the School of Biomedical and Sports Science, Faculty of Communications, Health and Science, Edith Cowan University, Perth, Western Australia.

Date of Submission 08/08/2000

ABSTRACT

During many athletic events, fatigue influences the physiological and biomechanical characteristics of performance. For optimal performance in events such as rowing, athletes must maintain a skillfully co-ordinated movement technique, which involves the major muscle groups contributing to force output. The ability to produce high force outputs during repetitive contractions is influenced by fatigue and dependent on a number of factors including neuromuscular activation.

Neuromuscular activation may be expressed by amplitude and frequency characteristics of the electromyographic signal (EMG) sampled from the muscle. During sustained isometric contractions, changes in EMG characteristics are related to changes in force, which may be useful in monitoring the fatigue process (Basmajian, 1974; De Luca, 1985). However, the force-EMG relationship is not as clear when applied to dynamic contractions such as those in rowing performance.

The central objective of this thesis is to assess the application of EMG in relation to biomechanical and physiological responses to rowing tasks. In particular, EMG characteristics of the quadriceps muscle in relation to total force output during a typical self-paced rowing ergometer performance. In order to reach the objective, five studies were undertaken in a systematic order. The studies had specific purposes, which included establishing force/torque-EMG relationships under controlled conditions and evaluating the transfer of force or torque output and EMG characteristics to less controlled performance conditions that were influenced by fatigue, pacing strategy, or both.

In this investigation, trials ($N = 117$) were conducted on three ergometers affording varying levels of control over muscle length, contraction velocity, and muscle contribution to force output. Subjects ($n = 11$) that participated were selected from trained rowing crews. Trials were performed on an isokinetic dynamometer with analogue outputs of angular rotation and torque recorded. In addition, trials were performed on a leg-only ergometer and a standard rowing ergometer with performance outcomes recorded using a potentiometer to measure handle position and a strain gauge to measure force output. Bipolar surface electrodes were used to record EMG activity of the rectus femoris and vastus lateralis muscles during all contractions. Biomechanical and EMG data were recorded on a data acquisition system (Amlab).

Results validated the force/torque-rmsEMG relationship during non-fatiguing isometric, isokinetic, and dynamic contractions. During fatiguing contractions performed on the rowing ergometer, strength of the force/torque-rmsEMG relationship was reduced and subject responses varied widely. Under the same condition, handle force and mean power frequency (MPF) of the vastus lateralis muscle showed a positive correlation, which might therefore be used to monitor fatigue during simulated rowing performance. EMG analysis was more appropriate when the exercise protocol was similar to that used during performance. Finally, a constant-pace strategy significantly reduced force loss and was associated with qualitative muscle activation changes that potentially might improve performance outcome. In conclusion, EMG analysis is constrained by methodological and confounding factors during dynamic exercise. Nevertheless EMG provides an insight into neural activation strategies during rowing fatigue and may be a useful tool for monitoring co-ordinated muscle activity and for devising strategies to improve performance.

DECLARATION

"I certify that this thesis does not incorporate without acknowledgement any material previously submitted for a degree or diploma in any institution of higher education; and that to the best of my knowledge and belief it does not contain any material previously published or written by another person except where due reference is made in the text."

Signed:

Date:

20/02/01

ACKNOWLEDGEMENTS

I would like to express my gratitude and acknowledge the considerable time and effort provided by Dr Paul Sacco during the last four years. His guidance and support enabled completion of the manuscript. This required considerable perseverance and sacrifice of Paul's personal time.

I would like to thank Professor Ross Sanders who provided biomechanical expertise, motivation to finish the project, and encouragement to continue with applied research in the future.

I am grateful to Associate Professor Barry Gibson and the School of Biomedical and Sports Science at Edith Cowan University for total support in every respect. In addition, for providing financial assistance and encouragement to present work at two international conferences and two national conferences, which resulted in two conference proceeding publications and one published referred article. Furthermore, I would like to thank the secretarial and technical support staff at the School of Biomedical and Sports Science for their open-ended assistance whenever requested.

Finally, I would like to express the deepest gratitude to my wife Sue and children Jamie, Kelly, Daniel, Marc, and Maria who have made countless sacrifices on my behalf on what has seemed a never ending road. I am truly indebted. Thank you.

TABLE OF CONTENTS

<u>ABSTRACT</u>	<u>II</u>
<u>DECLARATION</u>	<u>IV</u>
<u>ACKNOWLEDGEMENTS</u>	<u>V</u>
<u>LIST OF TABLES</u>	<u>XV</u>
<u>LIST OF FIGURES</u>	<u>XVI</u>
<u>CHAPTER 1</u>	<u>1</u>
 <u>INTRODUCTION</u>	
1.1 BACKGROUND TO THE PROBLEM	1
1.2 THE PROBLEM	6
1.3 PURPOSE	7
1.4 RATIONALE NEED AND SIGNIFICANCE	8
1.5 DEFINITION OF TERMS	10
1.5.1 Ergometer	10
1.5.2 Biomechanical	11
1.5.3 Physiological	12
1.6 ABBREVIATIONS	14

REVIEW OF THE LITERATURE

2.1	INTRODUCTION	16
2.2	ROWING EVENT	16
2.3	ROWING ERGOMETER USE AND DEVELOPMENT	17
2.4	FACTORS AFFECTING ROWING PERFORMANCE	23
2.4.1	Biomechanical	23
2.4.2	Physiological	26
2.5	PHYSIOLOGY OF MUSCLE FATIGUE IN RESPECT TO ROWING	28
2.5.1	Mechanism of Force Generation	28
2.5.2	Muscle Fatigue	31
2.5.3	Energy Metabolism in Rowing	32
2.6	PACING STRATEGY	35
2.7	BASIS OF EMG	38
2.8	USE OF EMG IN PHYSIOLOGY	41
2.8.1	Force-EMG Relationship	41
	<i>Force-EMG amplitude</i>	41
	<i>Force-EMG frequency</i>	44
2.8.2	Fatigue and EMG	45
	<i>Fatigue and EMG amplitude</i>	46
	<i>Fatigue and EMG frequency</i>	49
2.8.3	Timing of EMG in Relation to Muscle Force	53
2.9	METHODOLOGICAL ASPECTS OF EMG	55
2.9.1	Electrode Configuration	55
2.9.2	Stationarity	56
2.9.3	Muscle and Skin Temperature	58
2.9.4	Impedance	59

CHAPTER 3**60****METHODS**

3.1	EMG MEASUREMENT	60
3.1.1	Subject Preparation	60
3.1.2	Amlab System	62
3.1.3	EMG Signal Processing	63
	<i>EMG signal</i>	63
	<i>Root mean square of EMG (rmsEMG)</i>	63
	<i>Mean power frequency of EMG (MPF)</i>	64
3.1.4	Signal Noise	64
3.1.5	Data Recording	65
3.2	BIOMECHANICAL MEASUREMENT	66
3.2.1	Force	66
	<i>Rowing ergometer force</i>	66
	<i>Adapted leg-only ergometer force</i>	70
	<i>Isokinetic dynamometer torque</i>	70
3.2.2	Displacement	72
	<i>Rowing ergometer displacement</i>	72
	<i>Adapted leg-only ergometer displacement</i>	75
	<i>Isokinetic dynamometer displacement</i>	75
3.2.3	Calculated Biomechanical Parameters	76
	<i>Velocity</i>	76
	<i>Power</i>	76
3.3	BLOOD LACTATE MEASUREMENT	76
3.4	DATA TREATMENT	77

**THE RELATIONSHIP BETWEEN FORCE OR TORQUE OUTPUT AND EMG
OF LEG EXTENSORS DURING ISOMETRIC, ISOKINETIC, AND DYNAMIC
CONTRACTIONS**

4.1	INTRODUCTION	78
4.2	METHODS	80
4.2.1	Subjects	81
4.2.2	Apparatus	81
4.2.3	EMG	85
4.2.4	Procedure	86
4.2.5	Analysis	88
4.3	RESULTS	90
4.3.1	Force/torque-rmsEMG	90
4.3.2	Force/torque-MPF	98
4.3.3	Ergometer and Muscle Effects	98
4.4	DISCUSSION	99
4.4.1	Force/torque-rmsEMG Relationship	99
4.4.2	Force/torque-MPF Relationship	104
4.4.3	Ergometer and Muscle Effects	105
4.5	CONCLUSIONS	112

CHAPTER 5**113****THE RELATIONSHIP BETWEEN FORCE OR TORQUE OUTPUT AND EMG
OF LEG EXTENSORS DURING FATIGUING ISOKINETIC OR DYNAMIC
CONTRACTIONS**

5.1	INTRODUCTION	113
5.2	METHODS	115
5.2.1	Subjects	115
5.2.2	Apparatus	116
5.2.3	EMG	116
5.2.4	Procedure	117
5.2.5	Analysis	118
5.3	RESULTS	119
5.3.1	Force or Torque Loss and Final Blood Lactate	119
5.3.2	Force/torque-rmsEMG	121
5.3.3	Force/torque-MPF	128
5.3.4	Ergometer and Muscle Effects	129
5.4	DISCUSSION	132
5.4.1	Force or Torque Loss and Final Blood Lactate	132
5.4.2	Force/torque-rmsEMG Relationship	133
5.4.3	Force/torque-MPF Relationship	138
5.4.4	Ergometer and Muscle Effects	141
5.5	CONCLUSIONS	146

**COMPARISON OF RESPONSES IN FORCE AND EMG BETWEEN A
REPEATED MVC PROTOCOL AND A SELF-PACE PROTOCOL
PERFORMED ON A ROWING ERGOMETER**

6.1	INTRODUCTION	148
6.2	METHODS	150
6.2.1	Subjects	150
6.2.2	Apparatus	150
6.2.3	EMG	150
6.2.4	Procedure	151
6.2.5	Analysis	152
6.3	RESULTS	152
6.3.1	Performance Distance and Final Blood Lactate	152
6.3.2	Force Loss	153
6.3.3	Final MVC rmsEMG	153
6.3.4	Final MVC MPF	154
6.4	DISCUSSION	159
6.4.1	Performance Distance and Blood Lactate Response	159
6.4.2	Comparison of Force Changes	160
6.4.3	Comparison of MVC rmsEMG Changes	162
6.4.4	Comparison of MVC MPF Changes	165
6.5	CONCLUSIONS	166

**RELIABILITY OF FORCE AND EMG CHANGES DURING A SELF-PACE
ROWING ERGOMETER TASK**

7.1	INTRODUCTION	168
7.2	METHODS	170
7.2.1	Subjects	170
7.2.2	Apparatus	171
7.2.3	EMG	171
7.2.4	Procedure	171
7.2.5	Analysis	173
7.3	RESULTS	173
7.3.1	Performance Distance	173
7.3.2	Blood Lactate	175
7.3.3	Force Loss	175
7.3.4	Final MVC rmsEMG	176
7.3.5	Final MVC MPF	177
7.4	DISCUSSION	180
7.4.1	Reliability of Performance Distance	180
7.4.2	Reliability of Final Blood Lactate Concentration	181
7.4.3	Reliability of Force Loss	182
7.4.4	Reliability of Final MVC rmsEMG	184
7.4.5	Reliability of Final MVC MPF	187
7.5	CONCLUSIONS	190

CHAPTER 8**191****EFFECT OF PACING STRATEGY ON BIOMECHANICAL AND
PHYSIOLOGICAL RESPONSES TO A 2000-M ROWING ERGOMETER TASK**

8.1	INTRODUCTION	191
8.2	METHODS	193
8.2.1	Subjects	193
8.2.2	Apparatus	193
8.2.3	EMG	193
8.2.4	Procedure	194
8.2.5	Analysis	195
8.3	RESULTS	196
8.3.1	Performance Distance	196
8.3.2	Force and Power Output	196
8.3.3	rmsEMG	203
8.3.4	MPF	204
8.3.5	Blood Lactate, Heart Rate, and RPE	209
8.4	DISCUSSION	211
8.4.1	Pacing Effect on Performance	211
8.4.2	Pacing Effect on Force and Power Output	212
8.4.3	Pacing Effect on rmsEMG	214
8.4.4	Pacing Effect on MPF	217
8.4.5	Pacing Effect on Blood Lactate, Heart Rate, and RPE	218
8.5	CONCLUSIONS	221

CHAPTER 9	SUMMARY AND RECOMMENDATIONS	223
BIBLIOGRAPHY		231
APPENDIX A	AMLAB SCHEMATIC PROJECTS	250
APPENDIX B	STATEMENT OF DISCLOSURE	257
APPENDIX C	UNIVERSITY ETHICAL APPROVAL	259
APPENDIX D	WARM-UP PROTOCOL	261
APPENDIX E	TABLES	262
Table E1		262
Table E2		263
Table E3		264
Table E4		265

LIST OF TABLES

Table 1	Mean (SD) of Force/torque-rmsEMG Relationships During Isometric Random Force or Torque Contractions for Two Ergometer Types and Two Muscles	91
Table 2	Mean (SD) of Force/torque-rmsEMG Relationships During Dynamic or Isokinetic Random Force or Torque Contractions for Three Ergometer Types and Two Muscles	92
Table 3	Mean (SD) of Initial and Final Maximal Voluntary Contraction (MVC) Force or Torque Output and Final Blood Lactate in Response to Repeated MVCs for Three Ergometer Types	120
Table 4	Mean (SD) of Force/torque-rmsEMG Relationships During Repeated Maximal Voluntary Contractions (MVCs) for Three Ergometer Types and Two Muscles	123
Table 5	Mean (SD) of Force/torque-MPF Relationships During Repeated Maximal Voluntary Contractions (MVCs) for Three Ergometer Types and Two Muscles	130
Table 6	Comparison of Responses Between a Repeated Maximal Voluntary (MVC) Protocol and a Self-pace Protocol Performed on a Rowing Ergometer ($N = 8$)	155
Table 7	Reliability of Performance, Final Maximal Voluntary Contraction (MVC) Force and EMG Between 6-minute Self-pace Ergometer Trials ($N = 10$)	174
Table 8	Comparison of Performance Parameters Between Self-pace and Constant-pace 6-minute Ergometer Trials ($N = 7$)	198
Table 9	Comparison of EMG Parameters Between Self-pace and Constant-pace 6-minute Ergometer Trials ($N = 7$)	205

LIST OF FIGURES

Figure 1.	Chain of neural command (Loeb & Gans, 1986; Westerblad et al., 1991).	30
Figure 2.	Electrode sites for rectus femoris (RF), vastus lateralis (VL) and greater trochanter (GT).	61
Figure 3.	Arrangement of strain gauge attachment between rowing ergometer chain and handle.	67
Figure 4.	Calibration set-up for strain gauge.	69
Figure 5.	Modified stock for adapted leg-only ergometer.	71
Figure 6.	Ergometer potentiometer and pulley system (rear view, lateral view).	73
Figure 7.	Subject set-up for isokinetic dynamometer (Cybex 6000) used in T_{cyb} .	82
Figure 8.	Adapted leg-only ergometer used in T_{adap} .	83
Figure 9.	Subject set-up for standard rowing ergometer (Concept II) used in T_{stan} .	84
Figure 10.	Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and torque or force for the same subject during isometric contractions performed on a) the Cybex isokinetic dynamometer in T_{cyb} and b) the adapted leg-only ergometer in T_{adap} .	93
Figure 11.	Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and torque or force for the same subject during isokinetic or dynamic contractions performed on a) the Cybex isokinetic dynamometer in T_{cyb} , b) the adapted leg-only ergometer in T_{adap} , and (c) the standard rowing ergometer in T_{stan} .	94
Figure 12.	Relationship between torque output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) during a) isometric and b) isokinetic leg extensions performed on the dynamometer in T_{cyb} . Values are normalized and expressed as a percentage of value at maximal voluntary contraction (% MVC).	95

Figure 13.	Relationship between force output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) during a) isometric and b) dynamic leg extensions performed on the adapted leg-only ergometer in T_{adap} . Values are normalized and expressed as a percentage of value at maximal voluntary contraction (% MVC).	96
Figure 14.	Relationship between force and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) during dynamic contractions performed on the standard rowing ergometer in T_{stan} . Values are normalized and expressed as a percentage of value at maximal voluntary contraction (% MVC).	97
Figure 15.	Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and torque for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during 2-minutes of repeated MVCs performed on the Cybex isokinetic dynamometer in T_{cyb} .	124
Figure 16.	Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during 5-minutes of repeated MVCs performed on the adapted leg-only ergometer in T_{adap} .	125
Figure 17.	Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during 2-minutes of repeated MVCs performed on the standard rowing ergometer in T_{stan} .	126
Figure 18.	Relationship between torque or force output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) for all isokinetic or dynamic contractions performed on the a) Cybex dynamometer in T_{cyb} , b) adapted leg-only ergometer in T_{adap} , and c) standard rowing ergometer in T_{stan} . Values are averaged each five contractions then normalized and expressed as a percentage of value at initial maximal voluntary contraction (% MVC).	127
Figure 19.	Relationship between torque or force output and MPF of rectus femoris (RF) and vastus lateralis (VL) for all isokinetic or dynamic contractions performed on the a) Cybex dynamometer in T_{cyb} , b) adapted leg-only ergometer in T_{adap} , and c) standard rowing ergometer in T_{stan} . Values are averaged each five contractions then normalized and expressed as a percentage of value at initial maximal voluntary contraction (% MVC).	131

<u>Figure 20.</u>	Sample traces for EMG and rectified EMG of vastus lateralis (VL), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 2-minute repeated MVC effort performed on a rowing ergometer in T_{2-min} .	156
<u>Figure 21.</u>	Sample traces for EMG and rectified EMG of vastus lateralis (VL), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_{6-min} .	157
<u>Figure 22.</u>	Sample traces of the power frequency spectrum of vastus lateralis (VL) for the same subject recorded from initial and final maximal voluntary contraction (MVC) during a) the 2-minute repeated MVC effort in T_{2-min} and b) the 6-minute self-pace effort in T_{6-min} performed on a rowing ergometer.	158
<u>Figure 23.</u>	Sample traces for EMG and rectified EMG of vastus lateralis (VL), and force for the same subject as in T_2 recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_1 .	178
<u>Figure 24.</u>	Sample traces of EMG and rectified EMG of vastus lateralis (VL), and force for the same subject as in T_1 recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_2 .	179
<u>Figure 25.</u>	Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) performance distance and for b) final blood lactate for each subject. $N = 7$.	199
<u>Figure 26.</u>	Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in mean force output recorded each minute and b) final maximal force output for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by handle force > 50 N. $N = 7$, ** = $p < .01$.	200

Figure 20.	Sample traces for EMG and rectified EMG of vastus lateralis (VL), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 2-minute repeated MVC effort performed on a rowing ergometer in T_{2-min} .	156
Figure 21.	Sample traces for EMG and rectified EMG of vastus lateralis (VL), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_{6-min} .	157
Figure 22.	Sample traces of the power frequency spectrum of vastus lateralis (VL) for the same subject recorded from initial and final maximal voluntary contraction (MVC) during a) the 2-minute repeated MVC effort in T_{2-min} and b) the 6-minute self-pace effort in T_{6-min} performed on a rowing ergometer.	158
Figure 23.	Sample traces for EMG and rectified EMG of vastus lateralis (VL), and force for the same subject as in T_2 recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_1 .	178
Figure 24.	Sample traces of EMG and rectified EMG of vastus lateralis (VL), and force for the same subject as in T_1 recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_2 .	179
Figure 25.	Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) performance distance and for b) final blood lactate for each subject. $N = 7$.	199
Figure 26.	Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in mean force output recorded each minute and b) final maximal force output for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by handle force > 50 N. $N = 7$. ** = $p < .01$.	200

- Figure 27.** Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in mean power output recorded each minute and b) final maximal power output for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by handle force > 50 N. $N = 7$. ** = $p < .01$. 201
- Figure 28.** Sample traces of force for pre-maximal voluntary contraction (pre-MVC), initial contraction, and final MVC for the same subject during a 6-minute effort performed on a rowing ergometer using a) the self-pace strategy in T_{self} and b) the constant-pace strategy in T_{con} . 202
- Figure 29.** Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in rmsEMG of vastus lateralis (VL) recorded each minute and b) final rmsEMG (VL) for each subject. Data averaged each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by rmsEMG activation > .7 volts. $N = 7$. 206
- Figure 30.** Sample traces of EMG and rectified EMG of vastus lateralis (VL) for pre-maximal voluntary contraction (pre-MVC), initial contraction, and final MVC for the same subject during a 6-minute effort performed on a rowing ergometer using a) the self-pace strategy in T_{self} and b) the constant-pace strategy in T_{con} . 207
- Figure 31.** Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in MPF of vastus lateralis (VL) recorded each minute and b) final MPF (VL) for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by rmsEMG activation > 0.7 volts. $N = 7$, ** = $p < .01$. 208
- Figure 32.** Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for changes recorded each minute in a) mean heart rate (bpm) and b) rating of perceived exertion (RPE) on the Borg scale. $N = 7$. 210

CHAPTER 1

INTRODUCTION

1.1 BACKGROUND TO THE PROBLEM

Rowing performance may be influenced by many biomechanical factors (kinematic, kinetic, and neuromuscular). Most importantly, these include characteristics of the rowing stroke and the rower's ability to apply force at the handle (Dal Monte & Komor, 1989). Optimal force output measured at the handle is partly dependent on the co-ordination of muscle groups (leg, back, and arm). However, the leg extensor muscles contribute the initial and predominant power during the stroke (Nelson & Widule, 1983; Smith & Spinks, 1995). Efficient activation of the leg extensor muscles that contribute to force applied at the handle is partly dependent on optimal neural activation of the co-ordinated muscle groups involved (Sale, 1988).

During fatiguing contractions, metabolic changes associated with the decline of maximal voluntary force, affects both neural and contractile mechanisms of the muscle (Bigland-Ritchie & Woods, 1984). These fatigue-related changes affect biomechanical characteristics of the stroke (Smith & Spinks, 1995), and timing (Staniak, Nosarzewski, & Karpilowski, 1994) and co-ordination of muscle contributions (Henry, Clark, McCabe, & Vanderby, 1995) that influence optimal performance. In order to optimize performance outcome, it would be useful to monitor changes in neural activity associated with force output during rowing tasks. Moreover, it would be beneficial to

devise strategies that minimize disruption to neuromuscular functions and stroke characteristics associated with fatigue.

One such strategy that may influence neuromuscular functions and stroke characteristics during rowing performance is the pacing strategy. During maximal rowing efforts, athletes manipulate pacing intensity in an attempt to optimize energy contribution, minimize the effects of fatigue, and maximize performance outcome. However, the pacing strategy employed by the majority of rowing crews (Dal Monte & Komor, 1989) has been described as metabolically (Hagerman, 1986) and biomechanically (Zatsiorsky & Yakunin, 1991) inefficient. The effect of different pacing strategies on energy contribution and force output are well-documented (Foster, Schrage, Snyder, & Thompson, 1994), but the effect of pacing on neural activation in relation to force output has not been examined during rowing performance.

Rowing ergometers are used extensively for testing, training, and research purposes, although computerized equipment is available for on-water biomechanical assessment (Smith, Spinks, & Moncrieff, 1992). Researchers have reported physiological responses (Hahn, Ryan, Lawton, Kearns, & Bellenger, 1995b) and biomechanical characteristics (Dawson, Lockwood, Wilson, & Freeman, 1998; Vaughan, 1989) that differ between ergometer and on-water rowing performances. Nevertheless, it is generally accepted that rowing ergometers provide an appropriate simulation of rowing performance. Strain gauges and potentiometers have been used to record and measure force and displacement characteristics that enabled detailed examination of the rowing stroke during simulated rowing (Henry et al., 1995; Smith & Milburn, 1996; Staniak et al., 1994). Few researchers have examined muscle activation characteristics during ergometer rowing (Rodriguez, Rodriguez Cook, & Sandborn, 1990; Wilson, Robertson,

& Stothart, 1988) and even less have examined the relationship between changes in force output and muscle activation during fatiguing performance (Kyrolainen & Smith, 1999). Changes in muscle activation characteristics under these conditions are important to performance outcome but have received little attention to date, partly due to technical restraints associated with recording muscle activation patterns during dynamic movement.

Muscle activation patterns may be examined using electromyography (EMG). Neural drive regulates muscle activation and in its last stage involves the propagation of action potentials along muscle fibers, which initiates contractile force. Neural activation patterns are influenced by recruitment and firing rates of the active motor units and are frequently assessed by the signal of electrical activity sampled above the muscle (electromyogram). EMG signal activity may be expressed quantitatively as amplitude (e.g. root mean square [rmsEMG]) or as spectral power frequency (e.g. mean power frequency [MPF]).

EMG activity is a useful indicator of neuromuscular function during performance tasks (Mannion & Dolan, 1996b) and has been used to evaluate timing of muscle activation, (Arendt-Nielsen & Sinkjaer, 1991), force output (Gamet, Duchene, Garapon-Bar, & Goubel, 1990), and fatigue (Taylor, Bronks, Smith, & Humphries, 1997). EMG shows potential for assessing changes in muscle activation in relation to biomechanical markers, for example, timing and amplitude of muscle activation in relation to muscle force output. Changes in these factors are critical to rowing performance (Kyrolainen & Smith, 1999; Wilson et al., 1988). However, the use of EMG applied to dynamic movement such as rowing is controversial. Basmajian and De Luca (1985) stated that

analysis of the EMG signal relative to muscle force should be restricted to static contractions due to the recording problem: associated with signal stationarity during dynamic contractions. In contrast, Dowling (1988) proposed that the EMG signal might be analyzed during dynamic tasks and represented a non-invasive view of muscle force. Although most athletic tasks involve dynamic movements, relatively few researchers have investigated EMG characteristics under dynamic performance conditions (Arendt-Nielsen & Sinkjaer, 1991; Kyrolainen & Smith, 1999). In order that EMG analysis may be of use in applied situations, the force-EMG relationship should be investigated under these conditions.

The strength of the relationship between force or torque and EMG is partly dependent on the contraction type. During isometric contractions, force and EMG amplitude are well correlated (Basmajian, 1974). During isokinetic contractions when the limb moves at a controlled rate, the strength of the force/torque-EMG amplitude relationship is weaker (Muro, Nagata, & Moritani, 1983). However, during dynamic contractions that involve changes in muscle length, contraction velocity, or muscle contribution, the force-EMG correspondence is not clear. In most athletic situations, changes in these confounding factors affect force output and EMG characteristics (Rodriguez et al., 1990; Wilson et al., 1988).

During repetitive muscle contractions, fatigue affects the force/torque-EMG relationship (Arendt-Nielsen & Sinkjaer, 1991). Change in the EMG during fatiguing performance might reflect biochemical changes in the muscle that influence the pattern of voluntary muscle activation (Christensen, Sogaard, Jensen, Finsen, & Sjogaard, 1995; Hagberg, 1981). Fatigue, which can be defined as a reduction in maximal voluntary contraction

(MVC) force (Bigland-Ritchie & Woods, 1984), is accompanied by changes in characteristics of the EMG (Oda & Moritani, 1995; Tesch, Dudley, Duvoisin, Hather, & Harris, 1990). For example, in response to sustained or repeated MVCs, decreases in maximal voluntary force or torque output are generally accompanied by decreases in amplitude and mean power frequency of the EMG (De Luca, 1985; Fitts, 1994; Tesch, Komi, Jacobs, Karlsson, & Viitasalo, 1983). However, in response to fatiguing submaximal contractions, changes in EMG characteristics are less consistent (Gamet et al., 1990; Helal, Guezennec, & Goubel, 1987). Under submaximal conditions, decreases in maximal voluntary force or torque output are generally accompanied by initial increases followed by decreases in amplitude and mean power frequency of the EMG (Arendt-Nielsen & Mills, 1988). It is evident that EMG responses to fatiguing muscle contractions are influenced by characteristics of the task performed (Enoka & Stuart, 1992). Rowing performance tasks, unlike other commonly used laboratory based exercise protocols, involve self-pace contractions that attempt to reproduce optimal force outputs for the duration of the event. To date, EMG characteristics and related force outputs have not been quantified during self-pace maximal rowing efforts.

1.2 THE PROBLEM

Rowing performance outcomes are partly dependent on optimal muscle activation and resultant force output. There has been a lack of quantitative research investigating muscle activation and EMG changes under performance conditions. The paucity of EMG rowing research may be partly attributed to previous inconsistent EMG responses demonstrated during dynamic studies, particularly those that involved fatigue (Arendt-Nielsen & Sinkjaer, 1991; Van Dieen, Boke, Oosterhuis, & Toussaint, 1996). Under those conditions, subjects demonstrated a wide range of force/torque-rmsEMG and force/torque-MPF relationships (Helal et al., 1987; Komi & Tesch, 1979). The reliability of EMG changes during dynamic conditions has been questioned (Gamet et al., 1990; Kollmitzer, Ebenbichler, & Kopf, 1999).

It is evident that alterations in muscle length, contraction velocity, and relative muscle contribution to force output associated with dynamic contractions may potentially affect the force-EMG relationship. The relationship may be further influenced by fatigue during repeated contractions. To what extent these factors influence the force-EMG relationship is not clear. In order to establish the usefulness for force-EMG relationships during rowing performance, the influences of muscle length, contraction velocity, muscle contribution, and fatigue need to be systematically examined. Force/torque-EMG changes during fatiguing rowing tasks may provide an insight into testing or training protocols that optimize muscle functions and neural activation strategies, which may be used to enhance performance outcome.

1.3 PURPOSE

The central objective of this thesis is to assess the application of EMG in relation to biomechanical and physiological responses to rowing ergometer tasks. In particular, the investigation focuses on effects of fatigue on the relationship between total force output measured at the handle and EMG characteristics of the quadriceps muscle (e.g. force-EMG relationship) during rowing ergometer performance. In order to achieve the objective, five studies were undertaken in a systematic order. The general purpose was to establish force/torque-EMG relationships under controlled conditions and evaluate the transfer of force or torque output and EMG characteristics to less controlled performance conditions that were influenced by fatigue, pacing strategy, or both.

The first study established force/torque-EMG relationships during isometric, isokinetic, and dynamic contractions performed on different ergometer types. The second study examined effects of fatigue on force/torque-EMG relationships using the same three ergometer types that were used in the first study. The third study compared fatigue-related responses in force output and EMG between two different exercise protocols performed on a rowing ergometer. The fourth study examined the reliability of responses to a 6-minute self-pace maximal rowing ergometer effort. The final study compared biomechanical, physiological, and EMG characteristic changes between a self-pace and a constant-pace rowing ergometer performance.

1.4 RATIONALE NEED AND SIGNIFICANCE

Establishing force and EMG characteristic changes during rowing tasks may be of use in the assessment of performance strategies. For example, if changes in quadriceps EMG (rmsEMG, MPF) were related to decline in ergometer force output during rowing tasks, then changes in quadriceps EMG could be used to (a) predict performance outcome, (b) indicate potential contractile failure of the quadriceps muscle group, and (c) devise strategies to optimize muscle activation, force output, and performance outcome. In spite of these potential uses, there remains a lack of research into neural activation changes during rowing performance tasks.

In the initial study (chapter 4) it was necessary to establish force/torque-EMG relationships during non-fatiguing contractions in which there were no changes in muscle length, contraction velocity, or muscle contribution to force output. This was achieved by quantifying force/torque-EMG relationships during isometric contractions performed on an isokinetic dynamometer and on an adapted leg-only ergometer. In order that force/torque-EMG relationships might be applied in performance situations, it was necessary to determine whether the relationships obtained under controlled isometric conditions transferred to practical tasks such as ergometer rowing. For this purpose, trials were designed to investigate the effect of changes in muscle length, contraction velocity, and muscle contribution using three ergometer types. The ergometer types used for dynamic contractions were (a) an isokinetic dynamometer that involved changes in muscle length, (b) an adapted leg-only ergometer that involved changes in muscle length and contraction velocity, and (c) a standard rowing ergometer that involved changes in muscle length, contraction velocity, and muscle contribution.

In the second study (chapter 5), force/torque-EMG relationships were examined in response to fatigue. Force/torque-EMG relationships found in fresh muscle may alter during fatigue. Therefore, the relationships were examined during repeated MVCs performed on an isokinetic dynamometer. In order to examine whether force/torque-EMG relationships under isokinetic conditions transferred to less-controlled dynamic conditions such as rowing tasks, subjects performed repeated MVCs on an adapted leg-only ergometer and on a rowing ergometer on separate occasions.

In the third study (chapter 6), force and EMG responses to different fatigue protocols were examined. In fatigue studies researchers often use repeated MVC protocols, whereas in rowing performance studies self-pace protocols are preferred where contraction intensity is not controlled. In order to investigate whether force and EMG responses to these two exercise protocols were different, a two-minute repeated MVC task was compared to a six-minute self-pace task performed on a rowing ergometer.

In the fourth study (chapter 7), in order that changes in force and EMG characteristics be of practical use in performance assessment, the reliability of changes in force output and quadriceps EMG were examined under self-pace rowing performance conditions.

In the final study (chapter 8), pacing strategies that might minimize fatigue and maximize performance (Foster et al., 1994; Hagerman, Connors, Gault, Hagerman, & Polinski, 1978) were examined. Force output and quadriceps EMG were compared between a constant-pace and self-pace rowing strategy. EMG responses to pacing strategies may provide an indication of muscle activation strategies employed by the central nervous system that optimize force output and performance outcome.

1.5 DEFINITION OF TERMS

1.5.1 Ergometer

Catch: When the ergometer handle reaches the forward most point of travel with respect to the ergometer and is subsequently moved in the opposite direction.

Drive: Commences at the instant of the catch, continues during leg extension followed by back extension and finally arm flexion. The drive is completed prior to the hands moving away from the body.

Recovery: Commences at the end of the drive phase at the instant when the hands move away from the body and the body begins to move in the opposite direction to the drive movement. Continues up until the start of the catch phase.

Ergometer stroke time: The complete cyclic stroke time from the instant of the catch through the drive and recovery phases up until the instant prior to the next catch.

Stroke rate: The number of complete stroke cycles per minute inclusive of drive and recovery phases.

Pace: Normally expressed as time to complete 500 m. Measured continuously and averaged for each stroke, for each 500 m, or for duration of the effort. Pace may be manipulated by changes in stroke rate or stroke power (contraction force x velocity).

Rowing ergometer: Such as the standard Concept II rowing ergometer used for co-ordinated muscle actions involved in the rowing stroke.

Adapted leg-only ergometer: Such as the standard Concept II rowing ergometer adapted with frame attachment and harness, used for leg drive without the assistance of trunk or upper body.

Isokinetic ergometer: Such as the Cybex isokinetic dynamometer used for leg extension exercises.

1.5.2 Biomechanical

Maximal voluntary contraction (MVC): Defined as the average of three or five contractions (see each chapter method) performed with maximal voluntary effort.

Force: Generally describes ergometer handle force and is the component of force applied by the rower at the handle and assumed to be the main component of the propulsive force during the drive phase. Measured in Newtons (N).

Torque: Isokinetic dynamometer torque is the moment of force measured around the axis of rotation during leg extension exercises. Measured in Newton-meters (Nm)

Force/torque-EMG relationship: Describes the relationship between force measured at the handle or torque measured at the axis of rotation and EMG characteristic values of the quadriceps muscle group (see also 1.5.3)

Muscle force: Describes force or torque applied by muscle. In general or non-specific cases in the text, muscle force may describe both force and torque.

MVC force or torque: Defined as the average force or torque of three or five contractions (see each chapter method) performed with maximal voluntary effort.

Force-time curve: A graphical plot of the instantaneous propulsive linear force measured at the ergometer handle for the duration of each contraction.

Torque-time curve: A graphical plot of the instantaneous torque measured at the axis of rotation for the duration of each contraction.

Linear handle displacement: Defined as the horizontal distance moved by the ergometer handle from a fixed pre-calibrated point. Measured in meters (m). Handle displacement represents the distance over which force is applied at the ergometer handle and is required to calculate work done.

Angular displacement: As for linear handle displacement but measures the rotational arc of the dynamometer lever movement. Measured in degrees (deg).

Force-displacement curve: A graphical plot of the instantaneous propulsive linear force measured at the ergometer handle against the displacement of the handle during each contraction.

Torque-displacement curve: A graphical plot of the instantaneous torque measured at the axis of rotation against the rotational arc of the dynamometer lever movement during each contraction.

Work: Ergometer work (linear motion) is the product of force applied at the handle and the displacement of the handle. Measured in Joules (J).

Power: Ergometer power is the rate of doing work and is the product of force applied at the ergometer handle and positive velocity of the handle. Peak power per stroke is the maximum of force \times velocity per stroke. Average power per stroke is measured during the drive phase of the stroke. Measured in Joules/second = Watts (W).

Average handle velocity: The average distance the handle travels each stroke (positive direction) divided by the average time taken while the handle moves in a positive direction each stroke. Describes handle velocity during the drive phase.

Performance outcome: During ergometer tasks of fixed duration (6-min) performance outcome is the distance covered, or alternatively, during tasks over a set distance (2000-m) performance outcome is the time taken.

1.5.3 Physiological

Blood Lactate: The concentration of lactate in a sample of capillary blood drawn from the index finger or earlobe and analyzed with the Accusport. Measured in millimoles per liter (mmol/L).

Fatigue: Loss of MVC force.

Neural activation: The pattern of motor unit recruitment and motor unit firing rates of the muscle.

Neuromuscular activation: The combined mechanisms of neural activation and contractile processes that affect the function of muscle.

Electromyography, Electromyogram, or Electromyographical (EMG):

Electromyography is the recording and measurement of the electromyogram (the signal of electrical activity sampled above the muscle). EMG signal activity may be expressed quantitatively as amplitude (e.g. root mean square [rmsEMG]) or qualitatively as spectral power frequency (e.g. mean power frequency [MPF]).

Root mean square (rmsEMG): Is the root mean square of the recorded EMG waveform expressed as the average signal amplitude. RmsEMG reflects the amount of underlying muscle activity. Measured in millivolts (mV) and calculated for a specific window duration.

Mean power frequency (MPF): Defined as the average frequency of the EMG. MPF is measured in Hertz (Hz) after Fast Fourier Transformation (FFT).

1.6 ABBREVIATIONS

ANT	= anaerobic threshold
AP	= action potential
ATP	= adenosine triphosphate
bpm	= beats per minute
CNS	= central nervous system
diff.n.s.	= difference not significant
EMG	= electromyogram or electromyography or electromyographical
FFT	= Fast Fourier Transformation
Force-MPF	= relationship between force measured at the ergometer handle and mean power frequency (EMG) of the quadriceps muscle
Force-rmsEMG	= relationship between force measured at the ergometer handle and root mean square (EMG) of the quadriceps muscle
Force/rmsEMG	= force measured at the ergometer handle divided by rmsEMG of the quadriceps muscle expressed as a ratio
Force/torque-EMG	= relationship between force measured at the handle or torque measured at the axis of rotation and EMG characteristic values of the quadriceps muscle
IEMG	= integrated electromyography
MF	= median frequency of the EMG
MPF	= mean power frequency of the EMG
MU	= motor unit
MUAP	= motor unit action potential
MVC	= maximal voluntary contraction

rmsEMG	= root mean square of the EMG
RF	= rectus femoris muscle
ROM	= range of motion
RPE	= rating of perceived exertion
rpm	= revolutions per minute
T _{adap}	= adapted leg-only trial (chapter 4, 5)
T _{con}	= constant-pace rowing ergometer trial (chapter 8)
T _{cyb}	= Cybex dynamometer trial (chapter 4, 5)
Torque-MPF	= relationship between torque measured at the axis of rotation and mean power frequency (EMG) of the quadriceps muscle
Torque-rmsEMG	= relationship between torque measured at the axis of rotation and root mean square (EMG) of the quadriceps muscle
T _{self}	= self-pace rowing ergometer trial (chapter 8)
T _{stan}	= standard rowing ergometer trial (chapter 4, 5)
T ₁	= trial one self-pace (chapter 7)
T ₂	= trial two self-pace (chapter 7)
T _{2-min}	= 2-minute repeated MVC trial (chapter 6)
T _{6-min}	= 6-minute self-pace trial (chapter 6)
VE	= minute ventilation
$\dot{V}O_{2\max}$	= maximal oxygen uptake
VL	= vastus lateralis muscle

CHAPTER 2

REVIEW OF THE LITERATURE

2.1 INTRODUCTION

In order to assess the application of EMG in relation to biomechanical and physiological responses to fatiguing rowing tasks, it is necessary to review the literature that addresses the essential aspects of rowing, fatigue, and EMG. The literature review begins with aspects relevant to the rowing event, the development of rowing research, and the use of rowing ergometers for testing, training, and research purposes. The next section introduces biomechanical and physiological factors affecting rowing performance, focusing on the main performance determinants that are relevant to muscle activation. This is followed by discussions on the physiology of fatigue related to rowing performance and pacing strategies used to minimize fatigue. The remaining review sections outline the basis of EMG, discuss the uses of EMG in exercise physiology, and review the factors that affect EMG during performance tasks.

2.2 ROWING EVENT

The event focus of this project is sweep-boat rowing by males. There are various sweep-boat rowing events, each requiring a different combination of crew members and boat types. Sweep-boats are rowed with or without a coxswain in various weight categories. Males compete on a standardized 2000-meter straight river course that takes between six and seven minutes to complete (Secher, 1993). Each rower moves on a sliding seat and propels the boat through the water using a single oar in a co-ordinated

cyclic pattern of movement that involves all the major muscle groups. A race consists of several crews competing against each other using tactical and paced strategies with the objective of finishing the course the quickest. Often a competition involves several rounds of racing on one day, performed at near-maximal intensities with short intervals between races. This format stresses the rower's physiological capacities near to the limit. Testing and training of physiological and biomechanical capacities commonly involve the use of rowing ergometers that have been developed to simulate biomechanical movements and physiological responses of on-water rowing.

2.3 ROWING ERGOMETER USE AND DEVELOPMENT

The development of rowing ergometers came from the need to replicate the on-water rowing movement in a controlled environment. Prior to ergometer development, investigators examined stroke forces using instrumented oars during on-water tasks. Force data were sent to shore using a FM transmitter and then analyzed together with video recordings of the movement (Ishiko, 1971; Schneider, Angst, & Brandt, 1975). On-water rowing investigations accelerated with advances in technology and frequently examined biomechanical parameters of rowing performance. For example, Wing and Woodburn (1995) utilized on-water technology to investigate the co-ordination and consistency of rowers in a racing eight from recorded force profiles. They observed fluctuations in stroke force duration and inter-stroke intervals both in the short and long term. Smith et al. (1992) developed an on-water data analysis system (ROWSYS) capable of recording detailed biomechanical parameters that were instantly transmitted to a portable computer. These workers assembled torque transducers to measure strain on the oar, potentiometers to measure oar angle, and an on-board microcomputer to

record data. McBride and Elliott (1999) used real-time telemetry on-water to record biomechanical parameters related to movement of the rower in the boat and rowing technique in respect to boat velocity. They concluded that in order to achieve optimal boat velocity and minimize boat velocity fluctuations, stroke rate and stroke length should be maximized and oarlock forces attained near to the orthogonal position whilst maintaining movement technique. Previously, on-water rowing analysis was not always successful due to technological and environmental restraints. Ergometers were developed that simulated the rowing movement for testing and training purposes but avoided environmental restraints.

One such ergometer used in early rowing studies was the Gamut-Stanford mechanically braked ergometer, which provided task specificity to sweep rowing but was heavy and expensive (Hagerman, 1996). Recent investigations have used Gjessing ergometers that are mechanically braked or Concept II ergometers that are air-braked. Hahn, Tumity, Shakespear, Rowe, and Telford (1988) compared the Gjessing and Concept II ergometers. They concluded that both were equally useful for monitoring physiological characteristics of rowers and average power outputs were highly correlated between ergometers during simulated racing. Lormes, Buckwitz, Rehbein, and Steinacker (1993) compared the same two ergometers. These authors concluded that higher anaerobic effort was required and that lower maximum power output was achieved on the Gjessing compared to the Concept II. Both ergometers are currently used for research purposes. In addition to the on-water and ergometer evaluations, water tanks have been utilized in rowing studies. Henry et al. (1995) compared a water tank, a Concept II ergometer, and a Stanford ergometer for maximal power output in a 30-s test. They used strain gauges and potentiometers to measure forces and oar positions.

The authors concluded that water tank compared to ergometer tests provided more specific biomechanical data (e.g. technical efficiency, consistency) for interpreting aspects of skill.

Ergometer and on-water responses have not always been similar. For example, the Concept II was used in comparing ergometer and on-water physiological responses during 2000-meter maximal efforts (Hahn et al., 1995b). The investigators reported significantly higher blood lactate responses to ergometer compared to on-water efforts. They attributed the difference to either ergometer stability, greater propulsive muscular activity, or difference in gearing between the ergometer and the boat. Other workers have qualified or questioned the similarity between on-water and ergometer performance. For example, Seiler (1996) suggested that predictions of on-water rowing performance using ergometer tests were accurate within a weight class. Martindale and Robertson (1984) and Rodriguez et al. (1990) reported that the straight pull ergometer (with the self-returning handle) and the on-water sweep action did not require the same muscular effort to clear the oar during recovery.

It is evident that replication of movement actions and physiological responses between ergometer and on-water rowing are not always similar. Nonetheless, rowing ergometers are used extensively for testing and training athletes and provide certain advantages over on-water rowing. Ergometer advantages include (a) the ease and ability to monitor and measure biomechanical factors (e.g. force, velocity, displacement), (b) the control of environmental conditions (e.g. wind, water), (c) the stability of influencing variables for test-retest protocols (e.g. laboratory conditions), (d) the ease in administering physiological tests (e.g. blood lactates), (e) the ease and ability to monitor and control

target pace protocols (e.g. average power), and (f) the ability to change stroke resistance (e.g. drag).

Use of rowing ergometers for research purposes has focussed on three main aspects of performance, which may be grouped into biomechanical, physiological, and electromyographical studies. Rowing ergometers have been used to establish biomechanical characteristics of performance (Jensen, Freedson, & Hamill, 1996; Nelson & Widule, 1983; Smith & Spinks, 1995) that included models of force, velocity, and power (Hartmann, Mader, Wasser, & Klauer, 1993). In order to measure force and angle characteristics of the stroke during simulated rowing, ergometers have been modified with strain gauges and potentiometers (Henry et al., 1995). Smith and Spinks (1995), and Staniak et al. (1994) developed computerized systems with specifically designed software that controlled and processed data from ergometer force and displacement transducers, which enabled assessment of the rowing technique. For example, Smith and Spinks examined the biomechanical differences between novice, good, and elite rowers using an instrumented ergometer. They concluded that biomechanical variables such as consistency of the force profile could be used to discriminate between rowers of different skill levels.

Rowing ergometers have been used to establish physiological characteristics of performance (Hagerman, 1986), effects of training on physiological parameters (Womack et al., 1996), anaerobic thresholds (Beneke, 1995), cardio-respiratory, and metabolic responses (Steinacker, 1993). In one such study, Peltonen and Rusko (1993) investigated the relationships between energy metabolism and biomechanical outputs during leg-only ergometer tests. In that investigation, subjects performed a progressive

test and 2-, 6-, and 12-minute all-out tests. The authors reported that in the 6-minute all-out test average power output was highly correlated with maximal oxygen uptake and blood lactate concentration at the anaerobic threshold. In that study, rowers compensated for loss in force output by increasing the stroke rate, thus preventing a decrease in power output. In another physiological rowing study, Messonnier, Freund, Bourdin, Belli, and Lacour (1997) examined blood lactate kinetics during ergometer performance and concluded that the ability to row at high-intensity work rates was correlated with the increased removal of blood lactate.

Rowing ergometers have been used with EMG methods to establish muscle utilization patterns during the rowing movement (Bompa, 1980). Most EMG rowing studies have focussed on the muscles activated, timing of muscle activation, and co-ordination of various muscle groups. For example, Rodriguez et al. (1990) identified movement phases and muscle activation patterns during rowing ergometry. They identified six phases of the stroke and relative EMG activity levels of each muscle group. The authors concluded that strength of an individual muscle was not as important as the combined activity of two or more muscles. However, that study provided limited detail of EMG methods, limited stroke analysis, and no quantified EMG data. Gauthier (1985) found that timing and co-ordination of muscle actions could be improved during the rowing movement by providing feedback of the EMG signal. Wilson et al. (1988) investigated lower limb functions during the rowing movement performed on a Gjessing ergometer. These workers concluded that biceps femoris, gluteus maximus, gastrocnemius, and vastus lateralis were activated just prior to the catch and reached maximal excitation after peak force was attained. In addition, they reported that some muscles were isometrically coactivated and acted as a bi-articular link.

A study by Daireax Pottier, 1983 (as cited in Wilson et al., 1988) examined differences in muscle activation levels between skilled and novice rowers. During the recovery phase the author observed greater variance in EMG and elevated vastus lateralis activity in novice rowers compared to skilled rowers. This observation was attributed to increased eccentric contraction of the knee extensors in the novice rowers required to reduce the slide movement during recovery. Marr and Stafford (1982) reported that the EMG of experienced rowers compared to novice rowers revealed more activity, longer duration, and greater intensity of muscle contraction during the drive phase.

Recently, the data acquisition system and instrumented ergometer developed by Smith, Spinks, and Moncrieff in 1992 was modified to enable EMG signals from several muscle groups and biomechanical data from several transducers to be recorded synchronously. Kyrolainen and Smith (1999) used the modified system to investigate stroke rate effect on muscle activity patterns and mechanical variables during a 4-minute maximal effort. These workers found that knee extensor power and EMG activity increased, and that onset of EMG activity of the knee extensors became progressively earlier (nearer to the minimum handle position). The authors concluded that control of rowing technique decreased and, or the pattern of motor unit recruitment changed throughout the effort. Recent advances in data acquisition technology enable investigators not only to identify the relative activation of muscles, but in addition, to quantify muscle activation changes during rowing performance.

2.4 FACTORS AFFECTING ROWING PERFORMANCE

2.4.1 Biomechanical

Many factors influence the biomechanics of rowing performance. These factors include equipment (boat, oars, rigging), sweep (skills, experience), environment (wind, water depth, currents, waves), tactics (competition type, competitors draw) and most importantly, rowers stroke characteristics that are dependent on force, displacement, and velocity parameters (Dal Monte & Komor, 1989). Stroke characteristics, which include force at the handle and handle displacement, have been analyzed during ergometer and on-water rowing. Zatsiorsky and Yakunin (1991) suggested that there were six influential forces in rowing: (a) the pulling force at the handle, (b) the reactive force of the oar lock, (c) the reactive force of the water, (d) the force of air resistance, (e) the force of gravity, and (f) the force of hydrostatic pressure. Most biomechanical investigations of rowing ergometer forces have focussed on the rower's pulling force at the handle.

Secher (1993) reported that it took 0.3 - 0.4 s to reach maximal pulling force during the stroke which was independent of stroke frequency. Zatsiorsky and Yakunin (1991) suggested that peak force should be applied as quickly as possible after the catch and maintained at 90 % of maximum intensity throughout the drive phase. For optimum rowing performance, it is necessary for the rower to maximize the forces generated and effectiveness with which these forces are applied. For on-water rowing, optimal peak forces are attained while the blade is perpendicular and square to the boat (Smith & Spinks, 1995). For ergometer rowing, the equivalent handle position can be calculated from force and displacement data.

Effectiveness of the applied force and peak force displacement during ergometer rowing may be examined by viewing the propulsive force value plotted against linear displacement each stroke. The linear propulsive force applied to the ergometer handle in the direction of movement is considered equivalent to the major driving force on-water exerted by the rower in the direction of the boat. The instantaneous propulsive force value plotted against time during a rowing stroke provides a graphical force-time profile that allows comparison of timing, efficiency, and consistency of force output. In addition, the measurement and comparison of individual skill capabilities that include co-ordination of segmental forces, may be assessed using force-time curves (Smith et al., 1992). The optimal force-time curve demonstrates an initial rapid rise in force with peak force maintained as long as possible before a rapid decline. Fatigue affects the ability to maintain consistent and optimal force-time curves. For example, force-time curves may demonstrate force oscillations early in performance and progressively lower values later in performance due to fatigue. Consequently, changes in both intra- and inter-rower force-time curves are demonstrated during performance (Wing & Woodburn, 1995).

Skill and co-ordination may enhance the efficient application of forces by optimizing neural activation patterns (Sale, 1988). While the legs produce the initial major power output of each stroke, efficient power output is partly dependent on the co-ordinated activity of leg, back, and arm muscles (Nelson & Widule, 1983; Smith & Spinks, 1995). The co-ordination of agonistic, synergistic, and antagonistic muscle groups may be monitored using EMG (Arendt-Nielsen & Sinkjaer, 1991; Marr & Stafford, 1982; Rodriguez et al., 1990) and the resultant biomechanical output illustrated using force-angle, force-time, or power-time curves (Henry et al., 1995; Nelson & Widule, 1983;

Smith & Spinks, 1995). Rodriguez et al. (1990) examined co-ordinated muscle activity in upper and lower limbs using EMG and video analysis during ergometer rowing. They concluded that timing and sequencing of muscle activation during different phases of the stroke were more likely to influence performance than strength of individual muscles. Henry et al. (1995) found that rowers, who applied a steady force throughout the drive phase using the correct summation of co-ordinated forces, generated higher average power. These workers concluded that alterations in muscle group co-ordination corresponded to changes in slope and smoothness of the power-time curve.

Marr and Stafford (1982) examined technique differences that affect performance in junior and novice rowers. They used EMG to monitor timing and intensity of muscle activity, and videography to monitor body segment movements. Video analysis subjectively demonstrated an ineffective summation of joint forces in novice rowers compared to junior rowers, although details of force outputs were not provided. Temporal EMG patterns revealed longer, more intense patterns of quadriceps EMG activity during the drive phase and greater hamstring activation during the recovery phase, in junior compared to novice rowers. However, EMG activity was not quantified and the author's conclusions were based on visual representation of the size of the EMG, which could be erroneous due to methodological factors (e.g. electrode configuration, impedance). Smith and Spinks (1995) compared rowers of different standards during a 6-minute maximal ergometer test. Of the biomechanical variables investigated, stroke smoothness, which illustrated the smooth transfer of applied forces from each of the major muscle groups, best discriminated the rower's ability. The authors assumed that skilled rowers produced a force-time profile in the shape of the positive half of a sinewave. The closer that force-angle and force-time curves

consistently resembled the ideal biomechanical model, the better the performance outcome. Smith and Spinks suggested that rowers might maximize propulsive effort by improving technique, skill, and co-ordination, which might be assessed by stroke smoothness.

In summary, biomechanical factors that influence performance may be graphically illustrated using the force-time and force-angle curves attained during simulated rowing. They may be used to quantify force outputs and evaluate optimal stroke characteristics that affect performance outcome, and to assess rowers of differing abilities. In addition, timing and co-ordination of muscle group activity may be assessed using force curves together with EMG. Biomechanical characteristics of force output and associated muscle activation are central factors in rowing performance.

2.4.2 Physiological

Successful rowers show similar anthropometric characteristics (Morton, Lawrence, Blanksby, & Bloomfield, 1984) although anthropometric variables are poor predictors of a 2000-meter rowing race (Hahn, Ryan, Lawton, Kearns, & Bellenger, 1996). Hahn et al. (1996) demonstrated that maximum oxygen uptake ($\dot{V}O_{2\max}$) and ergometer performance outcome were the best predictors of on-water performance. Trained rowers tend to have large aerobic capacities and high anaerobic thresholds (85 - 95 % $\dot{V}O_{2\max}$) attributed to a high mitochondrial capillary density and enzymatic changes that help delay the accumulation of lactic acid (Steinacker, 1993). Most authors agree that during maximal 6-minute rowing efforts energy is provided by approximately 65 % aerobic and 35 % anaerobic metabolism (Hagerman, 1986; Secher, 1993) that involves

approximately 70 % of the total body muscle mass (Steinacker, 1993). Elite oarsmen have demonstrated significant anaerobic contribution by post-exercise blood lactate concentrations ranging between 14 - 18 mmol/L (Hagerman, 1986). The accumulation of blood lactate is associated with a loss of maximal voluntary force and is considered to have a negative affect on performance outcome (Hahn, Ryan, Bellenger, Lawton, & Kearns, 1995a)

Physiological and biochemical processes that affect rowing performance are partly influenced by the level of maximal force produced. During maximal efforts, the rower attempts to maintain very high force and power outputs (80 – 100 % MVC) for the duration of the task. During intense rowing efforts lasting approximately 6-minutes, decreases in maximal force and power outputs (Staniak et al., 1994) are associated with biochemical changes in the muscle (Bruton, Lannergren, & Westerblad, 1998; Fitts, 1996). The rate of fatigue, defined as the rate of reduction in MVC force (Bigland-Ritchie & Woods, 1984), is critical to performance outcome. For example, fatigue has been associated with inappropriate biomechanical alterations during maximal rowing. These include inefficient timing of muscle forces, reduced muscle group co-ordination, altered inter-rower synchronization of applied forces (Henry et al., 1995; Wing & Woodburn, 1995), and changes in optimal stroke characteristics (Smith & Spinks, 1995; Staniak et al., 1994). The mechanisms and biochemical processes of fatigue are discussed in the following section.

2.5 PHYSIOLOGY OF MUSCLE FATIGUE IN RESPECT TO ROWING

Fatigue is influenced by many interdependent factors including oxygen transport (Dempsey & Babcock, 1995), energy depletion (Vollestad, 1995), and biochemical change in the muscle (Westerblad, Lee, Lannergren, & Allen, 1991). No single process is responsible for muscle fatigue under all conditions but the principle processes are dependent on type and duration of the exercise (Bigland-Ritchie, Rice, Garland, Walsh, 1995; Enoka & Stuart, 1992). Mechanisms of force generation, muscle fatigue, and energy metabolism during rowing performance are described in this section. For detailed reviews of the cellular aspects of force generation and muscle fatigue, refer to Fitts (1996), Lee (1994), and Westerblad et al. (1991). A brief summary is presented here.

2.5.1 Mechanism of Force Generation

For graphical representation of the following text, see "Chain of neural command" in Figure 1. The initiation of a muscle contraction begins in the motor cortex of the brain where nerve impulses form to trigger movement. Central commands from the motor cortex lead to activation of motor neurons and conduction of nerve impulses (action potentials). Action potentials from the motor cortex propagate along nerve fibers into the spinal cord making synapses with lower motor neurons. Nerve fibers connect the lower motor neurons with synapses at the neuromuscular junction of the muscle where electrical and biochemical interactions occur which is known as excitation-contraction coupling. At the neuromuscular junction, action potentials initiate the release of acetylcholine from the nerve endings causing depolarization at the end-plates. Depolarization at the end-plates by voltage-gated sodium and potassium channels

initiates an action potential in the muscle cell. The action potential then propagates along the surface membrane of the muscle cell and into the transverse tubular system (called T-tubules) within the muscle cells. A sample of action potentials from the surface membrane of muscle cells and from the T-tubules can be measured on the skin surface using EMG. When the electrical impulse travels along the T-tubule, it alters the shape of a protein embedded in the sarcoplasmic reticulum, subsequently releasing calcium into the muscle cell. The main effect of calcium releasing into the muscle cell is to change the character of troponin, which allows muscle proteins (actin and myosin filaments) to interact.

Myosin filaments have cross-bridges that stretch out and interact with actin filaments when calcium and troponin combine. The cross-bridge filaments slide past each other before releasing and then repeating the sequence, thus the muscle contracts. Adenosine triphosphate (ATP) is the chemical fuel generated by burning of oxygen in the mitochondria and provides energy for cross-bridge cycling and muscle contraction. An ATP-dependent pump continuously transports calcium back into the sarcoplasmic reticulum. When activation stops and calcium is removed from troponin, the cross-bridge cycling ceases and the muscle relaxes.

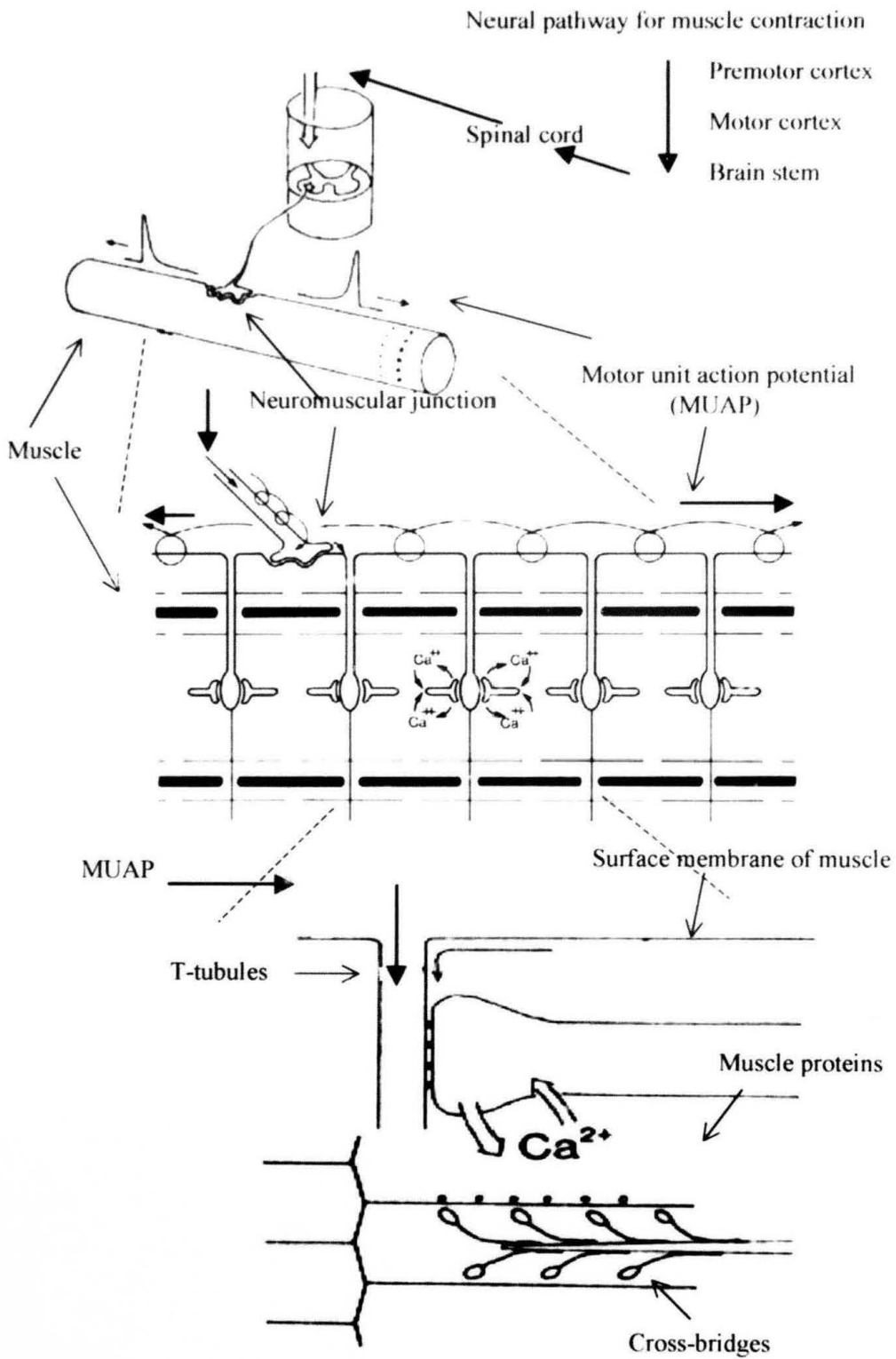


Figure 1. Chain of neural command.
(Loeb & Gans, 1986; Westerblad et al., 1991)

2.5.2 Muscle Fatigue

Muscle fatigue can be described as a failure to maintain force in a prolonged contraction or to re-attain force in repeated contractions (Lee, 1994), or as in this thesis, as a reduction in MVC force (Bigland-Ritchie, Furbush, & Woods, 1986). Fatigue is associated with a slowing of contraction and relaxation time of muscle and may be attributed to central or peripheral mechanisms. Central mechanisms are related to neural input to the higher brain center, central command, and the recruitment and activity of alpha motor neurones. Reduced central drive has been associated with loss of voluntary muscle force and related to decline in electrical activity of a muscle (Jones & Bigland-Ritchie, 1986). However, EMG changes associated with central factors may also be attributed to lack of motivation, to central regulation mechanisms that prevent contractile failure (Bigland-Ritchie & Woods, 1984), and to reduced thresholds for excitatory and inhibitory processes in the motor cortex (Gandevia, 1998). Peripheral fatigue is attributed to sites distal to the neuromuscular junction. The peripheral process includes excitation-contraction coupling, which involves activation of the surface membrane, propagation down the T-tubules, release of calcium, and activation of contractile elements and cross-bridge cycling to produce force (Fitts, 1996). Calcium release may be affected by ionic changes in the T-tubule system associated with the intra- and extra-cellular balance of sodium and potassium. The ionic balance affects efficient conduction of electrical impulses that stimulate the sarcoplasmic reticulum. During prolonged contractions, the amount of calcium released by the sarcoplasmic reticulum may begin to fall or the myofilaments may become progressively less responsive to calcium. During repeated contractions, release of calcium from the sarcoplasmic reticulum is reduced (Westerblad et al., 1991). Fitts (1996) suggests that metabolic processes associated with the breakdown of ATP and creatine phosphate,

produce adenosine diphosphate, phosphate ions, and creatine products that are involved in peripheral fatigue. The metabolic changes may also affect the calcium pump that returns calcium back into the sarcoplasmic reticulum. In addition, metabolic products (phosphate, lactic acid, hydrogen ions) that are liberated during transition from low to high force states, reduce the response of myofilaments to calcium. Inorganic phosphate causes (a) blocking of cross-bridge cycling by reversing the cross-bridge transition from the low to high force state, (b) decreases in the free energy of ATP hydrolysis, and (c) inhibition of calcium re-uptake into the sarcoplasmic reticulum (Westerblad et al., 1991). Lactic acid and hydrogen ions are involved with (a) reducing the rate of force development and the binding of troponin to calcium, (b) inhibiting the cross-bridge transition from the low to high force state, (c) inhibiting the cross-bridge cycle rate or velocity, and (d) prolonging the rate of calcium re-uptake (Lee, 1994).

2.5.3 Energy Metabolism in Rowing

Biomechanical factors of the task (speed, intensity, and duration of contraction) may influence the balance of energy systems employed and the associated metabolic changes in muscle, thereby affecting the development of fatigue. Furthermore, Enoka and Stuart (1992) suggested that neural strategy (pattern of muscle activation and motor command) might alter during some tasks in order to minimize fatigue. Hence, energy pathways employed in tasks such as rowing are influenced by variables of the task, which affect development of fatigue and may alter the neural strategy during repeated contractions.

A brief summary of energy metabolism as described by Powers & Howley (1994, pp. 51-68) is presented. The energy for muscle contraction is supplied by the chemical break down of ATP. The production of ATP by anaerobic metabolism involves both

the phosphagen and the glycolytic system. The phosphagen system provides the most rapid supply of ATP and is the major source of energy for high intensity, short duration exercise lasting approximately ten seconds. The glycolytic system (lactic system) provides the major supply of ATP for up to three minutes during intense activity. The lactic system partially breaks down carbohydrates and in the process produces metabolic by-products (including inorganic phosphate and lactic acid) which disrupt cross-bridge cycling and restrict the ability of muscle to produce force (refer to section 2.5.2). The accumulation of lactic acid and associated free hydrogen ions can have a negative effect on force output by inhibiting enzymes (e.g. phosphofructokinase) that slow ATP energy production. Neither of the anaerobic pathways is dependent on oxygen for their activity. For longer duration exercise, energy is produced by aerobic glycolysis with complete breakdown of carbohydrates or by beta-oxidation of fats. Both aerobic pathways are dependent on oxygen. Energy from aerobic metabolism may be influenced by substrate depletion in longer events. The contribution of each energy system may be viewed as a continuum from anaerobic to aerobic supply dependent on the duration and intensity of the exercise.

Optimal rowing efforts require a balance of ATP generation between the slower more enduring aerobic pathways and the more rapid anaerobic pathways (associated with metabolite accumulation and muscle fatigue). Each rower or crew seek the optimal blend of energy contribution that maximizes power output and minimizes the metabolic changes that affect muscle function (described previously in section 2.5.2). During performance situations it is not always practical to use muscle biopsy techniques in order to observe metabolic changes that occur within muscle. However, blood lactate concentration, although not necessarily representative of muscle lactate or intracellular

pH, may provide an indication of a metabolite change within the muscle that is associated with fatigue. In rowing situations the effects of fatigue are commonly monitored by the measurement of lactic acid concentration in the blood (Hagerman et al., 1978; Hahn et al., 1995b).

During maximal rowing efforts, physiological changes that indicate anaerobic involvement have been reported previously. For example, substantial increases in blood lactate (Hagerman, 1996; Hahn et al., 1995b; Steinacker et al., 1993), decreases in arterial acidity, and lower partial pressure of expired carbon dioxide (McLellan & Cheung, 1992) have been observed. Energy contributions during ergometer and on-water rowing performances may not be the same. Hahn et al. (1995b) found no significant correlation for post-exercise blood lactate or blood pH between 2000-meter ergometer and on-water performances. They suggested that the rowing task and specific rowing technique required for each situation could influence the energy pathway balance (anaerobic-aerobic). Interestingly, Hahn et al. (1995a) postulated that lactate metabolites might not be the major limiting factor to performance in highly trained athletes. These workers observed that sodium bicarbonate buffering did not enhance performance in highly trained rowers when compared to lower grade rowers, which indicated that performance was not limited by blood lactate accumulation in the former group. Other reasons, such as diffusion of buffering substances or lactate transport kinetics, that might explain why sodium bicarbonate buffering did not enhance performance in the highly trained subjects were not discussed in that study. For example, buffering substances such as sodium bicarbonate diffuse poorly into muscle fibers and varies with fiber type, which may limit intracellular pH buffering (Allen, Westerblad, & Lannergren, 1995). Thus, in the Hahn et al. (1995a) study rowers may

not have received significant buffering at the cellular level, which might explain the lack of performance improvement in the supplementation trial.

In conclusion, the balance of energy production from aerobic-anaerobic systems may affect rate and magnitude of metabolic change, which determines the development of fatigue. In performance situations, fatigue is often monitored by the measurement of blood lactate, and energy contribution to work from each metabolic system is influenced by variables of the performance task. In the case of rowing performance, the task variables may be changed by manipulation of contraction intensity, contraction velocity, and stroke rate known collectively as the pacing strategy.

2.6 PACING STRATEGY

The pacing strategy employed by the majority of rowing crews starts with a high stroke-rate at maximal power intensity, followed by a reduced power output and steady pace, before finishing with a maximal effort over the last minute (Dal Monte & Komor, 1989). Hagerman (1986) suggested that this pacing strategy was metabolically inefficient and produced an undesirable quantity of lactic acid. Hagerman et al. (1978) examined metabolic functions for 310 competitive rowers during a 6-minute maximal ergometer test employing the typical pacing strategy. The authors reported that rowers initially relied on anaerobic energy from the phosphagen and glycolytic energy systems with associated lactate build-up. Oxygen debt and lactate responses demonstrated the dominance of anaerobic metabolism during early stages of the effort. Aerobic and anaerobic metabolism were calculated to contribute 70 and 30 % respectively, to the total energy expenditure. For each subject the peak power output, the majority of

oxygen debt, and 90 % of blood lactate accumulation occurred during the first minute of exercise. Blood lactate concentration peaked at the second minute and remained elevated for the duration of the effort. It is evident that the typical pacing strategy causes a substantial accumulation in metabolic by-products that may have a negative affect on performance. The stroke rate and intensity at the start of a typical effort increases the metabolic cost of exercise compared to the lower steady-state stroke rate and reduced intensity performed for the majority of the race (Hagerman et al., 1978). It is likely that increased oxygen utilization and decreased anaerobic contribution at the start of an effort may delay the deleterious effects of increased lactic acid and other metabolites. On the other hand, the initial intense effort and high workload involved in the typical pacing strategy might serve to increase the rate of oxygen uptake earlier in performance compared to a less intense initial effort. Thus, earlier increased oxygen availability may allow a greater amount of total work to be completed in 6-minutes compared to that during a constant workload when the rate of oxygen increase may be delayed (Secher, 1993).

Zatsiorsky and Yakunin (1991) stated that the typical on-water pacing strategy caused variation in boat velocity, and therefore was biomechanically inefficient due to reduced drag efficiency. Maintaining a constant velocity reduces drag between the boat and the water, thereby reducing inertia of the rower-boat-oar system and increasing work efficiency. Hence, a constant velocity will require less energy expenditure than a fluctuating velocity because of the dependence of drag on velocity squared. This applies to intra-stroke and strategy type variations in speed. Dal Monte and Komor (1989) observed that constant boat velocities were associated with efficient rowing techniques. They reported significant changes in optimal efficiency of rowing

technique (demonstrated by changes in the force-time and force-angle curves) from the initial to the final strokes when rowers were unable to maintain power output due to fatigue. Consistent rowing technique and constant boat velocity may be influenced by the pacing strategy adopted.

Various workers have reported beneficial effects of controlled pacing during automated repeated movement events such as cycling. Cherry, Lakomy, Neville, and Fletcher (1997) found that work output achieved in an all-out effort on a cycle ergometer could be replicated in a constant-pace effort at cadences between 60 – 90 rpm. The constant-pace effort resulted in less reduction in peak power output and less disturbed post-exercise blood lactate. They explained these findings by the partial sparing of Type II muscle fibers during the constant-pace effort. Foster, Snyder, Thompson, Green, Foley, and Schrage (1993) examined various pacing strategies during 2-km time trials using a cycle ergometer. In one trial, subjects performed the first 50 % of the effort at a constant-pace (equivalent to average pace for a self-pace 2-km time trial) and the second 50 % of the effort at maximum pace. This pacing strategy produced the fastest time compared to the other strategies examined, although no systematic differences were found in serial oxygen uptake, oxygen debt, or post-exercise blood lactate. To the contrary, Burton (1996) found no benefits in performance parameters or physiological markers in controlled-pace compared to self-pace efforts. In Burton's study, a group of club-level rowers performed three 2000-meter rowing ergometer trials. Trial 1 was at a self-determined pace and trial 2 was at a constant-pace equal to the average-pace in trial 1. In trial 3 subjects attempted to complete a 2000-meter distance in a faster time than in trial 1, with a constant-pace (as in trial 2) for the first 1500-meters and a self-pace for the last 500-meters. Burton reported that final blood lactate concentration and mean

heart rate responses were not significantly different between self-pace or constant-pace efforts. In addition, all but one of the subjects reported that the constant-pace trial felt much harder than the self-pace trial, and only three out of nine subjects improved the performance time in trial 3.

It appears that constant-pace strategies may provide performance benefits for some athletes, but for other athletes self-pace strategies may optimize physiological and biomechanical mechanisms and maximize performance.

2.7 BASIS OF EMG

Basmajian and De Luca (1985), De Luca (1997), and Loeb and Gans (1986) have contributed extensively to our understanding of EMG theory and application. A brief summary is presented that describes muscle activation and the basis of EMG. The reader is referred to the aforementioned authors for more detail.

As muscles are activated, minute voltage gradients are created in the surrounding medium of the tissues. The voltage gradients may be recorded as a myoelectric signal. Surface electromyography is the measurement of the myoelectric signal activity sampled above the muscle and provides a means of monitoring associated physiological processes that initiate muscle contraction.

At the neuromuscular junction action potentials initiate the release of acetylcholine causing depolarization which affects sodium and potassium channels. Subsequently, changes in voltage-gated sodium and potassium channels initiate action potentials from

each of the muscle fibers, which then propagate along the sarcolemma. The sum of these single fiber action potentials from all the fibers in a given motor unit gives rise to the motor unit action potential (MUAP). MUAPs then propagate into the T-tubule system within the myofiber. Skin, adipose layer, and other tissues reduce the voltage amplitude of MUAPs recorded on the skin surface by absorbing some of the electrical energy. The electromyogram recorded at the skin surface is a sample of the action potentials propagating along the muscle cell membrane.

The basic unit of the nervous system is the motor unit (MU), which consists of the lower motor neuron, the lower motor neuron axon, and the muscle fibers that are activated. The fibers in a given motor unit are of similar size, have similar contractile properties, and fire simultaneously when stimulated to do so by the motor neuron. Fibers in different motor units will fire asynchronously so that fibers in some motor units are contracting while others are relaxing and this produces a smoother contraction. Neural activation may be described as a combination of the recruitment of MUs and the firing rates of those MUs.

Generally during a muscle contraction, recruitment of MUs is based upon the size principle (Henneman, Somjen, & Carpenter, 1965). According to De Luca, LeFever, McCue, and Xenakis (1982) smaller MUs are recruited first and subsequently, as neural drive increases up to 40 % MVC larger MUs are recruited. De Luca et al. (1982) suggested that muscles use various mechanisms to increase force output dependent on fiber composition and function of the muscle. Large muscles used for gross powerful contractions, such as the deltoid muscle, are composed of many MUs. In this type of muscle, MU recruitment is the major mechanism for generating force up to 80 % MVC.

Between 80–100 % MVC when all available MUs are recruited, firing rate increase becomes the major mechanism for generating extra force. Small muscles required for finely graded force contractions, such as the first dorsal interosseous muscle, are composed of fewer MUs. In this type of muscle, increase in MU firing rates is the major mechanism for generating more force.

EMG signal activity, which reflects MU recruitment and firing rates, may be processed and then expressed quantitatively as amplitude or as spectral power frequency. EMG signal processing involves a series of steps that include filtering, rectifying, and smoothing the raw EMG signal. There are several commonly used means of expressing EMG amplitude in the time domain, such as root mean square (rmsEMG), peak to peak, or the averaged rectified integrated value. In the current project, rmsEMG was used to quantify EMG amplitude as it represented the signal power with minimal distortion and had physical meaning. The rmsEMG was calculated by squaring the EMG data (after analogue to digital conversion), summing the squares, dividing this sum by the number of observations, and taking the square root of that value (refer to section 3.1.3 for signal processing methods). In addition, the energy from muscle activity has a frequency spectrum. In order to decompose the EMG signal into its various frequency components a mathematical technique called Fast Fourier Transformations (FFT) is performed on the EMG signal data. After FFT processing, characteristic values of the spectral power frequency, such as the median, mean, mode frequency, or the ratio of low to high frequency bandwidths may be calculated. In the current project, the mean power frequency (MPF) was used to represent the average frequency characteristic of the EMG signal as it provided a reliable and consistent parameter that was related to muscle fiber conduction velocity.

2.8 USE OF EMG IN PHYSIOLOGY

2.8.1 Force-EMG Relationship

Force-EMG amplitude

EMG amplitude expressed as a ratio of the MVC value has been used to indicate the level of muscle excitation (Bigland-Ritchie & Woods, 1984; Jones & Bigland-Ritchie, 1986). During isometric muscle contractions, force and EMG amplitude are strongly correlated (Basmajian, 1974; De Luca, 1997; Mannion & Dolan, 1996a). Esposito, Malgrati, Veicsteinas, and Orizio (1996) reported a significant correlation for force-EMG amplitude of the biceps brachii during isometric contractions (20 – 100 % MVC) in subjects differing in age and in gender. Similar force-EMG amplitude correspondences have been demonstrated in the erector spinae during isometric back extensions (Dolan & Adams, 1993) and in the quadriceps during isometric leg extensions (Hakkinen, 1994). At high force levels, variability of the force-rmsEMG relationship is increased. This has been attributed to the presence of agonistic co-contraction or to the possible development of fatigue (Yang & Winter, 1983). Furthermore, force-rmsEMG variability during near-maximal isometric contractions has been ascribed to inefficient central nervous control resulting in increased firing rates of active MUs while the force contribution of active MUs is maximal (De Luca, 1997).

Christensen et al. (1995) and Muro et al. (1983) both reported that the force-rmsEMG relationship of brachial biceps during static contractions and during slow isokinetic concentric contractions were not significantly different. However, Christensen et al.

noted that during isokinetic contractions the strength of the force-rmsEMG correspondence was dependent on applied torque and muscle length.

During dynamic contractions, variability in the force-rmsEMG relationship may be partly attributed to inefficient co-ordinated muscle contribution caused by inappropriate activation patterns of synergistic muscles (Arendt-Nielsen & Sinkjaer, 1991). Andersson, Nilsson, & Thorstensson (1997) observed that during human locomotion force-EMG amplitude relationships were different between individual muscles. The differences were attributed to difference in onset and relative duration of activation for each muscle. Andersson et al. concluded that in spite of regulated movement patterns during well-practiced cyclic contractions, EMG activity changed, particularly when contraction velocity or force was modulated.

During dynamic contractions, confounding factors affect conduction velocity, motor unit recruitment, and firing frequency mechanisms and therefore, alter the force-EMG amplitude relationship. The confounding factors include interactions between muscle length, contraction velocity, and synergistic muscle contribution that varies with the type of contraction performed (isometric, isokinetic, dynamic). During dynamic contractions, change in muscle length affects the area of sampled muscle activity, the spatial filtering of action potentials, and the muscle fiber diameter (and consequently the conduction velocity of action potentials). Such changes influence EMG characteristics and stability of the EMG signal (Doud & Walsh, 1995; Mannion & Dolan, 1996a). Stability of the EMG signal has been described in terms of "stationarity" of the signal and refers to stability of the electrodes with respect to active muscle fibers and stability of the motor unit activation pattern (De Luca, 1997). Both are modulated by the

confounding factors. The stationarity of the signal is also affected by contraction velocity. Contraction velocity influences the relative number of Type II (fast twitch) fibers, which have faster conduction velocities compared to Type I (slow twitch) fibers, that are engaged. In response to faster contractions relatively more Type II than Type I fibers are activated, which affects action potential conduction velocity and the EMG (Kupa, Roy, Kandarian, & De Luca, 1995). The effect of contraction velocity on the EMG is increased during non-constant velocity contractions when the time delay between the signal and the force influences the stationarity of the signal (Arendt-Nielsen & Mills, 1988; Potvin, 1997). During most muscle contractions, the measured force may not be exclusively attributed to any one muscle. Agonistic, antagonistic, and synergistic muscles may contribute to the measured force output depending upon the level of movement control. However, the recorded EMG reflects only the activity of muscle fibers sampled directly below, underlying, or adjacent to the electrode sites (Ebenbichler et al., 1998; Psek & Caffarelli, 1993).

Attempts have been made to minimize the effects of muscle length, contraction velocity, and synergistic muscle contribution on the EMG when examining dynamic contractions. For this purpose investigators have used intermittent isometric contractions (Dolan & Adams, 1998; Taylor et al., 1997) and restricted EMG data-windows (Bouissou, Estrade, Goubel, Guezennec, & Serrurier, 1989; Potvin & Bent, 1997) (discussed in section 2.9.2). Few studies have compared force-EMG relationships during isometric, isokinetic, and unrestricted performance contractions in which the confounding factors were under different levels of control.

Force-EMG frequency

Median frequency (MF) of the EMG has been shown to increase proportionally with increased force demand (Broman, Bilotto, & De Luca, 1985). The parallel increase in MF and force output is believed to be related to faster conduction velocities and higher firing rates of Type II muscle fibers compared to Type I muscle fibers recruited in response to increased force (Arendt-Nielsen & Mills, 1988). However, the force-frequency (EMG) relationship in non-fatigued muscle is not clear. De Luca et al. (1982) reported that the force-frequency relationship varied between muscles due to the considerably variation of MU recruitment and firing rate thresholds between muscles. Esposito et al. (1996) found that during isometric elbow flexion, MF of the biceps brachii was significantly related to the level of effort between 20 – 80 % MVC. However, the authors reported that MF leveled off and even decreased at force levels between 80 -100 % MVC, which indicated no further recruitment of large MUs or related increase in conduction velocity. In contrast, Petrofsky and Lind (1980) reported that the mean power frequency (MPF) was unrelated to power or intensity during fresh isometric muscle contractions. Furthermore, Mannion and Dolan (1996a) found that force output was significantly related to MF during isometric contractions of the back extensors, but the relationship varied according to muscle length. They suggested that an increase in muscle length would lead to a reduction in muscle fiber diameter, which may cause a reduction in conduction velocity and thereby contribute to a lower MF. In addition to the effects of fiber type, MU threshold level, and muscle length, methodological factors such as muscle temperature, skin impedance, and signal stability may also affect conduction velocity and EMG frequency characteristics (see section 2.9).

2.8.2 Fatigue and EMG

Fatigue may be defined as a reduction in maximal voluntary force (Bigland-Ritchie & Woods, 1984) and is accompanied by alterations in the EMG signal (Oda & Moritani, 1995; Tesch, Dudley, Duvoisin, Hather, & Harris, 1990). Evidence that voluntary contraction force and central nervous drive are maximal may be provided using twitch interpolation techniques (Bigland-Ritchie, Johansson, Lippold, & Woods, 1983). The twitch interpolation technique compares the force that can be elicited during superimposed electrical stimulation with the force that can be produced by voluntary contraction. The imposed electrical stimulus should produce the maximum twitch of a resting muscle and recruit all available motor units. If no additional force is produced by superimposed stimulation during a maximal contraction, then the contraction is accepted as a MVC. In addition, measurement of electrical response in the muscle (M wave) to imposed electrical stimulation has been used to indicate alterations in neuromuscular propagation (Bigland-Ritchie et al., 1986; Milner-Brown & Miller, 1986).

The loss of MVC force may be associated with failure of neural mechanisms, contractile processes, or both. Physiological changes may occur within the muscle before loss of force becomes apparent (Vestergaard-Poulsen, Thomsen, Sinkjaer, & Henriksen, 1995). For example, during a sustained submaximal isometric contraction, force output may be maintained while action potential conduction velocity and MF of the EMG decreases (De Luca, 1985; Erberstein & Beattie, 1985). Hence, the EMG may provide an indication of neural activity associated with biochemical changes that occur in individual muscle before loss of voluntary force can be detected.

In addition, the EMG may provide an indication of MU recruitment and firing rate patterns used in order to maintain optimal voluntary force output and avoid contractile failure of the muscle during fatiguing performance. Many studies have used EMG analysis to monitor fatigue during sustained isometric contractions (Bigland-Ritchie & Woods, 1984; De Luca, 1985). However, during fatiguing dynamic contractions EMG analysis is used less frequently due to problems associated with the confounding factors and the changing geometric relationship between electrodes and active muscle fibers.

Fatigue and EMG amplitude

Most investigations of muscle activation patterns during fatigue have used isometric contractions. During a sustained isometric MVC, IEMG declines in parallel with voluntary force from the onset of contraction. This phenomenon has been explained as a sign of central fatigue or a regulating mechanism to optimize force output (Basmajian, 1974; Bigland-Ritchie & Woods, 1984; Jones & Bigland-Ritchie, 1986). Moritani, Muro, and Nagata (1986) reported significant decreases in rmsEMG during sustained MVC of the biceps brachii muscle, but in contrast, increases in rmsEMG during submaximal contractions (50 % MVC) of the same muscle. During submaximal isometric contractions sustained at a constant load, increases in IEMG have been attributed to recruitment of additional MUs and increasing MU firing rates that may counteract contractile failure of slower MUs (Bigland-Ritchie & Woods, 1984). In addition, Bigland-Ritchie and Woods (1984) suggested that reduced conduction velocity and increasing synchronization of motor unit discharges might also increase the IEMG. Under the same conditions when the submaximal load is no longer maintained, voluntary force declines in relation to IEMG e.g. there is a decrease in the force/IEMG ratio. A decrease in the force/IEMG ratio has been attributed to peripheral factors

involved in contractile failure of the muscle (Bigland-Ritchie & Woods, 1984). If the submaximal contraction is continued, a parallel decrease in IEMG and voluntary force output occurs, indicating that additional MUs are not recruited or firing rates increased in order to prevent the decline in voluntary force (Taylor et al., 1997).

Hagberg (1981) compared fatiguing isokinetic and isometric submaximal contractions of the elbow flexors and found that the regression coefficients (slope and intercept) of EMG amplitude were not significantly different between the two contraction types. Furthermore, endurance time for maintaining force or torque levels in response to each type of contraction were similar. During repeated isokinetic MVCs of the leg-extensors, Tesch et al. (1983), and Komi and Tesch (1979) demonstrated that EMG amplitude changes were related to the percentage of fast twitch fibers within a muscle. During 150 isokinetic MVCs, Wretling and Henriksson-Larsen (1998) found significant decreases in rmsEMG (compared to pre-exercise MVC) for RF and for vastus medialis (but not for VL) similar to those found by Tesch et al. (1983). Wretling and Henriksson-Larsen reported that rmsEMG mostly increased for the first 10 contractions, then decreased with a level of variability for 60 – 70 contractions before stabilizing. In that study, it appears that the initial 10 MVCs were in fact submaximal when compared to pre-exercise MVCs. Subjects, who were untrained, may have lacked ability to perform consistent and maximal efforts, thereby adding to rmsEMG variability. In general during isokinetic MVCs, EMG amplitude declines with loss of voluntary force in a similar fashion to that found during isometric MVCs.

A number of studies have examined changes in EMG amplitude during repeated dynamic contractions when velocity was not controlled. Arendt-Nielsen and Sinkjaer

(1991) found significant increases in EMG amplitude for vastus lateralis, biceps femoris, and semitendinosus during submaximal endurance walking. They concluded that electrical activity (EMG amplitude) and timing of co-ordinated movement patterns of each muscle were affected by fatigue. Potvin (1997) reported significant increases in EMG amplitude in response to repeated biceps brachii contractions performed with a constant load at a self-determined velocity. The highest rmsEMG increases were observed at the higher contraction velocities. The findings of Potvin were in agreement with those of Gabriel and Boucher (1998) who reported increased EMG amplitude during similar contractions, however, these workers found that EMG amplitude was related to increased elbow flexion time. Kyrolainen and Smith (1999) observed that IEMG of the knee extensor muscles significantly increased during four minutes of near-maximal ergometer rowing. Most importantly, these authors found that knee extensor muscle activity was initiated progressively earlier during each stroke and that maximal knee extensor power increased over the duration of exercise. In Kyrolainen and Smith's study, data for average force each contraction during the performance task was not reported and therefore, the force-EMG amplitude relationship over time could not be assessed. In that investigation, voluntary knee extensor force may have declined but power output been maintained due to increases in knee extensor contraction velocity or to increasing contribution from other synergistic muscles.

In general, during fatiguing submaximal dynamic contractions when contraction velocity is not controlled, an increase in EMG amplitude occurs similar to that found during isometric and isokinetic contractions. However, when dynamic contractions are performed at near-maximal power with insufficient recovery between contractions, force output and EMG amplitude may eventually decline in parallel, as found during

repeated isokinetic MVCs, indicating the involvement of central regulation or central fatigue mechanisms. During uncontrolled dynamic contractions, subjects may increase contraction velocity and, or alter timing of muscle activity in order to maintain power, which may influence the force-EMG amplitude relationship.

Fatigue and EMG frequency

It has been suggested that during fatiguing contractions, change in frequency characteristics of the EMG may provide an appropriate representation of metabolic status within the muscle (De Luca, 1985; Mannion & Dolan, 1996b). Modification of EMG frequency characteristics reflect changing aspects of voluntary muscle activation, such as conduction velocity, motor unit recruitment, and motor unit firing rates (Erberstein & Beattie, 1985; Hagg, 1992).

During sustained submaximal isometric contractions, shifts of the EMG MPF or MF towards lower values are well documented. During submaximal isometric exercise causing fatigue, MF declines relative to decline in maximal force capacity (Mannion & Dolan, 1996b) and endurance time (Mannion, Dumas, Stevenson, & Cooper, 1998). Mannion and Dolan (1996b) suggested that rate of MF decline (MF gradient) was representative of decline in force generating capacity of the muscle during that type of contraction. They found that higher submaximal forces were related to increased MF gradient and greater decline in force generating capacity. In addition, correspondences between decline in MF and force were slightly better for RF than VL muscles and independent of submaximal force output of the contraction (20 – 60 % MVC). In a later study Mannion et al. (1998) examined the force-MF relationship of the erector spinae during standing submaximal isometric (60% MVC) back extensor contractions and

during prone isometric back extensions held to the limit of endurance. They observed that the MF gradient was more related to fiber size than to relative fiber type distribution of the muscle. Lower MF gradients were related to greater Type I compared to Type II muscle fiber area. Mannion et al. (1998) concluded that endurance (expressed by the MF gradient) or ability to maintain submaximal force of the erector spinae muscle was significantly related to the area occupied by Type I relative to Type II muscle fibers.

During fatiguing isokinetic contractions, the torque-frequency (MF, MPF) relationship is less pronounced than for isometric contractions. Van Dieen et al. (1996) compared isometric (pre- and post-exercise) and isokinetic EMG changes in response to fatiguing submaximal isokinetic contractions of the erector spinae performed at different velocities. Only a weak correlation was demonstrated between frequency change and force loss for both post-exercise isometric and isokinetic contractions. Furthermore, the authors reported considerable variation among subjects for the torque-MPF relationship and no effect of contraction velocity. Wretling and Henriksson-Larsen (1998) observed that during fatiguing maximal isokinetic contractions of the leg extensors, initial MPF increases were followed by MPF decreases. Decreases in MPF were greater for RF than VL and similar between males and females. These workers noted the importance of the selected reference contractions used for normalization, particularly in untrained subjects. The initial contractions used for normalization in that study were not MVCs and may have influenced the findings. Tesch et al. (1990) examined MF and maximal voluntary torque during bouts of both concentric and eccentric leg extensor MVCs. During each bout of concentric work, MPF and torque were significantly related and

declined from initial values. In contrast, MPF and torque did not significantly change from initial values in response to eccentric contractions.

Doud and Walsh (1995) examined the effect of muscle length on MPF during submaximal isokinetic contractions of the biceps brachii with a fixed load (50 % MVC) performed to task failure. The EMG was sampled for overlapping segments (1024 ms) corresponding to elbow angles of 150 – 50° of extension during a constant rate of shortening. Elbow angle was converted to biceps muscle length and expressed as a percentage of optimal muscle length. The authors reported that the magnitude of EMG frequency decline was greater for shorter muscle lengths. A strong inverse relationship was demonstrated between MPF decline and muscle length for the first 75 % of the task, which was diminished for the final 25 % of the task.

During fatiguing dynamic contractions when velocity is not controlled, change in muscle length and contraction velocity may both affect EMG frequency characteristics and influence the force-frequency relationship (De Luca, 1997). Arendt-Nielsen and Sinkjaer (1991) found inconsistent MPF decreases between muscles during high intensity repeated contractions performed on a treadmill. The hamstring muscles (biceps femoris, semitendinosus) showed significant MPF decline from start to finish of exercise, but for the quadriceps muscle (vastus lateralis) MPF decline was not significant. These workers suggested that the difference in timing and duration of EMG activity for each muscle was a possible cause of inconsistent MPF responses during fatigue. In addition, Arendt-Nielsen and Sinkjaer noted the opposite influences of intramuscular temperature and fatigue on conduction velocity and EMG frequency. During cycle ergometer contractions performed at varying exercise intensities, Gamet et

al. (1990) reported that change in MPF of the quadriceps was different between subjects and independent of exerted power. These workers concluded that MU recruitment had an inconsistent effect on the MPF during dynamic contractions. Potvin and Bent (1997) compared MPF responses between repeated dynamic elbow contractions with a fixed load and isometric MVCs (pre- and post-dynamic exercise). Dynamic EMG data-windows (250 ms segments) were recorded for each contraction. These workers found similar MPF decreases from pre-exercise values for isometric (25%) and dynamic (27%) contractions, and no interaction across time between contraction types. They concluded that MPF values recorded during dynamic movement could be used to quantify fatigue of the biceps brachii muscle. Bouissou et al. (1989) examined spectral frequencies related to muscle lactate during exhaustive dynamic exercise on a cycle ergometer. Exercise was maintained at a fixed power output (375 W) and pedaling rate (90 rpm) until exhaustion (when pedaling rate dropped by 5 %). They found that MPF decreased by 10.1 % from initial value until exhaustion and that decline in MPF was linearly related to muscle lactate concentration. However, no significant correlation was found between change in muscle lactate and change in MPF.

Spectral modification, mostly seen as a compression of the EMG signal to lower frequencies, is influenced by recruitment, firing rate, synchronization, and shape and velocity of MU action potentials (MUAP). Slowing of MUAP conduction velocity and decrease in the frequency content have been associated with intramuscular pH and accumulation of local metabolites such as hydrogen ions and inorganic phosphates (Horita & Ishiko, 1987; Miller et al., 1987; Tesch et al., 1983) (see section 2.5.2). Milner-Brown and Miller (1986) examined action potential conduction velocity during muscle fatigue from isometric MVCs and supramaximal nerve stimulation of the first

dorsal interosseous, adductor pollicis, and anterior tibialis. They concluded that excitability and action potential conduction velocity were reduced, with the magnitude of impairment dependent on the duration of exercise and level of fatigue as well as intrinsic properties of the muscle. In addition to slowing of conduction velocity during sustained MVCs, MU firing frequency and MU synchronization significantly affect the EMG frequency spectrum (Bigland-Ritchie, Donovan, & Roussos, 1981).

2.8.3 Timing of EMG in Relation to Muscle Force

EMG may be used in physiology to measure the onset of muscle activation, the timing sequence of co-ordinated muscle activation, and the duration of muscle activation during dynamic and static contractions. The onset of electrical activity of a muscle may be identified in the EMG by a rectified value above a threshold that excludes any signal noise or artifact not initiated by the muscle contraction. The duration of electrical activity during a muscle contraction may be quantified by the time that the rectified value remains above the noise threshold and cessation of activity identified by the instant the rectified value drops below the noise threshold. A timing factor that may affect reliability of EMG characteristic values is electromechanical delay (EMD), which is the time delay between EMG activity and contractile force output of the muscle (Sadoyama, Masuda, & Miyano, 1983). EMD depends on several factors including electrode placement, fiber-type, firing-rate dynamics, and viscoelastic properties of the muscle.

The effect of muscle activation timing has been reported during non-fatiguing and fatiguing dynamic conditions. Andersson et al. (1997) reported differences in relative duration of EMG and onset of activation between individual hip-flexor muscles during

non-fatiguing treadmill running. Similar differences were observed by Marr and Stafford (1982) and Rodriguez et al. (1990) between muscle groups employed during non-fatiguing contractions on a rowing ergometer. Timing changes related to fatigue affect force output curves and muscle activation patterns. Dal Monte and Komor (1989) reported variations in timing and area under the force curve during rowing that were attributed to fatigue. Similarly, Arendt-Nielsen and Sinkjaer (1991) found that during uphill walking, onset timing and duration of muscle activation were disturbed. These workers evaluated the period from onset of the EMG burst to heel contact as well as the duration of the EMG burst. Although the overall movement pattern remained stable, onset time and burst duration of the EMG altered in most muscles. In agreement, Kyrolainen and Smith (1999) observed that during a four-minute maximal effort on a rowing ergometer, the timing of knee extensor muscle activity became progressively earlier, possibly reflecting reduced control of rowing technique.

During well-practiced automated cyclic movements, such as rowing and cycling, it appears that timing (onset, duration, cessation, EMD) of muscle activity remains relatively stable, providing that the task is non-fatiguing and power output, contraction velocity, and duty cycle remain stable. In contrast, when the same activity is sufficiently intense to cause fatigue, then timing of muscle activity and movement technique may alter force and EMG characteristics.

2.9 METHODOLOGICAL ASPECTS OF EMG

Methodological factors that affect the EMG signal include electrode configuration, signal stationarity, muscle or skin temperature, and impedance (Basmajian & De Luca, 1985; De Luca, 1997; Enoka & Stuart, 1992).

2.9.1 Electrode Configuration

During EMG sampling using bipolar differential configurations, recording configuration and inter-electrode spacing may cause a substantial change in magnitude of the recorded MUAPs and effect volume conduction (Roeleveld, Stegeman, Vingerhoets, & Van Oosterom, 1997). The location and orientation of electrodes may affect the MUs detected and the EMG frequency bandwidth. Thus, electrode configuration may influence the measured action potential conduction velocity that affects amplitude and frequency characteristics of the signal (Basmajian & De Luca, 1985).

Mannion and Dolan (1996a) found that increased inter-electrode spacing (25 – 35 mm) significantly increased rmsEMG recorded from the erector spinae but had no effect on MF values. Potvin and Bent (1997) noted that during dynamic contractions inter-electrode spacing increased with the change of muscle length and influenced the degree of low-pass filtering. These workers limited this effect by restricting EMG data-windows to a 20° range of movement. Furthermore, they assumed that changes in electrode spacing from the EMG data-window were consistent between repeated contractions.

2.9.2 Stationarity

To date most EMG studies have used static contractions in order to control stability or stationarity of the signal, thereby limiting change in the MUs sampled. Dynamic contractions that involve changes in muscle length and contraction velocity affect stationarity of the EMG signal and alter filtering characteristics of the surrounding tissues (Potvin, 1997). During dynamic contractions, the force-length relationship of muscle varies non-linearly, affecting the MUs relative position between the electrode and contracting muscle fiber. De Luca (1997) suggested that under dynamic conditions, it was advisable to limit the EMG sample to small data-windows and synchronize the data-windows to the same time during each contraction. However, one disadvantage of quantifying the EMG over a restricted data-window is that the resolution of the Fourier analysis will fall with a decrease in the duration of the data-window. Investigations that quantify the EMG from restricted data-windows recorded during dynamic contractions are not extensive.

Potvin and Bent (1997) used the restricted data-window approach to compare isometric and dynamic contractions of the biceps brachii for frequency (MPF) and amplitude (average EMG) changes during fatigue. They restricted the EMG data-window to 250-ms samples during dynamic repetitive elbow flexion-extension movements with a fixed load. These workers found a significant increase in amplitude and decrease in MPF for isometric and for dynamic contractions, but no significant differences between contraction types. The authors concluded that restricted EMG data-windows could be used to monitor fatigue during dynamic movement. Bouissou et al. (1989) restricted EMG samples to 500-ms data-windows during cycle ergometry. They reported

decreases in MPF of the vastus lateralis that were similar to those found by Hakkinen (1994) for the same muscle during fatiguing isometric contractions.

Another approach to minimize the stationarity problems associated with dynamic contractions has been to use pre- and post-exercise isometric contractions to monitor force-EMG changes during fatigue. For example, Taylor et al. (1997) used pre- and post-exercise isometric MVCs on an isokinetic dynamometer to examine the effect of a 10-minute cycle ergometer effort performed at a constant load. In addition, they measured IEMG continuously. Taylor et al. observed that post-exercise isometric IEMG and MVC force were similar to pre-exercise values and therefore, no loss of isometric MVC force (fatigue) was shown. However, in order to maintain submaximal force during the trial, electrical activity of the muscle (IEMG) increased over time. A similar approach was used by Dolan and Adams (1998) who used submaximal isometric contractions (60 % pre-exercise MVC) to examine changes in MF of erector spinae muscles in response to a repetitive dynamic lifting task. The authors reported a significant decrease in isometric MF (5.5 % of pre-exercise value) following the dynamic exercise.

Although pre- and post-exercise isometric contractions have been used to minimize EMG recording problems during dynamic contractions, isometric contractions do not necessarily reflect dynamic forces or muscle activation patterns. Murphy and Wilson (1996) demonstrated that isometric measurement of force was a poor predictor of dynamic performance because of the different MU activation patterns between isometric and dynamic contractions. Therefore, it is preferable to examine the specific movement pattern under investigation. The restricted data-window approach has

limitations associated with signal stationarity, but for applied situations is more task-specific than the pre- and post-isometric contraction approach.

2.9.3 Muscle and Skin Temperature

During submaximal contractions of short duration, muscle temperature is unlikely to have any significant affect on the EMG signal (Bigland-Ritchie et al., 1981). However, in more intense or longer duration efforts, increases in muscle temperature may increase action potential conduction velocity (Masuda, Sadoyama, & Shirai, 1996) and thereby affect frequency characteristics of the EMG (Petrofsky, 1979). Arendt-Nielsen and Sinkjaer (1991) reported that intra-muscular temperature increased during treadmill running. The authors suggested that increased intra-muscular temperature affected MPF values and countered the effect of fatigue on frequency characteristics of the EMG. Skin temperature is often monitored during exercise, although changes in skin temperature are not necessarily reflective of changes in intra-muscular temperature. Jansen, Ament, Verkerke, & Hof (1997) observed an elevation of skin temperature (0 to 4° C) during incremental cycle ergometry and suggested the elevation was partially responsible for increased MF. Winkel and Jorgensen (1991) found that cooling of superficial tissues (32 to 21° C) during 10-minute bouts of concentric contractions of the soleus muscle caused a near doubling in EMG amplitude and reduced MPF (142 Hz to 83 Hz). The author's explanations for these findings were (a) reduction in muscle temperature that increased the duration of each action potential, (b) slowing of conduction velocity, (c) reduced force contribution from each active muscle fiber, (d) recruitment of more motor units at low temperatures, (e) simultaneous activity in the antagonist muscle, and (f) increased viscosity of muscle tissue. Shivering and central rectal temperature were not considered to affect the EMG in that study. The relevance

of Winkel and Jorgensen's (1991) study is that amplitude and frequency of the EMG may be altered when environmental or other factors cause a considerable change in skin ($> 11^{\circ}\text{C}$) or muscle temperature.

2.9.4 Impedance

Tissue resistance between active muscle fibers and electrodes may alter signal impedance and affect EMG amplitude. Impedance is related to thickness of the tissue between surface electrodes and muscle, and has an inverse relationship with EMG amplitude. Andersson et al. (1997) reported that up to 81% of between-subject variation for para-spinal EMG amplitude was attributed to variation in localized subcutaneous body tissue. Mannion and Dolan (1996a) found that rmsEMG displayed a significant inverse relationship with skinfold thickness in both thoracic and lumbar regions of the spine. They suggested that changes in muscle length altered the filtering effect of the skin, such that shorter muscle lengths and thicker tissues attenuated higher EMG frequency components. Skinfold and tissue impedance act as low pass filters reducing the EMG power content in the high frequency range and may alter in response to surface sweat and increases in muscle temperature during fatiguing contractions (Basmajian & De Luca, 1985).

CHAPTER 3

METHODS

In this chapter, methods common to the five experimental studies are outlined in order to reduce later repetition. In subsequent experimental chapters, details of subjects, apparatus, EMG, procedures, and data analysis are given where they differ from detail in this section and are briefly reviewed where appropriate.

3.1 EMG MEASUREMENT

3.1.1 Subject Preparation

Electromyographic signals were recorded from the skin surface using bipolar differential electrodes. Electrode pairs, 10 mm in diameter (disposable conductive adhesive electrodes, Ag/AgCl [Meditrace]) were placed on the midline of the long axis of the belly of vastus lateralis (VL) and rectus femoris (RF) muscles. Each electrode pair was separated from center to center by 30 mm. Palpation and anatomical landmarks were used to establish EMG electrode positions for VL and RF muscles from the right limb of all subjects. The electrode sites for VL and for RF were one-third and one-half respectively, of the perpendicular distance from the superior border of the patella to the greater trochanter. A ground reference electrode was placed on the greater trochanter of the right femur (see Figure 2 for electrode sites). In order to replicate EMG preparation and configuration as closely as possible between trials, the distances between anatomical landmarks and electrode sites were measured and recorded, and sites marked with indelible pen during the first trial.

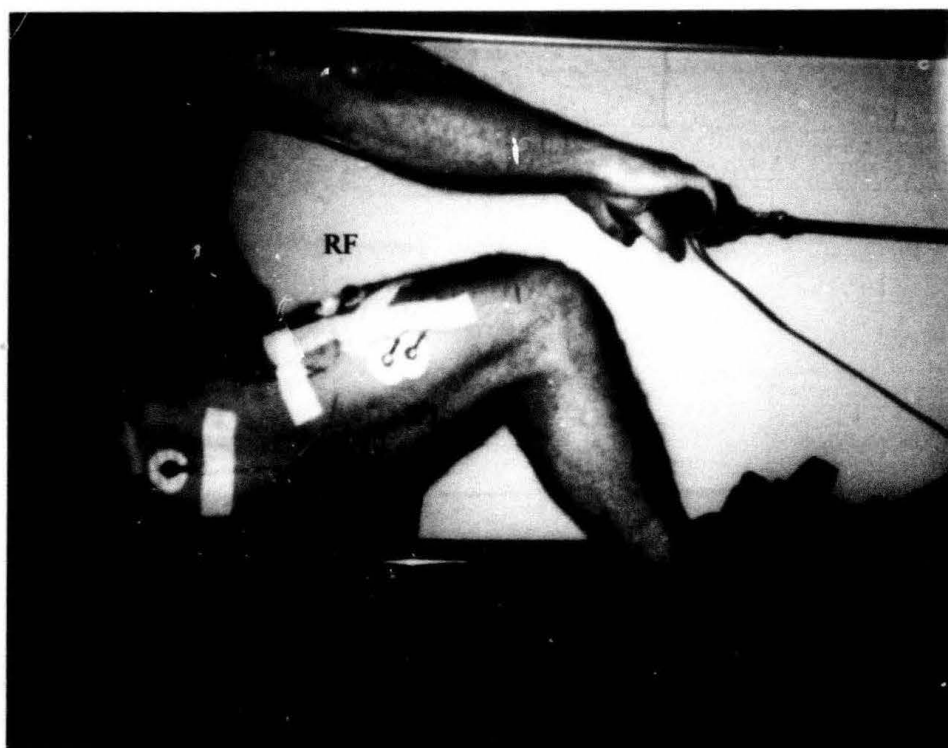


Figure 2. Electrode sites for rectus femoris (RF), vastus lateralis (VL), and greater trochanter (GT).

In order to reduce skin resistance and lower impedance between active muscle fibers and receiving electrodes, sites were thoroughly prepared by shaving (if necessary), abrading with emery paper, and swabbing the skin with alcohol. Impedance between electrode sites was monitored using a hand-held digital multimeter (ITT instruments, MX52). If skin impedance was greater than 10 k Ω , the site was re-abraded until impedance was less than 10 k Ω .

Foil shielded cables attached to the electrodes were of sufficient length (4 m) to reach the data acquisition system during full range of movement, but not so long as to cause unwanted noise or attenuation of the signal. The combined effect of cable capacitance and high EMG amplifier input were assessed and found not to attenuate high frequencies within the range emitted in response to quadriceps muscle activity (>300Hz). Electrodes and cables were taped firmly to the body in order to minimize signal artifact resulting from cable movement. EMG activity was relayed to the data acquisition system (Amlab) in real-time.

3.1.2 Amlab System

An Amlab computer based instrument (single digital signal processor, mini-rack interface, 18 channel isolated ground card) was used to record, store, and analyze data. The mini-rack system facilitated a configured selection of elements such as programmable isolated amplifiers, multi-channel A/D and D/A converters, and strain gauges. The Amlab system enabled software to be written into project schematics using a graphic compiler. Schematics were designed to detail the sequence of analogue signal processing (see section 3.1.3). After pre-amplification, the EMG signal was further amplified and processed within the project schematic. Data were recorded and

displayed in real time, and viewed in chart display windows. Schematics that specifically satisfied each study purpose and each ergometer requirement are referred to in each chapter and presented in Appendix A.

3.1.3 EMG Signal Processing

EMG signal

EMG activity was sampled at a frequency of 1024 Hz and 2000 Hz during isometric and dynamic contractions respectively. Pre-amplification gain was set at 8.988 before the signal was further amplified (within the signal processing schematic) by a gain of 3000 and multiplied by a scaling factor of 2. The EMG signal was band-pass filtered (above 300 Hz and below 5 Hz) and then displayed continuously on a computer monitor. The filter settings were selected to attenuate frequencies outside the spectrum of the EMG signal for the quadriceps muscle and electrode configuration as detailed by Mannion and Dolan (1996b). The band-pass filter consisted of a second order quasi-Butterworth low-pass filter in series with a second order quasi-Butterworth high-pass filter. The points of reference defining the band-pass were -3dB points. The schematic projects were designed to “trigger” recordings of the EMG for specific sample data-windows during each contraction, which were synchronized to either force output, handle velocity, handle displacement, or onset of muscle activation (see each chapter method).

Root mean square of EMG (rmsEMG)

After filtering, rmsEMG was calculated within the Amlab schematic for each 20 samples. Subsequently, rmsEMG data were exported to a statistical package (Microsoft Excel) and rmsEMG was averaged for the duration of each specific data-window.

RmsEMG was selected to express EMG amplitude because it represented the signal power with minimal distortion and had physical meaning (see section 2.7 for rmsEMG calculation).

Mean power frequency of EMG (MPF)

After filtering, the EMG signal was processed by a Fast Fourier Transform algorithm (FFT) each 1024 samples, using the mean value of the magnitude of the constant power signal similar to the method used by Potvin and Bent (1997). The MPF (the average frequency) was calculated (within the project schematic) for each 1024 samples within each data-window. MPF of the EMG was selected to express EMG frequency because it provided a reliable and consistent parameter that was related to muscle fiber conduction velocity (see section 2.7 for MPF calculation).

3.1.4 Signal Noise

Prior to exercise, the EMG signal-to-noise ratio was calculated for the required contraction type (isometric, isokinetic, dynamic) on the apparatus to be used (isokinetic dynamometer, leg-only ergometer, standard rowing ergometer). To record signal noise for isometric contractions, subjects relaxed the leg muscles at the pre-determined displacement (see section 4.2.3) and non-muscular signal noise was recorded. For isokinetic and dynamic contractions, subjects were pushed three times through the full range of motion on the selected ergometer while being completely relaxed. The pushed movement replicated the contraction movement required for each ergometer. Non-muscular signal noise was recorded during the externally pushed motions. To measure maximal EMG activity of the quadriceps muscle (VL, RF), subjects performed three MVCs of the contraction type on the selected ergometer under investigation. Average

peak signal noise values were compared to average peak EMG activity recorded during the MVCs and was expressed as the signal-to-noise ratio. Artifact (non-muscular noise) of less than 1% of peak MVC activity was accepted. If the signal-to-noise ratio was greater than 1%, the subject preparation procedure was repeated until the signal-to-noise ratio was less than 1%.

3.1.5 Data Recording

In order to synchronize recordings of biomechanical outputs (force, displacement, velocity) with corresponding EMG activity during the leg-drive phase, triggered data-windows of specific duration (epochs) were recorded for each contraction. The sample trigger and data-window duration were selected according to purpose of the study, contraction-type, and technical restrictions associated with each ergometer movement. Restricting data-windows to specific durations may have encompassed different fractions of the drive phase depending on the stroke and drive-recovery ratio. This method was not ideal. Some of the machines used in the current studies did not allow precise timing control of the leg extension/drive phase (e.g. rowing ergometer). Hence, in the ergometer studies subjects were asked to maintain a specific stroke rate and perform consistent leg extensions that matched, as near as possible, the duty cycle and leg extensions performed on the isokinetic dynamometer. The focus of the current project was the relationship between force and EMG during the majority of the leg extension/drive phase, and whether the relationship was similar in controlled and applied situations. Therefore, timing of the data-windows was mostly triggered by specific force levels, irrespective of possible deviations in the specific fraction of the drive phase. Nevertheless, restricted data-windows were a limitation of the methodology and may have been a confounding influence on the results. Details of the data triggers and

data-window duration are given in each experimental chapter together with the rationale for selection. Data were saved to disk and replayed later for subsequent analysis.

3.2 BIOMECHANICAL MEASUREMENT

3.2.1 Force

Rowing ergometer force

The force transducer (Radio Spares, model 021-300) for experiments using the rowing ergometer, was a foil (copper-nickel alloy) uni-axial strain gauge (resistance 120 Ω , Wheatstone bridge connection) attached between the ergometer chain and the handle (Figure 3). The force transducer received a constant analogue input voltage (± 5 V swing at 100 mA) from the Amlab system, which was subsequently modulated by the applied forces and relayed back to Amlab. There was no signal drift due to heating effects observed when the strain gauge was held at maximal excitation.

Forces applied to the ergometer handle resulted in corresponding alterations in the constant analogue signal received by Amlab and were displayed on a computer force chart. In order to display a signal in the region of ± 5 V, signal amplification was adjusted by manipulation of the gain or the scaling factor within the schematic amplifier. For rowing ergometer trials, the analogue force output signal of the strain gauge was relayed to Amlab via shielded cables. Project schematics were specifically designed for the rowing ergometer (see experimental methods). The sampling rate for force data was 200 Hz for all contraction types. The force signal was displayed on a chart for real-time viewing and saved to disk for later analysis.

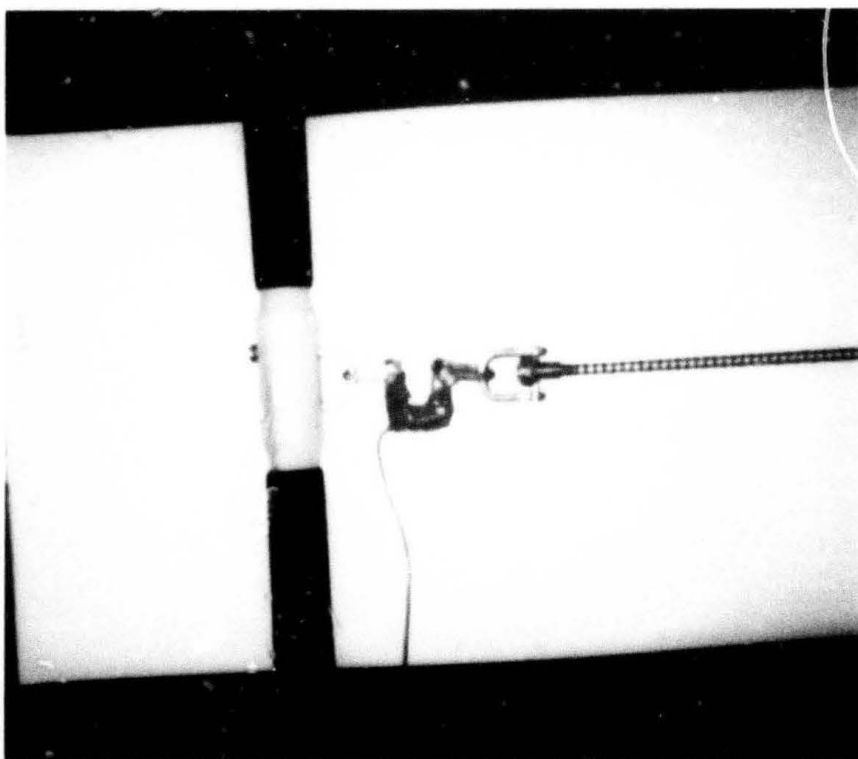


Figure 3. Arrangement of strain gauge attachment between rowing ergometer chain and handle.

For each transducer and each machine calibration data was observed, recorded and stored to disk (not reported here). Where required the transducers were recalibrated after the necessary adjustments. To calibrate the force transducer, the strain gauge attachment was removed from the ergometer chain and suspended from a metal frame (Figure 4). A range of calibration masses appropriate to the forces generated during rowing ergometry (100 - 1000 N), were suspended vertically from the strain gauge in increments of 11.34 kg. Each mass value, multiplied by the acceleration due to gravity (9.81 m/s^2), was calculated as equivalent to a weight force of 110.26 N. Signal amplification was adjusted until the signal trace for each suspended weight force corresponded to the force value (N) displayed on the schematic chart. Signal offset was adjusted until a signal zero base line was achieved when no mass was suspended. Schematic chart force value and suspended mass value were recorded each time mass was changed in both ascending and descending order. The procedure was repeated before and after each testing session, and the baseline value was checked for voltage-drift each trial. Minor adjustments in base line were made using a constant offset if necessary.

A force-voltage correlation graph was plotted each test day and data examined for linearity using calculated r^2 . The rms error for the force sensor calibration was recorded. A linear correlation greater than .98 was accepted as a stable force-voltage relationship. if r^2 was less than .98, then amplification and scaling factors of the voltage signal were adjusted and calibration procedure repeated.

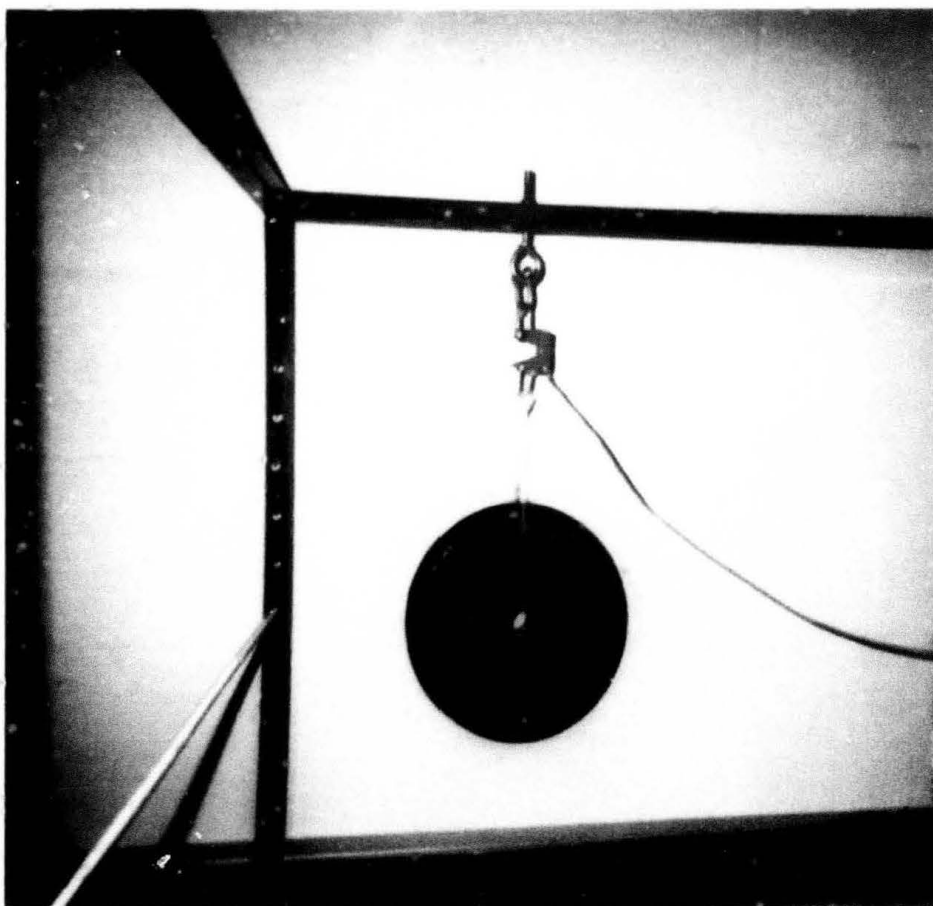


Figure 4. Calibration set-up for strain gauge.

Adapted leg-only ergometer force

The force transducer and calibration of procedures for the adapted leg-only ergometer were identical to those using the rowing ergometer except that the chain attachment was fixed to a modified stock rigidly attached to the seat (Figure 5). Modification to the leg-only ergometer is described in section 4.2.2. Amlab project schematics were specifically designed for the adapted leg-only ergometer (see experimental methods).

Isokinetic dynamometer torque

For isokinetic trials, the analogue torque output signal of the dynamometer (Cybex 6000, Lumex, Inc.) was relayed to Amlab via shielded cables. Project schematics were specifically designed for the dynamometer (see experimental methods). Procedures for amplification, sampling, display, and storage of dynamometer torque were similar to those for rowing ergometer force.

To calibrate torque (moment of force) output of the dynamometer load cell, the calibration procedure detailed in Cybex 6000 Extremity User's Guide pp. 4-31 (Cybex Division of Lumex, 1991) was performed. In addition, to verify calibration of torque displayed in the Amlab computer chart, a further procedure was performed prior to each testing session. The lever arm was positioned horizontally, adjusted with a spirit level, and securely locked.

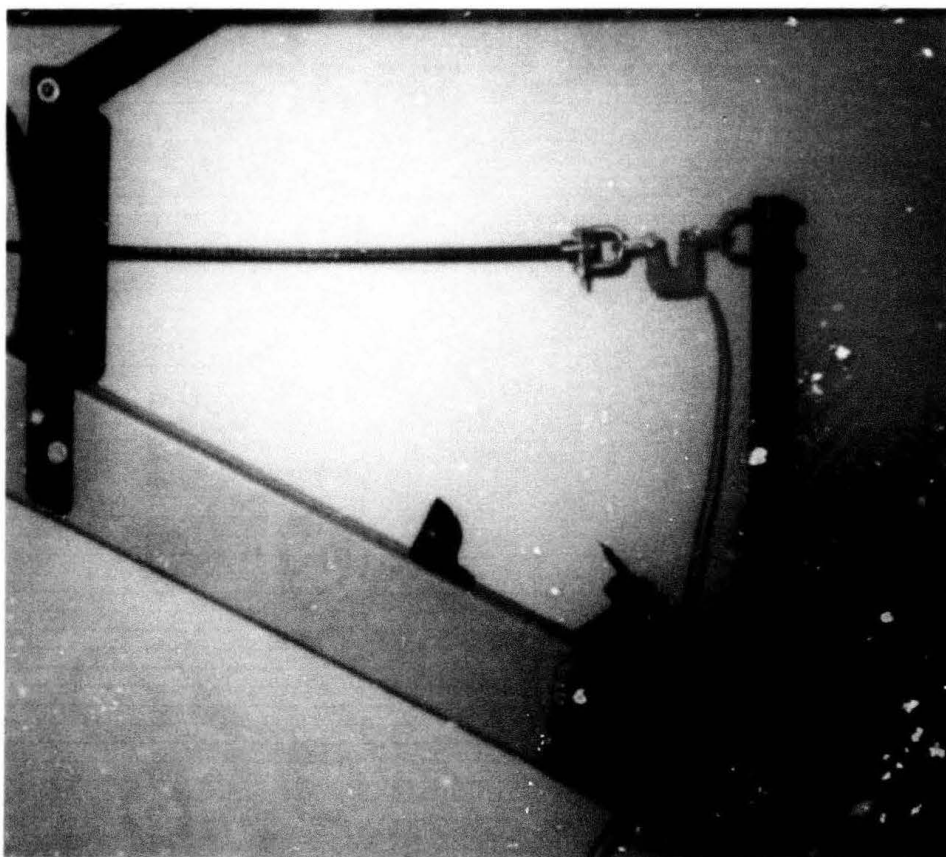


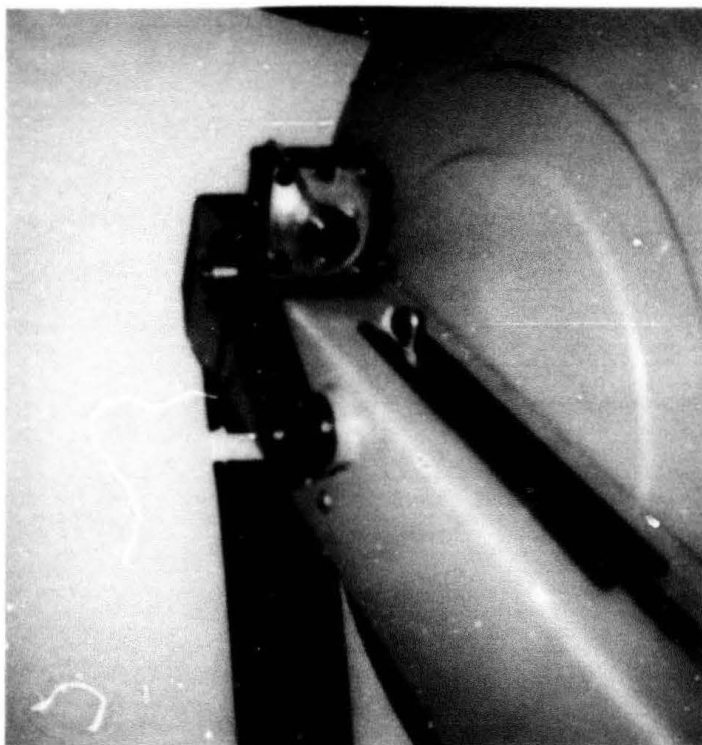
Figure 5. Modified stock for adapted leg-only ergometer.

The masses previously used were attached incrementally to the end of the lever arm. The length of the lever arm from the center position of the mass to the axis of rotation was measured. Calibration of the analogue torque signal was achieved by amplification adjustments until the torque value on the schematic chart corresponded to that produced by each suspended mass. Each mass was under the influence of gravitational acceleration with the known moment arm and with respect to the axis of rotation. Torque about the axis of rotation was calculated as the known mass multiplied by acceleration due to gravity, multiplied by the perpendicular distance from the axis of rotation to the mass.

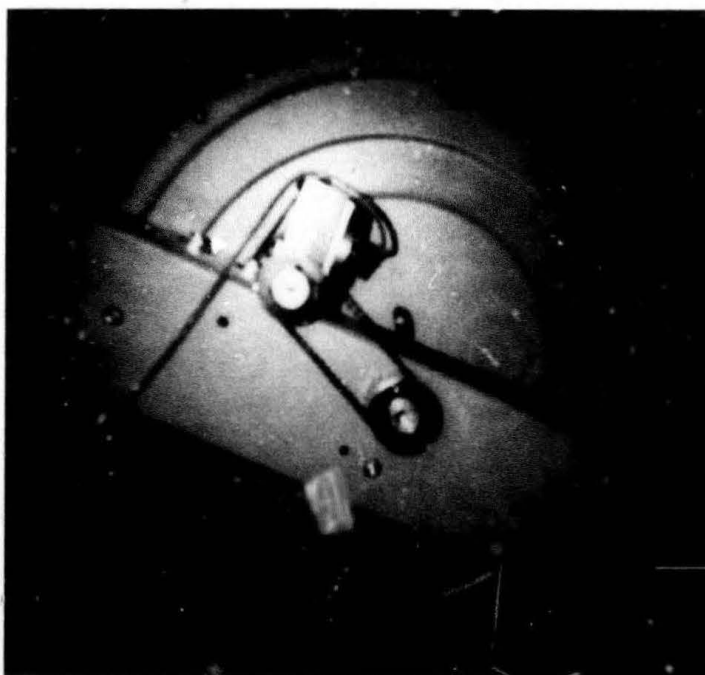
3.2.2 Displacement

Rowing ergometer displacement

The displacement transducer for experiments using the rowing ergometer was a 10 k Ω , 10 turn potentiometer (model RS 173-580, linearity $\pm 5\%$) fixed to a rotating spindle. The spindle was attached to a 20:1 gearing reduction, achieved by a constructed rubber pulley system encased in an aluminum box and firmly fixed to the rotating spindle of the ergometer fly wheel (Figure 6). Linear motion of the ergometer chain caused the flywheel to rotate and after 20:1 gearing reduction, simultaneously acted to rotate the potentiometer spindle (maximum 10 turns). The displacement potentiometer received a constant analogue output voltage (± 5 V swing at 100 mA) emitted from the Amlab system. Any displacement of the handle directly corresponded to the rotations of the potentiometer spindle that altered the internal resistors, thereby altering the signal relayed to Amlab. The Amlab signal received from the potentiometer was displayed on the computer displacement chart.



rear view



lateral view

Figure 6. Ergometer potentiometer and pulley gearing system (rear view, lateral view).

The potentiometer spindle was set to rotate a maximum of 10 turns and provide -5 to $+5$ mV values between zero and maximum displacement of the ergometer chain. Positive displacement values were related to quadriceps leg extension during the drive phase when the ergometer handle was moved away from the ergometer display unit (zero position). Negative displacement values represented displacement (from the zero position) when the handle was moved towards the zero position. In order to display a signal in the region of ± 5 V, amplification of the analogue displacement signal was adjusted by manipulation of the gain or the scaling factor in the schematic amplifier. The sampling rate of displacement data was 200 Hz. The displacement signal waveform was displayed on a chart for real-time viewing and stored on disk for later analysis.

To calibrate the displacement potentiometer, signal amplification was adjusted until a zero base line was achieved with handle and chain fully retracted (but not slack). This handle position was marked relative to the fixed ergometer display unit and defined as zero displacement. A calibrated two-meter ruler was positioned in the same horizontal plane as the chain sprocket, with the zero meter mark on the ruler aligned with the pre-designated zero handle displacement. A displacement range appropriate to expected handle movement (0 - 1.8 m) was calibrated at each 30 cm of handle displacement. Signal amplification was adjusted until the displacement signal trace corresponded to the value displayed in meters on the schematic displacement chart. The calibration procedure was repeated before and after each testing session, and the baseline was checked each session for voltage drift. Minor adjustments in base line were made using a constant offset if necessary. A displacement-voltage correlation graph was plotted each test day and examined for linearity using calculated r^2 .

Adapted leg-only ergometer displacement

For the adapted leg-only ergometer, the chain attachment was fixed to a modified stock rigidly attached to the seat. The calibration displacement procedure for the adapted leg-only ergometer was identical to that for the rowing ergometer except that displacement was measured by movement of the chain attached to the stock not to the handle (as for the standard ergometer).

Isokinetic dynamometer displacement

For isokinetic dynamometer (Cybex 6000) trials, the analogue signal for angular displacement was relayed from the dynamometer to Amlab via shielded cables. Procedures for amplification, sampling, display, and storage of dynamometer angular displacement were similar to those for rowing ergometer displacement.

In order to calibrate angular displacement of the isokinetic dynamometer, the calibration procedure detailed in Cybex 6000 Extremity User's Guide page 4-31 (Cybex Division of Lumex, 1991) was performed. In addition, to verify calibration of angular displacement of the lever arm for display in the Amlab computer chart, a further procedure was performed prior to each testing session. The lever arm was positioned at horizontal, at 45° below horizontal, and at vertical inclinations ascertained by spirit level, protractor, and Plumb bob respectively. Amplification was adjusted until the signal value on the chart corresponded to each position (0, 45, 90° from the horizontal plane) of the lever arm.

3.2.3 Calculated Biomechanical Parameters

Velocity

Handle linear velocity (m/s) of the ergometer and angular velocity ($^{\circ}$ /s) of the Cybex dynamometer arm were calculated within the project schematic using differentiation over time of linear and angular displacement respectively. Hence, positive velocity was related to the handle being moved away from the zero position on the ergometer or away from the vertical position on the isokinetic dynamometer, both associated with the leg extension drive phase. Negative velocity was related to the handle being moved towards the zero position on the ergometer or towards the vertical position on the isokinetic dynamometer, both associated with the leg flexion recovery phase.

Power

Instantaneous power (W) on the ergometers was calculated within the project schematic as the product of instantaneous force output at the handle and instantaneous handle velocity. Power was then averaged for the data-window duration.

3.3 BLOOD LACTATE MEASUREMENT

Blood lactate concentration was measured using a blood-lactate analyzer (Accusport, Boehringer Mannheim). The method used to collect blood samples followed manufacturer instructions and was similar to that used by Hagerman et al., (1996). Skin at the sample sites was thoroughly swabbed using isopropyl alcohol and dried with cotton wool. A sterile lancet inserted in a pricking device (Autoclix) was used to puncture the skin on the fleshy part of the thumb. A large suspended drop of capillary

blood was allowed to develop and applied to the test pad before the pad was inserted into the lactate analyzer without being touched. Lactate content was determined by reflectance photometry (via a lactate-oxidase mediator reaction) within the Accusport. Blood lactate value (mmol/L) was digitally displayed after 60 s.

Calibration of the Accusport was performed weekly during test periods, at temperatures between +15 and 35° C and at relative humidity less than 90 %. The Control-Lactate solutions (Boehringer Mannheim) for high and low concentrations were used according to manufacturer instructions. Calibration data for the Accusport were recorded and saved to disk (not reported here).

3.4 DATA TREATMENT

Numerical values of the data-window sample for each contraction were exported to a statistical package (Microsoft Excel) and averaged for data-window duration. In some studies, data from either three or five consecutive windows were averaged to provide a mean value for consecutive contractions. For MVC value in non-fatiguing trials (chapter 4), three consecutive data-windows were averaged and for MVC value in fatigue trials (chapter 5, 6, 7, 8), five consecutive data-windows were averaged. In all trials, pre-exercise MVC or initial MVC served as the reference value. Data were normalized to mean reference value of the same contraction type performed on the same ergometer and represented percentage change from pre- or initial MVC. Statistical analysis was performed in the Statistical Package for Social Sciences (see detail in each chapter).

CHAPTER 4

THE RELATIONSHIP BETWEEN FORCE OR TORQUE OUTPUT AND EMG OF LEG EXTENSORS DURING ISOMETRIC, ISOKINETIC, AND DYNAMIC CONTRACTIONS

4.1 INTRODUCTION

The relationship between force output and EMG activity may be a useful indicator of neuromuscular function during performance tasks (De Luca, 1997; Mannion & Dolan, 1996b). The amplitude of the EMG signal reflects both motor unit recruitment and the firing rate of active motor units. During controlled muscle contractions, such as those in isometric conditions, force output and EMG amplitude are positively correlated, their relationship being well-established (Basmajian, 1974; Mannion & Dolan, 1996a). Under the same conditions, the relationship between force output and frequency characteristics of the EMG signal is not as clear. For example, Petrofsky and Lind (1980) found no relationship between power output and MPF. In contrast, Broman et al. (1985) reported that MPF increased proportionally with force output which they ascribed to increased usage of Type II muscle fibers with faster conduction velocities. Furthermore, Mannion and Dolan (1996a) reported a non-linear force-MF relationship that differed in direction at different recording sites in the same muscle.

The level of movement restriction, which alters with type of contraction performed (isometric, isokinetic, dynamic), has a marked effect on the force/torque-EMG

relationship. The force/torque-EMG relationship during isometric and isokinetic (when movement is less restricted) contractions is similar, but the latter relationship is weaker (Christensen et al., 1995). In tasks such as rowing, movement is further unrestricted and contractions involve the contribution from multiple muscle groups, changes in muscle length, and varying velocities of muscle contraction (Rodriguez et al., 1990; Wilson et al., 1988). These so-called confounding factors that affect the EMG (De Luca, 1997) may be examined using different ergometer types. Few studies have systematically investigated the effect of movement restriction on force/torque-EMG relationships in the same subjects. In order to evaluate whether EMG analysis might be useful in rowing performance assessment, it is important to determine whether the force/torque-EMG relationship obtained under restricted movement conditions, transfers to more practical situations such as rowing ergometer tasks.

The purpose of this study was to establish force/torque-EMG (rmsEMG, MPF) relationships during isometric, isokinetic, and dynamic contractions performed on different ergometers. A further purpose was to examine for effect of ergometer type on the force/torque-EMG relationship under isometric and under less restricted conditions (isokinetic, dynamic).

It was hypothesized that:

- 1) There would be a significant relationship between voluntary force or torque output and EMG (rmsEMG, MPF) of the quadriceps muscle (RF, VL) during contractions of varying intensities under the following conditions:
 - a) isometric contractions performed on an isokinetic dynamometer.
 - b) isometric contractions performed on an adapted leg-only rowing ergometer.

- c) isokinetic contractions performed on an isokinetic dynamometer.
 - d) dynamic contractions performed on an adapted leg-only rowing ergometer.
 - e) dynamic contractions performed on a standard rowing ergometer.
- 2) There would be no significant effect of ergometer type on the force/torque-EMG (rmsEMG, MPF) relationship of the quadriceps muscle (RF, VL) between condition (a) and (b) during isometric contractions.
- 3) There would be a significant effect of ergometer type on the force/torque-EMG (rmsEMG, MPF) relationship of the quadriceps muscle (RF, VL) between condition (c), (d), and (e) during isokinetic or dynamic contractions

4.2 METHODS

4.2.1 Subjects

Male rowers ($n = 11$), mean (SD) age 28.2 (9.9) years, height 181.3 (8.4) cm, and weight 89.5 (7.8) kg with a minimum of two seasons training experience, consented to participate in the study (see Appendix B for informed consent information). The University Ethical Committee gave approval for the research (see Appendix C for ethical approval). Subjects performed leg extension exercises during three trials (T_{cyb} , T_{adap} , T_{stan}) on different ergometers, which were completed in the same order during separate visits to the laboratory. The first (T_{cyb}) was performed on an isokinetic dynamometer, the second (T_{adap}) on an adapted leg-only ergometer, and the third (T_{stan}) on a standard rowing ergometer. Eight of the subjects who participated in the study completed all three trials (T_{cyb} $n = 10$), (T_{adap} $n = 9$), (T_{stan} $n = 11$).

4.2.2 Apparatus

For simplicity of description, the term “torque” may be referred to as force or muscle force in this study, except where torque alone is specific to the measurement. For T_{cyb} , an isokinetic dynamometer (Cybex 6000) was modified to export analogue torque and angular displacement data during quadriceps leg extensions. Force applied at the ankle-pad was measured as torque at the knee-joint axis of rotation and displacement measured as angular displacement of the lever arm (Figure 7). For T_{adap} , an adapted leg-only rowing ergometer (Concept II) was fitted with a strain gauge between the stock (rigidly attached to the seat) and ergometer chain to measure linear propulsive force of the leg extensors, and a potentiometer to measure linear displacement of the stock (Figure 8). Transmission of force for this ergometer was achieved by means of a harness positioned around the subject’s lower back and firmly attached to the stock. For T_{stan} , a standard rowing ergometer (Concept II) was modified by addition of a strain gauge and a potentiometer to measure handle force and displacement respectively (Figure 9). For all three trials, the computerized data acquisition system (Anlab Systems) and specifically designed schematic projects were used for analogue to digital conversion and for calibration of force, torque, and displacement data (as previously described in section 3.2). The schematic projects used in this study were (a) “tcybisom.prw” for isometric contractions on the Cybex dynamometer (Appendix A1), (b) “tcybisok.prw” for isokinetic contractions on the Cybex dynamometer (Appendix A2), (c) “tadap.prw” for isometric and dynamic contractions on the adapted leg-only ergometer (Appendix A3), and (d) “tstan.prw” for dynamic contractions on the standard rowing ergometer (Appendix A4).



Figure 7. Subject set-up for isokinetic dynamometer (Cybex 6000) used in T_{cyb} .

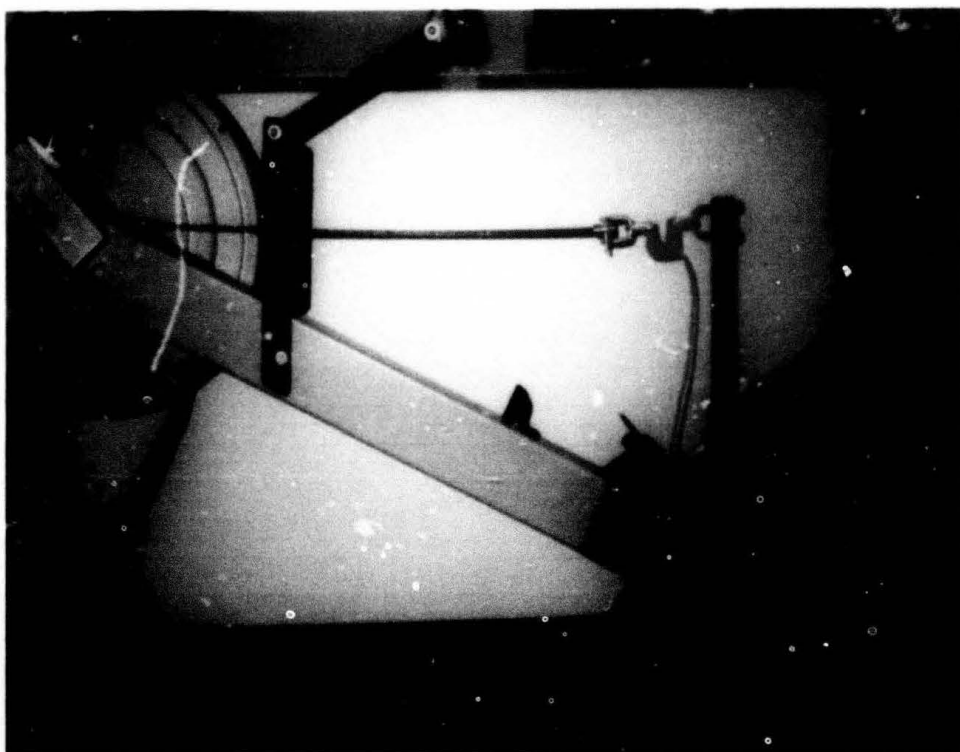


Figure 8. Adapted leg-only ergometer used in T_{adap} .

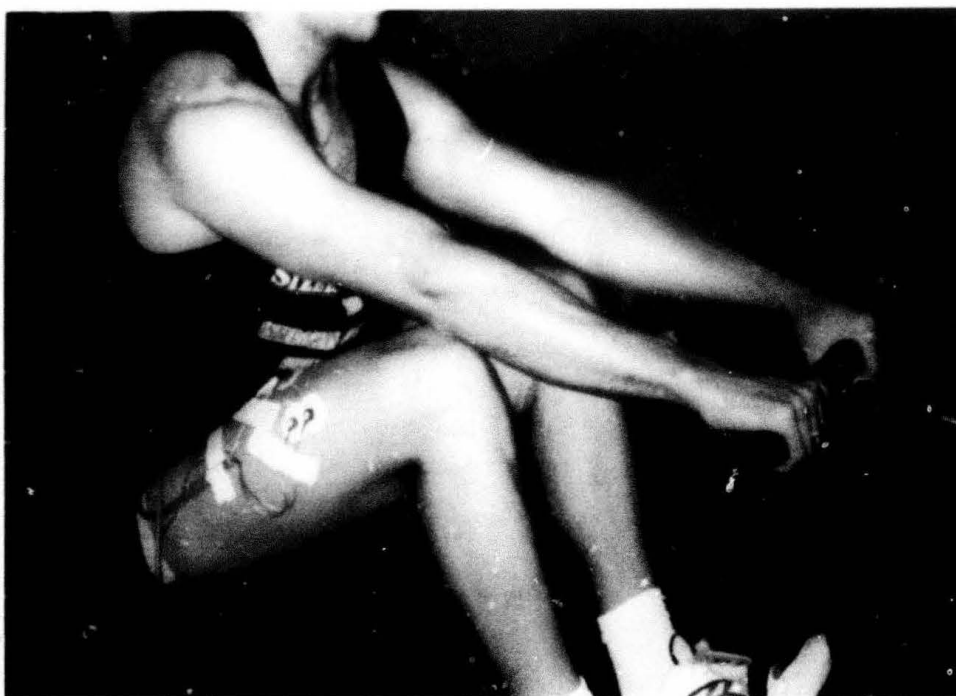


Figure 9. Subject set-up for standard rowing ergometer (Concept II) used in T_{stan} .

4.2.3 EMG

EMG activity was recorded from the right VL and RF muscles using bipolar surface electrodes placed over prepared sites. For all trials, EMG preparation and configuration were the same (see section 3.1). Impedance between the electrodes was maintained less than 10 k Ω and level of electrical signal noise was accepted at less than 1% of maximal excitation level.

The triggers and data-window duration for each contraction type were designed to sample similar force or torque windows inclusive of peak force or peak torque and the majority of leg extension activity. For isometric contractions in T_{cyb} , data collection was triggered at an angular displacement of 45° from the vertical. Subjects were asked to slowly extend the leg from the vertical position until an angular displacement of 45° was attained at which time the dynamometer arm automatically locked ready for the isometric contraction and the data trigger was initiated. This displacement was previously calculated to be the average angular displacement at which peak torque was attained from isokinetic trial-contractions. In order to exclude minor oscillations of the dynamometer arm during attainment of steady-state torque, data collection was delayed 1-s post-trigger. A 1.0-s data-window (1024 data point) of EMG was sampled at a frequency of 1024 Hz during each leg extension. The EMG signal was band-pass filtered (below 5 and above 300 Hz) to exclude artifact. EMG data were recorded on the computerized acquisition system and averaged (by taking the rmsEMG for each 20 samples) before being exported to a statistical package for further calculations. In addition, after band-pass filtering the EMG signal was processed by a FFT each 1024 samples before calculation of the MPF. For isokinetic contractions in T_{cyb} , the trigger for data collection (15° displacement from the vertical) was selected to exclude limb

acceleration prior to constant velocity (calculated from average trial-data). EMG data for isokinetic contractions were sampled for a 0.3-s window (660 data-point) at a frequency of 2000 Hz in order that only positive leg extension (drive) parameters were recorded. In T_{adap} , for both isometric and dynamic contractions data acquisition was triggered by forces greater than 50 N. Data-window duration was 1.0-s for isometric and 0.3-s for dynamic contractions. For dynamic contractions in T_{stan} data acquisition was triggered by positive handle velocity greater than 0.2 m/s and sampled for a longer period (0.5 s) in order to accommodate the increased time required to reach peak force. The changes from data triggering in T_{cyb} were necessary in order to record similar data-windows of the force/torque-time curve for each of the different movement patterns.

4.2.4 Procedure

The trials were performed in the same order (T_{cyb} , T_{adap} , and T_{stan}) on separate occasions. Trials were not counter-balanced as subjects were trained rowers and well practiced with the leg extension movements. Therefore, it was unlikely that any learning effect influenced performance between trials. Prior to each testing session, subjects completed a 5-minute non-fatiguing warm-up on a standard rowing ergometer at a pace that was approximately 65 % of MVC power output. Submaximal trials were performed to familiarize subjects to the contraction type, which included isometric leg extensions in T_{cyb} and T_{adap} , and in addition, isokinetic or dynamic leg extensions in T_{cyb} , T_{adap} , and T_{stan} . At completion of the isometric and prior to the isokinetic (in T_{cyb}) or dynamic protocol (in T_{adap}), subjects were released from the equipment and performed a series of stretching exercises. These were completed in order to increase the blood flow and relax the muscles.

For T_{cyl} , subjects were positioned and firmly secured on the dynamometer with the trunk at 90° to the thigh. The seat and lever arm were adjusted until the right knee joint was in line with the axis of rotation of the lever arm, and the ankle pad was fixed at the correct lever length above the medial malleolus of the right ankle. In T_{cyl} , the knee angle was 135° for isometric contractions and the range of knee movement was 90° for isokinetic contractions. For T_{adapt} , subjects were securely fitted with the back harness, which was attached to the stock. Displacement of the stock for isometric contractions in T_{adapt} was calculated as for T_{cyl} (described previously in section 4.2.3). In order to prevent any movement of the seat stock or subject, a non-extendible strap was fixed from the ergometer stock to the seat stock. For isometric and dynamic contractions in T_{adapt} , subjects were requested to remain upright, grip the seat, and perform consistent leg-drive technique without the use of the upper body. For dynamic contractions in T_{stan} , subjects were asked to maintain a consistent movement technique that involved the co-ordinated synergistic action of several muscle groups employed during rowing.

For all contractions, force or torque outputs were sampled at a frequency of 200 Hz. Force or torque and EMG data were triggered simultaneously and recorded for specific data-window duration (see section 4.2.3) then averaged for that duration. Due to technical delay in providing real-time averaged values during dynamic contractions, subject visual feedback was provided for peak target force instead of average target force each contraction. In addition, visual feedback was provided for actual peak force achieved.

Three maximal contractions were performed at commencement (pre-MVC) and completion (post-MVC) of the random force or torque protocols. Pre-MVC force or

torque was measured as the mean data-window value for three maximal efforts performed prior to the random protocol. Post-MVC force or torque (mean of three contractions) was compared to pre-MVC value in order to confirm that there was no loss of maximal voluntary force or torque during the protocol. MVC was calculated as the "mean of three contractions" as opposed to the "best contraction" because it was considered a more representative measure during applied dynamic conditions. The MVC calculation has been described previously (Cram, Kasman, & Holtz, 1998).

Subjects matched contractions to a peak target force or torque (20, 40, 60 or 80 % MVC). Each target force or torque contraction was repeated three times and target values were attempted in a random order with a 60-s recovery between each three contractions. The protocols were identical for each ergometer type, except that subjects were asked to maintain a movement velocity (200° s^{-1}) of the stock or handle and duty cycle (2 s) for T_{adap} and T_{stan} , similar to that in T_{cyclic} . These values were chosen to replicate the timing of leg drive and recovery movement during simulated rowing, equivalent to 30 strokes per minute.

4.2.5 Analysis

Computerized recordings of the contractions in each trial were replayed on the data acquisition system, with each series of triggered data-windows of the leg extension phase tagged, digitized, and exported for further calculations. EMG and force or torque data were averaged for each contraction data-window for each muscle. Corresponding pairs of EMG and force or torque data were analyzed for each series and each subject. Force or torque and EMG were initially collated in absolute terms, but in order to account for inter-subject variations in EMG signal and leg extensor strength, the EMG

and force or torque data were normalized to respective MVC values. In order to examine whether rmsEMG could predict force or torque output during isometric contractions, common linear variance scores (r^2) for force/torque-rmsEMG were calculated for each subject, for each muscle, and for each of the two ergometers under isometric conditions. In addition, common linear variance scores (r^2) for force/torque-rmsEMG were calculated for each subject, for each muscle, and for each of the three ergometers under isokinetic or dynamic conditions. Common linear variance scores (r^2) for force/torque-MPF were calculated for each subject and for each muscle under isometric conditions in T_{cys} . The twelve scores (r^2) for each subject became the dependent variables on subsequent analysis in Statistical Package for Social Sciences (SPSS). In order to examine whether the force/torque-rmsEMG and the force/torque-MPF scores were significantly ($p < .05$) different to zero, twelve one-sample t-tests were performed.

In order to examine for ergometer and muscle effects for the eight subjects who participated in all trials, fully repeated measures Analysis of Variance (ANOVA) was performed on force/torque-rmsEMG r^2 scores for those eight subjects. Hence, for isometric force/torque-rmsEMG r^2 scores, a 2 (ergometer type: T_{cys} , T_{adap}) x 2 (muscle: RF, VL) fully repeated measures ANOVA was performed. For isokinetic and dynamic force/torque-rmsEMG r^2 scores, a 3 (ergometer type: T_{cys} , T_{adap} , T_{stan}) x 2 (muscle: RF, VL) fully repeated measures ANOVA was performed. As a result of the previous one-sample t-tests, ANOVA was not performed on force/torque-MPF r^2 scores.

4.3 RESULTS

4.3.1 Force/torque-rmsEMG

A positive linear relationship was found between force output and rmsEMG (RF, VL) during isometric, isokinetic, and dynamic contractions. The mean r^2 scores for all conditions were statistically significant at $p < .01$ and similar for RF and VL muscles. Table 1 shows mean (SD) isometric force/torque-rmsEMG relationships (r^2) and Table 2 shows mean (SD) isokinetic and dynamic force/torque-rmsEMG relationships (r^2). Sample EMG traces together with the corresponding force or torque traces for RF and VL are shown for isometric contractions in Figure 10, and for isokinetic and dynamic contractions in Figure 11.

There was a clear increase in variability of the force/torque-rmsEMG association when forces were greater than 80 % MVC with a resultant loss in linearity of the relationship (refer to Figures 12, 13, and 14 for all the individual data for each trial performed by each of the subjects). Variability in the force/torque-rmsEMG relationship was greater for high intensity contractions, and greater during isokinetic and dynamic contractions than during isometric contractions. Furthermore, variability was greater for all intensities during dynamic contractions in T_{adap} and T_{stan} than during isokinetic contractions in T_{cyc} . On frequent occasions, particularly at near-maximal effort, normalized amplitude of the EMG was greater than demonstrated at pre-MVC without a concomitant increase in force output. EMG amplitude greater than the value at 100 % pre-MVC was more common in T_{stan} than in the other trials.

Table 1

Mean (SD) of Force/torque-rmsEMG Relationships During Isometric Random Force or Torque Contractions for Two Ergometer Types and Two Muscles

Muscle	Ergometer type	
	T _{Cyb} ^a	T _{Adap} ^a
	m (SD)	m (SD)
RF	.94 (.04)**	.91 (.08)**
VL	.94 (.04)**	.95 (.02)**

Note. Values are Linear Regression Coefficient "least squares" (r^2) scores

rmsEMG = root mean square of the electromyogram.

T_{Cyb} = Cybex dynamometer trial, T_{Adap} = adapted kcy-only ergometer trial.

RF = rectus femoris muscle, VL = vastus lateralis muscle.

$\eta^2 = 10$, $\eta^2 = 9$.

** = $p < .01$.

Table 2

Mean (SD) of Force/torque-rmsEMG Relationships During Dynamic or Isokinetic Random Force or Torque Contractions for Three Ergometer Types and Two Muscles

	Ergometer type		
	T_{cyb}^a	T_{adap}^b	T_{stan}^c
Muscle	m (SD)	m (SD)	m (SD)
RF	.87 (.05)**	.72 (.18)**	.75 (.09)**
VL	.87 (.05)**	.68 (.13)**	.82 (.11)**

Note. Values are Linear Regression Coefficient "least squares" (r^2) scores

rmsEMG = root mean square of the electromyogram, T_{cyb} = Cybex dynamometer trial,

T_{adap} = adapted leg-only ergometer trial, T_{stan} = standard rowing ergometer trial.

RF = rectus femoris muscle, VL = vastus lateralis muscle.

$n^a=10$, $n^b=9$, $n^c=11$.

** = $p < .01$.

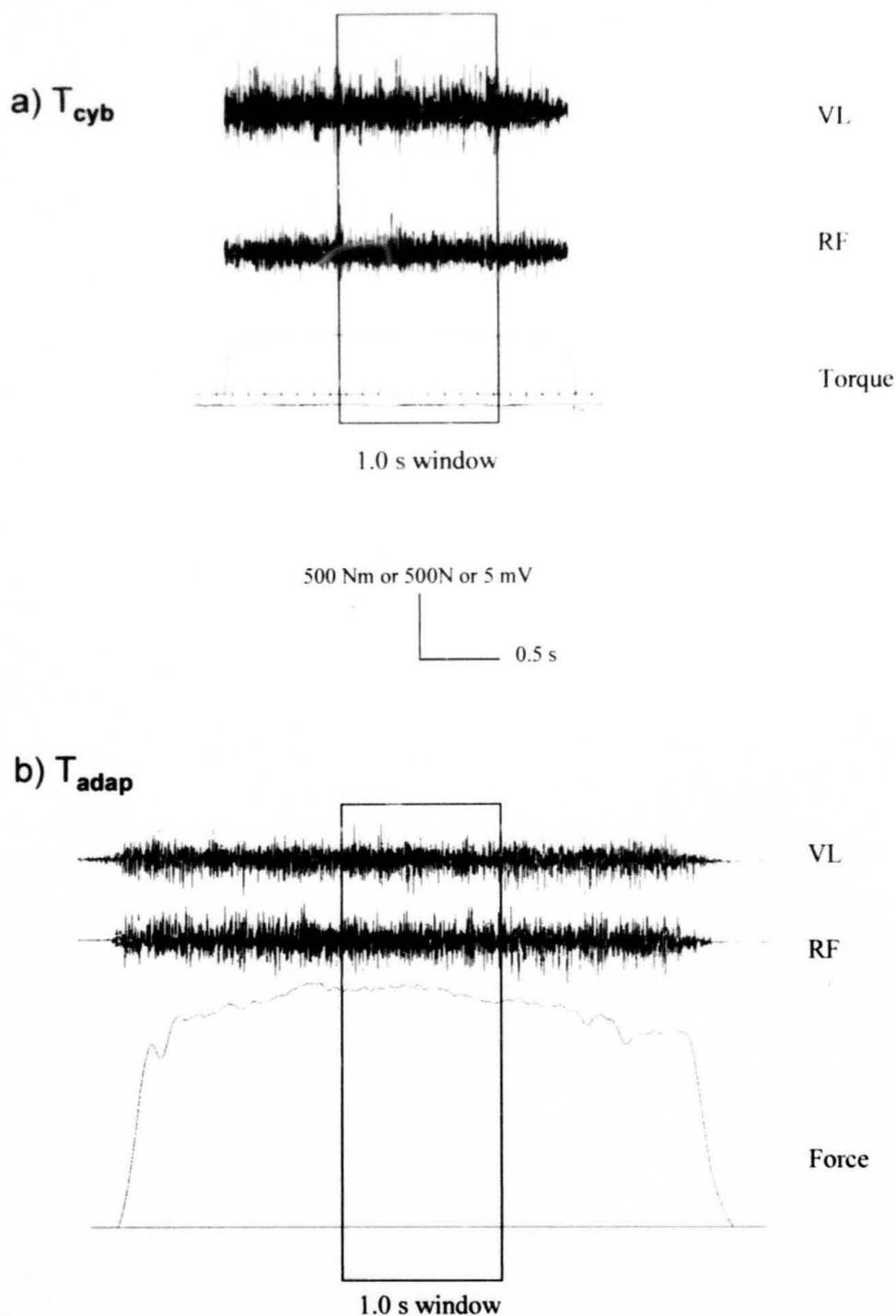


Figure 10. Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and torque or force for the same subject during isometric contractions performed on a) the Cybex isokinetic dynamometer in T_{cyb} and b) the adapted leg-only ergometer in T_{adap} .

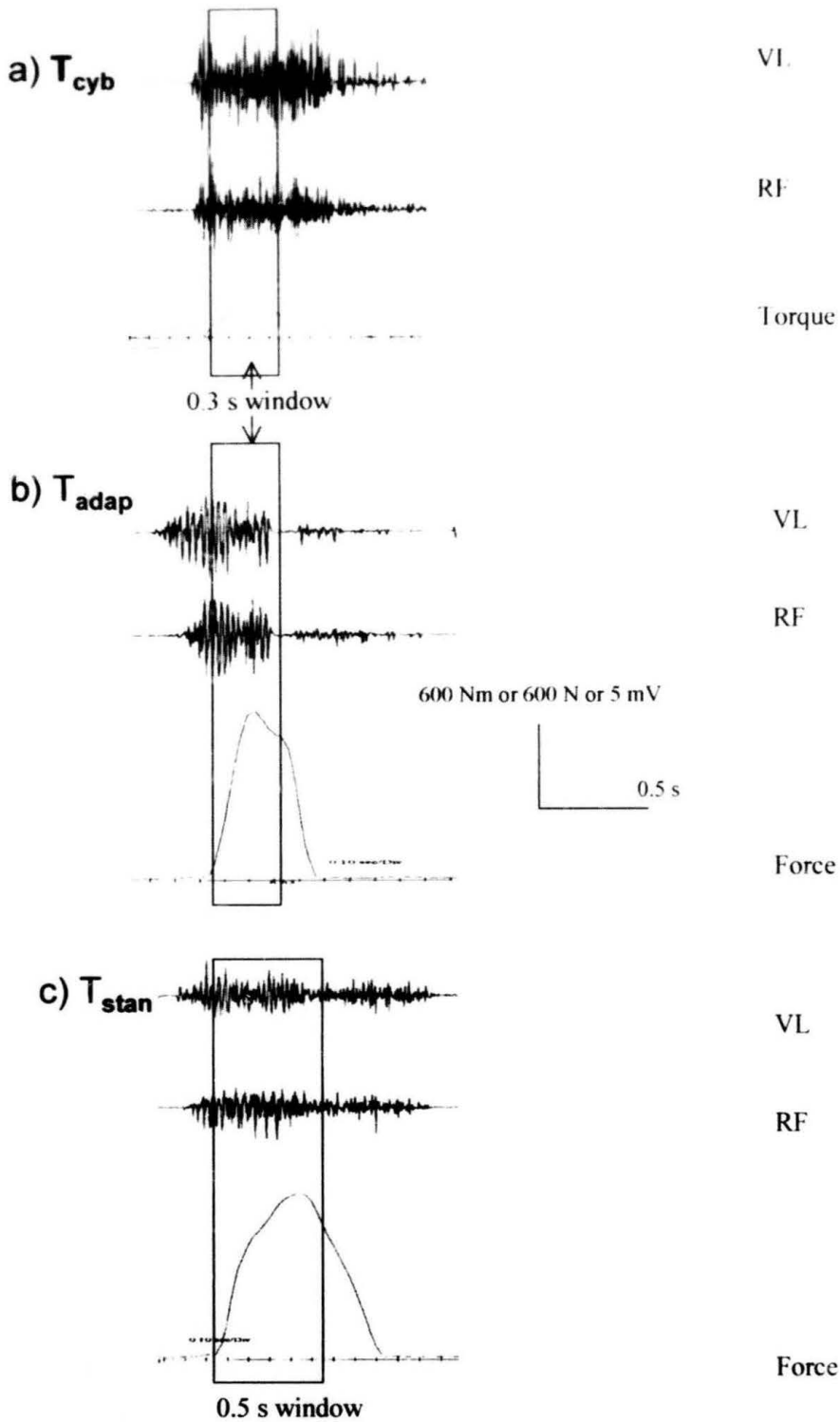
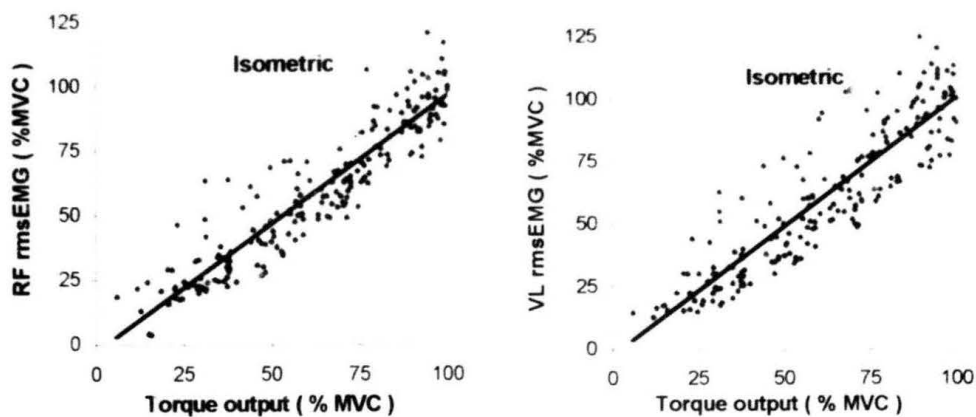


Figure 11. Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and torque or force for the same subject during isokinetic or dynamic contractions performed on a) the Cybex isokinetic dynamometer in T_{cyb} , b) the adapted leg-only ergometer in T_{adap} , and c) the standard rowing ergometer in T_{stan} .

T_{cyb}

a)



b)

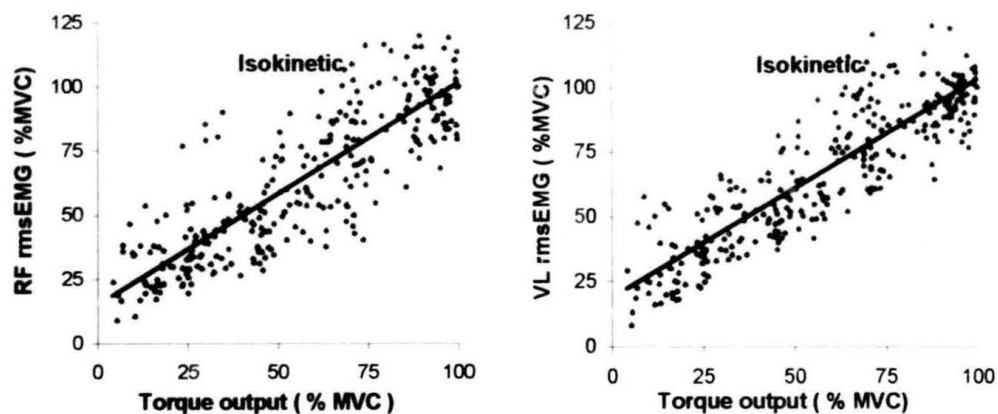
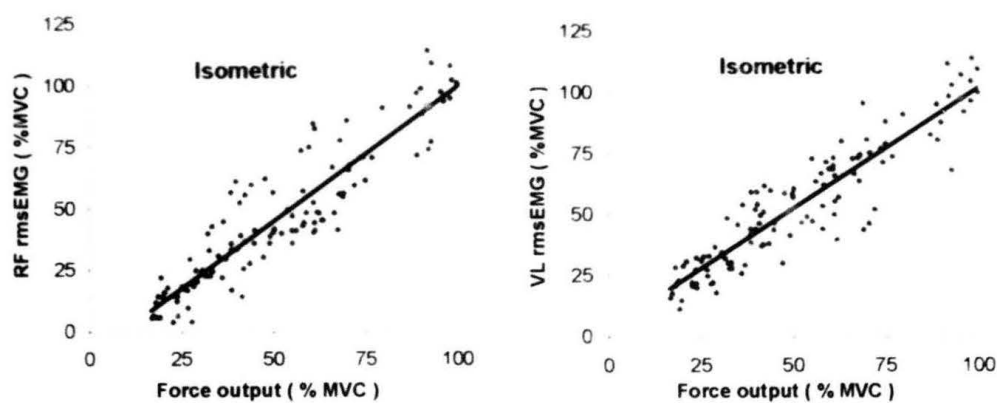


Figure 12. Relationship between torque output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) during a) isometric and b) isokinetic leg extensions performed on the dynamometer in T_{cyb} . Values are normalized and expressed as a percentage of value at maximal voluntary contraction (MVC).

T_{adap}

a)



b)

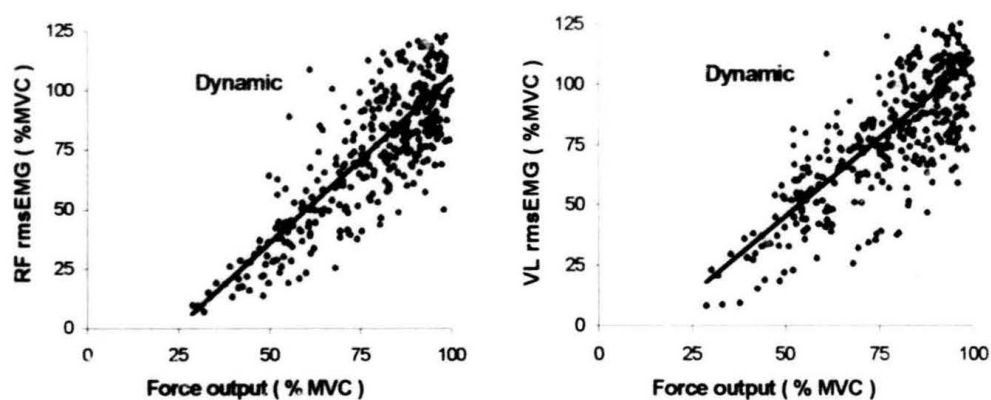


Figure 13. Relationship between force output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) during a) isometric and b) dynamic leg extensions performed on the adapted leg-only ergometer in T_{adap} . Values are normalized and expressed as a percentage of value at maximal voluntary contraction (MVC).

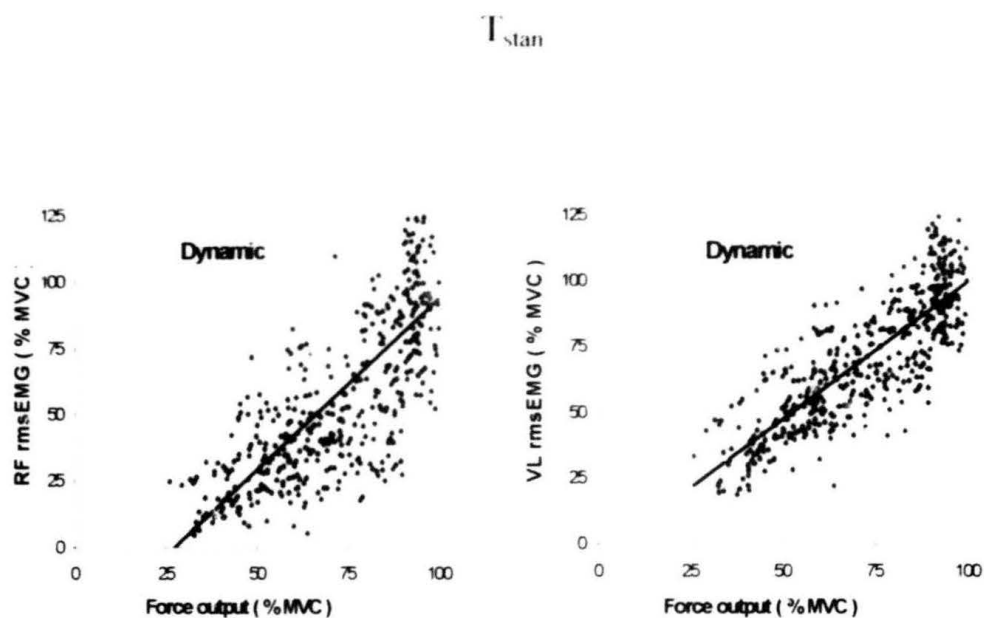


Figure 14. Relationship between force output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) during dynamic leg extensions performed on the standard rowing ergometer in T_{stan} . Values are normalized and expressed as a percentage of value at maximal voluntary contraction (MVC).

4.3.2 Force/torque-MPF

For isometric contractions in T_{cyl} , the mean torque-MPF correlation scores for RF were $r^2 = .119$ and for VL, $r^2 = .118$. There were no significant relationships found between leg extensor force and MPF (RF, VL) during isometric contractions under the most restricted movement condition. Therefore, no further analysis of MPF was completed for the isokinetic or dynamic contractions where the degree of movement was less restricted.

4.3.3 Ergometer and Muscle Effects

Means (SD) force/torque-rmsEMG scores (r^2) for subjects who performed in both isometric trials, or in all isokinetic and dynamic trials are shown in Table E1 and Table E2 respectively (Appendix E). The means (SD) displayed in Table E1 (Appendix E) and Table 1 (section 4.3) were not identical because some subjects did not complete both isometric trials and therefore, the populations used for each analysis were not identical. For similar reasons the means (SD) force/torque-rmsEMG scores (r^2) displayed in Table E2 (Appendix E) and Table 2 (section 4.3) were not identical.

For subjects ($N = 9$) who completed both isometric trials, repeated measures ANOVA showed no significant effects for ergometer type, muscle, or the interaction of muscle group with ergometer type. Comparison of the force/torque-rmsEMG relationship between the two leg extensor muscles (RF, VL) revealed similar r^2 scores for isometric conditions. For subjects ($N = 8$) who completed all three isokinetic and dynamic trials, repeated measures ANOVA showed a significant effect ($p < .05$) for ergometer type, $F(2,14) = 6.42$, $p = .01$, but no significant effect of muscle group, $F(1,7) = 0.37$, $p = .56$.

or the interaction of muscle group with ergometer type, $F(2,14) = 2.03$, $p = .17$. Post-hoc comparisons for ergometer effect during isokinetic and dynamic trials revealed that the strength of the force/torque-rmsEMG relationship was significantly lower for both T_{adap} ($p = .012$) and T_{stan} ($p = .018$) than T_{cyc} . A non-significant, but marginally stronger r^2 score occurred in VL compared to RF in T_{stan} . To determine whether there was any force or torque decline associated with the testing procedure, paired t-tests between pre-MVC and post-MVC were carried out for each subject. There was no significant force loss for any subject.

4.4 DISCUSSION

The purpose of this study was to establish force/torque-rmsEMG and force/torque-MPF relationships during isometric, isokinetic, and dynamic contractions. A further purpose was to examine the effect of ergometer type on the force/torque-EMG relationship.

4.4.1 Force/torque-rmsEMG Relationship

The hypothesis that predicted a significant relationship between random voluntary force or torque output and quadriceps rmsEMG (RF, VL) during isometric, isokinetic, and dynamic contractions was supported under all conditions. During isometric contractions of the quadriceps, an increase in rmsEMG (RF, VL) resulted in a predictable increase in force or torque output, particularly when contractions were below 80 % MVC. Importantly, the relationship was maintained during isokinetic or dynamic contractions performed on each ergometer, in spite of a number of factors which could impinge on the force/torque-EMG relationship under these conditions.

These factors include changing muscle length, changing contraction velocity, and multiple muscle contribution (discussed later in section 4.4.3).

The current findings were in agreement with other workers. For example, Esposito et al. (1996) found a significant force-rmsEMG correspondence during isometric contractions (20 - 100 % MVC) of the biceps brachii in subjects differing in age and gender. Dolan & Adams (1993) found a similar force-rmsEMG correspondence during isometric contractions of the erector spinae to those found during isometric contractions of the quadriceps in the current study. Weir, Wagner, and Housh (1992) reported linearity (r^2) of the isometric IEMG- torque relationship ranging from $r^2 = 0.77 - 0.96$ for the forearm flexors and $r^2 = 0.81 - 0.98$ for the leg extensors. These workers reported test-retest intra-class correlation coefficients at $R = 0.86$ and $R = 0.97$ for the forearm flexors and leg extensors respectively. In agreement with present findings, Christensen et al. (1995) found that force/torque-rmsEMG correspondence was similar during isometric and isokinetic concentric contractions ($7-10^\circ \text{ s}^{-1}$) of the brachial biceps. Most studies have reported strong relationships between EMG amplitude and force or torque output during isometric and during isokinetic contractions, similar to those demonstrated in the present study. Most importantly, the force-rmsEMG relationships previously reported under isometric and isokinetic conditions were similar, although stronger, to those found under rowing ergometer conditions in the current study. Hence, EMG amplitude of the leg extensors may be used to predict total force output during non-fatiguing ergometer rowing.

The current data demonstrated greater variability and reduced linearity in the force/torque-rmsEMG relationship for contractions above 80 % MVC compared to

weaker efforts. On frequent occasions, particularly at near-maximal efforts, EMG amplitude increased without concomitant increase in force or torque output. This phenomenon was described by De Luca (1997) who noted the EMG was unstable at force levels greater than 80 % MVC. Several explanations for this phenomenon have been suggested. These include inappropriate central nervous control (De Luca, 1997), the presence of agonistic co-contraction (Yang & Winter, 1983), the possible effect of fatigue (De Luca, 1985), and methodological factors (Basmajian & De Luca, 1985).

The first explanation for EMG variability was supported by De Luca et al. (1982) who reported that EMG instability at high force levels was related to central mechanisms. They noted that EMG instability increased when most motor units were recruited and additional force output was dependent on motor unit synchronization and increasing motor unit firing rates. These workers suggested that motor unit recruitment and firing rates might be inappropriate at near-maximal efforts because firing rate increased but force contribution remained constant.

Furthermore, EMG variability associated with inappropriate central drive may be related to task familiarity. Enoka and Stuart (1992) suggested that lack of practice in an activity resulted in an inadequate central drive to the appropriate motor neurons. Interestingly, greater variation and reduced force/torque-rmsEMG correspondence was found during dynamic contractions in T_{adap} compared to isokinetic contractions in T_{cyl} and dynamic contractions T_{stan} , although the differences were not significant. Subjects were more familiar with the task in T_{stan} compared to T_{adap} , and achieved a more consistent correspondence between activation of the leg extensors and force output. This may be related to the fact that motor control is influenced by familiarity of the task.

For example, Marr and Stafford (1982) found a significant difference in appropriate muscle activation patterns between well-practiced junior rowers and less-experienced novice rowers that was attributed to training and familiarity of the task. Similarly, the consistency of force profiles associated with appropriate muscle activation patterns has been used to discriminate between rowers of differing ability (Smith & Spinks, 1995). The greater force/torque-rmsEMG variation found in T_{adap} was likely associated with inappropriate fine-tuning of central nervous control required to achieve a particular target force. At contractions below 80 % MVC, the major mechanism that modulates force or torque in large powerful muscles such as the quadriceps is motor unit recruitment. However, fine tuning of force or torque output occurs by the regulation of motor unit firing frequency, which has less influence in this type of muscle at force levels below 80 % MVC (De Luca et al., 1982).

The second explanation for instability of the force/torque-rmsEMG relationship was co-contraction of agonistic and synergistic muscles that contributed to the measured force or torque output but did not contribute to the measured EMG. In T_{cyb} , agonistic and synergistic co-contraction was restricted to leg extensor muscles. However, it is likely that the hamstring muscle group had some antagonistic effect and contributed negative torque during leg extension exercises. In addition, the hamstring muscle group may have contributed to muscle activity crosstalk that was recorded in the quadriceps EMG signal. Psek and Cafarelli (1993) observed that co-activation of the hamstring muscle group (biceps femoris) increased during repeated leg extension tasks and concluded that correlation between antagonist and agonist EMG indicated a common drive. In the current study, the hamstrings antagonistic effect would have been greater in T_{adap} and T_{stan} than in T_{cyb} because this muscle group served to control the sliding movement at

the end of the leg extension drive phase. Furthermore, in I_{adap} and I_{stan} multiple synergistic muscle contributions influenced the quadriceps EMG (see section 4.4.3) and possibly increased variability of the force/torque-rmsEMG relationship.

A third possible explanation for EMG variability was fatigue. It is well known that fatigue influences the EMG characteristics of a muscle group. In the current study, comparison of pre- and post-MVC force or torque revealed no loss of force or torque and therefore, it was unlikely that effects of fatigue were responsible for the force/torque-rmsEMG variability observed.

Finally, methodological factors may affect variability of the EMG signal and include electrode positioning, skin temperature, and skin impedance. The effect of these were minimized or controlled in the current study and remained similar between trials (see section 3.1 for EMG methods). Methods of normalization may affect force/torque-EMG variability (Dolmage & Cafarelli, 1991). In the current study, it was considered appropriate that EMG data should be normalized to the MVC value of the same contraction type as recommended by Basmajian and De Luca (1985). Another methodological factor that may affect force/torque-EMG variability is crosstalk. Crosstalk from muscle fibers (other than the muscle investigated) more likely occurred during dynamic than isometric contractions, when change in muscle length altered the relative position of electrodes and MUs (De Luca, 1985; Jones & Bigland-Ritchie, 1986; Westerblad et al., 1991).

4.4.2 Force/torque-MPF Relationship

The hypothesis that predicted a significant relationship between random voluntary torque output and quadriceps MPF (RF, VL) during isometric contractions of varying intensity performed on an isokinetic dynamometer was not supported. During isometric contractions in T_{cvt} , which was the most restricted movement condition, there was no correlation between torque output and MPF of the quadriceps. Since the isometric protocol was the most controlled condition and confounding factors were minimized, analysis of the force/torque-MPF relationship was not performed for conditions when confounding factors such as changing muscle length, changing contraction velocity, and multiple muscle contribution would further reduce correspondence.

Under isometric conditions, investigators have reported contrasting force/torque-MPF correspondences. In agreement with present observations, Petrofsky and Lind (1980) found no relationship between power and MPF. In contrast, Broman et al. (1985) and Esposito et al. (1996) reported that MF increased proportionally with force and associated recruitment of more powerful fiber types. However, Mannion et al. (1998) suggested that increases in MF, which corresponded to increases in force, might be more influenced by fiber size than fiber type. The force/torque-MPF relationship is even less clear during dynamic movement when the EMG may be affected by the confounding factors.

EMG frequency characteristics of a muscle are related to action potential conduction velocity and firing rate (De Luca et al., 1982; Lindstrom, Magnusson, & Petersen, 1970). Mannion et al. (1996a) found that during isometric contractions of the erector spinae, the force/torque-MF relationship altered at different muscle lengths, possibly

reflecting a reduction in conduction velocity of the elongated muscle fibers. However, during isometric contractions in T_{cyb} in the present study, muscle length remained fixed at the displacement for optimal force for each subject and therefore, did not influence conduction velocity. The MPF provided no evidence that supported an increase in recruitment of Type II muscle fibers, or associated increase in conduction velocity, or increase in firing frequency that corresponded with increase in torque output.

4.4.3 Ergometer and Muscle Effects

The hypothesis that predicted no significant effect of ergometer type on the force/torque-rmsEMG (RF, VL) relationships during isometric contractions performed on an isokinetic dynamometer and an adapted leg-only ergometer was supported. Under isometric conditions in T_{cyb} , muscle contribution was restricted to the quadriceps muscle group, predominantly RF and VL, although other muscles of the quadriceps group likely contributed to force generation. In addition, muscle length remained fixed during the contractions, and contraction velocity was zero. Conditions for isometric contractions in T_{adap} were similar, except that the plantar flexors might have additionally contributed to force output. Under current isometric conditions, it was unlikely that muscle length, contraction velocity, and muscle contribution factors influenced the force/torque-rmsEMG relationship between ergometer types. However, under isometric conditions joint position influences the muscle biomechanics and the force/torque-rmsEMG relationship (Dolan & Adams, 1993).

The isometric angular displacement selected in T_{cyb} and T_{adap} replicated relative hip, knee, and ankle joint positions at or near dynamic peak force or torque production found on the respective equipment. Isometric force measured at this displacement position has

been suggested as an accurate predictor of force output during performance tasks (Murphy & Wilson, 1996). This was not the case in the present study as MVC peak torque or force was notably greater for isometric than for isokinetic contractions in both T_{cyb} and T_{adap} for all subjects. In T_{cyb} , the hip joint position was fixed and the ankle joint position had no influence on results because the applied torque was above the ankle. However in T_{adap} , both the hip joint and the ankle joint position were different to those in T_{cyb} and could have influenced muscle force outputs. Nevertheless, the force/torque-rmsEMG (RF, VL) relationship during isometric contractions was strong and not significantly different between ergometer types.

The hypothesis that predicted a significant effect of ergometer type during isokinetic and dynamic contractions performed in the three trials was supported. In response to these contractions, the force/torque-rmsEMG correlation was significantly greater in T_{cyb} than in T_{adap} and T_{stan} . Furthermore, the correlation in T_{stan} was stronger (diff.n.s.) than in T_{adap} . However, the significant force/torque-rmsEMG relationship found under all conditions indicated that the effects of methodological factors, which may influence variability of the EMG, were minimized. Strength of the force/torque-rmsEMG relationship during isokinetic and dynamic contractions was more likely related to a number of confounding factors that interfere, to different degrees, with the relationship for each ergometer type. These include multiple muscle contribution, change in muscle length, and varying contraction velocities. The relevance of these confounding factors to each trial is discussed.

The first confounding factor, muscle contribution, was considerably different for each ergometer. During isokinetic leg extensions performed on a dynamometer, the RF and

VL muscles are the major agonistic muscles that contribute to torque output (Solemonow et al., 1987). In T_{cyb} , muscle contribution to torque output was restricted to the leg extensor muscles. In T_{adap} , hip extensors, plantar flexors, as well as leg extensors may have contributed to the measured force output, while back extensors contributed to stabilizing forces at the trunk. Under that condition, additional synergistic muscle contribution to force output measured at the stock was restricted by the harness effect. During dynamic contractions in T_{adap} , hip angle increased during the leg extension movement, which closely replicated the muscle activity and movement pattern seen in the rowing leg drive. The movement pattern during the drive phase on the rowing ergometer replicated the on-water rowing action (Rodriguez et al., 1990). In T_{stan} , the movement pattern of the drive phase allowed complete co-ordinated synergistic muscle activation of plantar flexors, leg extensors, hip extensors, back extensors, and arm flexors to contribute to force output (Wilson et al., 1988). However, during the leg extension phase of ergometer rowing the main agonistic muscles are RF, VL, and gastrocnemius (Gauthier, 1985; Wilson et al., 1988). In movements such as rowing that involve synergistic muscle contributions, approximately halfway into the drive phase the knee moment becomes a flexion moment, which has a differential effect on the activation of both RF and VL. Furthermore, biarticular muscles such as RF are less resistant to fatigue than single joint muscles such as VL and therefore, increases in IEMG over time would be greater (Kyrolainen & Smith, 1999). The differential effect between RF and VL is apparent in T_{adap} (see Figure 11), although less obvious in T_{stan} and not present in T_{cyb} , which involved only the quadriceps group. It is evident that the dynamics of the different machines had an influence on the activation patterns of RF and VL.

It has been suggested that during high intensity rowing efforts, the leg extensors contribute the majority of propulsive force (Nelson & Widule, 1983; Smith & Spinks, 1995). The estimated contribution of the leg extensors to propulsive force output during rowing has varied (70 – 90%), partly due to the different intensities, stroke rates, and modes used for data collection (Hagerman, 1986; Rodriguez et al., 1990). In the current study, restricted leg extensor contribution to MVC force output in T_{adap} represented 85% of MVC force output from total synergistic muscle contribution in T_{stan} . However, the relatively high leg extensor contribution would have been reduced had the entire duration of force output and handle displacement for the drive phase been included in analysis. In the current study, the data-window was restricted to identify leg extension activity not including force output or displacement that occurred after full leg extension. Synergistic contribution to force output from multiple muscle groups appeared to be a factor that increased measured force output during dynamic contractions in T_{adap} and even more so in T_{stan} compared to in T_{cyb} . However, in this study multiple muscle group activation was not assessed. Only EMG of the quadriceps muscle group (RF, VL) was measured and therefore, strength of the force/torque-rmsEMG (RF, VL) relationship was affected by the different ergometer types dependent on synergistic muscle involvement for each ergometer type.

It should be noted that in T_{cyb} , the torque-rmsEMG relationship represented unilateral EMG and torque contribution from the right leg, whereas in T_{adap} and T_{stan} , although EMG was recorded from the right leg, bilateral force contributions were measured. The anticipated unilateral-bilateral influence between T_{cyb} and the other ergometers had a significant effect on force and torque outputs and possibly on rmsEMG. For example, bilateral MVC forces in T_{adap} and T_{stan} were considerably greater than right leg

unilateral MVC torque in T_{cyb} . Other workers have observed modulation of the EMG between bilateral and unilateral contractions. Vandervoort, Sale, and Moroz (1984) noted significantly less EMG activity in the dominant leg during bilateral MVCs compared to that during unilateral MVCs. These workers suggested that responses were due to lesser utilization of fast-twitch motor units during bilateral contractions. In the current study, the EMG was not compared between bilateral and unilateral contractions because this was not the purpose of the investigation, nevertheless it was possible that there was a bilateral-unilateral effect on the EMG. In spite of multiple muscle group and bilateral contributions to force output, force output at the handle and unilateral quadriceps rmsEMG were significantly related in T_{stan} , though not as strongly as in T_{cyb} .

The second confounding factor that influenced the EMG during isokinetic or dynamic contractions on each ergometer was the effect of changes in muscle length. Changes in muscle length during dynamic contractions alter the area of muscle sampled (De Luca, 1997; Mannion & Dolan, 1996a) and the action potential conduction velocity (Doud & Walsh, 1995), which influence the EMG characteristics. During isokinetic contractions in T_{cyb} , change in muscle length was controlled by the range of motion (ROM) between 0 - 90° of knee extension and remained consistent between contractions and between subjects. In T_{adap} and T_{stan} , ROM (and change in muscle length) varied between subjects dependent on biomechanical limitations and task automation for each subject. Thus, under the latter two trial conditions, change in muscle length may have influenced the EMG sample, the action potential conduction velocity, and consequently, the force/torque-rmsEMG relationship. The slopes and intercepts of the force/torque-rmsEMG relationships illustrated in Figures 12-14 were quite different. These findings probably reflect differences in muscle length and speed of muscle shortening (discussed

below) at the time of sampling under the different test conditions because both of these factors independently influence the torque-EMG relationship. However, two reasons indicate that comparisons between ergometers were valid. Firstly, changes in quadriceps muscle length were the same within subject in T_{cyb} , T_{stan} , and T_{adap} . Secondly, the effect of changes in muscle length on the EMG signal in T_{stan} and T_{adap} was minimized by data-sampling windows that were consistent within each subject. Furthermore, the trigger and duration of the data-window were designed to capture the complete leg drive phase and were similar for each contraction within each ergometer condition.

The third confounding factor to influence the force/torque relationship was variation in contraction velocities. During isokinetic contractions in T_{cyb} , leg movement velocity was constant (200°/s) and therefore, acceleration and deceleration did not effect force/torque-rmsEMG. In T_{adap} and T_{stan} , subjects were requested to maintain similar movement velocities to that in T_{cyb} . However, leg extension velocities were not constant in the former conditions and involved leg acceleration and deceleration during each contraction. Movement velocity was least controlled in T_{adap} compared to other trials, which was reflected by reduced strength of the force/torque-EMG relationship. In T_{adap} and T_{stan} , the leg extension movements more closely replicated the rowing action than leg extensions performed on the isokinetic dynamometer.

Change in contraction velocity and power output may involve change in relative use of fast-twitch muscle fibers that influence action potential conduction velocity, and thereby alter motor unit recruitment and firing rates. The effect of motor unit recruitment and firing rate changes may be increased during non-constant velocity contractions (Arendt-

Nielsen & Mills, 1988). Furthermore, changes in contraction velocity may affect stationarity of the EMG signal. As suggested by De Luca (1997), data-windows for dynamic contractions in the present investigation were of minimal duration and limited to include peak force or torque together with corresponding EMG data, thus minimizing the effect of sampling stationarity. Nonetheless, stationarity of the EMG sample may have influenced the strength of force/torque-rmsEMG relationships compared between isometric, isokinetic, and particularly non-constant velocity contractions. The reduced strength of force/torque-rmsEMG relationships during non-constant velocity contractions in T_{adap} and T_{stan} , was likely due to increased variability in motor unit recruitment and firing rates, associated with changes in action potential conduction velocity and stationarity of the EMG signal.

In the present study, the hypothesis that predicted a significant ergometer effect on the force/torque-MPF relationship was not investigated. The reason for this was the complete lack of torque-MPF correspondence found in response to the most restricted movement condition (isometric contractions in T_{cyl}). The force/torque-MPF correspondence was unlikely to improve during isokinetic or dynamic situations where confounding factors would further reduce strength of the relationship.

There was no effect of muscle (RF, VL) on the force/torque-rmsEMG relationship. Force/torque-rmsEMG correlation was similar for both muscles for each ergometer trial except for T_{stan} , where VL was stronger than RF (diff.n.s.). This finding for T_{stan} indicated a more consistent relationship between force output and VL than RF during rowing ergometry. The pattern of VL and RF muscle activation during rowing ergometer contractions reported by Wilson et al. (1988) was in agreement with the

current findings. However, these workers did not compare force-EMG relationships between muscles. The current study indicated that neuromuscular contribution to force output from the RF and VL muscles were not significantly different during isometric, isokinetic, or dynamic contractions. This finding was anticipated because muscles in the quadriceps group have been shown to act synergistically during leg extension tasks (Tesch et al., 1983). However, synergistic muscle activation (RF, VL) and EMG characteristics may differ when influenced by fatigue (Horita & Ishiko, 1987).

4.5 CONCLUSIONS

In the current study, a significant positive force/torque-rmsEMG (RF, VL) relationship was established during leg extensor contractions under isometric and isokinetic conditions. Although diminished in strength, this relationship was maintained under dynamic leg-only and standard rowing ergometer conditions. However, a significant effect for ergometer type was shown between isokinetic contractions performed on the isokinetic dynamometer and dynamic contractions performed on the leg-only and standard rowing ergometers. In contrast, the force/torque-MPF relationship was not significant during leg extensor contractions under isometric conditions. In conclusion, quadriceps muscle activation quantified by rmsEMG during non-fatiguing contractions may be used to predict force output during isometric, isokinetic, and dynamic contractions. Quadriceps muscle activity (rmsEMG of RF and VL) may be used to predict total force output measured at the ergometer handle during non-fatiguing rowing contractions. The force-rmsEMG analysis was valid for unrestricted ergometer rowing, however qualitative characteristics of the EMG (MPF) were not related to isometric torque during the most restricted condition.

CHAPTER 5

THE RELATIONSHIP BETWEEN FORCE OR TORQUE OUTPUT AND EMG OF LEG EXTENSORS DURING FATIGUING ISOKINETIC OR DYNAMIC CONTRACTIONS

5.1 INTRODUCTION

In chapter 4, significant positive relationships were demonstrated for force/torque-rmsEMG during contractions performed on an isokinetic dynamometer, a leg-only ergometer, and a standard rowing ergometer. Nevertheless, a significant effect for ergometer type was demonstrated where the force/torque-rmsEMG correlation was greater in isokinetic compared to dynamic trials. The current study aimed to determine the effect of fatigue on EMG responses, and whether EMG responses to fatigue differed between ergometer types.

Fatigue may be defined as a reduction in maximal voluntary force or power (Bigland-Ritchie & Woods, 1984) and is accompanied by characteristic changes in the EMG pattern (Bigland-Ritchie et al., 1981; DeVries, 1968). In general, during sustained isometric (Moritani et al., 1986; Oda & Moritani, 1995) and repeated isokinetic (Tesch et al., 1990) MVCs, decreases in maximal voluntary force or power are associated with decreases in rmsEMG and MPF. It is unclear whether the change in EMG observed during fatiguing isometric and isokinetic contractions, also occur in less restricted dynamic conditions. EMG responses may differ during less restricted dynamic conditions such as rowing. For example, in the previous chapter the strength of the

force/torque-rmsEMG relationship was reduced during non-fatiguing ergometer contractions compared to those found under isometric and isokinetic conditions. Furthermore, a number of authors have questioned the consistency of EMG response patterns during fatiguing dynamic contractions (Arendt-Nielsen & Sinkjaer, 1991; Merletti & Roy, 1996; Van Dieen et al., 1996; Wretling & Henriksson-Larsen, 1998). For example, Arendt-Nielsen and Sinkjaer (1991) found inconsistent rmsEMG and MPF responses between muscles and between contractions during high intensity cyclic contractions, which they attributed to inconsistent muscle activation patterns resulting from fatigue.

In order to assess the use of the force/torque-EMG relationship in rowing performance situations that involve fatigue, there is a need to systematically examine whether changes that occur under controlled isokinetic conditions, also occur under dynamic conditions when movement and muscle activation patterns are similar to those used during rowing performance tasks.

The purpose of this study was to determine the effect of fatigue on EMG responses during repeated isokinetic or dynamic contractions performed on three different ergometers. A further purpose was to determine whether EMG responses to fatigue differed between ergometer types where muscle contribution, muscle length, and shortening velocity were under different levels of control.

It was hypothesized that:

- 1) There would be a significant relationship between voluntary force or torque output and EMG (rmsEMG, MPF) of the quadriceps muscle during repeated MVCs performed on (a) an isokinetic dynamometer, (b) an adapted leg-only rowing ergometer, and (c) a standard rowing ergometer.

- 2) There would be a significant effect of ergometer type for the force/torque-EMG (rmsEMG, MPF) relationship during repeated MVCs performed on (a) an isokinetic dynamometer, (b) an adapted leg-only rowing ergometer, and (c) a standard rowing ergometer.

5.2 METHODS

5.2.1 Subjects

Subjects who participated in the preceding study were invited to take part in the current study. Male rowers ($n = 11$), mean (SD) age 28.2 (9.9) years, height 181.3 (8.4) cm, and weight 89.5 (7.8) kg with a minimum of two seasons training experience, consented to participate in the study (see Appendix B for informed consent). The University Ethical Committee gave approval for the research (see Appendix C for ethical approval). Trials performed in the same order during separate visits to the laboratory were (a) Cybex isokinetic dynamometer trial (T_{cyb}), (b) adapted leg-only ergometer trial (T_{adap}), and (c) standard rowing ergometer trial (T_{stan}). Of the subjects who participated in each trial ($T_{cyb} \ n = 8$, $T_{adap} \ n = 7$, $T_{stan} \ n = 11$) five completed all three trials.

5.2.2 Apparatus

The apparatus was identical to that described in chapter 4. Briefly, in T_{cyb} an isokinetic dynamometer (Cybex 6000) was modified to export analogue torque and angular displacement data during quadriceps leg extensions (Figure 7). In T_{adapt} , an adapted leg-only rowing ergometer was fitted with a strain gauge (between the stock and the ergometer chain) to measure linear propulsive force, together with a potentiometer to measure linear displacement of the stock (Figure 8). In T_{stan} , a standard rowing ergometer (Concept II) was modified by addition of a strain gauge and a potentiometer to measure handle force and displacement respectively (Figure 9). In all three trials, the computerized data acquisition system (Amlab system) and specifically designed schematic projects were used for analogue to digital conversion and, calibration of force and displacement data (as previously described in section 3.2). The schematic projects used in this study were (a) "tcybisok.prw" for isokinetic contractions on the Cybex dynamometer (Appendix A2), (b) "tadapt.prw" for dynamic contractions on the adapted leg-only ergometer (Appendix A3), and (c) "tstan.prw" for dynamic contractions on the standard rowing ergometer (Appendix A4). A blood lactate analyzer (Accusport) was used to measure blood lactate concentration in capillary blood samples (see section 3.3).

5.2.3 EMG

Preparation, configuration, sampling frequency, and triggered sample windows of the EMG were replicated for each trial and were similar to those for isokinetic and dynamic contractions in the preceding study (see section 4.2.3). Briefly, EMG activity was recorded from the right VL and RF muscles using bipolar surface electrodes placed over prepared sites. Impedance between the electrodes was maintained less than 10 k Ω . The EMG signal was band-pass filtered to exclude artifact and sampled at a frequency of

2000 Hz. EMG data was recorded on the computerized acquisition system and averaged by taking the rmsEMG for each 20 samples. In addition, after band-pass filtering the EMG signal was processed by a FFT each 1024 samples before calculation of the MPF. EMG data were triggered by 15° leg displacement from the vertical in T_{cyb} , by force greater than 50 N in T_{adapt} , and by positive handle velocity greater than 0.2 m/s in T_{stan} . EMG data were sampled for a 0.3-s data-window in T_{cyb} and T_{adapt} , and for a 0.5-s data-window in T_{stan} . The changes for data triggering and data-window duration between trials were necessary to facilitate the different ergometer movement patterns and to record similar data-windows of the quadriceps force/torque-time curves. Each series of data-windows were exported to a statistical package for further calculations (SPSS).

5.2.4 Procedure

Prior to each trial subjects completed a non-fatiguing warm-up (Appendix D) on a standard rowing ergometer. Impedance between EMG electrodes was recorded and motion artifact monitored. Subject positioning and movement patterns for each ergometer were similar to those in chapter 4. In order to familiarize subjects to the ergometer type, submaximal practice contractions were performed in each trial. One-minute prior to the exercise protocol, a blood sample was taken from the fleshy part of the thumb and analyzed for blood lactate concentration (see section 3.3). Visual feedback of force or torque output and verbal encouragement were provided. Subjects were asked to maintain a movement velocity and duty cycle in T_{adapt} and T_{stan} similar to that in T_{cyb} (200° s^{-1} , 2 s). Movement velocity and duty cycle values were chosen to replicate relative leg drive and recovery movement during simulated rowing performed at 30 strokes per minute. Subjects were requested to perform repeated MVCs for two

minutes in T_{cyb} and T_{stan} , and for five minutes in T_{adap} . Protocol duration in T_{cyb} was selected to allow the maximum single bout of repetitions for the dynamometer at the prescribed contraction velocity (Cybex Division of Lumex, 1991). This involved approximately 60 MVCs that were estimated to produce 30 – 40 % loss of MVC force (demonstrated in a pilot study). Protocol duration in T_{adap} and T_{stan} were selected with the aim of producing similar final perceived exertion as that in T_{cyb} determined in a pilot study. MVC force or torque was defined as the average force or torque achieved during five consecutive triggered data-windows recorded during the initial maximal efforts. Within 30 seconds of completion of the exercise task a second blood sample was taken. The subjects were allowed to warm-down on the ergometer at a self-selected intensity until five minutes post-exercise, when a third blood sample was taken and analyzed for blood lactate concentration. At this point subjects were encouraged to rest while electrodes and cables were removed. Biomechanical and EMG data were stored to disk for subsequent analysis.

5.2.5 Analysis

For each ergometer trial, computerized recordings for each subject were replayed on the data acquisition system. The series of triggered data-windows for all contractions were tagged, digitized, and exported for calculations as described previously (see section 3.4). For each data series, force or torque and corresponding rmsEMG and MPF from each five consecutive data-windows were averaged for each muscle. In order to standardize for individual variations in absolute force or torque and EMG, all averaged data were normalized to respective values at initial MVC on the same ergometer. In order to examine whether rmsEMG or MPF could predict force or torque output,

common linear regression scores (r^2) for force/torque-rmsEMG and for force/torque-MPF were calculated for each subject, for each of the three ergometers, and for each muscle. The twelve r^2 scores for each subject became dependent variables for subsequent analysis in a statistical package for social sciences (SPSS). In order to examine whether the force/torque-rmsEMG and force/torque-MPF r^2 scores were significantly different ($p < .05$) to zero, twelve one-sample t-tests were performed on the r^2 scores. In addition, for the five subjects who participated in all three trials, r^2 scores were analyzed for ergometer and muscle effects. To examine for ergometer and muscle effects on the force/torque-rmsEMG and force/torque-MPF relationships, two (3 [ergometer type: T_{cyb} , T_{adap} , T_{stan}] \times 2 [muscle: RF, VL]) fully repeated measures for Analysis of Variance were performed.

5.3 RESULTS

5.3.1 Force or Torque Loss and Final Blood Lactate

There was a significant decrease in MVC force or torque from initial values for each ergometer trial (refer to Table 3 for mean [SD] of initial MVC force or torque, final MVC force or torque, and loss in force or torque [% MVC] for three ergometer types). Mean initial MVC force or torque was lowest in T_{cyb} (range, 102.2 - 151.3 Nm) compared to T_{adap} (range, 685.1 - 847.5 N) and T_{stan} (range, 738.4 - 960.6 N). Over the course of the exercise task the greatest decrease in mean (SD) force or torque was 48.6 (7.7) % in T_{cyb} compared to 19.2 (8.7) % in T_{adap} and 18.9 (6.9) % in T_{stan} . The final strength values (% initial MVC values) were quite variable between individuals and ranged between 46.6 - 69.6 % in T_{cyb} , 66.8 - 92.7 % in T_{adap} , and 67.2 - 90.9 % in T_{stan} .

Table 3

Mean (SD) of Initial and Final Maximal Voluntary Contraction (MVC) Force or Torque Output and Final Blood Lactate in Response to Repeated MVCs for Three Ergometer Types

	Ergometer type		
	T_{cyb}^a	T_{adap}^b	T_{stan}^c
	m (SD)	m (SD)	m (SD)
Initial MVC force (N)		777.7 (50.7)	853.5 (80.2)
Initial MVC torque (Nm)	120.9 (16.4)		
Final MVC force (N)		629.3 (80.0)	689.0 (57.8)
Final MVC torque (Nm)	63.5 (11.4)		
Final force or torque (% MVC)	52.5 (7.7)	80.9 (8.7)	80.7 (6.9)
Final blood lactate (mmol/L)	7.4 (2.8)	9.7 (1.9)	13.0 (1.9)

Note. Initial and Final force or torque represented the average values of initial five and final five MVCs respectively. T_{cyb} = Cybex dynamometer trial, T_{adap} = adapted leg-only ergometer trial, T_{stan} = standard rowing ergometer trial.

$n^a = 8$. $n^b = 7$. $n^c = 11$.

Blood lactate after warm-up and prior to testing was less than 2 mmol/L for all subjects in each trial. Final mean blood lactate immediately at completion of exercise (Table 3) was lowest in T_{cyb} (range, 4.5 – 11.6 mmol/L) compared to T_{adap} (range, 7.7 - 15.5 mmol/L) and T_{stan} (range, 9.7 - 15.7 mmol/L).

5.3.2 Force/torque-rmsEMG

During repeated contractions the force/torque-rmsEMG relationships for RF and VL in T_{cyb} and for VL in T_{stan} were significant at the $p < .01$ level (refer to Table 4 for mean [SD] force/torque-rmsEMG linear regression scores [r^2] for each ergometer trial). The force/torque-rmsEMG relationship for RF in T_{stan} was significant at an alpha level of $p < .05$. However, there was no significant force/torque-rmsEMG correlation for RF or for VL in T_{adap} . Sample traces of initial and final MVC force or torque over time together with corresponding traces of EMG for VL and RF muscles are shown for T_{cyb} , T_{adap} , and T_{stan} in Figures 15, 16, and 17 respectively. The group linear trends for force/torque-rmsEMG relationships were strongest for T_{cyb} compared to T_{adap} and T_{stan} (see Figure 18 scattergraphs for force or torque plotted against rmsEMG for all subjects for each ergometer and muscle).

All subjects in each trial demonstrated decreases in MVC force or torque. However, the responses in rmsEMG were more variable than decreases in MVC force or torque. Thus, force/torque-rmsEMG relationships demonstrated negative and positive trends for different individuals and no consistent rmsEMG response to the protocols among subjects. For example, in T_{cyb} individual force/torque-rmsEMG values (r^2) for RF ranged between .11 - .81 and demonstrated 6 positive and 2 negative linear trends. For

VL, values (r^2) ranged between .16 - .89 and demonstrated 7 positive and 1 negative linear trend. In T_{adap} individual force/torque-rmsEMG values (r^2) for RF ranged between .00 - .36 and for VL between .00 - .82, neither muscle demonstrated any linear trends. In T_{stan} individual force/torque-rmsEMG values (r^2) for RF ranged between .00 - .56 and demonstrated 2 positive and 9 negative linear trends, and for VL ranged between .01 - .86 and demonstrated 3 positive and 8 negative linear trends.

Table 4

Mean (SD) of Force/torque-rmsEMG Relationships During Repeated Maximal Voluntary Contractions (MVCs) for Three Ergometer Types and Two Muscles

	Ergometer type		
	T_{cyb}^a	T_{adap}^b	T_{stan}^c
<u>Muscle</u>	<u>m</u> (<u>SD</u>)	<u>m</u> (<u>SD</u>)	<u>m</u> (<u>SD</u>)
RF	.40** (.28)	.11 (.14)	.22* (.23)
VL	.59** (.23)	.21 (.30)	.33** (.26)

Note. Values are Linear Regression “least squares” Coefficient (r^2) scores. Scores were calculated from force or torque and rmsEMG values that were averaged for each five MVCs and normalized to initial five MVCs.

rmsEMG = root mean square of the electromyogram, T_{cyb} = Cybex dynamometer trial,

T_{adap} = adapted leg-only ergometer trial, T_{stan} = standard rowing ergometer trial,

RF = rectus femoris muscle, VL = vastus lateralis muscle.

$n^a = 8$. $n^b = 7$. $n^c = 11$.

** = <.01. * = <.05.

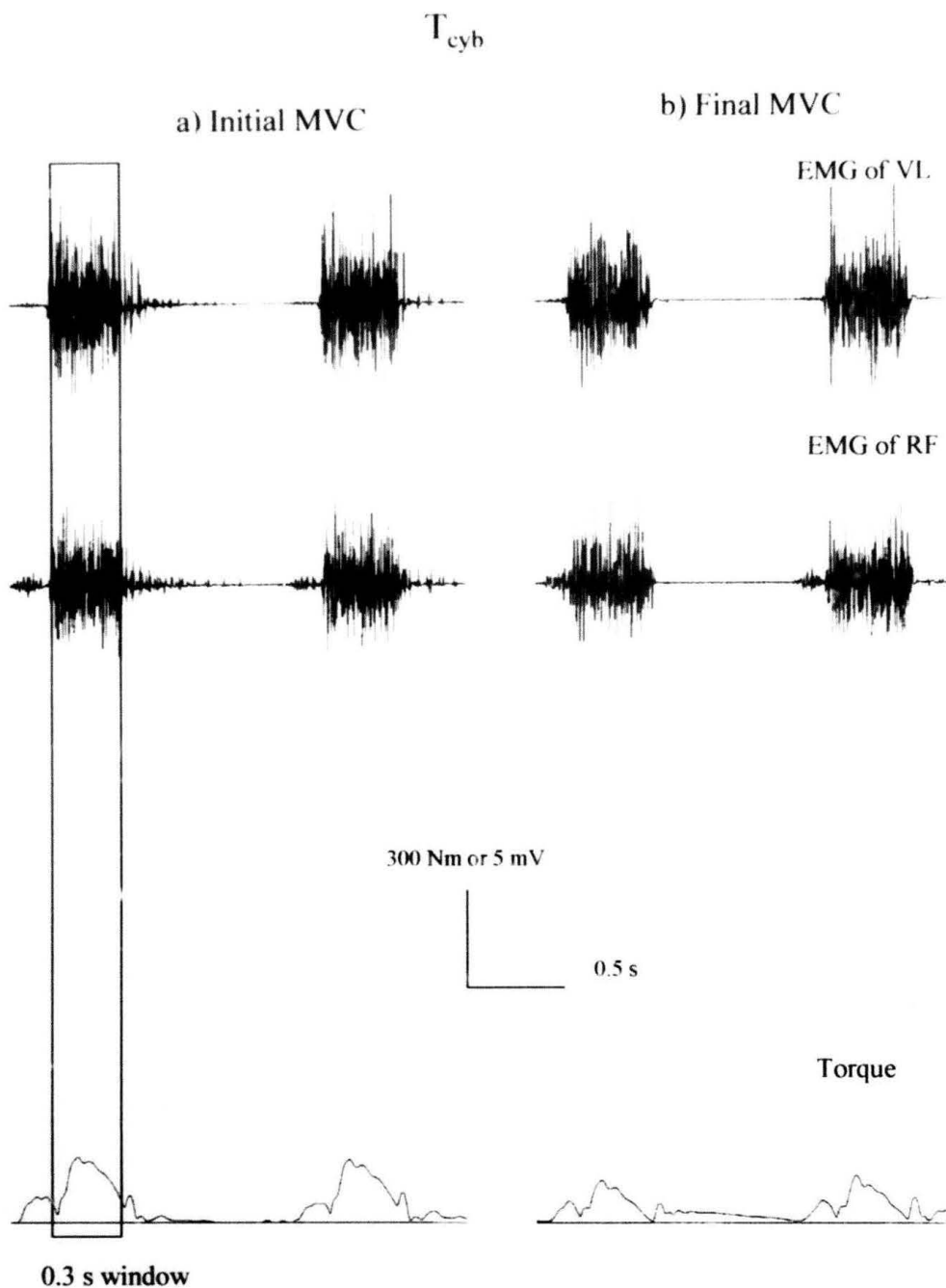


Figure 15. Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and torque for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during 2-minutes of repeated MVCs performed on the Cybex isokinetic dynamometer in T_{cyb} .

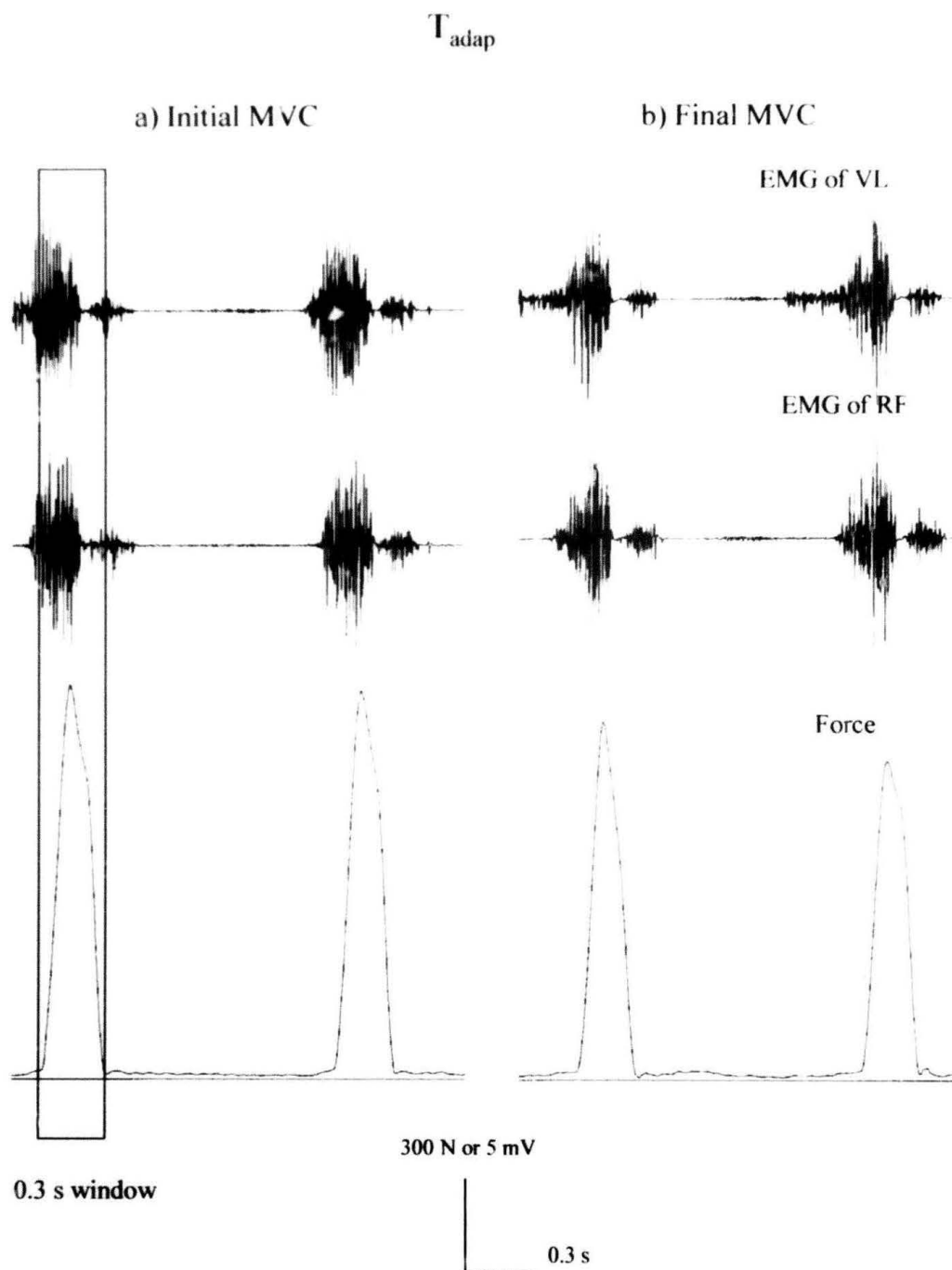


Figure 16. Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during 5-minutes of repeated MVCs performed on the adapted leg-only ergometer in T_{adap} .

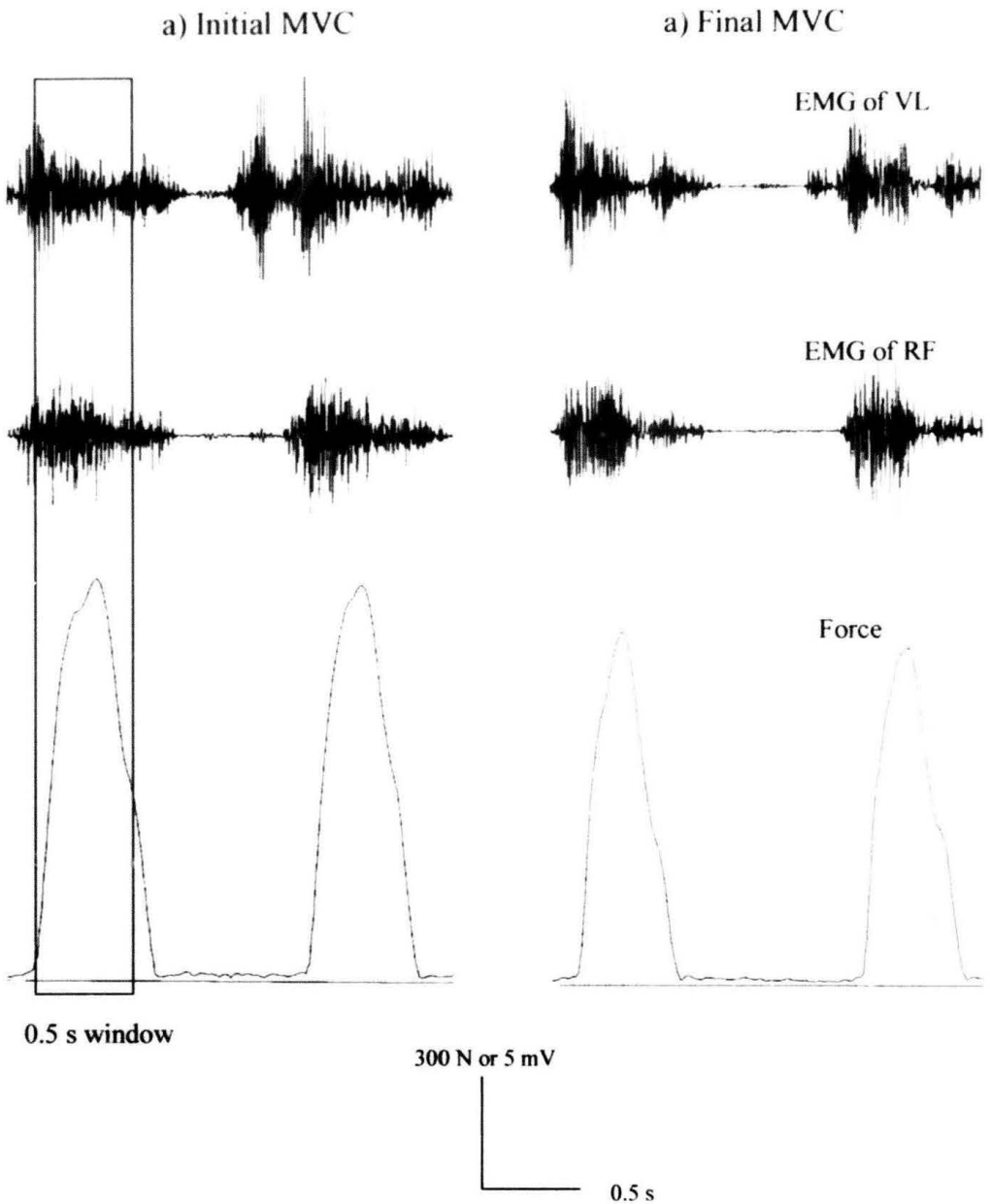
T_{stan}


Figure 17. Sample traces of EMG of vastus lateralis (VL) and rectus femoris (RF), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during 2-minutes of repeated MVCs performed on the standard rowing ergometer in T_{stan} .

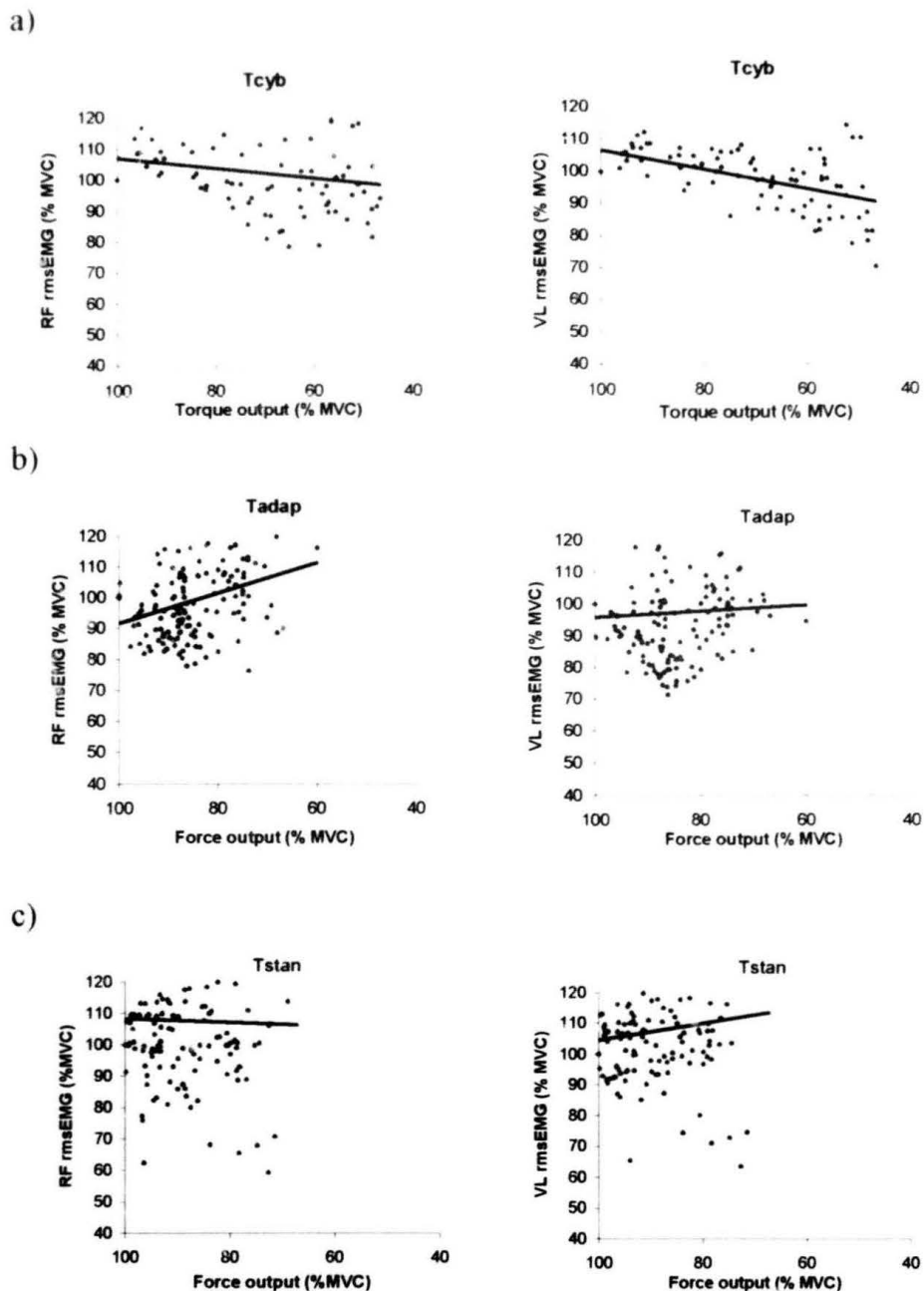


Figure 18. Relationship between torque or force output and rmsEMG of rectus femoris (RF) and vastus lateralis (VL) for all isokinetic or dynamic contractions performed on the a) Cybex dynamometer in T_{cyb} , b) adapted leg-only ergometer in T_{adapt} , and c) standard rowing ergometer in T_{stan} . Values are averaged each five contractions then normalized and expressed as a percentage of value at maximal voluntary contraction (MVC).

5.3.3 Force/torque-MPF

During repeated contractions the force/torque-MPF relationships for RF and VL in T_{cyb} and for VL in T_{stan} were significant at the alpha level of $p < .01$ (refer to Table 5 for mean [SD] force/torque-MPF linear regression scores [r^2] for each ergometer trial). The force/torque-MPF relationships for RF and VL in the remaining trials were significant at the alpha level of $p < .05$. For RF, the force/torque-MPF relationship was strong in T_{cyb} , not as strong in T_{adap} , and weak in T_{stan} . For VL, there was a significant but weaker force/torque-MPF relationship compared to RF in T_{cyb} , which was similar in T_{adap} and T_{stan} . In T_{stan} , VL demonstrated a stronger force/torque-MPF relationship compared to that for RF. The group linear trends for force/torque-MPF relationships were strongest for T_{cyb} compared to T_{adap} and T_{stan} (see Figure 19 scattergraphs for force or torque plotted against MPF for all subjects for each ergometer and muscle).

Most subjects showed a decrease in MVC MPF. In T_{cyb} , MPF decrease was 8.1 % for VL and 12 % for RF. In T_{adap} and T_{stan} , MPF decreases demonstrated similar trends. In T_{adap} , MPF decrease was 7.7 % for VL and 16.3 % for RF and in T_{stan} , MPF decrease was 10.7 % for VL and 8.7 % for RF. For most individuals, the force/torque-MPF relationship was positive although some subjects showed negative trends. Subject force/torque-MPF r^2 scores were variable in strength, as was the case for force/torque-rmsEMG r^2 scores. For example, in T_{cyb} individual force/torque-MPF values (r^2) for RF ranged between .36 - .88 and for VL, between .07 - .78 and all demonstrated positive linear trends. In T_{adap} individual force/torque-MPF values (r^2) for RF ranged between .00 - .36 and demonstrated 5 positive and 2 negative linear trends, and for VL ranged between .00 - .82 and demonstrated 4 positive and 3 negative linear trends. In T_{stan}

individual force/torque-MPF values (r^2) for RF ranged between .00 - .56 and demonstrated 7 positive and 4 negative linear trends, and for VL ranged between .01 - .86 and all demonstrated positive linear trends.

5.3.4 Ergometer and Muscle Effects

For the five subjects who performed all three trials, the force/torque-rmsEMG r^2 score in T_{cyb} was slightly stronger than in T_{stan} and substantially stronger (diff.n.s.) than in T_{adap} . The force/torque-rmsEMG r^2 score for VL was greater than for RF (refer to Appendix E Table E3 for mean [SD] force/torque-rmsEMG linear regression scores [r^2] for each ergometer trial). ANOVA showed no significant effects of ergometer type, $F(2,8) = 2.55$, $p = .14$, muscle, $F(1,4) = 7.33$, $p = .05$, or their interaction, $F(2,8) = .05$, $p = .947$ on the force/torque-rmsEMG relationship.

For the five subjects who performed all three trials, the force/torque-MPF r^2 score in T_{cyb} was slightly stronger than in T_{stan} and substantially stronger (diff.n.s.) than in T_{adap} . The force/torque-MPF r^2 score in VL was similar to that in RF (refer to Appendix E Table E4 for mean [SD] force/torque-MPF linear regression scores [r^2] for each ergometer trial). ANOVA showed no significant effects of ergometer type, $F(2,8) = 1.89$, $p = .21$, muscle, $F(1,4) = .01$, $p = .94$, or their interaction, $F(2,8) = 2.08$, $p = .19$ on the force/torque-MPF relationship. Although ANOVA revealed no ergometer effects for the five subjects who performed all three trials, a significant ergometer effect ($p = .047$) was shown in pairwise comparison for the seven subjects who performed both in T_{cyb} and in T_{adap} . The force/torque-MPF r^2 value for the seven subjects was greater in T_{cyb} than in T_{adap} .

Table 5

Mean (SD) of Force/torque-MPF Relationships During Repeated Maximal Voluntary Contractions (MVCs) for Three Ergometer Types and Two Muscles

Muscle	Ergometer type		
	T_{cyb}^a	T_{adap}^b	T_{stan}^c
	m (SD)	m (SD)	m (SD)
RF	.63** (.20)	.36* (.27)	.18* (.21)
VL	.38** (.22)	.30* (.28)	.32** (.23)

Note. Values are Linear Regression "least squares" Coefficient (r^2) scores. Scores were calculated from force or torque and MPF values that were averaged for each five MVCs and normalized to initial five MVCs.

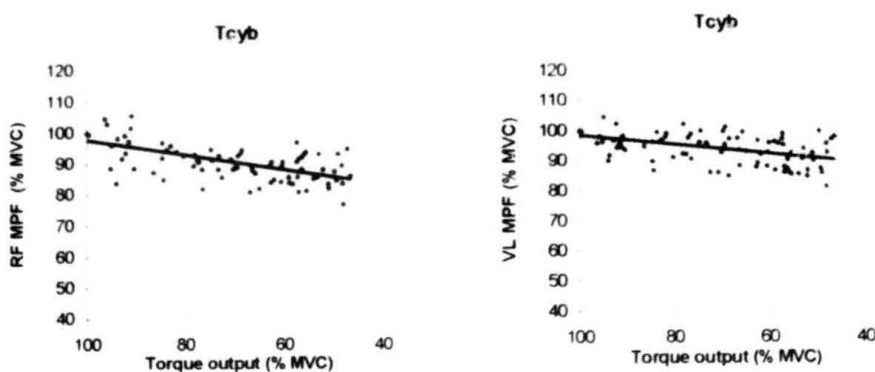
MPF = mean power frequency of the electromyogram. T_{cyb} = Cybex dynamometer trial, T_{adap} = adapted leg-only ergometer trial, T_{stan} = standard rowing ergometer trial.

RF = rectus femoris muscle. VL = vastus lateralis muscle.

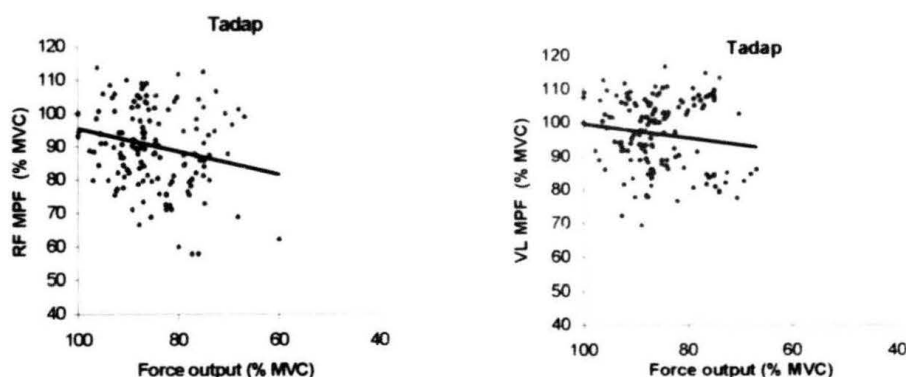
$\underline{n}^a = 8$. $\underline{n}^b = 7$. $\underline{n}^c = 11$.

** = <.01. * = <.05.

a)



b)



c)

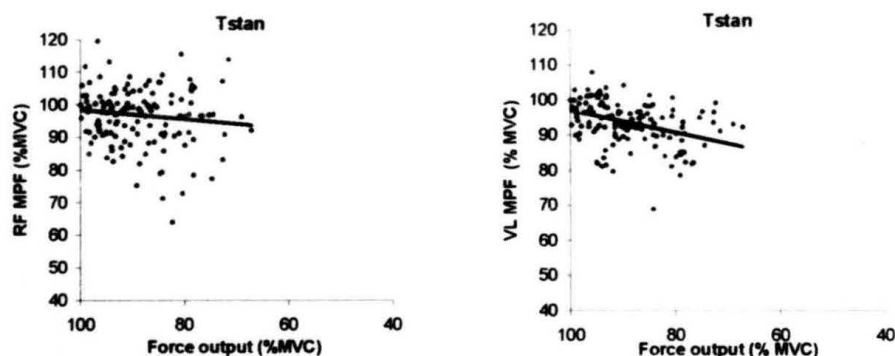


Figure 19. Relationship between torque or force output and MPF of rectus femoris (RF) and vastus lateralis (VL) for all isokinetic or dynamic contractions performed on the a) Cybex dynamometer in T_{cyb} , b) adapted leg-only ergometer in T_{adap} , and c) standard rowing ergometer in T_{stan} . Values are averaged each five contractions then normalized and expressed as a percentage of value at maximal voluntary contraction (MVC).

5.4 DISCUSSION

The purpose of this study was to determine the effect of fatigue on EMG responses during repeated isokinetic or dynamic contractions performed on three different ergometers where muscle contribution, muscle length, and shortening velocity were under different levels of control. A further purpose was to determine whether EMG responses to fatigue differed between ergometer types.

5.4.1 Force or Torque Loss and Final Blood Lactate

Strength decrease was greatest for T_{cyb} compared to T_{adap} and T_{stan} , but final blood lactate taken at completion of exercise was lowest. This was likely due to the relative muscle mass and, or the duration of exercise involved in each trial. It has been reported that blood lactate increases in relation to the muscle mass used (Westerblad et al., 1991) and the duration and intensity of exercise (Dolmage & Cafarelli, 1991). In T_{cyb} the leg extensors were used, whereas in T_{adap} (plantar flexors) and T_{stan} (plantar flexors, back extensors, arm flexors) additional muscle masses were involved. In T_{cyb} and T_{stan} the duration of exercise (2 minutes) and number of MVCs performed (60) were similar and therefore, did not influence fatigue or final blood lactate levels. However in T_{adap} , the duration of exercise (5 minutes) was greater than the other trials. It would seem that duration did not have a great effect upon blood lactate because the blood lactate concentration for T_{adap} was intermediate between the other two trials. It is more likely that lactate production and clearance reached a steady state in all the trials, which was dependent upon the relative workload for the different muscle masses involved. The increase in blood lactate levels at completion of each trial indicated considerable anaerobic involvement in each case. However, blood lactate levels were not necessarily

reflective of lactate concentration in the active quadriceps muscle (Mills & Edwards, 1984). During fatiguing contractions, the accumulation of blood lactate, hydrogen ions, and other metabolites are associated with peripheral mechanisms and contractile processes that are involved in the loss of maximal voluntary force or torque output (De Luca, 1985; Jones & Bigland-Ritchie, 1986; Westerblad et al., 1991). Under current conditions, it appears that fatigue of the quadriceps muscle group was considerably greater in T_{cyb} than in the other trials. In T_{cyb} only the leg extensors contributed to torque output and for this reason final blood lactate levels were associated with quadriceps force loss in T_{cyb} . However in T_{adap} and T_{stan} , bilateral and synergistic muscles contributed to force output and therefore, final blood lactate levels were not necessarily reflective of quadriceps force loss.

5.4.2 Force/torque-rmsEMG Relationship

The hypothesis that predicted a significant relationship would exist between voluntary force or torque output and rmsEMG of the quadriceps during 2-minutes of repeated MVCs, was supported in the isokinetic dynamometer (T_{cyb}) and standard rowing ergometer (T_{stan}) trials, but not the adapted leg-only trial (T_{adap}). The lack of force-rmsEMG relationship in T_{adap} was in part due to inconsistent muscle activation patterns associated with the unfamiliar movement and inconsistent contraction velocity demonstrated on this type of ergometer and discussed previously in chapter 4. The inconsistency of muscle activation patterns on this ergometer type was likely increased by fatigue.

In response to repeated isokinetic MVCs in T_{cyb} , decreases in both torque and EMG amplitude were well correlated for RF and for VL muscles. The trends in EMG

amplitude in T_{cyb} were in agreement with previous reports. Wretling and Henriksson-Larsen (1998) observed a decrease in EMG amplitude (VL, RF) during 150 repeated isokinetic MVCs of the quadriceps muscle and noted that both torque and rmsEMG tended to plateau after 60 - 80 contractions. The plateau was attributed to reduced influence of Type II (FT) muscle fibers and a greater reliance on Type I (ST) muscle fibers after the initial 60 MVCs, which was demonstrated by minimal decrease in rmsEMG and MPF after the plateau level. In the current study, a similar plateau was not observed during 60 MVCs in T_{cyb} . It appears that in T_{cyb} a plateau that represented the fatigue threshold level for FT fibers was not reached at completion of 60 MVCs. However, it is likely that torque and rmsEMG might have reached a plateau level had further contractions been performed.

Other authors have concurred that changes in EMG amplitude (IEMG) during fatigue might be related to fiber type (Komi & Tesch, 1979; Tesch et al., 1983). FT fibers, which fatigue quickly, are innervated by large motor neurons with fast action potential conduction velocities. In comparison ST fibers, which are more fatigue resistant, are innervated by smaller motor units with slower conduction velocities. During isokinetic leg extensions Komi and Tesch (1979) found a significant decline in quadriceps IEMG corresponding with decline in torque, which was dependent on fiber type ratios. These authors suggested that fatigue might affect MU recruitment patterns more severely in FT fibers compared to ST fibers. Hence, fiber type ratio of the predominate muscle in a task, may influence muscle activation patterns during fatigue. Trained ~~runners~~ tend to have a greater percentage of ST compared to FT fibers (Hagerman, 1986). Therefore, the effect of fatigue on muscle activation patterns for the majority of quadriceps muscle fibers (ST) may be less severe and minimize torque loss for these subjects. In addition

to fiber type ratio, the relative area occupied by specific fiber types may influence the EMG. For example, Mannion et al. (1998) examined EMG (MF) changes related to relative area of the muscle occupied by specific fiber types during fatiguing back extensor activity. They observed that the rate of MF decline was related to the relative area of the muscle occupied by Type I fibers.

In the current study, force or torque and rmsEMG might have been influenced by subject fiber type ratios or relative area of specific fiber types. However, any influence would likely have affected the force/torque-rmsEMG correspondence similarly during each ergometer condition as mostly the same subjects participated in each trial. Muscle biopsies were not taken and percentage fiber types not estimated in the current study, therefore the influence on the torque-rmsEMG relationship of fiber type ratios and areas occupied could not be established.

In contrast to current findings for T_{cyb} , the torque-EMG amplitude relationship found during isokinetic MVCs has not always been positive. For example, Tesch et al. (1990) reported that IEMG of RF and VL increased when torque declined in response to maximal isokinetic concentric leg extensions. The difference in findings may be attributed to rest periods. In the study by Tesch et al., subjects were allowed rest periods between bouts of exercise, which may have reduced fatigue of active MUs and allowed an increase in recruitment and firing rates of rested MUs. Whereas in T_{cyb} , the exercise protocol was continuous and without rest periods, except for that during passive leg flexion between each contraction.

In spite of decreases in force or torque output under each ergometer condition that was greatest in T_{cyb} , subjects demonstrated a wide range of force/torque-rmsEMG relationships. For most subjects in T_{stan} , individual force/torque-rmsEMG correlation was moderate compared to that in T_{cyb} , but most importantly, the relationship trend was opposite to that in T_{cyb} . Most subjects in T_{stan} (8 out of 11 for VL, 9 out of 11 for RF) showed an increase in rmsEMG corresponding with reduced force output. This was not the case during non-fatiguing random force contractions performed on the same ergometer in chapter 4. In that study, the force/torque-rmsEMG relationship was positive and very strong for each subject in T_{cyb} and maintained in T_{adap} and T_{stan} .

In the present trials, rmsEMG frequently increased above the value (100%) found during initial MVCs before decreasing with decline in force. Motor unit recruitment is the major mechanism that influences force or torque output in large powerful muscles such as the quadriceps at contractions below 80 % MVC. Above 80 % MVC, where force outputs of recruited motor units are near maximal, the major mechanism that influences force or torque output is increase in motor unit firing rates (De Luca et al., 1982). In the current study, all contractions involved maximal effort (> 80% MVC) and therefore, increase in MU firing-rate without concomitant force increase was most likely responsible for the increases in EMG amplitude. A similar feature was observed during random force contractions in chapter 4, when variability in the force/torque-rmsEMG relationship was greater at higher levels of force (> 80% MVC).

Researchers have used changes in EMG characteristics related to changes in force or torque in order to identify the sites of fatigue. Horita and Ishiko (1987) and Bigland-Ritchie et al. (1983) suggested that peripheral fatigue (distal to the neuromuscular

junction) was indicated when EMG amplitude increased relative to force. This criterion was used by Taylor et al. (1987) to identify peripheral fatigue in the VL muscle in response to cycle ergometry. In the current study and in accordance with the same criterion, the force/torque-rmsEMG relationship observed in T_{stan} and T_{adap} was indicative of peripheral fatigue. In T_{stan} and T_{adap} quadriceps muscle activity (rmsEMG) increased during the protocol for most subjects despite loss of force for all subjects. Thus for most subjects in T_{stan} and T_{adap} , neural input at the muscle cell membrane was maintained and loss of force was likely associated with mechanisms distal to the neuromuscular junction. This finding was in agreement with Bigland-Ritchie and Woods (1984) who suggested that increases in EMG activity relative to force output may reflect maintenance of central drive despite loss of force output associated with peripheral mechanisms involving failure of contractile processes. However, this was not the case for T_{cyb} as central drive was not maintained and rmsEMG decreased with decrease in force or torque output. Under these conditions, central fatigue in addition to peripheral fatigue may be indicated.

Horita and Ishiko (1987) stated that central fatigue (proximal to the neuromuscular junction) was indicated during MVCs when EMG amplitude decreased relative to force. In T_{cyb} but not in the other trials, the decrease in rmsEMG with decrease in torque output indicated loss of central drive. However, Tesch et al. (1983) suggested that decrease in EMG amplitude corresponding to decrease in force or torque outputs, should not be interpreted to mean that the site of fatigue was only proximal to the neuromuscular junction. Hence, in T_{cyb} both central and peripheral mechanisms of fatigue were likely to have contributed to loss of torque. Alternatively, the decrease in rmsEMG with decrease in torque output in T_{cyb} may be a response to central regulation.

Changes in EMG activity associated with fatiguing exercise may also result from a reduced motor drive in response to lack of motivation or pain, or from regulation of central drive (Bigland-Ritchie & Woods, 1984). Subjects in the present study were experienced rowers accustomed to exercise pain and capable of self-motivation in order to maintain maximal effort. Therefore, it is unlikely that decline in force or torque output was due to intolerance to pain or to lack of motivation. Bigland-Ritchie et al. (1983) postulated that parallel decreases in EMG amplitude and force might be central regulation in order to optimize force and prevent failure of neuromuscular transmission. Their explanation was that during development of fatigue maximum voluntary force declined due to contractile failure and consequently, a lower motoneuron firing rate was required to keep motor units fully activated. Thus, under fatigue conditions, central regulation and modulation of motoneuron discharge rates prevented failure of neuromuscular transmission. Central regulation may be influenced by sensory system feedback affected by peripheral processes in the muscle cell, such as changes in cellular metabolites. In the current study and in accordance with the criteria of Horita and Ishiko (1987) and Bigland-Ritchie et al. (1983), the torque-rmsEMG relationship observed in T_{cyb} may have been a response to central regulation. However, it was likely that mechanisms of central fatigue and contractile failure were also involved under that condition. For further discussion on central and peripheral mechanisms of fatigue that influence the EMG see section 2.5.2.

5.4.3 Force/torque-MPF Relationship

The hypothesis that predicted a significant relationship between voluntary force or torque output and MPF of the quadriceps during repeated MVCs was supported under each trial condition.

For the RF muscle, decrease in MPF was strongly related to decrease in torque output in T_{cyb} , but the strength of the relationship was diminished in T_{adap} and T_{stan} . The strength of the force/torque-MPF relationship for RF appeared to be associated with the level of synergistic muscle contribution to force or torque output for each ergometer type. However, synergistic muscle group contribution to force or torque output appeared to have less influence on the force/torque-MPF relationship for VL, which was similar in T_{cyb} , T_{adap} , and T_{stan} . These findings demonstrate that in T_{cyb} , muscle torque was more dependent on RF than VL as indicated by the good correspondence between MPF and torque decline. On the other hand in T_{stan} , which involved all the muscle groups employed in rowing, force output was less dependent on RF compared to VL.

Current observations of force/torque-MPF relationships in response to fatigue were consistent with those previously demonstrated during isometric (Mannion & Dolan, 1996b; Moritani et al., 1986), isokinetic (Doud & Walsh, 1995; Tesch et al., 1983), and dynamic (Bouissou et al., 1989; Potvin & Bent, 1997) contractions. Mannion and Dolan (1996b) supported the use of EMG frequency change to indicate fatigue during isometric contractions. They suggested that rate of decline (gradient) in MF and MVC force were linearly related dependent on submaximal force output during isometric contractions. Dolan, Mannion, and Adams (1995) found that increase in frequency components in the lower frequency band (5 – 30 Hz) provided the best linear regression of the EMG power spectra and the most reliable index of change during isometric contractions. Changes in frequency bands were not examined in the current study, however the saved EMG data may be replayed through the Amlab data acquisition system for frequency band analysis at a later stage.

In response to repeated isokinetic MVC's in T_{cyc} , mean torque decrease was 47.2 % and MPF decrease was 8.1 % for VL and 12 % for RF, which were similar to those found by others. Tesch et al. (1983) demonstrated a 27 % decline in torque and a decline 10 % in MPF in response to isokinetic MVC's of the leg extensors. Under similar conditions, Komi and Tesch (1979) reported a 38 – 51 % decline in torque and 12 – 25 % decline in MPF. Potvin and Bent (1997) compared MPF changes during isometric and dynamic elbow flexion using a fixed 7-kg load until volitional exhaustion. These workers found similar MPF decreases (25 – 29 % of initial MVC) for both types of contractions. However, individual correlation between isometric and dynamic change in MPF was weak, which the authors attributed to technique inconsistency during the dynamic task.

In general, investigators have reported a greater range of characteristic frequency changes during dynamic compared to isometric and isokinetic contractions. For example, Bouissou et al. (1989) found a 10 % MPF shift of VL to lower frequencies at fatigue (determined by a greater than 5 % decline in pedaling rate) in response to intense cycling at a fixed load. Lee, Minamitani, Wakano, Onishi, & Yamazaki (1996) found a MF decline of 31 – 38 % (compared to initial MVC value) of lumbar muscles during repeated dynamic trunk extensions. In T_{stan} in the current study, MPF decrease (10.7 % for VL, 8.7 % for RF) was similar to that found Bouissou et al., but considerably less than that found by Lee et al. and Potvin and Bent.

The wide range of EMG frequency characteristic changes found in response to fatigue during dynamic tasks may be attributed partly to methodological factors (see section 2.8.1) and partly to differences in variables of the task (see section 2.5.3). In spite of the wide range of responses, frequency characteristics of the EMG have been used as an

indication of fatigue and may be better related than EMG amplitude to decline in force output (Potvin and Bent, 1997). The current results demonstrated that decline in MPF provided an indication of fatigue during controlled isokinetic contractions, but a weaker indication of fatigue during less controlled performance contractions.

5.4.4 Ergometer and Muscle Effects

The hypothesis that predicted there would be a significant effect of ergometer type for force/torque-rmsEMG and force/torque-MPF relationships in response to fatiguing contractions was not supported. For the five subjects who completed all three trials there was no significant effect of ergometer type. However, correlation for force/torque-rmsEMG and for force/torque-MPF was noticeably better for T_{cyb} than for the other two conditions. The lack of significant ergometer effect during fatiguing contractions was in contrast to previous findings (chapter 4) for non-fatiguing contractions performed on the same ergometers. In the current study, ergometer effect may have been influenced by the small number of subjects (five) who participated in all three trials. Furthermore, when ergometer effect was examined for the seven subjects who completed both T_{cyb} and T_{adap} , a significant difference was found between ergometer trials for force/torque-MPF, although not for force/torque-rmsEMG. The noticeable but non-significant difference in the force/torque-EMG (rmsEMG, MPF) relationship between the three ergometer types, may be partly explained by the influence of muscle length, multiple muscle contribution, and muscle contraction velocity discussed previously (see section 4.4.3).

In the current study, for each subject changes in quadriceps muscle length during each contraction remained similar during the course of each trial. Minor reductions in

muscle length changes were not included in the restricted data-window and therefore, did not influence the analysis.

Muscle contributions were different for each ergometer trial. In T_{cyb} , torque contribution was restricted to knee extensor muscles and decline in torque output was strongly related to changes EMG activity of RF and VL. This was not the case for the other ergometers where additional muscles contributed to force output (plantar flexors in T_{adap} and plantar flexors, back extensors, arm flexors in T_{stan}). In comparing pre-MVC force between T_{adap} and T_{stan} , subjects produced on average an extra 15 % force output when using the back extensors and arm flexors in addition to the leg extensors during the latter trial. However, values were calculated for restricted data-windows that did not necessarily include all force output during the contraction and excluded some force contribution from the back extensors and arm flexors. Nonetheless, 85 % of the propulsive force measured at the handle during ergometer rowing was a result of bilateral leg extension. Most authors agree that the quadriceps muscle group provides the majority of the drive force during the rowing movement (Hagerman, 1996; Secher, 1993). During the rowing ergometer task in T_{stan} , it was likely that synergistic muscle activation was increased in order to compensate for reduced quadriceps muscle contribution resulting from fatigue. Hence, the force/torque-rmsEMG (RF, VL) relationship may have been different between trials due to synergistic muscle contribution that helped maintain measured force output in T_{adap} and T_{stan} , which was not available in T_{cyb} . One purpose of the current study was to quantify quadriceps muscle activation that contributed to total force output at the handle. However, force contribution and EMG activation specific to other muscle groups involved in the rowing

movement, could be examined by additional force transducers and EMG recordings from different muscle groups.

In the present study, subjects were able to achieve more consistent movement velocity in T_{cyb} compared to T_{stan} and T_{adap} . This was partly achieved by experimentally controlled limb velocity, duty cycle, and ROM in the single joint movement in T_{cyb} . In T_{stan} and T_{adap} , the movement task required co-ordinated muscle action and self-pace control over contraction velocity, duty cycle, and ROM. Change in muscle shortening velocity is associated with change in action potential conduction velocity that influences the MPF (Arendt-Nielsen & Mills, 1988). Therefore, it was likely that changes in movement velocity T_{stan} and T_{adap} influenced the MPF.

The timing of quadriceps muscle activity relative to measured force or torque output may have been different on each ergometer. For example, in T_{stan} the correspondence of handle force and quadriceps EMG may have been affected by the timing of synergistic muscle group activation or by fatigue in specific muscle groups. These factors would have affected the relationship between measured quadriceps EMG activity and measured force output more in T_{stan} than in T_{cyb} , due to involvement of multiple muscle groups in T_{stan} . In T_{stan} , the timing of co-ordinated movement may have affected muscle activation patterns, EMG data-window sampling, and strength of the force/torque-rmsEMG relationship. For example, the quadriceps muscle group was activated in order to stop the slide forward at the end of the recovery phase, but did not necessarily contribute to propulsive force output. Furthermore, activation of the quadriceps muscle group may have continued as an isometric contraction at completion of the dynamic drive phase, while the back extensors and arm flexors continued to contribute to

propulsive force output. Evidence of quadriceps muscle activation and force output outside the recorded data-window was shown by the EMG and force trace in Figure 17. In T_{stan} , EMG measurement was triggered by handle velocity (greater than 0.2 m/s) that did not necessarily coincide with quadriceps muscle activation. Quadriceps activation may have occurred before and, or after the sampled triggered data-window. Therefore, the measured data-window of quadriceps activity during each contraction was representative of muscle activity at that time of force output during positive handle movement, but was less representative of total quadriceps activation.

For some subjects in the rowing ergometer trial, timing and synchronization of quadriceps activity with force output appeared to deteriorate with decline in maximal voluntary force. Visual inspection of the EMG and force trace in T_{stan} revealed minor alterations in timing of quadriceps activation from initial to final MVCs, most likely in response to fatigue. Furthermore, the differential effect of knee moment changes during the drive phase on the activation of RF and VL (discussed 4.4.3) may have increased due to fatigue in the different muscle fibers. Hence, synchronization of force output and quadriceps EMG activity during fatigue may have influenced the force/torque-EMG correlation. Arendt-Nielsen and Sinkjaer (1991) found changes in rmsEMG and duration of EMG activity for VL during uphill walking, similar to those found in T_{stan} . These authors reported that co-ordination of movement patterns and muscle performance were disturbed by fatigue. They suggested that to maintain the required movement pattern during fatigue, the central nervous system might control individual and co-ordinated muscle motor recruitment in order to optimize neuromuscular activity.

The influence of timing and duration of EMG activity was not analyzed in the current study. Furthermore, the possible effect of electromechanical delay (discussed 2.8.3) may have influenced the measurements because electrical activation precedes force development by about 80 to 100msec depending upon the muscle (Gabriel & Boucher, 1998). However, the findings were valid because the study objective was to measure quadriceps muscle activity relative to total force output measured at the handle during the major force-producing phase. In future studies it would be interesting to address timing aspects by measuring total muscle activation using a muscle activation threshold trigger to record EMG and corresponding force output (see chapter 8).

No muscle effect on the force/torque-rmsEMG relationship was demonstrated. VL showed a stronger torque-rmsEMG relationship than RF in T_{cyb} and both muscles demonstrated a significant parallel decrease in rmsEMG and torque output. In T_{adap} , the force/torque-rmsEMG relationship was weak and non-significant for both muscles. This was likely due to inconsistent activation patterns associated with the less familiar movement. In T_{stan} , the decrease in MVC force was accompanied by an increase in rmsEMG for VL and a slight decrease for RF. No muscle effect on the force/torque-MPF relationship was demonstrated. The similar decreases in MPF for RF and VL in response to isokinetic MVCs in T_{cyb} , were in agreement to those reported by Tesch et al.(1990). Current observations indicated that RF fatigued (expressed by MPF decrease) more quickly than VL (diff.n.s.) during isokinetic leg extensions, but VL fatigued more quickly than RF (diff.n.s.) during dynamic ergometer rowing. The MPF responses for each muscle may be related to different movement patterns on each ergometer, which influence the agonistic and synergistic force output demands on each muscle. Furthermore, the fiber type ratio may influence RF and VL differently. The

proportions of FT and ST fibers in the RF and VL muscles have not been established. However, the RF muscle tends to have a higher proportion of FT compared to ST fibers, which is reversed in the VL muscle (Chwalbinska-Moneta, Hanninen, & Penttila, 1994). Hence, it would be expected that RF would fatigue more quickly than VL resulting in greater EMG amplitude decrease in RF compared to VL during repeated MVCs. There was no evidence in the present study that fatigue, expressed by rmsEMG or MPF decreases, was significantly greater in the RF compared to the VL muscle.

5.5 CONCLUSIONS

There was a significant positive relationship between decrease in MVC torque and decrease in EMG (rmsEMG, MPF) of the quadriceps muscle (RF, VL) during repeated isokinetic MVCs performed on a dynamometer. This finding implicated loss of central drive and involvement of peripheral fatigue. There was a significant relationship between MVC force and EMG (rmsEMG, MPF) on a standard rowing ergometer, although for most subjects the association between force and rmsEMG revealed opposite trends that indicated maintenance of central drive or central regulation, as well as peripheral fatigue. Subjects exhibited a wide range of force/torque-rmsEMG and force/torque-MPF correlation scores, which were least in T_{cyb} . There were no significant effects of ergometer type on force/torque-rmsEMG or force/torque-MPF relationships, although correlation was stronger in the isokinetic trial compared to the other ergometer trials. It can be concluded that during fatiguing contractions, the change in neuromuscular activation was more consistent in controlled isokinetic situations compared to performance situations and EMG responses to fatigue varied between subjects. Furthermore, mechanisms and sites of fatigue may be different

during less-restricted dynamic rowing compared to isokinetic tasks. Nevertheless, force-rmsEMG and force-MPF relationships were valid for fatiguing rowing ergometer performance. It would be interesting to investigate if EMG responses to repeated MVCs were the same in a typical self-pace rowing performance.

CHAPTER 6

COMPARISON OF RESPONSES IN FORCE AND EMG BETWEEN A REPEATED MVC PROTOCOL AND A SELF-PACE PROTOCOL PERFORMED ON A ROWING ERGOMETER

6.1 INTRODUCTION

EMG changes during fatiguing exercise are often examined using repeated isokinetic MVC tasks. In the previous study during a 2-minute repeated isokinetic MVC task, decreases in rmsEMG and MPF for RF and VL muscles were associated with decreases in torque output. This was consistent with the observations of others under similar task conditions (Bigland-Ritchie & Woods, 1984; Horita & Ishiko, 1987). To the contrary, in response to the same protocol performed on a rowing ergometer, eight out of eleven subjects demonstrated increases in rmsEMG of VL associated with decreases in force output. The eight subjects were able to fully maintain VL muscle activation for the duration of that protocol. However, in a longer duration effort such as a 6-minute rowing performance, it is unlikely that maximal force output or muscle activation could be maintained. In order to optimize muscle activation and force output under these conditions, contractions are often submaximal and the effort paced. During submaximal contractions, increases in both rmsEMG (Taylor & Bronks, 1994) and MPF (Jansen et al., 1997) have been reported. However, in performance situations when force demands are near maximal, contractile function and central drive may decrease with duration of effort. This may be indicated by decline in EMG amplitude (Bigland-Ritchie & Woods 1984). When the ability to maintain force output is impaired, decreases in rmsEMG and

MPF are dependent on intensity and duration of the contractions (Gamet, Duchene, & Goubel, 1996). It has been pointed out that EMG responses to fatigue are specific to the performance task and dependent on variables of that task (e.g. intensity, duration) (Dolmage & Cafarelli, 1991; Enoka & Stuart, 1992).

For changes in EMG to be useful in analysis of rowing performance, conditions that replicate a typical rowing task, such as pattern of muscle activation, contraction velocity, contraction intensity, and duration of effort, should be examined. A typical strategy during a 2000-m (approximately 6-minute) rowing performance requires maximal effort at the start and finish with optimal self-pace submaximal efforts for the remainder of the event (Hagerman, 1986; Hahn et al., 1995b). Fatigue related changes in muscle activation patterns resulting from a typical fatigue protocol (2-minute repeated MVCs) may differ to those resulting from a typical rowing performance (6-minute self-pace contractions) in which intensity and duration are not the same.

The purpose of this study was to compare force output and EMG responses between a typical repeated MVC protocol and a typical self-pace rowing strategy performed on a standard rowing ergometer.

It was hypothesized that:

There would be a significantly greater decrease in final MVC rmsEMG and MPF of VL in response to a 6-minute optimal self-pace protocol compared to a 2-minute repeated MVC protocol performed on a standard rowing ergometer.

6.2 METHODS

6.2.1 Subjects

Subjects who participated in previous studies (chapters 4, 5) were invited to take part in the present study. Trained male rowers ($n = 8$) with a minimum of two seasons training experience, mean (SD) age 28.3 (9.5) years, height 184.9 (6.1) cm, and weight 91.3 (7.5) kg gave consent. Subjects performed rowing exercises at each of two ergometer trials (T_{2-min} , T_{6-min}) which were conducted approximately 2-weeks apart during separate visits to the laboratory.

6.2.2 Apparatus

The equipment was identical to that described in chapter 4. Briefly, a standard Concept II rowing ergometer was modified by addition of a strain gauge and potentiometer to measure handle force and displacement respectively (Figure 9). The computerized data acquisition system (Amlab systems) and specifically designed project "tstan2.prw" (Appendix A5) were used for analogue to digital conversion and calibration of force and displacement data.

6.2.3 EMG

In the previous study (chapter 5), no muscle effect was demonstrated in the rowing ergometer trial, although VL showed a stronger relationship than RF, with force output. In the present study, it was considered that the VL muscle would best represent quadriceps activation during rowing ergometry and therefore, EMG of the RF muscle was not analyzed. EMG configuration and preparation for VL was replicated between T_{2-min} and T_{6-min} and was the same as the previous study in T_{stan} with one exception. The

data collection was triggered by positive handle force (greater than 50 N) which was shown more accurate than handle velocity (previously used) in relation to quadriceps activation during ergometer rowing. The preferred trigger for this ergometer type was established during pilot trials. In brief, impedance between EMG electrodes was recorded and motion artifact monitored. EMG data collection was sampled at a frequency of 2000 Hz for a 0.5-s data-window (1000 data-points). For each triggered data-window rmsEMG was calculated after the EMG signal was processed (refer to section 3.1.3). In addition, average handle force and corresponding rmsEMG activity (VL) for each triggered data-window were expressed as a ratio (force/rmsEMG) that represented neuromuscular efficiency of the VL muscle. MPF for the triggered data-window was calculated after the EMG signal was passed through a FFT for 1024 samples.

6.2.4 Procedure

Prior to each trial, subjects completed a non-fatiguing warm-up (Appendix D) on a standard rowing ergometer. One-minute prior to starting the exercise protocol, a blood sample was taken from the fleshy part of the thumb (see section 3.3 for blood lactate measurement) and analyzed for blood lactate concentration (Accusport). Visual feedback of power output was provided from the ergometer readout. In the first trial (T_{2-min}), subjects were requested to perform 2-minutes of repeated maximal rowing strokes on a standard rowing ergometer. Verbal encouragement was provided during all contractions. Within 30-s of completion a second blood sample was taken and performance distance recorded from the ergometer display unit. Subjects were allowed to warm-down on the ergometer at a comfortable self-selected pace until five minutes post-exercise, when a third blood sample was taken and analyzed for blood lactate

concentration. In the second trial ($T_{6\text{-min}}$), subjects were requested to perform an optimal self-pace effort with the objective being to obtain the greatest distance in a 5-minute period, including five initial and five final maximal efforts. In each trial, initial and final MVC force was defined as the mean value of five contractions.

6.2.5 Analysis

Computerized recordings were replayed on the data acquisition system with the initial five and final five triggered data-windows of the leg extension phase tagged, digitized, exported to a statistical package (Microsoft Excel), and then averaged for each data-window duration. Subsequently, the five initial and the five final consecutive data-windows were further averaged to provide a mean MVC value for initial five and for final five consecutive contractions. In both trials mean initial MVC value served as a reference value for normalization. Normalized final MVC values became dependent variables for subsequent analysis in a statistical package for social sciences (SPSS). Performance distance and blood lactate value (recorded at completion of exercise) were dependent variables and examined in absolute terms. The dependent variables were examined for significant difference ($p < .05$) between trials ($T_{2\text{-min}}$, $T_{6\text{-min}}$) using paired samples t -tests.

6.3 RESULTS

6.3.1 Performance Distance and Final Blood Lactate

Initial mean [SD] blood lactate values were similar in $T_{2\text{-min}}$ (3.0 [0.4] mmol/L) and in $T_{6\text{-min}}$ (2.9 [0.3] mmol/L). All subjects demonstrated a significant increase in blood lactate concentration resulting in final values that were always greater in $T_{6\text{-min}}$ (range,

12.2 – 17.2 mmol/L) compared to T_{2-min} (range, 8.9 – 14.8 mmol/L). Paired samples t -tests revealed a significant but small systematic difference between trials for final blood lactate, $t(7) = -4.169$, $p = .004$. Means (SD) performance distance and blood lactate at completion of exercise for each trial are shown in Table 6.

6.3.2 Force Loss

Initial mean (SD) MVC force output values were similar in T_{2-min} (871.5 [69.9] N) and in T_{6-min} (853.5 [80.2] N). All subjects demonstrated a significant decrease in final MVC force in T_{2-min} (range, 72.7 – 86.9 % initial MVC) and in T_{6-min} (range, 67.8 – 81.8 % initial MVC). Five out of the eight subjects had greater force losses in T_{6-min} compared to T_{2-min} . Mean force loss was greater in T_{6-min} compared to T_{2-min} but the difference was not significant, $t(7) = 2.118$, $p = .072$. Means (SD) for final force output for each trial are shown Table 6. Sample force traces are shown for T_{2-min} and T_{6-min} in Figures 20 and 21 respectively.

6.3.3 Final MVC rmsEMG

Paired samples t -tests revealed a significant difference between trials for final MVC rmsEMG, $t(7) = 3.378$, $p = .012$. Final MVC rmsEMG of VL demonstrated a small increase from initial MVC in T_{2-min} but a large decrease in T_{6-min} . Means (SD) for final rmsEMG for each trial are shown Table 6. Subjects demonstrated considerable variation in final rmsEMG responses for T_{2-min} (range, 63.5 – 123.7 % initial MVC) and for T_{6-min} (range, 43.7 – 117.5 % initial MVC). The subject who showed the greatest decrease for final rmsEMG in T_{2-min} showed the greatest decrease in T_{6-min} . Six of the eight subjects demonstrated a greater decrease in final rmsEMG in T_{6-min} compared to in

T_{2-min} . Sample traces of EMG and rectified EMG of VL from initial and final MVCs are shown for T_{2-min} and T_{6-min} in Figures 20 and 21 respectively.

There was a significant difference between trials for the force/rmsEMG ratio change, $t(7) = -2.481$, $p = .042$. The final force/rmsEMG ratio (% initial MVC) of VL decreased (80.1 %) from initial value in T_{2-min} and increased (101.6 %) from initial value in T_{6-min} . The force-rmsEMG ratio change in each trial demonstrated considerable subject variation, which ranged between 68.5 – 114.4 % in T_{2-min} and 57.7 – 171.0 % in T_{6-min} . The subject who showed the greatest increase for the force/rmsEMG ratio in T_{2-min} showed the greatest increase in T_{6-min} . This subject differed from all others in both trials. The force/rmsEMG ratio change compared between trials was more influenced by change in rmsEMG than change in force.

6.3.4 Final MVC MPF

Final MVC MPF demonstrated a decrease from initial MVC in both trials, which was greater in T_{2-min} compared to T_{6-min} . However, there was no significant difference between trials for final MVC MPF, $t(7) = -2.257$, $p = .059$. Means (SD) for final MPF for each trial are shown in Table 6. Seven of the eight subjects showed a greater decrease in MPF in T_{2-min} compared to T_{6-min} . The range in final MVC MPF was 68.9 – 94.9 % of initial MVC in T_{2-min} and 83.6 – 107.3 % of initial MVC in T_{6-min} . All subjects in T_{2-min} and seven out of eight subjects in T_{6-min} demonstrated a decrease in final MVC MPF. Sample traces of EMG power frequency spectrum of VL from initial and final MVCs are shown for T_{2-min} and T_{6-min} in Figure 22.

Table 6

Comparison of Responses Between a Repeated Maximal Voluntary Contraction (MVC) Protocol and a Self-pace Protocol Performed on a Rowing Ergometer (N = 8)

	Ergometer Trial			
	T _{2-min}	T _{6-min}	T _{2-min} - T _{6-min}	
	m (SD)	m (SD)	m	(SD)
Performance distance (m)	688.4 (40.5)	1787.5 (89.0)	-879.3**	(467.2)
Final Blood lactate (mmo/L)	12.6 (1.9)	14.7 (1.5)	-2.1**	(1.4)
Final force (% MVC)	80.8 (5.6)	76.6 (4.5)	4.3	(5.7)
Final rmsEMG VL (% MVC)	103.8 (18.2)	81.2 (21.4)	22.6*	(18.9)
Final MPF VL (% MVC)	88.0 (8.8)	93.7 (6.7)	-5.7	(7.1)

Note. Data are presented as mean (SD) for trials (T_{2-min}, T_{6-min}) and as mean difference (T_{2-min} - T_{6-min}) between trials. Performance distance and blood lactate at completion of exercise represent absolute values. Data represent the average of final five MVC values normalized to initial five MVC values. T_{2-min} = 2 minute repeated MVC trial, T_{6-min} = 6 minute self-pace maximal effort trial, rmsEMG = root mean square of the electromyogram, MPF = mean power frequency of the electromyogram, VL = vastus lateralis muscle.

* = $p < .05$. ** = $p < .01$.

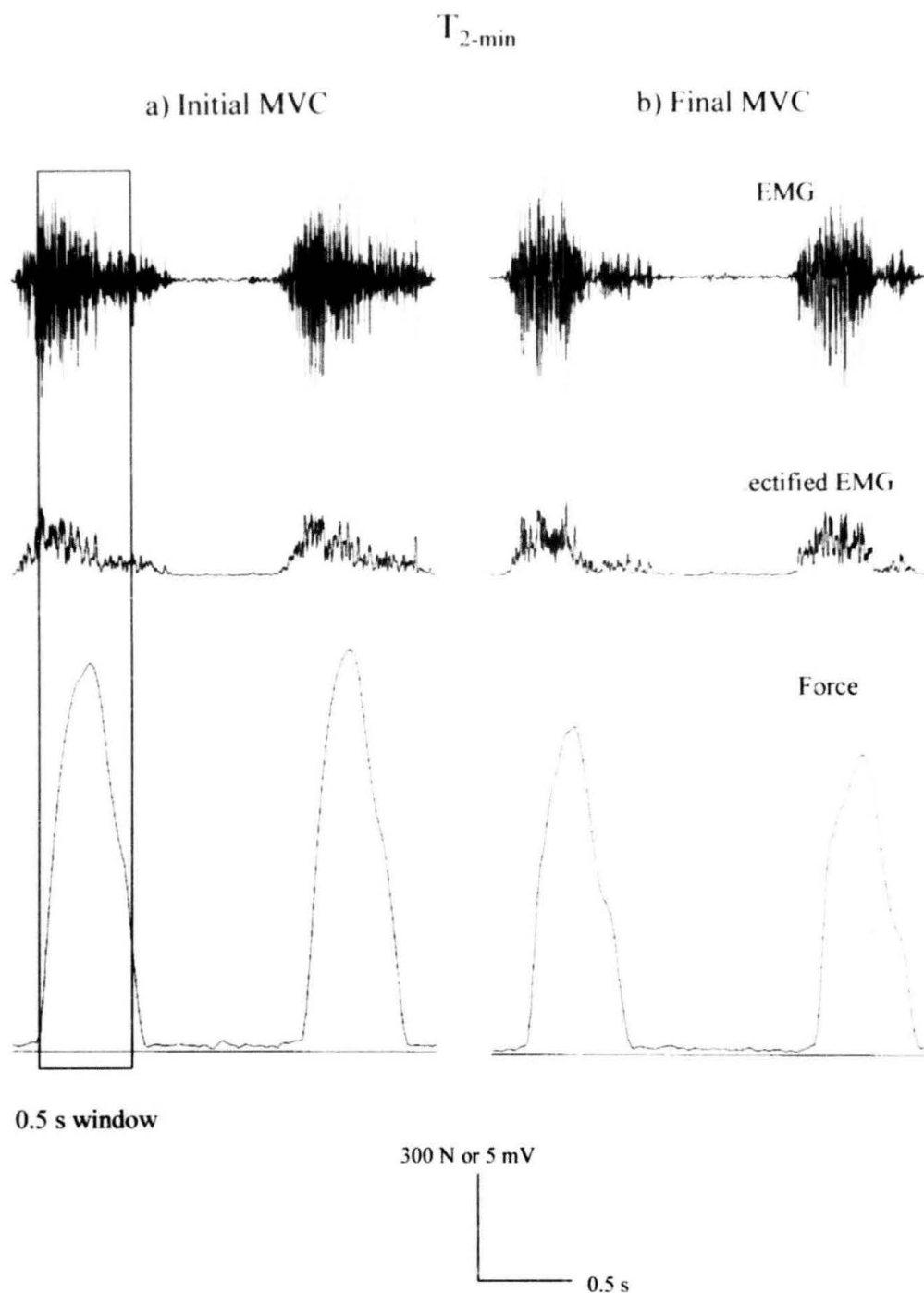


Figure 20. Sample traces of EMG and rectified EMG of vastus lateralis (VL), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 2-minute repeated MVC effort performed on a rowing ergometer in $T_{2\text{-min}}$.

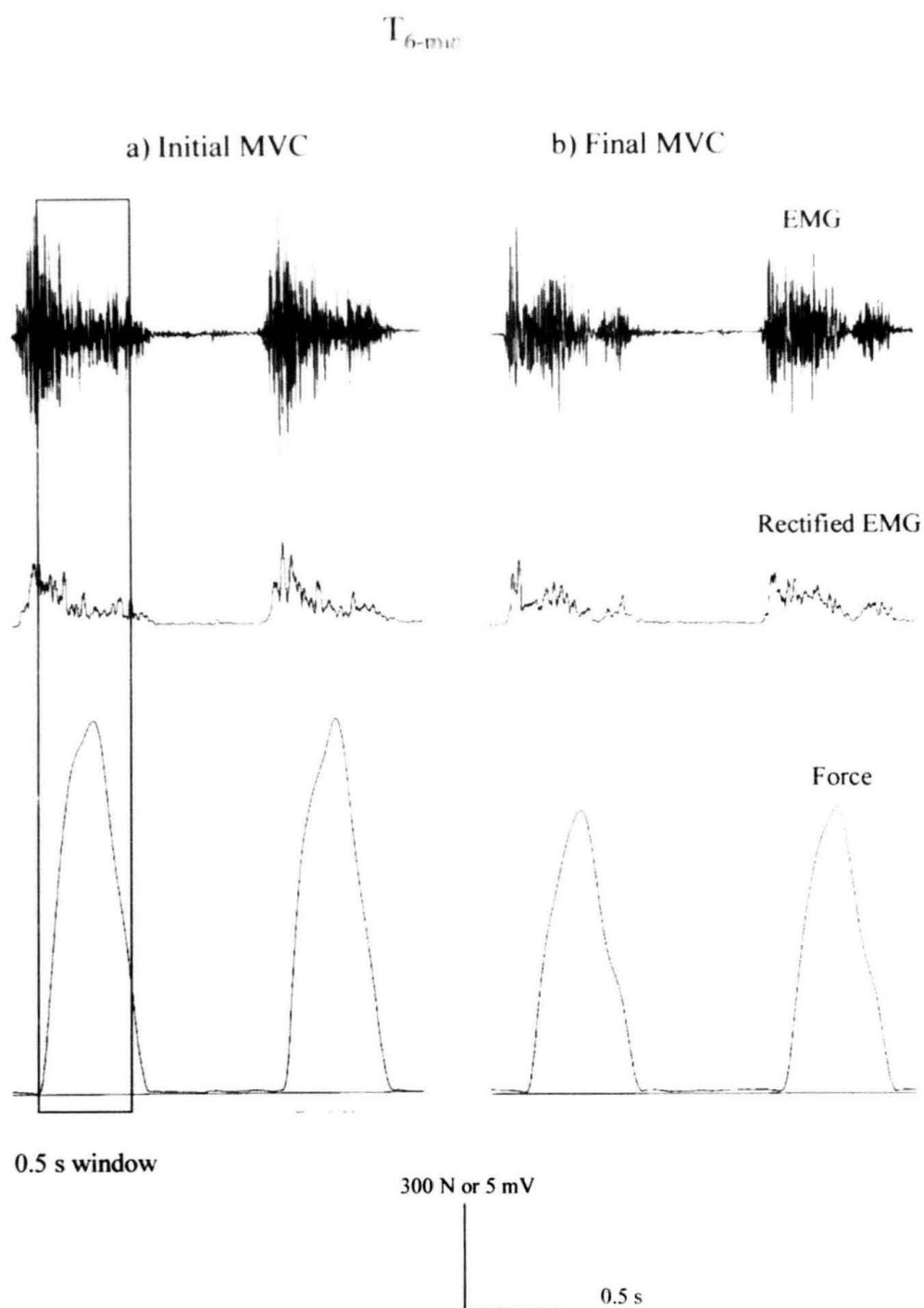


Figure 21. Sample traces of EMG and rectified EMG of vastus lateralis (VL), and force for the same subject recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-paced effort performed on a rowing ergometer in $T_{6\text{-min}}$.

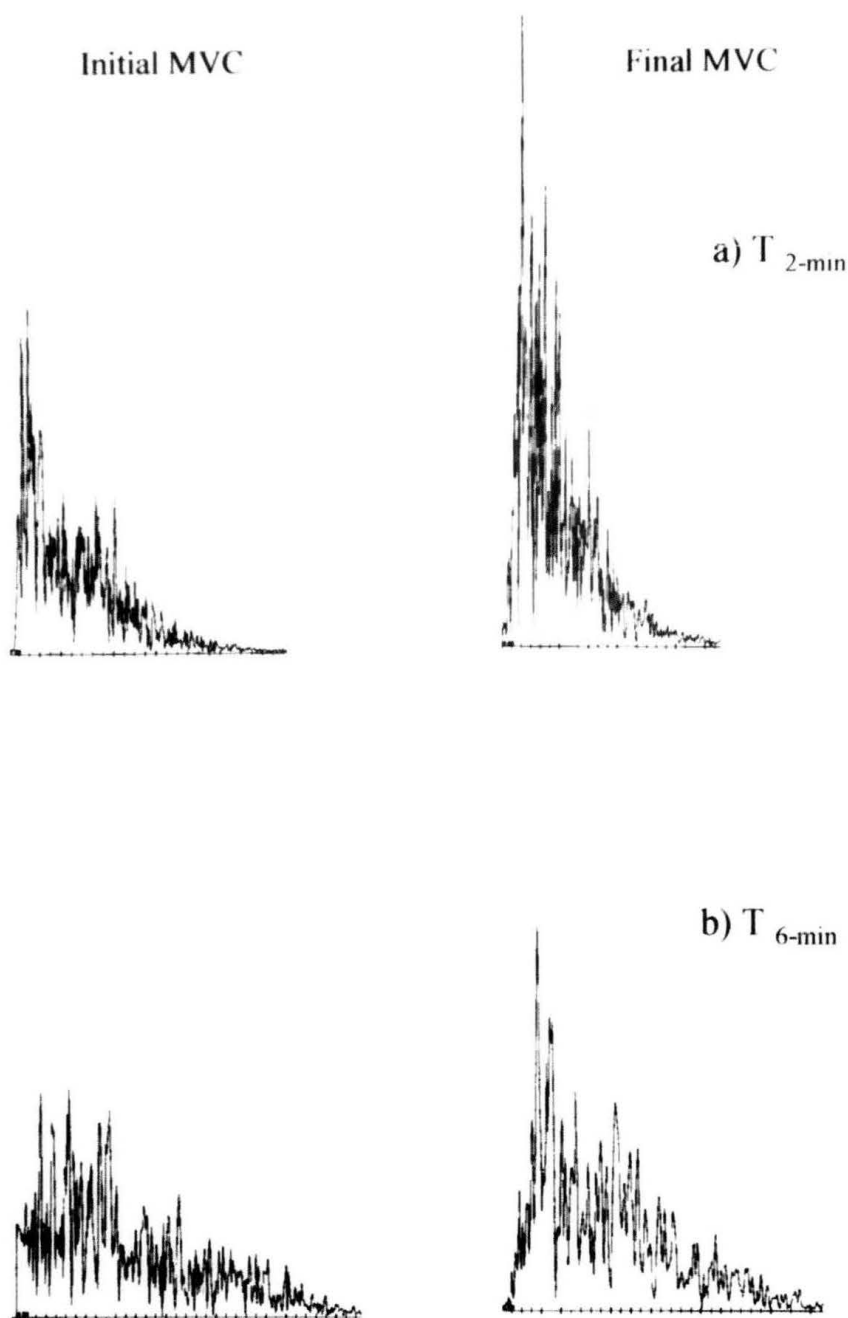


Figure 22. Sample traces of the power frequency spectrum of vastus lateralis (VL) for the same subject recorded from initial and final maximal voluntary contraction (MVC) during a) the 2-minute repeated MVC effort in T_{2-min} and b) the 6-minute self-pace effort in T_{6-min} performed on a rowing ergometer.

6.4 DISCUSSION

6.4.1 Performance Distance and Blood Lactate Response

Responses were compared between a repeated MVC protocol and a self-pace protocol performed on a standard rowing ergometer. The amount of work completed was relative to performance distance achieved in each trial, which equated to approximately 260 % greater work completed in $T_{6\text{-min}}$ compared to $T_{2\text{-min}}$. Blood lactate in $T_{6\text{-min}}$ compared to $T_{2\text{-min}}$ was greater in response to the greater amount of work completed, although the rate of work was less, and reflected a substantial anaerobic energy contribution in both trials. The rate and duration of work may influence anaerobic contribution and accumulation of blood lactate (Dolmage & Cafarelli, 1991). In both current trials, final blood lactate was expected to be high due to intensity and duration of muscle work. Other workers have reported high blood lactate concentrations during maximal rowing efforts, which occur during the first minute and are maintained for the duration of a race (Hagerman et al., 1978). Final blood lactate values that were influenced by the intensity, duty cycle, and duration of effort, were reflective of buffering substances together with influx and active removal of blood lactate (Westerblad et al., 1991). It should be noted that blood lactate concentration was not necessarily reflective of muscle lactate, which may affect muscle force and EMG differently (Bangsbo, Johansen, Graham, & Saltin, 1993). It was unlikely that the marginally greater blood lactate concentration in $T_{6\text{-min}}$ compared to $T_{2\text{-min}}$ was responsible for the significant difference observed for final muscle activation (rmsEMG) between trials. Muscle lactate, hydrogen ions, and other metabolites that influence muscle activation were not examined in this study but their influence was discussed previously (see section 2.5.3). However, it is reasonable to assume that blood

lactate values were sufficiently high in both current trials to be implicated in the loss of force. Any systematic associations between blood lactate concentration and change in EMG for each trial were beyond the scope of this study.

6.4.2 Comparison of Force Changes

In spite of greater work performed and longer exercise duration in $T_{6\text{-min}}$ compared to $T_{2\text{-min}}$, decreases in MVC force were similar between trials, which indicated an optimizing strategy during fatigue. The self-pace strategy in $T_{6\text{-min}}$ replicated a typical race performance, which has been described previously (Hagerman et al., 1978; Secher, 1993). The final force values (% MVC) in $T_{6\text{-min}}$ (range, 67 – 81 %) were similar to final force values (range, 75 – 90 %) observed by others during 2000-m (approximately 6-minute) rowing ergometer trials (Burton, 1996; Hagerman, 1996; Peltonen et al., 1994).

In the present study, change in force output from initial to final MVCs was examined. The MVCs were not confirmed as maximal and therefore, MVC force loss and change in muscle activity might be attributed to lack of effort. However, it was assumed that contractions were performed with maximal effort, as trained rowers are considered well motivated and capable of producing optimal performances (MacFarlane, Edmond, & Walmsley, 1997). Twitch interpolation methods (refer to section 2.8.2) that confirm maximal central drive and MVC force were not used due to technical difficulties associated with dynamic movement. Nevertheless, trained athletes that are well motivated, familiar with the task, and injury free are normally capable of fully activating large muscles in order to produce MVC force. The fact that rowers did not match initial MVCs later in performance provided support to the assumption that efforts

were maximal. Furthermore, maximal efforts were encouraged verbally by the investigator and assisted visually by feedback from the ergometer display. During $T_{6, \text{min}}$, with exception of initial and final contractions, force output ranged between 85–95 % MVC during the majority of the effort. Muscle forces applied at the handle throughout the effort were not consistent (data not included). This was expected, as intensity of each contraction was self-determined. Visual feedback was for average power output, not average force output each contraction and therefore, did not necessarily assist rowers in maintaining consistent force outputs.

The degree of force loss associated with fatigue is partly dependent on the task performed. Task variables, such as pattern of muscle activity and motor command, intensity and duration of effort, and speed and duty cycle of contraction have been cited as important (Dolmage & Cafarelli, 1991; Enoka & Stuart, 1992). In order to maintain average power output and achieve best performance outcome rowers manipulate the task variables. For example rowers manipulate pacing intensity, which involves alterations in force output, contraction velocity, and stroke-rate during the performance task (Hartmann et al., 1993; Peltonen & Rusko, 1993). This was evident from the ergometer display and from biomechanical data recorded each stroke during the self-pace task. In spite of force loss, rowers try to maintain pace or average power output by increasing stroke rate and contraction velocity (Dal Monte & Komor, 1989; Zatsiorsky & Yakunin, 1991). Force output data for each stroke were not included in this chapter as the purpose was to examine final change in response to each protocol, but are the subject of interest in a later study (chapter 8).

6.4.3 Comparison of MVC rmsEMG Changes

The hypothesis that predicted a significantly greater decrease in final MVC rmsEMG (VL) in response to a 6-minute optimal self-pace protocol compared to a 2-minute repeated MVC protocol performed on a standard rowing ergometer was supported. However, final MVC rmsEMG showed considerable variation between subjects in both trials. Variations in rmsEMG have been associated with subject characteristics, forces greater than 80 % MVC, confounding factors (muscle length, contraction velocity, synergistic muscle contribution), methodological factors, and data-window triggers (discussed in chapter 5).

In spite of a substantial loss of force in both ergometer trials, rmsEMG increased during the repeated MVC protocol in T_{2-min} and decreased during the self-pace protocol in T_{6-min} . Tesch et al. (1990) reported increases in EMG amplitude of VL during repeated bouts of isokinetic MVCs with intermittent recovery, similar to those in T_{2-min} . In contrast, Tesch et al. (1983) observed decreases in rmsEMG of VL during repeated isokinetic MVCs performed continuously. Tesch et al. (1990) attributed the different findings to differences in contraction velocities and recovery intervals. Increases in EMG amplitude have been demonstrated during high intensity (80 % $\dot{V}O_{2max}$) cycling (Kostka & Cafarelli, 1982; Moritani, Takaishi, & Matsumoto, 1993) and during incremental cycling protocols (Chwalbinska-Moneta et al., 1994; Housh et al., 1996).

Generally, during fatiguing dynamic exercise at submaximal levels EMG amplitude increases as additional MUs are recruited or firing rates increase in an attempt to maintain force output. During dynamic exercise at maximal or near-maximal intensity

with insufficient recovery between contractions, EMG amplitude declines. The decline in EMG amplitude under these conditions is indicative of reduced central drive or afferent inhibition of motor output. The increase in rmsEMG with decrease in force output in $T_{2-\min}$ and the contrasting situation in $T_{6-\min}$, where decrease in rmsEMG corresponded with decrease in force output, indicated that sites of fatigue may not be the same in each protocol.

The sites of fatigue and related EMG responses were discussed previously (see section 5.4.2). Briefly, according to the criterion of Horita and Ishiko (1987), the increase in rmsEMG relative to decrease in MVC force observed in $T_{2-\min}$ was more likely related to peripheral rather than central mechanisms. Thus in $T_{2-\min}$, loss of voluntary force was unrelated to motivational causes or central fatigue, but more likely related to contractile failure in the quadriceps muscle. One subject in $T_{2-\min}$ was an exception to the group trend and demonstrated a marked decrease in final rmsEMG of VL (63 % initial MVC). For this subject, the decrease in final rmsEMG may be attributed to reduced central drive associated with lack of motivation. In $T_{6-\min}$, all but one subject demonstrated a decrease in rmsEMG similar to that in force. The same subject demonstrated increased rmsEMG in both trials, indicating that MVCs were not maximal efforts and were influenced by motivational factors. For most subjects in $T_{6-\min}$, the similar decrease in rmsEMG and force output indicated a loss of central drive and possible central fatigue. Although, according to Tesch et al. (1983) this response did not necessarily indicate that the site of fatigue was solely proximal to the neuromuscular junction. Alternatively, reduced EMG amplitude in $T_{6-\min}$ may have reflected regulatory control from central command that optimized force output during maximal efforts and minimized failure of neural transmission and contractile processes as previously described by Bigland-

Ritchie and Woods (1984). The observed decline in rmsEMG may have reflected failure of action potential propagation, neural transmission, or contractile processes (see section 2.5.2). It was beyond the scope of this study to identify the site of failure.

The force/rmsEMG ratio has been used to describe the neuromuscular efficiency of muscle at a particular time (Enoka & Stuart, 1992; Miller et al., 1987). In T_{6-min} , the final force/rmsEMG ratio was similar to initial value demonstrating little change in response to the self-pace effort. Hence, in T_{6-min} neuromuscular efficiency was maintained in spite of fatigue. In T_{2-min} the final force/rmsEMG ratio decreased from initial value due to an increase in EMG (VL) activity relative to decrease in handle force. Hence, in T_{2-min} neuromuscular efficiency decreased. It appears that in self-determined maximal efforts such as in T_{6-min} , neural input is self-regulated and modulated, in order to maintain neuromuscular efficiency throughout the task. Thus, central command provides sufficient neural input to optimize force output dependent on the contractile state of the muscle. This is in agreement with regulatory central control proposed by Bigland-Ritchie and Woods (1984).

It is evident that in order to optimize force output during fatigue, EMG activity may be reduced and neuromuscular efficiency maintained, dependent on the task performed. Due to intensity and duration of the self-pace maximal rowing efforts, it was likely that rowers experienced both central and peripheral fatigue that affected neuromuscular activation. To what extent neuromuscular activation was regulated under those conditions in order to optimize force output, was not clear.

6.4.4 Comparison of MVC MPF Changes

The hypothesis that predicted a significantly greater decrease in final MVC MPF (VL) in response to a 6-minute optimal self-pace protocol compared to a 2-minute repeated MVC protocol performed on a standard rowing ergometer was not supported. Although final MPF was not significantly different between trials, seven of the eight subjects showed a greater MPF decrease in response to the repeated MVC protocol compared to the self-pace protocol. All but one subject demonstrated a decrease in MPF in both trials. For this subject, MPF increase in T_{6-min} could be explained by inconsistent timing of quadriceps activation with handle force, lack of motivation in producing optimal performance, or submaximal effort.

In response to fatiguing contractions in the current study, decreases in MPF of VL were greater in T_{2-min} (12 %) than in T_{6-min} (6 %), although decreases in force output were less in T_{2-min} (19 %) than in T_{6-min} (23 %). A wide range of MPF changes has been reported in response to fatiguing contractions, which vary with intensity and type of contraction. For example during a 60-s isometric MVC, decreases of 35 % (Oda & Moritani, 1995) and 50 % MPF (Moritani et al., 1986) have been observed. During 120 isokinetic MVCs of the leg extensors Tesch et al., (1983) reported a 10 % decline in MPF and 27% decline in force. However, Van Dieen et al. (1996) reported that during 250 isokinetic MVCs of the erector spinae muscle, MPF decreases differed significantly between subjects and MPF increases were seen for some subjects. During submaximal dynamic contractions of the biceps brachii, Potvin and Bent (1997) found that MPF decreased 25 – 29 %. In contrast, Arendt-Nielsen and Sinkjaer (1991) found that from the several muscles examined during submaximal uphill walking, only MPF of biceps femoris decreased significantly. They suggested that increases in muscle temperature

may have increased MPF and countered the decrease in MPF found in response to fatigue.

One possible explanation for less MPF decrease in the self-pace compared to the repeated MVC trial in the present study, could be the partial sparing of FT fibers. However, it is unlikely there was a reduced dependency on FT fibers in the self-pace compared to the repeated MVC trial. During repeated MVCs, large FT motor units are maximally recruited. FT fibers have greater action potential conduction velocities and are less fatigue resistant than smaller ST fibers (Komi & Tesch, 1979). During the repeated MVC protocol in T_{2-min} , conduction velocity of the FT fibers may have slowed which was reflected in MPF decrease. During the majority of self-pace submaximal contractions in T_{6-min} , force output was between 85 – 95 % MVC. It has been suggested that for large limb muscles in contractions greater than 80 % MVC, motor units are maximally recruited (De Luca et al., 1982). Therefore, it is reasonable to assume that all fibers were fully recruited and partial sparing of FT fibers did not influence either condition.

6.5 CONCLUSIONS

Loss of maximal voluntary force was similar in response to a typical fatigue protocol (2-minute repeated MVC) and a typical rowing task (6-minute optimal self-pace effort) performed on a standard rowing ergometer. This indicated some form of optimization mechanism that regulated force output under fatigue conditions. One such mechanism that regulates, and may optimize force output, is neural activation. Different neural strategies were adopted in response to fatigue in each task condition. This was evident

from the final MVC rmsEMG responses that were significantly different between protocols. EMG amplitude increased (relative to initial MVC) under the repeated MVC protocol, which indicated contractile failure and maintenance of central drive. However, EMG amplitude decreased under the self-pace protocol, which indicated the involvement of central fatigue or central regulation to minimize fatigue, as well as contractile fatigue. MPF declined similarly under both conditions, although decline was less during the self-pace task, which indicated a partial sparing of FT fibers.

The current findings indicate that sites of fatigue and neural optimization strategies may be influenced by the task performed. Therefore, analysis of EMG activity would be more appropriate when the task examined was specific to the condition of performance, such as in a 6-minute self-pace rowing performance.

CHAPTER 7

RELIABILITY OF FORCE AND EMG CHANGES DURING A SELF-PACE ROWING ERGOMETER TASK

7.1 INTRODUCTION

Trained rowers are generally capable of producing reliable rowing performances (Hahn et al., 1988; Secher, 1993), consistent movement characteristics, and consistent force-time profiles (Henry et al., 1995; Smith & Spinks, 1995). In competition, rowers seek the optimal balance of force, contraction velocity, and stroke rate in order to maximize performance and minimize fatigue (Dal Monte & Komor, 1989). Under these conditions, high blood lactate and low pH in the muscles (Hahn et al., 1995a) may affect muscle activation patterns, resulting in alterations in technique and consistency of movement. It has been suggested that muscle activation patterns and EMG characteristics may be unreliable during non-constant velocity dynamic contractions (Basmajian & De Luca, 1985; De Luca, 1997) and are further disturbed by fatigue (Taylor et al., 1997). The reliability of EMG characteristic changes during co-ordinated muscle movement under rowing performance conditions has not been reported.

During high-intensity self-pace ergometer performance (chapter 6), it was found that decreases in EMG amplitude (rmsEMG) and frequency (MPF) of VL corresponded with decreases in force output. However, one feature of that study was the considerable individual variation of rmsEMG responses. Similar variations in rmsEMG responses were found by Komi and Tesch (1979) during repeated isokinetic MVCs. These

authors suggested that amplitude variations were related to subject fiber type characteristics. Arendt-Nielsen and Sinkjaer (1991), and Potvin and Bent (1997) have both reported that EMG activity was inconsistent during cyclic movements and attributed the variations to changes in movement technique.

Generally, workers have reported better reliability for EMG frequency compared to EMG amplitude changes during fatigue (Arendt-Nielsen & Mills, 1988; Mannion & Dolan, 1996b). During dynamic ergometer contractions (chapters 6), seven out of eight subjects demonstrated decreases in MPF (VL) that were positively related to decreases in force output, although the magnitude of force-MPF correlation varied widely between subjects. The force-MPF results to date are in agreement with other dynamic studies (Bouissou et al., 1989; Potvin & Bent, 1997) and indicate that MPF may be useful in monitoring fatigue.

Various statistical methods have been reported for assessing the reliability of EMG and biomechanical indices during fatigue. These include intraclass correlation coefficients (Finucane, Rafeei, Kues, Lamb, & Mayhew, 1998; MacFarlane et al., 1997) Bland-Altman-plots (Horstmann, 1998), Pearsons correlation (Sherwood, Priebe, & Graves, 1997), and combinations of statistical methods (Kollmitzer et al., 1999; Sale, 1991). In the current study, a combination of statistical methods was used to confirm reliability. Evidence of reliability between replicated trials was accepted if there was (a) no significant difference ($\alpha = .05$), (b) a significant correlation ($\alpha = .05$), and (c) less than 10 % coefficient of variation between trials.

The purpose of this investigation was to determine reliability of performance distance, blood lactate concentration, force loss, and EMG (VL) characteristics during final MVCs in response to a 6-minute self-pace maximal rowing ergometer effort.

It was hypothesized that:

- 1) Performance distance and final blood lactate concentration resulting from a 6-minute self-pace rowing ergometer performance would be reliable.
- 2) MVC force loss resulting from a 6-minute self-pace rowing ergometer performance would be reliable.
- 3) Final MVC rmsEMG (VL) resulting from a 6-minute self-pace rowing ergometer performance would not be reliable.
- 4) Final MVC MPF of the EMG (VL) resulting from a 6-minute self-pace rowing ergometer performance would be reliable.

7.2 METHODS

7.2.1 Subjects

Subjects who participated in the previous studies (chapter 4, 5, 6) were invited to take part in the current study. Trained male rowers ($n = 10$) with a minimum of two seasons ergometer training experience, mean (SD) age 27.2 (8.6) years, height 183.1 (8.3) cm, and weight 91.9 (8.1) kg gave consent.

7.2.2 Apparatus

The equipment was identical to that described in chapter 4. Briefly, a standard Concept II rowing ergometer was modified by addition of a strain gauge and potentiometer to measure handle force and displacement respectively (Figure 9). The computerized data acquisition system (Amlab systems) and specifically designed project "tstan2.prw" (Appendix A5) were used for analogue to digital conversion and calibration of force and displacement data.

7.2.3 EMG

In order to replicate EMG preparation and configuration as closely as possible between trials, the distances between anatomical landmarks and electrode sites were measured (see section 3.1.1) and sites marked with indelible pen during the first trial. Skin impedance was kept below 10 k Ω , motion artifact monitored, and transducers re-calibrated. Data collection was triggered by positive handle force (greater than 50 N) and sampled at a frequency of 1024 Hz for 0.5-s duration (500 data-point window). Measurement and calculation of rmsEMG and MPF for the VL muscle, were identical to that described previously chapter 6. Triggered data-windows of EMG from five initial MVCs and from five final MVCs were averaged.

7.2.4 Procedure

Subjects were instructed to perform 6-minute maximal self-pace efforts at each of two ergometer trials (T_1 , T_2) during separate visits to the laboratory, more than two days, but less than two weeks apart (mean interval = 8 days). Prior to each trial subjects completed a non-fatiguing warm-up on a standard rowing ergometer (Appendix D). One-minute prior to each trial, a blood sample was taken from the fleshy part of the

thumb (see section 3.5) and analyzed for blood lactate concentration (Accusport). Subjects were then requested to perform a maximal self-pace ergometer effort with the objective of obtaining the greatest distance in a 6-minute period. The ergometer effort consisted of repeated optimal leg extensions (with the assistance of back extension and arm flexion). Visual feedback of power output was provided. Subjects were instructed to include five initial MVCs at commencement of the task and five final MVCs during the final 10 seconds of the task. Initial and final maximal efforts are part of normal pacing strategy used by rowers during time-trials. Initial MVC force was defined as the average force achieved during the initial five triggered data-windows after reaching a steady stroke rate (normally reached within two or three strokes). Final MVC force was defined as the average force achieved during the final five triggered data-windows. Within 30 seconds of completion of the task a second blood sample was taken and performance distance was recorded from the ergometer display. Subjects were allowed to warm-down on the ergometer at a comfortable self-selected pace until five minutes post-exercise, when a third blood sample was taken and analyzed for blood lactate concentration. On a subsequent occasion for the second ergometer trial, subject preparation, procedure, and conditions from the first trial were replicated as closely as possible. Subjects were requested to rest the day prior to each trial, so that pre-exercise activity was the same on both occasions. Instructions in both trials were the same and subjects were not informed of their performance results. However, the rowers were capable of gauging their performance from the visual feedback (power output) that was provided in both trials.

7.2.5 Analysis

Computerized recordings were replayed on the data acquisition system and all triggered data-windows of the leg extension phase tagged, digitized, and exported for further calculations (Excel). Exported data-windows for each contraction were averaged and mean values for the initial five and the final five MVCs were calculated. Final MVC data were normalized to the value at initial MVC and represented percentage change. Percentage change in each measure became dependent variables for subsequent analysis in a Statistical Package for Social Sciences (SPSS) and mean (SD) for force, rmsEMG, and MPF were calculated for each trial. A combination of statistical methods was used to confirm reliability. Dependent variables were examined for significant difference between trials (T_1 , T_2) using paired t -tests. To examine for variation and correlation between trials, method error expressed as the coefficient of variation (V) and Pearson's correlation (r) were used respectively for all dependent variables.

7.3 RESULTS

7.3.1 Performance Distance

Paired samples t -tests for mean performance distance showed no significant difference ($p = .83$) between T_1 and T_2 . Mean (SD), correlation score (r), and coefficient of variation (V) for performance distance achieved in T_1 and T_2 are shown in Table 7. Performance distances achieved were very similar in range between T_1 (1654 - 1905 m) and T_2 (1638 - 1896 m) demonstrating less than 1 % variation and significant ($p = .001$) correlation between trials. Individual performance distance for T_1 expressed as a percentage of T_2 , ranged from 98.7 - 103.4 %. In relation to preset reliability criteria, the performance distance achieved was reliable under current conditions.

Table 7

Reliability of Performance, Final Maximal Voluntary Contraction (MVC) Force and EMG Between 6-minute Self-Pace Ergometer Trials (N = 10)

	Trial							
	T ₁		T ₂		T ₁ - T ₂			
	<u>m</u>	(<u>SD</u>)	<u>m</u>	(<u>SD</u>)	<u>m</u>	(<u>SD</u>)	<u>V</u>	<u>r</u>
Performance (m)	1779.0	(80.2)	1781.5	(76.8)	-1.7	(24.9)	0.9	.95**
Final Blood lactate (mmol/L)	15.1	(1.7)	14.6	(1.9)	0.5	(2.8)	13.3	.22
Final force (% MVC)	77.0	(4.1)	79.3	(4.7)	-2.2	(4.8)	4.4	.40
Final rmsEMG VL (% MVC)	78.2	(21.3)	76.5	(33.3)	1.7	(27.0)	24.7	.59
Final MPF VL (% MVC)	95.1	(8.5)	97.8	(9.6)	-2.6	(7.8)	5.7	.64*

Note. Data are presented as mean (SD) for trials (T₁, T₂), mean difference (T₁ - T₂), coefficient of variation (V), and correlation coefficient (r) between trials. Performance distance and blood lactate at completion of exercise represent absolute values. Data represent the average of final five MVCs normalized to average of initial five MVCs. rmsEMG = root mean square of the electromyogram, MPF = mean power frequency of the electromyogram, VL = vastus lateralis muscle.

** = $p < .01$. * = $p < .05$.

7.3.2 Blood Lactate

For all subjects pre-exercise blood lactate values were less than 3.5 mmol/L in both trials. Paired samples *t*-tests for mean final blood lactate concentration showed no significant difference ($p = .62$) between T_1 and T_2 . Mean (SD), correlation score (*r*), and coefficient of variation (*V*) for final value in blood lactate for T_1 and T_2 are shown in Table 7. The range in final blood lactate concentration in T_1 (11.9 - 17.3 mmol/L) and T_2 (12.6 - 17.8 mmol/L) demonstrated a substantial increase above pre-exercise values in both trials, which indicated a significant anaerobic energy contribution for all subjects. Variation in blood lactate concentration between trials was moderately high (*V* = 13.3 %) and greater than that for performance distance (*V* = 0.9 %). Correlation between trials for final blood lactate concentration was low (*r* = .22) and not significant ($p = .55$). Individual final blood lactate concentration for T_1 expressed as a percentage of T_2 , ranged from 66.8 - 123.7 %. In relation to preset reliability criteria, results indicated that final blood lactate concentration was not reliable.

7.3.3 Force Loss

Paired samples *t*-tests for mean final force output showed no significant difference ($p = .18$) between T_1 and T_2 . Mean (SD), correlation score (*r*), and coefficient of variation (*V*) for final value in force (% MVC) for T_1 and T_2 are shown in Table 7. The ranges in final force output normalized to initial MVC were similar between T_1 (70.7 - 85.8 %) and T_2 (64.9 - 82.4 %) and represented a decrease from initial values. All subjects in both trials demonstrated a force output decrease from initial MVC value, which showed little variation (*V* = 4.4 %) between trials, but the correlation between trials did not reach significance ($p = .25$). Individual final force output for T_1 , expressed as a

percentage of T_2 , ranged from 94.3 - 114.1 %. In relation to preset reliability criteria, results indicated that final force output (percentage change) was not reliable.

7.3.4 Final MVC rmsEMG

Paired samples t -tests for mean final rmsEMG (VL) revealed no significant difference ($p = .84$) between T_1 and T_2 and represented a decrease from initial values in both trials. Mean (SD), correlation score (r), and coefficient of variation (V) for final value in rmsEMG (VL) for T_1 and T_2 are shown in Table 7. Sample traces of initial and final EMG of VL (EMG, rectified EMG) together with corresponding force are shown for T_1 and T_2 in Figures 23 and 24 respectively. The range in final rmsEMG (VL) was similar between T_1 (38.0 - 146.0 %) and T_2 (40.5 - 111.7 %). In T_1 , eight subjects demonstrated a final rmsEMG (VL) decrease from initial MVC values and two an increase. In T_2 , nine subjects demonstrated a final rmsEMG (VL) decrease from initial MVC values and one an increase. The subject in T_1 , who had the greatest increase from initial MVC value compared to other subjects, also had the greatest increase from initial MVC value in T_2 . Individual final rmsEMG for T_1 expressed as a percentage of T_2 , ranged from 48.93 - 165.2 %. Final rmsEMG showed considerable variation (V = 24.7%) between trials that was greater than the variation in force (V = 4.4 %) in spite of similar decreases in mean final rmsEMG and mean final force from initial MVC values. The correlation for final rmsEMG (VL) between trials did not reach significance ($p = .07$). In relation to preset reliability criteria, results indicated that final rmsEMG of VL (percentage change from initial MVC value) was not reliable.

Change in the force/rmsEMG ratio showed no significant difference ($p = .39$) between T_1 and T_2 and represented an increase from initial values in both trials. The change in

the force/rmsEMG ratio from initial value showed considerable subject variation. In T_1 , four subjects demonstrated a force/rmsEMG ratio decrease from initial MVC value and six an increase. In T_2 , six subjects demonstrated a force/rmsEMG ratio decrease from initial MVC value and four an increase. Compared to all other parameters, the change in force/rmsEMG ratio showed the greatest variability ($V = 34\%$) and no significant correlation ($p = .45$) between trials. In relation to preset reliability criteria, results indicated that change in the force/rmsEMG ratio (from initial MVC values) was not reliable.

7.3.5 Final MVC MPF

Paired samples t -tests for mean final MPF showed no significant difference ($p = .31$) between T_1 and T_2 and represented a small decrease from initial value compared to that in force. Mean (SD), correlation score (r), and coefficient of variation (V) for final value in MPF of VL for T_1 and T_2 are shown in Table 7. The range in final MPF was similar between T_1 (86.2 – 115.2 %) and T_2 (84.2 – 112.1 %). In T_1 , six subjects demonstrated a MPF decrease from initial MVC value and four an increase. In T_2 , eight subjects demonstrated a MPF decrease from initial MVC value and two an increase. Individual final MPF for T_1 , expressed as a percentage of T_2 , ranged from 94.4 - 114.1%. Change in MPF showed little variation ($V = 5.7\%$) and significant correlation ($r = .64$, $p = .04$) between trials. In relation to preset reliability criteria, results indicated that final MPF (percentage change from initial MVC value) was reliable.

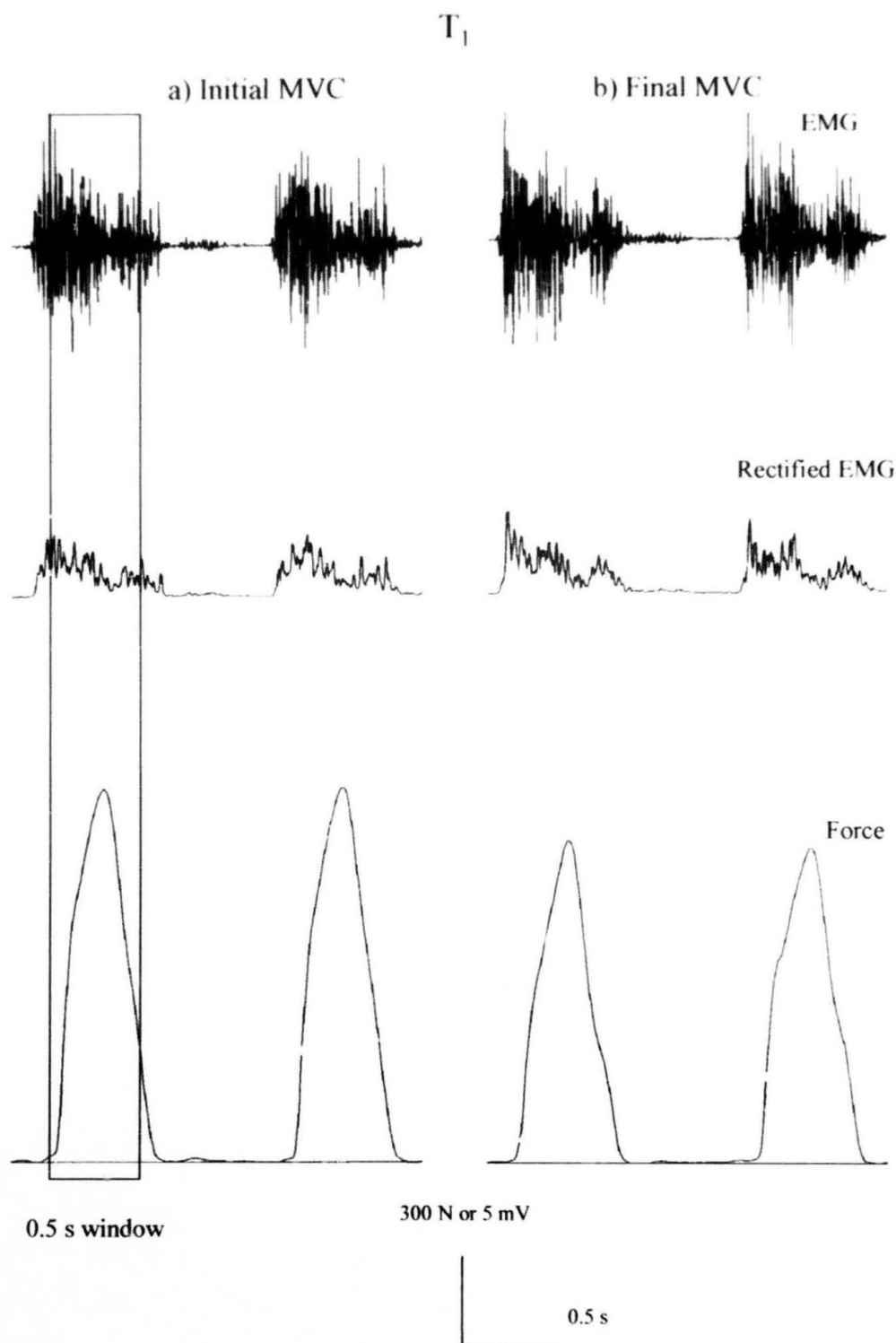


Figure 23. Sample traces of EMG and rectified EMG of vastus lateralis (VL) and force for the same subject as in T_2 recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-pace effort performed on a rowing ergometer in T_1 .

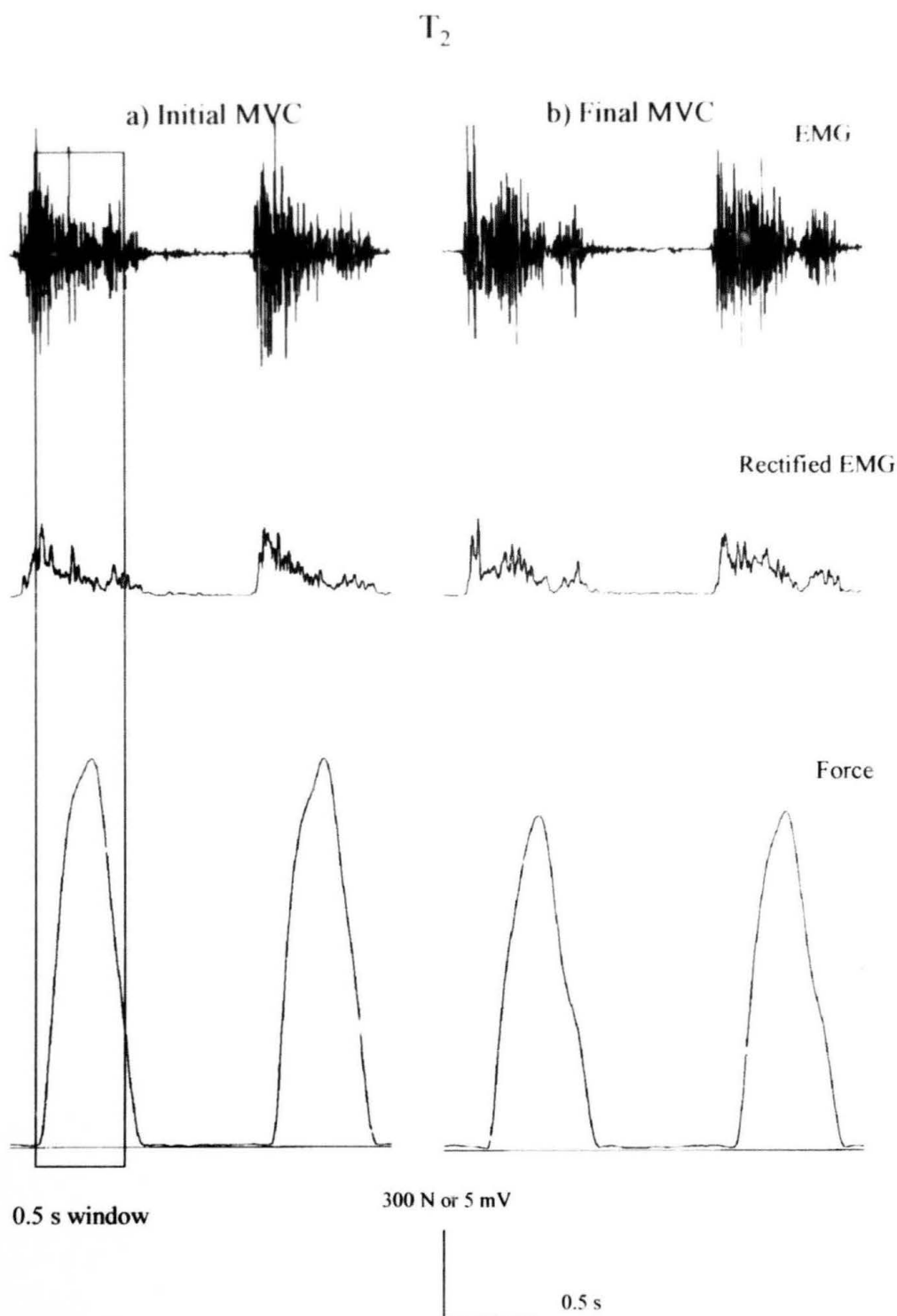


Figure 24. Sample traces of EMG and rectified EMG of vastus lateralis (VL), and force for the same subject as in T_1 recorded from a) initial and b) final maximal voluntary contraction (MVC) during a 6-minute self-paced effort performed on a rowing ergometer in T_2 .

7.4 DISCUSSION

The purpose of this investigation was to determine the reliability of performance distance, blood lactate concentration, force loss, and EMG characteristics during final MVCs in response to a 6-minute self-pace maximal rowing ergometer effort.

7.4.1 Reliability of Performance Distance

The hypothesis that predicted performance distance resulting from a 6-minute self-pace rowing ergometer performance would be reliable was supported in the current study. Subjects participating in this study were trained rowers and practiced in self-pace rowing ergometer trials. The performance distance achieved by each subject showed little variation between trials, although neither pace, force, or contraction velocity were controlled during the protocol. Trained rowers are generally capable of producing reliable performance, as measured by time to complete 2000-m or by distance attained in 6-minutes (Hahn et al., 1988; Secher, 1993). Performance reliability has been demonstrated during 2000-m ergometer trials performed on separate days over a two-week period and reported by Hahn et al. (1995a). Twenty elite rowers participated in three trials (control, placebo, and sodium bicarbonate supplementation) and the mean performance times (min:sec) were 6:25.4, 6:25.4, and 6:24.1 respectively. Thus, excellent performance reliability was demonstrated in highly trained rowers, even though, in one trial a buffer was used with the aim of improving performance. In the current study the performance distance (mean, range) achieved in 6-minutes was in T₁ (1779, 1654 – 1905 m) and in T₂ (1781, 1638 – 1896 m). These distances, extrapolated to the slightly longer duration 2000-m ergometer trial, represent (mean and range in min:sec) (6:44.7 and 6:17.9 - 7:15.3) in T₁ and (6:44.2, 6:19.7 - 7:19.5) in T₂, which are

on average 4.9 % slower than the elite athletes tested by Hahn et al. (1995a). For the trained rowers in the current study the reliability of performance outcome was marginally inferior to that for the elite rowers tested by Hahn et al. (1995a).

During maximal efforts, reliable performance outcome is related to consistent and repeatable stroke characteristics. In a study designed to discriminate between novice, good, and elite rowers Smith and Spinks (1995) found that stroke-to-stroke consistency was a biomechanical feature of consistent and better performances. In that study, consistent stroke profiles (propulsive work consistency, mean propulsive power, stroke smoothness) for elite rowers demonstrated little variation during ergometer contractions. Biomechanical stroke data were recorded in the current trials but stroke profile analysis was not the purpose of this study, and was therefore saved for later investigation.

7.4.2 Reliability of Final Blood Lactate Concentration

The hypothesis that predicted final blood lactate concentration resulting from a 6-minute self-pace rowing ergometer performance would be reliable was not supported in the current study. The findings for blood lactate concentration (immediately post-exercise) in response to rowing ergometer performances were similar to those values reported by others under similar conditions. For example, values ranging between 10.1 – 18.0 mmol/L (Hahn et al., 1995b) and 14 – 18 mmol/L (Hagerman, 1986), and \bar{m} (SD) value of 15.1 (2.7) mmol/L (Lutoslawska, Klusiewicz, Sitkowski, & Krawczyk, 1996). In spite of significant final blood lactate values that indicated anaerobic energy contribution in both current trials, there was no systematic trend in final blood lactate values that was replicated between trials. For each subject, exercise history, pre-

exercise blood lactate concentration, and exercise task were similar between repeated trials and presumably, these factors did not influence final blood lactate values differently between trials.

Subject variations in blood lactate responses to ergometer rowing have been observed in similar efforts. Male rowers ($n = 25$) participated in two 2000-m ergometer trials, two 1750 m on-water time trials, and two 2000-m on-water races and were tested pre- and post-exercise for blood lactate concentration (Hahn et al., 1995b). The investigators found no significant difference in post-exercise mean blood lactate either between the two ergometer trials, or between the four on-water trials. However, data revealed considerable subject variation between each of the ergometer tests, each of the 2000-m races, and each of the 1750-m time trials. Interestingly, there was a significant difference in post-exercise mean blood lactate concentration between the ergometer and on-water tests. Subject variability in blood lactate responses observed in the Hahn et al. (1995b) study may be explained by individual variations in self-determined pacing strategy, or in subject exercise and diet prior to each trial. In addition, variation may have been influenced by differences in subject blood borne buffering substances and lactate transport kinetics between trials (Bangsbo et al., 1993).

7.4.3 Reliability of Force Loss

The hypothesis that predicted MVC force loss resulting from a 6-minute self-pace rowing ergometer performance would be reliable was not supported in the current study. In the current trials mean percentage decreases in MVC force ($T_1 = 77\%$, $T_2 = 79\%$) were similar to those observed (75 – 90 %) in other self-pace rowing ergometer trials of similar duration (Hagerman, 1986; Peltonen et al., 1994). Burton (1996) found that

rowers used similar pacing intensities during self-pace 2000-m performances to those reported by Secher (1993) and described previously (see section 2.6). Burton observed that in the final 500 m of performance, average power output was 87.1 % of each individual's maximum power. Burton noted that power increase in the final 500 m (compared to the previous 500 m) was mainly influenced by increase in stroke-rate and not in force output.

MacFarlane et al. (1997) demonstrated that during shorter ergometer performances (30s), trained rowers were capable of reproducing maximal force and power between trials within the same training period. In order to examine loss of MVC force in the present study, subjects were asked to perform initial and final MVCs. However, results were dependent on the subjects performing maximal efforts. Twitch interpolation methods (described in section 6.4.1) that may be used to determine maximal voluntary activation of the muscle were not used in the current study. However, to assist subjects perform maximal efforts, verbal encouragement and visual feedback were provided. Furthermore, subjects were well motivated and familiar with rowing ergometer tasks.

In order to achieve the performance objective (greatest distance in 6-minutes), subjects attempted to optimize power output by manipulation of force output, displacement, contraction velocity, and stroke rate. Under these conditions, it was unlikely that muscle forces applied at the handle were consistent or maximal for each contraction. During self-pace maximal efforts, rowers manipulate these biomechanical parameters in order to minimize fatigue and achieve optimal performance outcome (Peltonen & Rusko, 1993). In the current study, decrease in mean MVC force output was similar between trials and the coefficient of variation was small. However, decrease in MVC

force output was not considered reliable because the correlation between trials was not significant. This finding may be partly attributed to subject manipulation of contraction velocity and stroke-rate. Contraction velocity, stroke rate, and force output may have individual effects on the EMG signal, although resultant biomechanical output of these parameters may be determined by examining power output. Power output data will be presented in a subsequent chapter.

7.4.4 Reliability of Final MVC rmsEMG

The hypothesis that predicted final MVC rmsEMG (VL) resulting from a 6-minute self-pace rowing ergometer performance would not be reliable was supported by the current findings. Mean final rmsEMG (VL) represented a decrease from initial values in both T₁ and T₂, although two subjects in T₁ and one subject in T₂ demonstrated a final rmsEMG (VL) increase. In T₁ and T₂, decreases in both rmsEMG (VL) and force output were similar to those found in response to isokinetic MVCs (Komi & Tesch, 1979). Importantly, final rmsEMG showed considerable variation ($V = 24.7\%$) between current trials.

The poor reliability of rmsEMG change observed in the current study was similar to that found by Kollmitzer et al. (1999) during sustained isometric contractions repeated on different occasions. These workers found that rmsEMG correlation between muscles and between trials was low, and that rmsEMG changes were not reliable (ICC = 45 %). In contrast, Kadaba, Ramakrishnan, Wootten, Gainey, Gorton, & Cochran, (1989) reported that smoothed and rectified EMG for gait analysis was repeatable both within and between days. These workers used an adjusted coefficient of multiple correlation and a coefficient of variation to evaluate reliability. Bosco and Viitasalo (1982) found

that reproducibility of IEMG (expressed as a correlation coefficient) of the lower limb muscles during vertical jumping was good ($r = .79$) and that reproducibility of the IEMG-force relationship was greater ($r = .90$).

Kollmitzer et al. (1999) observed high variance of rmsEMG changes during fatigue, although variance was not significantly different between trials. In addition, these workers found that EMG variability increased at force levels greater than 80 % MVC and reliability improved at lower submaximal contractions. Subject variability at high force levels has been attributed to the presence of agonists co-contraction, to the possible effect of fatigue and to inefficient central nervous control (De Luca, 1997; Yang & Winter, 1983), which were discussed previously (see section 4.4.1). In the current study, most contractions were greater than 80 % MVC and therefore, reliability may have been influenced by these factors. In spite of performance distance and force loss showing little variance between trials, muscle activation of VL during final MVCs behaved inconsistently between trials and was not reliable under current conditions.

Possible explanations for the inconsistent rmsEMG changes may be related to methodological factors (see section 3.1), confounding factors (see section 4.4.3), and subject characteristics such as co-ordination and muscle activation patterns (see section 5.4.3). During rowing tasks the ability to co-ordinate muscle group action and manage efficient force contribution is essential to performance outcome. Subjects in the current study had automated rowing skills and at least two years competitive experience. In these subjects, muscle group co-ordination to produce optimal synergistic force at the handle is relatively consistent (Smith & Spinks, 1995). However during intense rowing (85 - 100 % MVC), optimal timing that involves the selective activation and relaxation

of muscle groups is influenced by fatigue. In the present study, the timing of handle force output and quadriceps activation (rmsEMG) may have been affected by synergistic action of other muscle groups, co-ordination of the contributing muscle groups, or by fatigue in specific muscle groups. In addition, changes in technique may have occurred that altered muscle activation patterns and affected the sampled EMG data-window. Arendt-Nielsen and Sinkjaer (1991) reported that duration and timing of EMG activity of VL changed significantly in response to fatigue during walking, although the pattern of movement remained constant. These authors observed that co-ordination of movement patterns and muscle performance were disturbed by fatigue and therefore, EMG parameters were not always reliable under these conditions.

During each contraction in the present study, the measured window of quadriceps rmsEMG represented muscle activity related to force output at that time, but was less representative of total quadriceps activation. In some subjects, synchronization of quadriceps activity with force output appeared to deteriorate with increased loss of force. Hence, the synchronization of measured force output with quadriceps EMG may have influenced reliability of the recorded EMG. Quadriceps activity from onset to offset of activation was not measured in this study because the objective was to examine quadriceps activity related to handle force output. Evidence of quadriceps muscle activation outside the recorded data-window was visible in the EMG and force trace (see Figure 17) and discussed previously (see section 5.4.4). In the final study (chapter 8), quadriceps muscle activation irrespective of force output is examined

The force/rmsEMG ratio has been used to describe the neuromuscular efficiency of muscle (Miller et al., 1987). In both T_1 and T_2 , the final force/rmsEMG ratio marginally

increased above initial values, thus neuromuscular efficiency was maintained during fatigue although not reliably between trials. Subjects demonstrated a decrease in electrical activity of the muscle similar to that in force output, which indicated central fatigue, although the same changes have also been attributed to central regulation. The maintenance of neuromuscular efficiency found in the present investigation was in agreement with Bigland-Ritchie et al. (1983) who suggested that parallel decreases in EMG amplitude and force indicated central regulation that optimized force and prevented electrical transmission failure. Tesch et al. (1983, 1990) found neuromuscular efficiency responses (force/EMG) that varied between fatigue trials performed on different occasions (although procedures were not replicated between trials). In agreement, the present study found that changes in rmsEMG and force-rmsEMG ratio in response to fatigue were not reliable following self-paced ergometer rowing.

7.4.5 Reliability of Final MVC MPF

The hypothesis that predicted final MVC MPF (VL) of the EMG resulting from a 6-minute self-paced rowing ergometer performance would be reliable was supported by the current findings. Decreases in MPF were weakly related to decreases in force. MPF percentage decreases (\bar{m} [SD] %) from initial MVC values for T₁ and T₂ (4.9 [8.5] %, 2.2 [7.8] %) were small compared to corresponding force decrements. There was a weak systematic trend in MPF change, although not all subjects (six out of ten in T₁ and eight out of ten in T₂) demonstrated a decrease in MPF. For some subjects MPF showed significant decreases (16 % from initial MVC) that were similar between trials and corresponded with decreases in force output. For these subjects, MPF provided a good indication of fatigue in the quadriceps muscle (VL). However for other subjects,

MPF changes were minimal and not always consistent between trials. Subject MPF responses to fatigue vary. For example, Gamet et al. (1996) examined reproducibility (week to week) of MPF of the quadriceps muscle during incremental cycle ergometry and found that subject MPF trends were inconsistent during exercise. Some subjects demonstrated a continuous MPF increase, some a slight MPF decrease, and some a MPF increase followed by a decrease. The reliability from week to week was considered satisfactory for five of the seven subjects (range, ICC = 49 – 94 %). One explanation for MPF intra-subject variation is the different fiber type ratios of muscle (Komi & Tesch, 1979). Subjects with a greater ratio of FT to ST fibers may demonstrate a greater slowing of action potential conduction velocity reflected in greater decline in MPF. The effect of fiber type on the EMG was discussed previously (see sections 5.4.2 and 6.4.4)

MPF responses during fatigue have demonstrated various degrees of change. In the current rowing trials, MPF decreases were less than those reported by others during dynamic exercise. In response to repeated dynamic biceps contractions with a fixed load, Potvin and Bent (1997) reported substantially greater MPF decreases (range, 25 - 29 %) compared to the present results. In response to exhaustive dynamic contractions performed on a cycle ergometer at constant power output, Arendt-Nielsen and Sinkjaer (1991) reported marginally greater MPF decreases (\bar{m} [SD] = 10.1 [0.9] %) compared to the present results. Furthermore, in chapter 6 under identical conditions, MPF showed a greater decrease (7 % from initial MVC) than those currently observed.

The repeatability of EMG during dynamic MVCs has been questioned due to the non-stationary nature of the EMG signal that involves changes in the motor units sampled

(De Luca, 1997). Stationarity of dynamic EMG was reported previously (see section 6.4.3). Reliability studies of MPF change during fatigue are difficult to assess due to variations in analysis, EMG methodologies, protocols, muscles, and subjects used. Kollmitzer et al. (1999) examined reliability of the quadriceps median frequency (MF) during sustained static submaximal (50 % MVC) and maximal contractions at 90-minute and 6-week intervals. These workers assessed reliability of EMG data using Bland-Altman-plots, Pearson's coefficient of correlation, and intraclass correlation of variance (ICC). They observed that shift in MF did not differ significantly from one assessment to another and that intra-subject variance of the MF shift did not significantly differ between intervals. They also found that the reliability for MF shifts under these conditions was excellent for the RF muscle (ICC = 87.3 %), but lower for the VL muscle (ICC = 34 %). The reliability of MF shift between days was lower (range, ICC = 49 - 61 %) than the shift within the same day (range, ICC = 85 - 99 %).

In the current investigation, MPF (VL) was found to be reliable between trials and may provide an indication of quadriceps fatigue during self-paced ergometer efforts. However, the force-MPF relationship was weak. It was likely that for some subjects, for whom MPF decrease was more pronounced, qualitative EMG changes were related to quadriceps fiber type ratios. It would be interesting to investigate the reliability of MPF decrease for only the subjects who showed significant MPF decreases in the current study. In addition, to examine the same subjects for fiber type ratios using muscle biopsy techniques.

7.5 CONCLUSIONS

Performance distance achieved and final MVC MPF of VL in response to 6-minute self-pace ergometer performance were considered reliable. Blood lactate at completion of exercise, force loss, and final MVC rmsEMG were not considered reliable under these conditions. MPF decreases in the current trials were not of sufficient magnitude to be of practical value for assessing fatigue of the VL muscle during self-pace rowing performance. However, some individuals demonstrated a significant decrease in MPF with decrease in force output. For these subjects decrease in MPF of VL might be useful to indicate quadriceps fatigue, and to predict loss of force output during performance. The literature indicates that MPF changes may be related to fiber type ratios. Further research is needed to confirm fiber type ratios in subjects that demonstrated significant MPF changes and to confirm reliability of the MPF changes in the same subjects over replicated performances. The findings to date provide no evidence as to whether EMG reliability would improve if EMG data-windows were to include all electrical activity of the quadriceps muscle irrespective of force output measured at the handle. This will be addressed in the next chapter. Previously (chapter 6) it was demonstrated that EMG responses to fatigue were dependent on variables of the task performed. If pacing intensity of the task is controlled, then force loss and EMG characteristic changes, such as those demonstrated in the current study, may be minimized. Pacing strategy is the focus of the final experimental study.

CHAPTER 8

EFFECT OF PACING STRATEGY ON BIOMECHANICAL AND PHYSIOLOGICAL RESPONSES TO A 2000-m ROWING ERGOMETER TASK

8.1 INTRODUCTION

During a 2000-m race performance, rowing strategy typically consists of intense effort for the first minute, maintenance of steady pace for four minutes, and then maximal effort in the last minute (Secher, 1993). This pacing strategy has been described as biomechanically inefficient because of the inconsistent forces applied at the oar that affect boat velocity and increase drag factors (Zatsiorsky & Yakunin, 1991). Furthermore, the typical race strategy is metabolically inefficient because of the high initial dependence on anaerobic energy systems and the early increase in blood lactate concentration, which might adversely affect performance (Hagerman, 1986). In addition, the neuromuscular changes associated with a typical self-pace strategy may affect performance. For example, in chapter 7 final MVC rmsEMG and MPF decreased with reduced maximal force output.

For the above reasons, a number of investigations have examined constant-pace efforts. The effects of pacing strategy on performance outcomes and physiological parameters have been examined during cycling tasks (Cherry et al., 1997; Foster et al., 1993) and during rowing tasks (Burton, 1996; Hagerman et al., 1978). In addition, the effect of

pacing strategy on neuromuscular activation has been examined during cycling tasks (Takaishi, Yasuda, Ono, & Moritani, 1996) but not as yet, during rowing tasks.

If rowing pace is maintained at a constant level so as to produce an equivalent performance as that seen using a typical pacing strategy, then neuromuscular fatigue may be minimized and latent force or power output increased. Hence, the purpose of this study was to compare force and power loss, change in EMG characteristics, and final blood lactate concentration between a typical self-pace and a controlled constant-pace rowing ergometer performance.

It was hypothesized that:

- 1) There would be no significant difference in performance distance achieved following a constant-pace compared to a self-pace 6-minute rowing ergometer performance.
- 2) There would be significantly greater MVC force and power output following a constant-pace compared to a self-pace 6-minute rowing ergometer performance.
- 3) There would be significantly greater MVC rmsEMG and MPF following a constant-pace compared to a self-pace 6-minute rowing ergometer performance.
- 4) There would be significantly less blood lactate concentration following a constant-pace compared to a self-pace 6-minute rowing ergometer performance.

8.2 METHODS

8.2.1 Subjects

Individuals who had participated in any of the previous studies (chapters 4, 5, 6, 7) were invited to take part in the current study. Trained male rowers ($n = 7$) with a minimum of two seasons training experience, mean (SD) age 27.6 (6.7) years, height 185.1 (9.4) cm, and weight 95.6 (6.1) kg gave informed consent.

8.2.2 Apparatus

The equipment used was identical to that described in chapter 4. Briefly, a standard Concept II rowing ergometer was modified by addition of a strain gauge and potentiometer to measure handle force and displacement respectively (Figure 9). The computerized data acquisition system (Amlab systems) and specifically designed project “tstan3.prw” (Appendix A6) were used for analogue to digital conversion and calibration of force and displacement data (see section 3.2). Blood lactate samples were analyzed using an Accusport and heart rates recorded using a Polar 4000 monitor. The Borg Scale (15 point) was used for ratings of perceived exertion (RPE).

8.2.3 EMG

EMG configuration and preparation for VL were replicated as closely as possible for the two protocols. Impedance between EMG electrodes was recorded and motion artifact monitored. Acquisition and analysis of EMG were similar to that described previously (see section 7.2.3), with the major exception that EMG data-windows were triggered by muscle activation of rmsEMG of VL greater than 0.7 volts. The Amlab schematic was “tstan3.prw” (Appendix A6). The amended data trigger allowed analysis of VL muscle

activation independent of force or velocity levels, and thus included onset of EMG prior to threshold levels of force output or handle velocity being achieved. Data were sampled at a frequency of 1024 Hz for a duration of 0.7 s each data-window. The data-window duration was increased from that in the previous study (0.5 s) in order to include muscle activation independent of positive handle force or velocity. Data-windows of EMG triggered by rmsEMG activation were recorded, processed, and normalized to pre-MVC values. In addition, a second series of EMG data-windows were triggered by positive handle force (greater than 50 N) using the “tstan2.prw” schematic project (Appendix A5) and sampled at a frequency of 1024 Hz for 0.5-s duration. The second series of EMG data-windows (handle force trigger) was recorded to enable comparison with the first series of EMG data windows (VL muscle activation). Unless indicated otherwise text, data, and graphs refer to the first series EMG data-windows triggered by minimum VL activation.

8.2.4 Procedure

Subjects visited the laboratory on separate occasions more than two days, but less than two weeks apart (mean interval = 7 days) to perform 6-minute self-pace (T_{self}) and constant-pace (T_{con}) ergometer trials. Prior to each trial, subjects were fitted with a heart rate monitor and completed a non-fatiguing warm-up on a standard rowing ergometer (Appendix D). Subjects then performed five MVCs prior to exercise (pre-MVC), followed by a two-minute recovery period. Prior to the 6-minute protocol a blood sample was taken from the fleshy part of the thumb (see section 3.3) and analyzed for blood lactate concentration (Accusport). In the first trial (T_{self}), subjects performed a maximal self-pace effort with the objective of attaining the greatest distance within 6 minutes that included five final MVCs. Rowing pace (velocity, force, and stroke rate)

was self-determined and assisted by visual feedback from the pacing-display (pace per 500 m updated each stroke). Subjects were familiar with the Borg Scale and received standardized instructions for RPE as described by (Borg, 1985). RPE and heart rate were recorded each minute. The investigator provided verbal encouragement particularly during the five final MVC's. Force and power data-windows were triggered by positive handle force greater than 50 N and recorded at a sampling frequency of 200 Hz for a duration of 0.5 s each contraction. Pre- and final MVC were taken as the average force of five consecutive triggered data-windows recorded during maximal efforts performed prior to exercise and during final exercise contractions. In addition, five consecutive contractions were recorded at the end of each minute during the performance task. Second and third blood samples were taken on completion of exercise and five minutes post-exercise then analyzed for blood lactate concentration. On the second visit the procedure was repeated except that subjects were instructed to perform at a target constant-pace calculated as the average power output achieved in the first trial. Subjects were encouraged to match power output (displayed on the ergometer) with the target power output as closely and for as long as possible during the 6-minute effort. Five final MVCs were performed and recorded immediately prior to the end of the 6-minute effort.

8.2.5 Analysis

Computerized recordings were replayed on the data acquisition system with all triggered data-windows of the leg extension phase tagged, digitized, and saved. Data was exported to a spreadsheet package (Excel) for further calculations. Final performance distance and blood lactate value at completion of exercise were compared in absolute terms. Final MVC (force, power, rmsEMG, MPF) data were normalized to

respective values at pre-MVC (in the same trial) and expressed as percentage change, which became dependent variables for subsequent analysis in a Statistical Package for Social Sciences (SPSS). The dependent variables were examined for significant difference between trials (T_{self} , T_{con}) using paired samples t-tests. Mean (SD) of the dependent variables was calculated for each trial and the method error was expressed as a coefficient of variation (V) calculated between each trial.

8.3 RESULTS

8.3.1 Performance Distance

Paired samples t-tests revealed that performance distance achieved in T_{self} was significantly greater ($p = .009$) than in T_{con} . Table 8 shows comparison of T_{self} and T_{con} means (SD) and V for performance distance. Distance achieved ranged between 1720 – 1880 m in T_{self} and 1712 - 1865 m in T_{con} , and was marginally greater for each subject in T_{self} compared to T_{con} (see Figure 25a).

8.3.2 Force and Power Output

Paired samples t-tests revealed that final MVC force output was significant greater ($p = .002$) in T_{con} (75 %) than in T_{self} (67 %), and that final MVC power output was significantly greater ($p = .001$) in T_{con} (65 %) than in T_{self} (56 %). Table 8 shows comparison of T_{self} and T_{con} means (SD) and V for final MVC force and power output. For all subjects final force and power outputs were greater in T_{con} compared to T_{self} , although individual responses varied (see Figures 26b and 27b). Most subjects showed between 5 –10 % greater final force and power output in T_{con} compared to T_{self} , but for

one subject that difference was 15 %. Sample force traces of pre-MVC, initial contraction, and final MVC for T_{self} and T_{con} are shown in Figure 28.

For both trials, with the exception of the first and last minute, force and power outputs were maintained relatively constant (see Figures 26a and 27a). In the first minute, force and power outputs were greater in T_{self} than in T_{con} . During the final minute in both trials, force and power outputs increased compared to the previous minute in response to the final maximal effort. However, a significantly greater increase in force and power output was demonstrated in T_{con} compared to T_{self} . The pattern of change in power output recorded each minute was similar to that for force output for both trials but at lower percentage pre-MVC.

Table 8

Comparison of Performance Parameters Between Self-Pace and Constant-Pace

6-minute Ergometer Trials (N = 7)

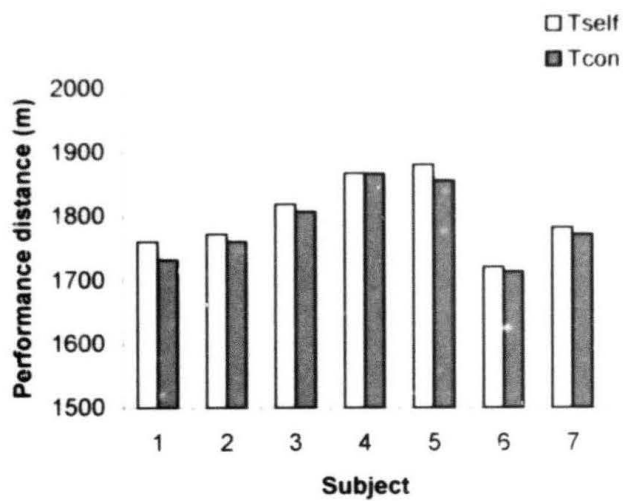
	Ergometer Trial			
	T _{self}		T _{con}	
	m	(SD)	m	(SD)
Performance distance (m)	1799.7	(58.0)	1786.0	(58.7)
			13.7	(9.5)**
Final force output (% MVC)	67.1	(7.0)	74.7	(5.3)
			-7.6	(3.7)**
Final power output (% MVC)	56.1	(9.0)	64.6	(8.0)
			-8.5	(2.7)**
Final blood lactate (mmol/L)	11.9	(1.3)	11.3	(1.3)
			0.6	(1.8)
Final heart rate (bpm)	184	(16)	186	(13)
			-2	(4)
Final RPE (Borg Scale)	19	(0.5)	19	(0.5)
			0	(0.5)

Note. Data are presented as mean (SD) for trials (T_{self}, T_{con}), mean difference

(T_{self} - T_{con}), and coefficient of variation (V) between trials. Performance distance at completion of exercise and final blood lactate represent absolute values. Data represent the average of final five maximal voluntary contractions (MVCs) normalized to the average of initial five MVCs. T_{self} = maximal self-pace trial, T_{con} = constant-pace trial (pace calculated to be the average pace from T_{self}).

** = $p < .01$. * = $p < .05$.

a)



b)

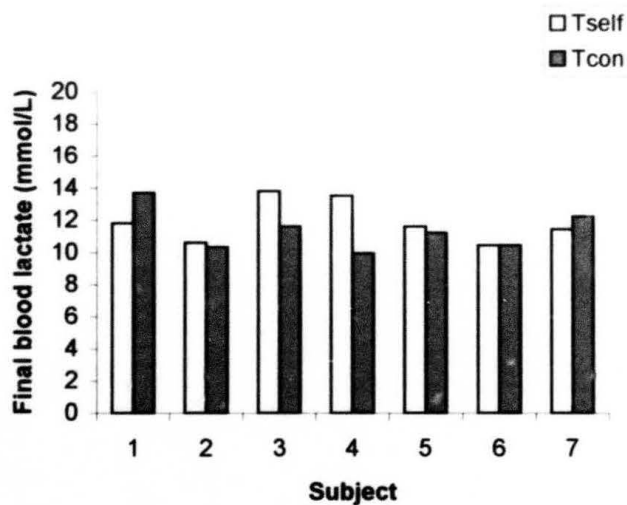
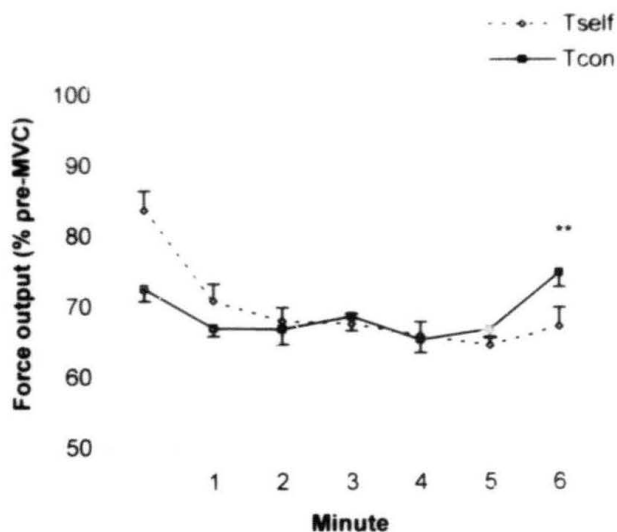


Figure 25. Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) performance distance and b) final blood lactate for each subject. $N = 7$.

a)



b)

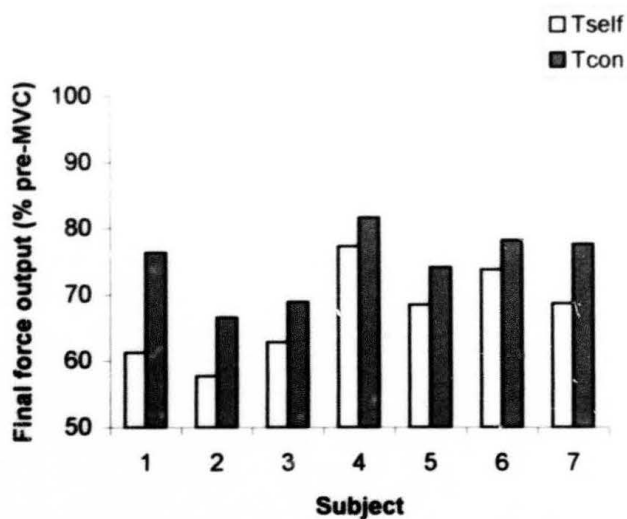
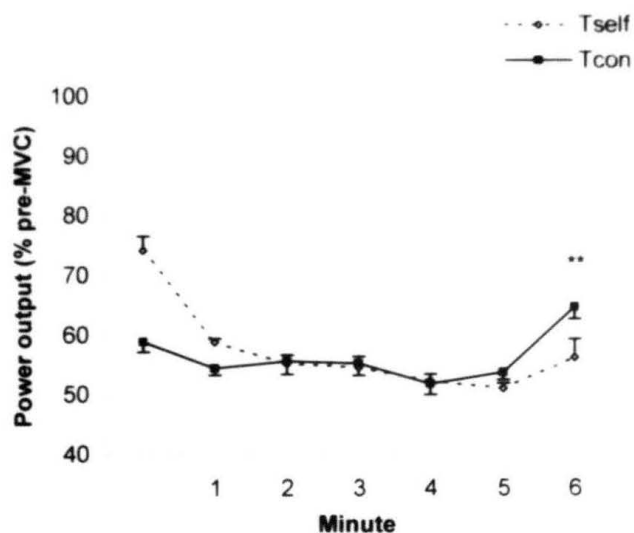


Figure 26. Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in mean force output recorded each minute and b) final maximal force output for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by handle force output > 50 N.

$N = 7$, ** = $p < .01$.

a)



b)

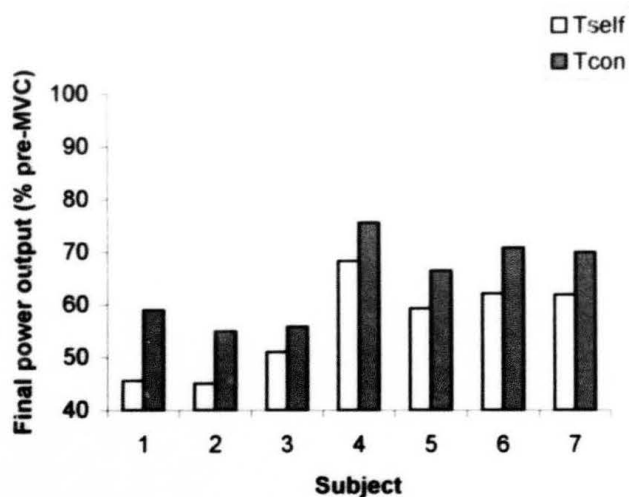


Figure 27. Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in mean power output recorded each minute and b) final maximal power output for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by handle force output > 50 N. $N = 7$, ** = $p < .01$.

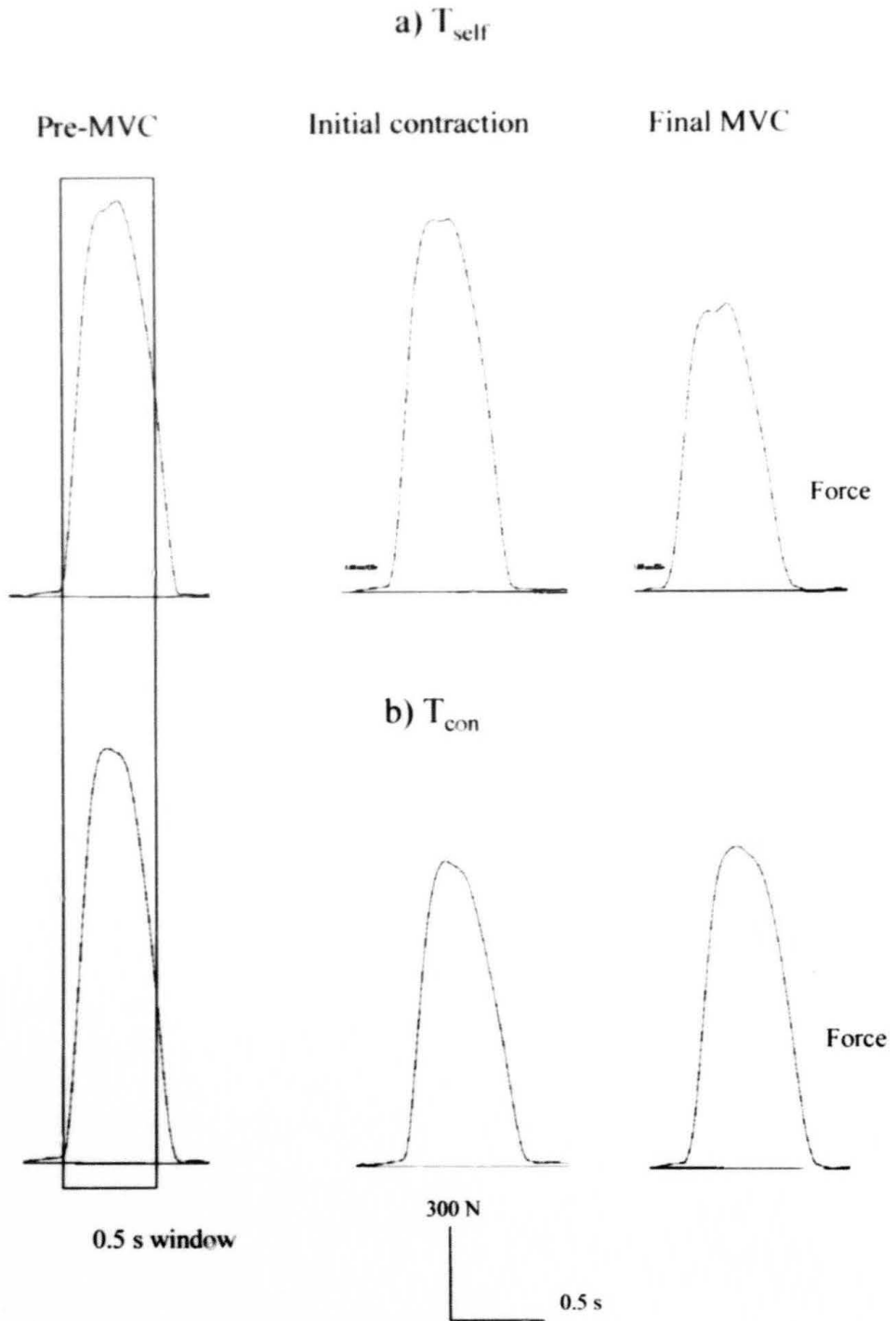


Figure 28. Sample traces of force for pre- maximal voluntary contraction (MVC), initial contraction, and final MVC for the same subject during a 6-minute effort performed on a rowing ergometer using a) the self-pace strategy in T_{self} and b) the constant-pace strategy in T_{con} .

8.3.3 rmsEMG

The first series of rmsEMG data-windows were triggered by minimum VL muscle activation ($\text{rmsEMG} > 0.7 \text{ V}$) for a fixed duration (0.7 s) and represented most VL EMG activity irrespective of force or velocity values. The second series of rmsEMG data-windows were triggered by positive handle forces ($>50 \text{ N}$) for 0.5 s duration and represented only VL activation related to positive handle force (as in chapter 7).

Final rmsEMG of VL (triggered by minimum VL muscle activation) decreased in a similar fashion in both T_{self} and T_{con} with values of 90.4 % and 91.5 % of pre-MVC respectively (refer to Table 9). There was no significant difference between trials for final rmsEMG. In both trials there was a smaller decrease but a greater variation in final rmsEMG triggered by rmsEMG activation (refer to Table 9) compared to final force output (refer to Table 8). Final rmsEMG (VL) triggered by muscle activation demonstrated a significantly smaller decrease than that triggered by handle force (refer to Table 9).

For each subject final rmsEMG (VL) responses varied between T_{self} and T_{con} . Two subjects had substantially greater final rmsEMG (VL) in T_{con} compared to T_{self} (Figure 29b), and one of these subjects (subject 4) was the best performer in the group. Sample traces of pre-MVC, initial contraction, and final MVC for EMG and rectified EMG (VL) in T_{self} and T_{con} are shown in Figure 30.

In T_{self} , the rmsEMG of VL (% pre-MVC) recorded each minute showed initial submaximal activation (88 %), subsequent lower activation (80 %), and increase in activation (90 %) during final MVCs (see Figure 29a). In this trial, the pattern of

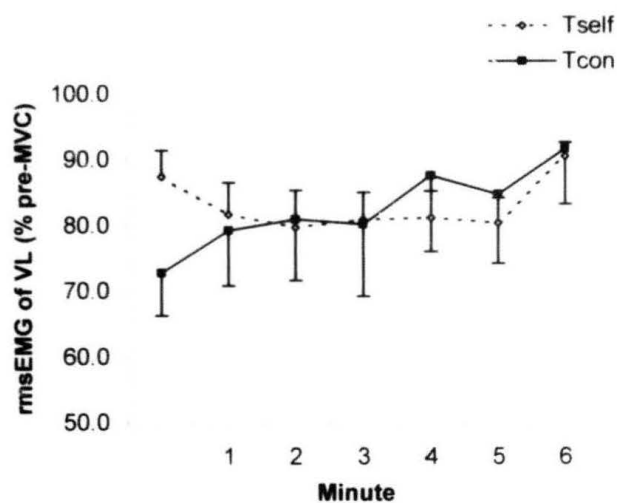
change each minute for rmsEMG (VL) showed a very similar profile to that for force and power outputs (see Figures 26a and 27a). In T_{con} , rmsEMG of VL (% pre-MVC) recorded each minute showed initial submaximal activation (73 %), steady increase throughout the 6-minute task and rapid increase (91 %) that corresponded to final MVCs. Initial rmsEMG (VL), force, and power values were lower in T_{con} than in T_{self} . However, greater final force and power outputs in T_{con} compared to T_{self} , were not accompanied by correspondingly greater rmsEMG (VL) in T_{con} .

8.3.4 MPF

There was a significantly greater decrease ($p = .001$) for final MPF demonstrated in T_{self} than in T_{con} (refer to Table 9). The mean difference in final MPF between trials was 8.3%. For each subject final MPF (VL) was greater in T_{con} compared to T_{self} (Figure 31b) that ranged between 81 – 98 % and between 74 – 91 % of pre-MVC values respectively. Most subjects demonstrated between 5 – 10 % greater final MVC MPF in T_{con} compared to T_{self} and for one subject this was 15 %.

The pattern of MPF (VL) recorded each minute remained at higher percentage pre-MVC in T_{con} than in T_{self} throughout the task (see Figure 31a). In T_{self} , initial MPF of VL (% pre-MVC) was high (93 %) with subsequent slow decrease (83 %) that did not increase when subjects attempted the final maximal effort. In T_{con} , initial MPF of VL (% pre-MVC) was near maximal (98 %) and greater than the initial value in T_{self} , with a subsequent slow decrease (91 %) that did not increase when subjects attempted the final maximal effort. Unlike rmsEMG during the final MVCs, MPF did not increase with increase in force output in either trial.

a)



b)

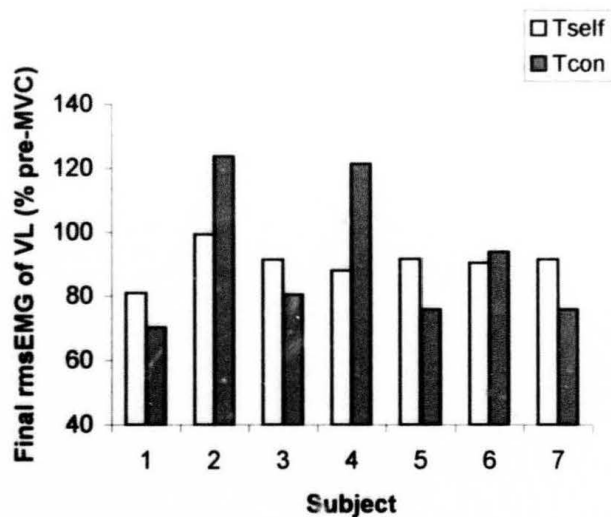


Figure 29. Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in rmsEMG of vastus lateralis (VL) recorded each minute and b) final rmsEMG (VL) for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by rmsEMG activation > 0.7 V. $N = 7$.

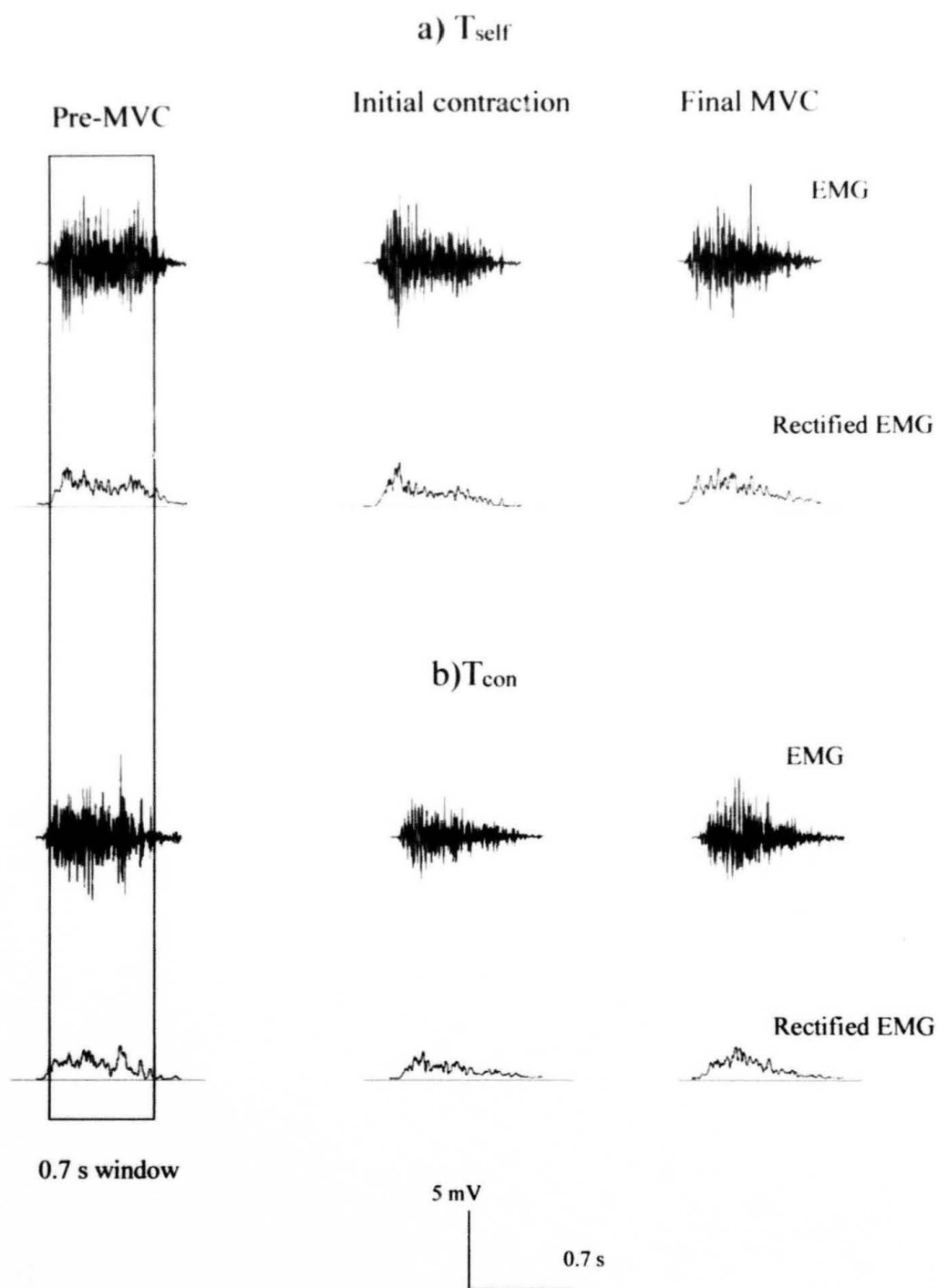
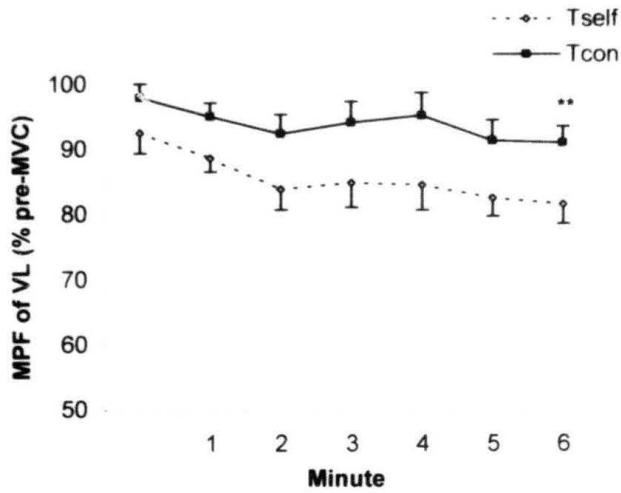


Figure 30. Sample traces of EMG and rectified EMG of vastus lateralis (VL) for pre-maximal voluntary contraction (pre-MVC), initial contraction, and final MVC for the same subject during a 6-minute effort performed on a rowing ergometer using a) the self-pace strategy in T_{self} and b) the constant-pace strategy in T_{con} .

a)



b)

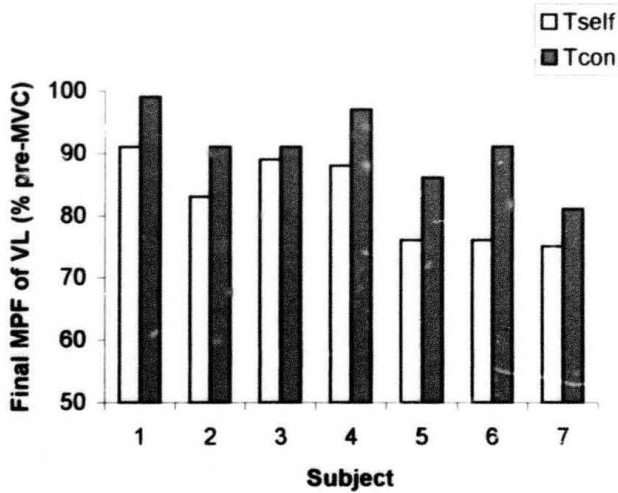


Figure 31. Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for a) changes in MPF of vastus lateralis (VL) recorded each minute and b) final MPF (VL) for each subject. Data averaged for each five contractions and normalized to the value at pre-maximal voluntary contraction (% pre-MVC). Data triggered by rmsEMG activation > 0.7 V. $N = 7$, $** = p < .01$.

8.3.5 Blood Lactate, Heart Rate, and RPE

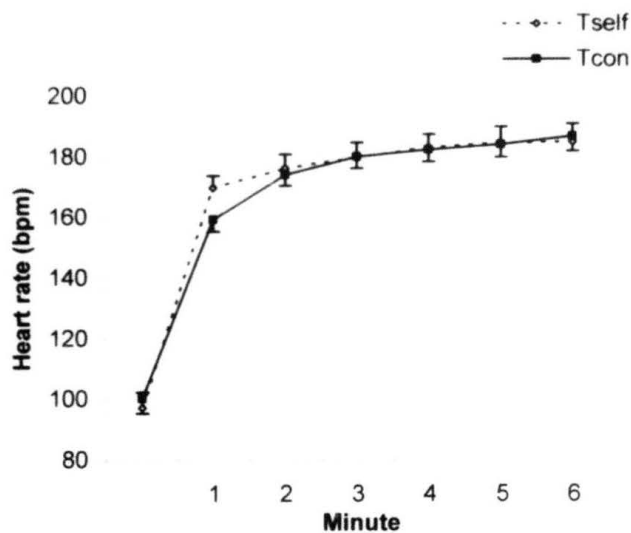
There was no significant difference between trials for final blood lactate concentration at completion of exercise (refer to Table 8). Initial blood lactate was similar between T_{con} ($\bar{m} = 2.2$ mmol/L) and T_{self} ($\bar{m} = 2.4$ mmol/L). Final blood lactate concentration showed a substantial increase above initial values in both trials that was less in T_{con} (11.3 mmol/L) compared to T_{self} (11.9 mmol/L), but no systematic difference was observed between trials. For five out of seven subjects, final blood lactate taken at completion of exercise was lower in T_{con} than in T_{self} (refer to Figure 25b). Post 5-minute mean blood lactate was 9.5 mmol/L in T_{con} and 10.3 mmol/L in T_{self} . For all subjects, post 5-minute blood lactate was equal or lower than final blood lactate.

Final heart rate (mean) was not significantly different in T_{self} (184 bpm) compared to T_{con} (186 bpm). The pattern of change for heart rate each minute is shown in Figure 32a.

The heart rate pattern showed an initial rapid increase followed by a slow upward drift that was similar between both trials except for the first minute, when heart rate was lower (diff.n.s.) in T_{con} than in T_{self} .

Final RPE (mean) was the same in T_{self} and T_{con} (19). The pattern of change in RPE each minute (refer to Figure 32b) was similar in T_{self} compared to T_{con} showing a curvilinear trend in both trials, which was marginally lower at the first minute in T_{con} .

a)



b)

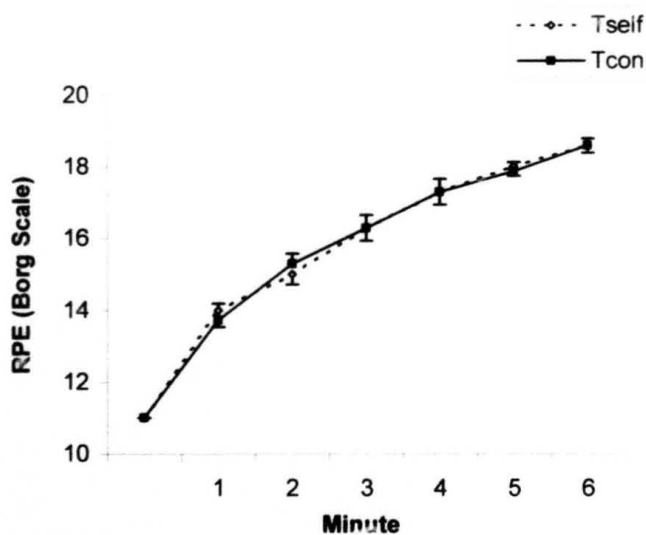


Figure 32. Comparison between self-pace (T_{self}) and constant-pace (T_{con}) 6-minute ergometer trials for changes recorded each minute in a) mean heart rate (bpm) and b) mean rating of perceived exertion (RPE) on the Borg Scale. $N = 7$.

8.4 DISCUSSION

8.4.1 Pacing Effect on Performance

The hypothesis that predicted performance distance achieved would not be significantly different between a constant-pace and a self-pace 6-minute rowing ergometer performance was not supported, with all subjects producing inferior performance in T_{con} compared to T_{self} . The difference in performance distance was marginal (0.4 %), however this difference proved statistically significant. In T_{con} , subjects were requested to maintain but not to exceed a target average-pace (average power output) calculated as the average power output from T_{self} . These instructions resulted in force and power outputs for each contraction in T_{con} being equal or slightly below the required target output. Subjects were not encouraged to increase power output in order to reduce target output deficit, therefore the shorter distance achieved in T_{con} was a result of subjects slightly undershooting the target output. This could be avoided in future experiments by encouraging subjects to compensate for target short falls by briefly increasing power output during the effort.

The current statistical findings for performance were not in agreement with others. For example, Cherry et al. (1997) demonstrated that during constant-pace cycling (equivalent to the average pace of an all-out effort) power could be maintained and external work replicated providing the pedaling rate was controlled. Furthermore, Foster et al. (1993) found that during 2-km cycling time-trials, an even-pace strategy produced the fastest performance time when compared to other pacing strategies that used 7 % faster and 7 % slower splits than the average pace. During race situations negative splits and even-pace performances have often demonstrated superior final

times (Anderson, 1997). In the current study, although the difference for performance distance between trials was statistically significant, the size of the deficit was not great (0.4 %). However, the resulting difference in work output (or distance achieved) in each trial may have influenced the measured outcomes between trials. Reports from other pacing trials suggest that in most situations, performance distance or time may be maintained using a constant-pace strategy compared to a self-pace strategy.

8.4.2 Pacing Effect on Force and Power Output

The hypothesis that predicted significantly greater MVC force and power output following a constant-pace compared to a self-pace 6-minute rowing ergometer performance was supported. Final MVC force and power outputs were greater (by 8 % and 9 % respectively) in T_{con} compared to T_{self} , but were less than pre-MVC values in both trials. These results indicated that fatigue (loss of MVC force) was reduced in T_{con} compared to T_{self} . It appears that provided an average-pace strategy is employed for the majority of a 6-minute effort, there is potential for additional increase in force and power output during the latter stages that may improve performance outcome. Similar findings were reported by Cherry et al. (1997) who investigated cycling tasks at high intensities. They found a lower level of fatigue, demonstrated by significantly less reduction in peak power output after a paced effort compared to an all-out effort with similar work produced.

Rowers manipulate force, handle velocity, and stroke-rate in order to maintain power output during fatiguing efforts (Hartmann et al., 1993). Under fatigue conditions, it is likely that rowers compensate for force loss with increases in stroke rate. Furthermore, Kyrolainen & Smith (1999) demonstrated that during fatiguing ergometer efforts,

maximal knee extensor power and knee extensor activity (IEMG) increased over time and that knee extensor activity was initiated earlier during the stroke. They found these changes were unaffected by stroke rate. In the current study, although subjects were asked to maintain a consistent stroke rate, the stroke rate could not be strictly controlled using the standard rowing ergometer. Any variation in stroke rate may have influenced the timing of RF and VL activity in relation to measured force output, thereby affecting the force-rmsEMG relationship. In addition, performance variables of force output and handle velocity were not controlled and therefore, the effect of each variable could not be determined. However, the effect could be assessed for the combined influences of these variables i.e. power output. In the constant-pace trial, target power outputs and stroke rates were set and monitored on the rowing ergometer display. In spite of the attempt to maintain constant stroke rate and handle velocity in this trial, quadriceps contraction velocity may have varied dependent on the drive and recovery phase of each stroke. The restricted MVC data-windows may have been confounded by any changes in stroke rate or the drive/recovery ratio, which effect the shape of the force/time curve (discussed in detail 3.1.5). There was some evidence that force output, handle velocity, and power output varied during the constant-pace trial. For example, for the best performer in T_{con} (subject 4), the mean (SD) values recorded each minute (not including final MVCs) were 499.04 (10.32) N for force output, 1.84 (0.04) m/s for handle velocity, and 928.27 (26.98) W for power output. Furthermore, power output decreased more than force output and therefore, it was likely that decreases in contraction velocity were partly responsible for decreases in power output (assuming stroke rate was maintained). It was not possible to obtain evidence of changes in contraction velocity of the leg extensors using the current transducers. Evidence of these changes would require force and displacement transducers that measure propulsive seat force and

displacement related to leg extension similar to those for the adapted leg-only ergometer used previously (chapters 4, 5).

During a typical race performance, initial intensity has been reported to range between 87 - 100 % of maximal effort (Hagerman et al., 1978; Millward, 1987). In the present study, when rowers were set a self-pace performance task with the objective of achieving the greatest distance in 6 minutes, initial contractions were paced and less than the maximal effort seen during pre-MVC. The intensity of effort during the first minute of performance determines the extent to which physiological changes, which are associated with fatigue, influence the loss of maximal force. Hence, optimal intensity during the initial contractions of a maximal self-pace effort are crucial to performance and may vary between subjects. The effect of initial contraction intensity, velocity, and stroke rate requires further investigation.

8.4.3 Pacing Effect on rmsEMG

The hypothesis that predicted significantly greater MVC rmsEMG following a constant-pace compared to a self-pace 6-minute rowing ergometer performance was not supported. Initial muscle activation (rmsEMG) was lower in T_{con} compared to T_{self} , which corresponded with lower force and power output, and was sufficient to maintain the target force output. In T_{con} muscle activation increased steadily throughout the effort in order to maintain target force output as contractile mechanisms fatigued. Whereas, following the initial and prior to the final contractions in T_{self} , muscle activation appeared to be regulated at a steady state, possibly optimizing neural input. Nevertheless, final muscle activation (rmsEMG % pre-MVC) was similar in both trials irrespective of pacing influence. During the last 10 seconds of each trial, subjects were

encouraged to perform maximal efforts. The final efforts resulted in increases in both rmsEMG (VL) and force output that were less than respective pre-MVC values. At this point, it was likely that most motor units were recruited and that firing rate increases were responsible for the increase in rmsEMG and force output.

The correspondence between force and rmsEMG (VL) during final MVCs in both current trials was in agreement with previous findings under similar conditions (chapters 6 and 7), and with other investigations during fatiguing dynamic contractions (Potvin, 1997). Most importantly, final MVC rmsEMG (VL) was similar between trials, but force output was significantly greater in T_{con} compared to T_{self} . The difference in final force output resulting from each pacing strategy was more likely related to difference in contractile processes or in qualitative properties of muscle activation, than difference in quantitative properties of muscle activation. The current evidence indicates that ability to increase final force output in each trial was mostly related to the level of contractile fatigue. One possible explanation for greater final force output in T_{con} compared to T_{self} , could be the partial sparing of FT fibers associated with lower initial contraction intensity in the former trial (see section 8.4.4).

The strong positive force-rmsEMG relationship found during initial muscle contractions was similar in both trials. Similar values (% pre-MVC) were shown for initial force and initial rmsEMG in each trial (see figures 26a & 29a, respectively), and was in agreement with previous results for fresh muscle contractions performed on a rowing ergometer (discussed in section 4.4.1). Following the initial contractions in T_{self} , rmsEMG was maintained at a steady level in relation to the slow decline in force and power output. During that period of the trial, modulation of EMG activity may have

been a result of central regulation that optimized force output (discussed previously section 6.4.3). Despite an increase in rmsEMG during final MVC, the value was significantly less than pre-MVC value, which was in agreement with previous findings (Chapter 6 for T_{6-min}). Reduced final rmsEMG (VL) was indicative of reduced central drive or afferent inhibition of motor output. In T_{con} , the pattern of steady increase in rmsEMG (VL) was similar to that observed by Potvin and Bent (1997) during repetitive elbow flexion-extension with a constant load. Increases in EMG amplitude under similar conditions have been attributed to recruitment of additional MUs or increases in MU firing rates in spite of fatigue (De Luca, 1985; Merletti & Roy, 1996).

Four out of seven subjects showed reduced rmsEMG and greater force and power output in final MVC in T_{con} compared to T_{self} . Hence, for the same four subjects greater final force and power output in T_{con} was not associated with greater VL muscle activation, which was more likely associated with reduced contractile fatigue. However, it was also possible that in order to increase force output these subjects used relatively more contribution from muscle groups that were not measured. In contrast, the best performer in the group (subject 4) had greater rmsEMG (VL) activation in T_{con} compared to T_{self} during final maximal contractions. For this subject, the ability to increase central drive indicated less central fatigue, less central regulation, or less contractile fatigue in the constant-pace compared to the self-pace strategy.

Importantly, results indicated that timing triggers for EMG data collection influenced the force-rmsEMG correspondence under present conditions. In both trials, final rmsEMG (% pre-MVC) was substantially greater in EMG data-windows triggered by muscle activation compared to those triggered by handle force. The latter trigger

method recorded very similar decreases in rmsEMG and force output, whereas the former trigger method recorded substantially less decrease in rmsEMG compared to force output. Thus, rmsEMG (VL) triggered by handle force was more related to handle force output, and rmsEMG (VL) triggered by VL activity was more representative of complete VL activation each stroke. It is evident from the force-EMG traces and the trigger effect on rmsEMG values, that some VL muscle activity did not directly contribute to force output at the handle. Hence, contractions may have involved unnecessary VL activation that did not improve performance outcome, for example post-drive isometric activation of the quadriceps muscle. Each trigger method would have different advantages dependent on the purpose of the experiment. The implication of this observation is the specificity of data triggers to experimental purpose. For example, when the relationship between force output measured at the handle and EMG of the leg extensors is examined, a handle force trigger should be used to determine the data-window. However, when fatigue related changes of leg extensor muscle activity is examined, a rmsEMG trigger is preferable to determine the data-window.

8.4.4 Pacing Effect on MPF

The hypothesis that predicted significantly greater MVC MPF following a constant-pace compared to a self-pace 6-minute rowing ergometer performance was supported. In both trials, decline in MPF was slow and did not increase when subjects attempted final MVCs. Decreases in MPF and the rate of decline in MPF have been associated with fatigue under various contraction conditions and were reported previously (see section 5.4.3). In the current study, the rate of MPF decline was not examined but decrease in MPF was represented as a percentage of the value at pre-MVC. MPF (VL) remained higher throughout the 6-minute effort and was greater during final MVCs in

T_{con} compared to T_{self} . One explanation for MPF differences between trials might be the partial sparing of FT fibers in T_{con} . FT fibers tend to be more powerful, possess greater conduction velocities, and fatigue more quickly than ST fibers. During fatigue a slowing of action potential conduction velocity is associated with decline in MPF (Sadoyama et al., 1983) and occurs more rapidly dependent on the relative use of FT fibers. The greater MPF decrease seen in T_{self} compared to T_{con} indicated that FT fibers made a greater contribution in T_{self} . The lower initial intensity in T_{con} compared to T_{self} , likely reduced dependence on FT fibers during the early contractions in T_{con} . However, in agreement with early findings (see 4.3.2) the lower initial intensity in T_{con} compared to T_{self} was not accompanied by lower initial MPF. Evidence of the sparing of FT fibers in T_{con} was provided by the lower final blood lactate value observed in this trial. However, considerable subject variation in final blood lactate was noted (see section 8.4.5). The constant-pace strategy may have reduced physiological stress over the first part of the trial, thus minimizing the effect on contractile and electrical properties of the muscle, which allowed increased activation of larger motor units later in the performance task.

8.4.5 Pacing Effect on Blood Lactate, Heart Rate, and RPE

The hypothesis that predicted significantly less blood lactate concentration following a constant-pace compared to a self-pace 6-minute rowing ergometer performance was not supported. In both trials, significant involvement of the anaerobic energy systems was indicated by blood lactate values greater than 4.0 mmol/L at completion of exercise. Reduced final blood lactate concentration in T_{con} compared to T_{self} (diff.n.s.) indicated that anaerobic involvement was less T_{con} , although considerable individual variation was evident. The three best performers (subjects 3, 4, and 5) demonstrated the greatest

reduction in final blood lactate between T_{con} and T_{self} (Figure 28b). This finding suggests that better performers may benefit from reduced blood lactate accumulation more than less-able performers in response to constant-pace rowing. However, Hahn et al. (1995a) reported that more-able rowers were not limited by acid build-up and were more able to tolerate a high acid environment in the muscles, together with associated blood lactate accumulation. The ability to tolerate high levels of muscle lactate and acidic environments may influence the ability to maintain central nervous drive or to inhibit sensory feedback mechanisms that modulate muscle activation (Bigland-Ritchie & Woods, 1984; Garland, Garner, & McComas, 1988).

In agreement with the current findings, Foster et al. (1993) found no systematic difference in 3-minute post-exercise blood lactate or any other physiological parameter that could account for improved performance during constant-pace efforts. In contrast, Cherry et al. (1997) found that 3-minute post-exercise blood lactate, pH, and ammonia were less disturbed in response to an even-pace effort compared to an all-out effort. These workers speculated that differences in blood lactate concentrations between trials were associated with the partial sparing of FT fibers during initial submaximal contractions in the constant-pace trial. They suggested that lower blood lactate values in the constant-pace trial were related to slower muscle action speed and to submaximal activation of some MUs. The different findings of Foster et al. and Cherry et al. may be partly attributed to the duration of exercise for each study, which influenced the contribution from anaerobic and aerobic energy systems as well as blood lactate kinetics. The beneficial effect of constant pacing on blood lactate concentration appears to be greater in events that maximally stress the anaerobic system and is reduced in events that involve relatively greater aerobic contribution.

It was expected that mean heart rate would be significantly lower in response to constant-pace compared to self-pace ergometer performance, but this was not the case. The rapid increase in heart rate during the first minute was less in T_{con} compared to T_{self} , reflecting the difference in contraction intensity, before a slow upward drift that was replicated in both trials. The trend in heart rate was reflected by RPE. During the 6-minute rowing ergometer performance, peak heart rate was similar in response to both pacing strategies. This finding was in agreement with Cherry et al. (1997) who found similar peak heart rates between even-pace and all-out pace performances on a cycle ergometer. The heart rates observed in the current trials were comparable to those reported previously in response to 2000-m ergometer rowing performances (Hagerman et al., 1978; Hahn et al., 1995b).

It was expected that mean RPE would be significantly lower in response to the constant-pace compared to the self-pace ergometer performance. This was not the case. It appears that subjects did not find the rowing task any easier to complete when the pace was constant. Subject mean RPE were alike in both trials, although RPE reported for the first two minutes were marginally less in T_{con} than in T_{self} . Mahler, Andrea, & Andresen (1984) found that peak RPE values were comparable between a 6-minute all-out rowing ergometer trial and a progressive incremental trial. Their findings suggest that under conditions that involve near-maximal intensities, peak RPE tends to be similar irrespective of earlier intensities during the task. Marriott and Lamb (1996) reported that changes in RPE were strongly related to increased intensity when measured in terms of heart rate or power output during rowing performance. The current findings were not in total accordance with Marriott and Lamb, in that final RPE was similar between trials but corresponding final power output was significantly

different. Final and mean RPE were similar between the current trials and reflective of average power output for each trial. Sense of effort expressed as RPE reflects integrated signals from both central (heart and lungs) and local (exercising muscles) regions. At exercise levels above 50 % $\dot{V}O_{2\text{ max}}$, minute ventilation (VE) strongly influences the central signal of perceived exertion (Robertson, 1982) and arguably, muscle lactate strongly influences the local signal of perceived exertion (Cafarelli, 1982). In the present study, RPE and total work produced were similar in control-pace and self-pace trials. RPE that reflected physiological (central and local) and psychological signals was not influenced by pacing strategy.

8.5 CONCLUSIONS

In comparing a constant-pace with a self-pace rowing effort, MVC force and power outputs were significantly greater following the former strategy. The findings indicated a potential latent power that may be used during the latter stages of an effort to improve performance outcome. The differences in force and power outputs between pacing strategies appeared not to be related to quantitative differences in VL muscle activity (rmsEMG), but were more likely related to qualitative differences (MPF) in that muscle. The potential for increases in power output following constant-pace efforts may be due to partial sparing of FT fibers and better maintained action potential conduction velocity. This was reflected by reduced decline in the MPF. Blood lactate concentration, heart rate, and RPE are useful parameters for monitoring and assessing rowing performance, and provide an indication of the physiological stress during rowing performance. In the current trials, these parameters provided no evidence of reduced physiological stress using the constant-pace strategy. It appears that constant-

pace strategies may marginally reduce blood lactate accumulation and significantly increase latent power output during the final strokes of rowing performance, particularly in highly trained rowers. Constant-pace strategies may potentially improve performance outcomes during ergometer trials, however racing situations are more involved due to tactics and psychological implications.

CHAPTER 9

SUMMARY AND RECOMMENDATIONS

EMG has the potential to monitor changes in muscle activation characteristics during rowing performance and may provide important data on neuromuscular strategies adopted in response to fatigue. Muscle activation patterns in relation to magnitude, loss, and timing of force output are critical to rowing performance. The aim of this thesis was to assess the application of EMG in relation to biomechanical and physiological responses to rowing tasks, in particular the effects of fatigue on force output and EMG characteristics of the quadriceps muscle. Each of the five experimental studies performed had specific purposes. These included establishing force/torque-EMG relationships under controlled conditions and evaluating the transfer of force or torque output and EMG characteristics to less controlled performance conditions that were influenced by fatigue, pacing strategy, or both.

During rowing events, fatigue influences the physiological and biomechanical characteristics of performance. For optimal performance, rowers must maintain a coordinated muscle action that reproduces near-maximal force outputs for approximately 6-minutes. The ability to produce high force outputs during repetitive contractions is influenced by fatigue and dependent on a number of factors including neuromuscular activation, which may be expressed by amplitude and frequency characteristics of the EMG. EMG characteristics are related to changes in force during sustained isometric contractions, which may be useful in monitoring the fatigue process. The problem is that the force-EMG relationship is not as clear when applied to dynamic movements

such as those in rowing performance. In these situations, dynamic contractions involve changes in muscle length, contraction velocity, or muscle contribution, which may influence the force-EMG relationship. In order to evaluate optimal rowing performance, it would be useful to monitor changes in neural activity associated with force output during rowing ergometer tasks. Moreover, it would be beneficial to devise strategies that minimize disruptions to neuromuscular function (and thus stroke characteristics) associated with fatigue. EMG analysis is constrained by methodological and confounding factors during dynamic movement, but may provide an insight into neural activation strategies during rowing fatigue.

In this investigation, trials were conducted on three ergometers affording varying levels of control over muscle length, contraction velocity, and muscle contribution to force output. Subjects were selected from trained rowing crews. Trials were performed on an isokinetic dynamometer, an adapted leg-only ergometer and a standard rowing ergometer with performance outcomes recorded using a potentiometer to measure handle position and a strain gauge to measure force output. Bipolar surface electrodes were used to record EMG activity of the quadriceps muscle during all contractions. Biomechanical and EMG data were recorded on a data acquisition system (Amlab).

The purpose of the first study was to examine the transfer of force/torque-EMG relationships from well-controlled situations such as isometric contractions, to less-controlled situations such as dynamic rowing contractions. Force/torque-EMG relationships were examined for significance and compared between ergometer types during isometric, isokinetic, and dynamic random force contractions in fresh quadriceps muscle. Results indicated a strong positive linear correlation between force or torque

and EMG amplitude (rmsEMG) under non-fatiguing isometric and isokinetic conditions. Although diminished in strength, the force/torque-rmsEMG relationship was maintained under dynamic leg-only and standard ergometer conditions. However, there was a significant effect for ergometer type on the force/torque-rmsEMG relationship. In addition, no correlation was found between force or torque and EMG frequency (MPF) under controlled isometric conditions. These findings support the use of quadriceps EMG amplitude, although not EMG frequency, analysis in relation to total force output measured during simulated rowing contractions. Hence, during non-fatiguing contractions, quadriceps muscle activation (rmsEMG) may be used to predict force output and the relationship applied to ergometer rowing. However, characteristic force/torque-EMG changes are not so clear during fatiguing tasks.

EMG may provide an insight into muscle functions and neural activation strategies that influence performance outcome during fatigue. The purpose of the second study was to examine the effects of fatigue on force/torque-EMG relationships using the same ergometer types as in the previous study. Repeated isokinetic and dynamic MVCs were performed on the three ergometers. Force/torque-EMG relationships were examined for significance and compared between ergometer types. Results indicated that during fatiguing isokinetic dynamometer contractions, there was a significant positive relationship for torque-rmsEMG and torque-MPF of the quadriceps muscle. The strength of relationships was maintained but reduced for dynamic contractions on a standard rowing ergometer, with no significant effect for ergometer type. However, some subjects demonstrated opposite force-rmsEMG and force-MPF associations in response to fatiguing contractions under these conditions. It can be concluded that during fatiguing contractions, the pattern of neuromuscular activation was more

consistent in controlled isokinetic situations compared to performance situations. Nevertheless, the force-rmsEMG relationship was valid for fatiguing ergometer tasks. EMG responses to fatigue varied between subjects and possible sites of fatigue were discussed. The rmsEMG and MPF changes demonstrated in response to repeated MVCs may not be the same in response to a typical self-pace rowing performance.

Rowing performance tasks, unlike other commonly used laboratory based exercise protocols, involve self-pace contractions that attempt to reproduce optimal force output for the duration of the event. The purpose of the third study was to investigate whether responses in force output and EMG characteristics to a commonly used fatigue protocol were similar to a typical rowing performance task. Subjects performed a 2-minute repeated MVC protocol and on a subsequent occasion, a 6-minute self-pace protocol on the same rowing ergometer. Results demonstrated that in spite of a similar loss of MVC force and decrease in MPF under both conditions, final MVC rmsEMG was significantly different between trials. EMG amplitude (rmsEMG) increased (relative to initial MVC) under the repeated MVC protocol and decreased (relative to initial MVC) under the self-pace protocol. The findings indicated that sites of fatigue and neural optimization strategies might be influenced by the task performed. Therefore, analysis of EMG activity would be more appropriate when the task is specific to the condition of performance.

In order that changes in force and EMG characteristics be of practical use in rowing performance assessment, the reliability of change in force output and quadriceps EMG was examined under self-pace performance conditions. The purpose of the fourth study was to determine reliability of performance responses to a 6-minute self-pace maximal

rowing ergometer effort. Under these conditions, performance distance and final MVC MPF of VL were reliable. Blood lactate at completion of exercise, force loss, and final MVC rmsEMG were not reliable in this study. These findings support the analysis of quadriceps MPF in relation to handle force output during simulated rowing performance. However, MPF decreases in the current trials were not of sufficient magnitude to be of practical value for assessing fatigue of the VL muscle during self-pace rowing performance.

EMG found in response to pacing strategies may provide an indication of muscle activation strategies employed by the central nervous system that optimize force output and performance outcome. The purpose of the fifth study was to compare force and power loss, and change in EMG characteristics between a typical self-pace and a controlled constant-pace rowing ergometer performance. Self-pace and constant-pace trials were performed and compared with a view to minimizing the physiological changes associated with fatigue. Results indicated that during final MVCs, force and power outputs were significantly greater, and MPF was significantly higher in response to constant-pace compared to self-pace efforts. In conclusion, constant-pace strategies significantly reduced the loss in MVC force and power output near completion of performance, which potentially may improve performance outcome. The greater force or power output following the constant-pace task appeared not to be related to quantitative differences in muscle activity (rmsEMG) of VL, but were more likely related to qualitative differences (MPF) in that muscle. EMG data-windows that included all quadriceps muscle activation, irrespective of force output, demonstrated less magnitude of change compared to EMG data-windows triggered by force. It appears that constant-pace strategies may marginally reduce blood lactate accumulation

and significantly increase latent power output during the final strokes of rowing performance, particularly in highly trained rowers. Although constant-pace strategies may improve performance outcomes during ergometer trials, racing situations are more complicated due to tactics and psychological implications.

This series of experiments attempted to establish force and muscle activation changes during the rowing movement by monitoring EMG characteristic changes. The findings were in agreement with well-established significant force/torque-EMG relationships during isometric and isokinetic contractions, but strength of the relationships was reduced under rowing ergometer conditions. Changes in quadriceps EMG (rmsEMG, MPF) were related to decline in maximal handle force output during rowing ergometer tasks. However, the changes either were inconsistent between trials, or varied between subjects, or insufficient in magnitude so that EMG could not be used in practical situations to predict performance outcome. Nonetheless, EMG provided an indication of neural activation changes related to central and peripheral mechanisms of fatigue. It appears that during self-pace maximal efforts, neuromuscular mechanisms may regulate central drive to optimize muscle activation, force output, and performance outcome. In addition, it is likely that pacing strategies may minimize contractile failure of the muscle, and thereby maximize performance outcome. In spite of the current limitations of EMG to provide quantified valid assessment of muscle activation during rowing performance, EMG shows potential to monitor fatigue and timing of muscle activation in relation to force output.

In the future, further investigations are required to substantiate if inter-subject variations in force/torque-rmsEMG and force/torque-MPF relationships are a result of subject,

protocol, or method variability. It is evident that there are technological deficiencies related to signal stationarity, which limit the quantification of EMG during dynamic movement. In the near future it is likely that micro-chip technology will enable snapshot measurement of the EMG signal that capture stable samples of motor unit activity during dynamic movement. This would allow the prediction of force loss in specific muscle groups during applied situations such as rowing and allow manipulation of strategies to minimize the effect of fatigue.

One interesting and less problematic use of EMG to improve performance is the assessment of muscle activation timing. Currently, the onset, duration, and cessation of muscle activity may be accurately measured. Timing pattern changes may be related to fatigue and are critical to rowing performance. Most importantly, knowledge of the timing pattern changes between synergistic muscle groups would enable examination of co-ordinated muscle action that should maximize force output measured at the handle. Changes in timing of muscle activation sequence influence the application of efficient rowing technique and may be detrimental to performance outcomes. Much time is spent during training to optimize technique, yet little data is available on alterations of muscle activation timing during fatigue. Timing of all muscle group activation employed in the rowing movement could be monitored and relative amplitude between muscle groups assessed by using EMG recordings from several different muscle sites. In addition, other force transducers that isolate force outputs together with EMG could be used to monitor specific muscle contribution to force output. For example, forces at the foot stretcher or at the seat might be more related than forces at the handle to leg extensor activity.

Identifying the sites of fatigue during maximal rowing provides an interesting challenge for research. To be able to identify failure of central drive or contractile mechanisms or the influence of central regulation during rowing performance would encourage the development of training strategies to minimize the effect of fatigue. Twitch interpolation techniques, used during intermittent isometric contractions throughout a maximal ergometer effort, could provide evidence of the loss of central drive and could be used in self-paced rowing performance to assess the level of central regulation.

Analysis of on-water muscle activation patterns provides the greatest challenge for rowing researchers and may provide the key to optimal levels of neuromuscular excitation and relaxation that minimizes fatigue and maximizes rowing performance.

BIBLIOGRAPHY

- Allen, D. G., Westerblad, H., & Lannergren, J. (1995). The role of intracellular acidosis in muscle fatigue. *Advances in Experimental Medicine and Biology*, 384, 57-68.
- Anderson, O. (1997). Moderate starts, consistent pacing, and negative splits can carry you to new PBs. *Peak Performance*, 96, 1-3.
- Andersson, E. A., Nilsson, J., & Thorstensson, A. (1997). Intramuscular EMG from the hip flexor muscles during human locomotion. *Acta Physiologica Scandinavica*, 161, 361-370.
- Arendt-Nielsen, L., & Mills, K. R. (1988). Muscle fibre conduction velocity, mean power frequency, mean EMG voltage and force during submaximal fatiguing contractions of human quadriceps. *European Journal of Applied Physiology*, 58, 20-25.
- Arendt-Nielsen, L., & Sinkjaer, T. (1991). Quantification of Human Dynamic Muscle Fatigue by Electromyography and Kinematic Profiles. *Journal of Electromyography and Kinesiology*, 1(1), 1-8.
- Bangsbo, J., Johansen, L., Graham, T., & Saltin, B. (1993). Lactate and H⁺ effluxes from human skeletal muscles during intense dynamic exercise. *Journal of Physiology*, 462, 115-133.
- Basmajian, J. V. (1974). *Muscles Alive: Their functions revealed by Electromyography*. (3rd ed.). Baltimore: Williams and Wilkins.
- Basmajian, J. V., & De Luca, C. J. (1985). *Muscles alive*. (5th ed.). Baltimore: Williams & Wilkins.

- Beneke, R. (1995). Anaerobic threshold, individual anaerobic threshold and maximal lactate steady state in rowing. *Medicine and Science in Sports and Exercise*, 27(6), 863-867.
- Bigland-Ritchie, B., Donovan, E. F., & Roussos, C. S. (1981). Conduction velocity and EMG power spectrum changes in fatigue of sustained maximal efforts. *Journal of Applied Physiology*, 51 (5), 1300-1305.
- Bigland-Ritchie, B., Furbush, F., & Woods, J. J. (1986). Fatigue of intermittent submaximal voluntary contractions: central and peripheral factors. *Journal of Applied Physiology*, 61 (2), 421-429.
- Bigland-Ritchie, B., Johansson, R., Lippold, O. C. J., & Woods, J. J. (1983). Contractile Speed and EMG Changes During Fatigue of Sustained Maximal Voluntary Contractions. *Journal of Neurophysiology*, 313-324.
- Bigland-Ritchie, B., Rice, C. L., Garland, S. J., & Walsh, M. L. (1995). Task-Dependent Factors in Fatigue of Human Voluntary Contractions. In S. C. Gandevia, R. M. Enoka, A. J. McComas, D. G. Stuart, & C. K. Thomas (Eds.), *Advances in Experimental Medicine and Biology: Fatigue Neural and Muscular Mechanisms* (Vol. 384, pp. 360-380).
- Bigland-Ritchie, B., & Woods, J. J. (1984). Changes in muscle contractile properties and neural control during human fatigue. *Muscle and Nerve*, 7, 691-699.
- Bompa, T. O. (1980). Technique and Muscle Force. *Canadian Journal of Applied Science*, 5(4), 245-249.
- Borg, G. (1985). *An Introduction to Borg's RPE Scale*. Michigan: Movement Publications.

- Bosco, C., & Viitasalo, J. T. (1982). Potentiation of myoelectrical activity of human muscles in vertical jumps. *Electromyography and Clinical Neurophysiology*, 22, 549-562.
- Bouissou, P., Estrade, P. Y., Goubel, F., Guezennec, C. Y., & Serrurier, B. (1989). Surface EMG power spectrum and intramuscular pH in human vastus lateralis muscle during dynamic exercise. *Journal of Applied Physiology*, 67(3), 1245-1249.
- Broman, H., Bilotto, G., & De Luca, C. J. (1985). Myoelectric signal conduction velocity and spectral parameters: influence of force and time. *Journal of Applied Physiology*, 58, 1428-1437.
- Bruton, J. D., Lannergren, J., & Westerblad, H. (1998). Mechanisms underlying the slow recovery of force after fatigue: importance of intracellular calcium. *Acta Physiologica Scandinavica*, 162, 285-293.
- Burton, G. (1996). *Free and fixed paced strategies in 2000 meter rowing*. Unpublished Masters, Loughborough University, Loughborough, England.
- Cafarelli, E. (1982). Peripheral contributions to the perception of effort. *Medicine and Science in Sport and Exercise*, 14(5), 382-389.
- Cherry, P. W., Lakomy, H. K. E., Neville, M. E., & Fletcher, R. J. (1997). Constant External Work Cycle Exercise - The Performance and Metabolic Affect of All-out and Even-Paced Strategies. *European Journal of Applied Physiology and Occupational Physiology (Berlin)*, 75(1), 22-27.
- Christensen, H., Sogaard, K., Jensen, B. R., Finsen, L., & Sjogaard, G. (1995). Intramuscular and Surface EMG Power Spectrum from Dynamic and Static Contractions. *Journal of Electromyography and Kinesiology*, 5(1), 27-36.

- Chwalbinska-Moneta, J., Hanninen, O., & Penttila, I. (1994). Relationship Between EMG and Blood Lactate Accumulation During Incremental Exercise in Endurance and Speed Trained Athletes. *Clinical Journal of Sports Medicine*, 4(1), 31-38.
- Cram, J. R., Kasman, G. S., & Holtz, J. (1998). *Introduction to Surface Electromyography*. Maryland: Aspen Publishers.
- Cybox Division of Lumex, New York. (1991). *Cybox 6000 Extremity Testing and Rehabilitation System. Users Guide*. New York.
- Dal Monte, A., & Komor, A. (1989). Rowing and Sculling Mechanics. *Biomechanics of Sport*. Florida: CRC Press Inc.
- Dawson, R. G., Lockwood, N. J., Wilson, J. D., & Freeman, G. (1998). The Rowing Cycle: Sources of Variance and Invariance in Ergometer and On-the-Water Performance. *Journal of Motor Behavior*, 30(1), 33-43.
- De Luca, C. J. (1985). Myoelectrical manifestations of localized muscular fatigue in humans. *Critical Reviews in Biomechanical Engineering*, 11(4), 251-276.
- De Luca, C. J. (1997). The Use of Surface Electromyography in Biomechanics. *Journal of Applied Biomechanics*, 13, 135-163.
- De Luca, C. J., LeFever, R. S., McCue, M. P., & Xenakis, A. P. (1982). Behaviour of human motor units in different muscles during linearly varying contractions. *Journal of Physiology*, 329, 113-128.
- Dempsey, J. A., & Babcock, M. A. (1995). An integrative view of limitations to muscular performance. *Advances in Experimental Medicine and Biology*, 384, 393-399.

- DeVries, H. A. (1968). Method for evaluation of muscle fatigue and endurance from electromyographic fatigue curves. *American Journal of Physical Medicine*, 47, 125-135.
- Dolan, P., & Adams, M. A. (1998). Repetitive lifting tasks fatigue the back muscles and increase the bending moment acting on the lumbar spine: *Journal of Biomechanics*, 31, 713-721.
- Dolan, P., & Adams, M. A. (1993). The relationship between EMG activity and extensor moment generation in the erector spinae muscles during forward bending activities. *Journal of Biomechanics*, 26, 513-522.
- Dolan, P., Mannion, A. F., & Adams, M. A. (1995). Fatigue of the erector spinae muscles. A Quantitative Assessment using "Frequency Banding " of the Surface Electromyographic Signal. *SPINE*, 20(2), 149-159.
- Dolmage, T., & Cafarelli, E. (1991). Rate of fatigue during repeated submaximal contractions of human quadriceps muscle. *Canadian Journal of Physiology and Pharmacology*, 69, 1410-1415.
- Doud, J. R., & Walsh, J. M. (1995). Muscle fatigue and muscle length interaction: effect on the EMG frequency components. *Electromyography and Clinical Neurophysiology*, 35, 331-339.
- Dowling, J. J. (1988). *Comparison of muscular and segmented work estimates during dynamic elbow movement*. Paper presented at the 5th Biennial Conference of the Canadian Society for Biomechanics.
- Ebenbichler, G. R., Kollmitzer, J., Glockler, L., Bochdansky, T., Kopf, A., & Fialka, V. (1998). The role of the biarticular agonist and co-contracting antagonist pair in isometric muscle fatigue. *Muscle and Nerve*, 21, 1706-1713.

- Enoka, R. M., & Stuart, D. G. (1992). Neurobiology of muscle fatigue. *Journal of Applied Physiology*, 72, 1631-1648.
- Erberstein, A., & Beattie, B. (1985). Simultaneous measurement of muscle conduction velocity and EMG power spectrum changes during fatigue. *Muscle and Nerve*, 8, 768-773.
- Esposito, F., Malgrati, D., Veicsteinas, A., & Orizio, C. (1996). Time and frequency domain analysis of electromyogram and sound myogram in the elderly. *European Journal of Applied Physiology*, 73(503-510).
- Finucane, S. D. G., Rafeei, T., Kues, J., Lamb, R. L., & Mayhew, T. P. (1998). Reproducibility of electromyographic recordings of submaximal concentric and eccentric muscle contractions in humans. *Electroencephalography and Clinical Neurophysiology*, 109, 290-296.
- Fitts, (1994). Cellular mechanisms of muscle fatigue. *Physiological Review*, 74(1).
- Fitts, R. H. (1996). Muscle fatigue: The Cellular aspects. *The American Journal of Sports Medicine*, 24(6), S9-S13.
- Foster, C., Schrager, M., Snyder, A. C., & Thompson, N. (1994). Pacing Strategy and Athletic Performance. *Sports Medicine (Auckland)*, 17(2), 77-85.
- Foster, C., Snyder, A. C., Thompson, N. N., Green, M. A., Foley, M., & Schrager, M. (1993). Effect of pacing strategy on cycle time trial performance. *Medicine and Science in Sports and Exercise*, 25(3), 383-388.
- Gabriel, D. A., & Boucher, J. P. (1998). Effects of repetitive dynamic contractions upon electromechanical delay. *European Journal of Applied Physiology and Occupational Physiology*, 79(1), 37-40.

- Gamet, D., Duchene, J., Garapon-Bar, C., & Goubel, F. (1990). Electromyogram power spectrum during dynamic contractions at different intensities of exercise. *European Journal of Applied Physiology*, 61, 331-337.
- Gamet, D., Duchene, J., & Goubel, F. (1996). Reproducibility of kinetics of electromyogram spectrum parameters during dynamic exercise. *European Journal of Applied Physiology*, 74, 504-510.
- Gandevia, S. C. (1998). Neural control in human muscle fatigue: changes in muscle afferents, moto neurones and moto cortical drive. *Acta Physiologica Scandinavica*, 162, 275-283.
- Garland, S. J., Garner, S. H., & McComas, A. J. (1988). Reduced voluntary electromyographic activity after fatiguing stimulation of human muscle. *Journal of Physiology*, 401, 547-556.
- Gauthier, G. M. (1985). Visually and acoustically augmented performance feedback as an aid in motor control learning: a study of selected components of the rowing action. *Journal of Sports Science*, 3, 3-26.
- Hagberg, M. (1981). Muscular Endurance and Surface Electromyography in Isometric and Dynamic Exercise. *Journal of Applied Physiology*, 51(1), 1-7.
- Hagerman, F. C. (1986). Applied Physiology of Rowing. *International Journal of Applied Medicine and Science in Sport and Exercise*, 1 (4), 303-326.
- Hagerman, F. C. (1996). Physiology and Nutrition for Rowing. *Perspectives in Exercise Science* (pp. 221-302).
- Hagerman, F. C., Connors, M. C., Gault, J. A., Hagerman, G. R., & Polinski, W. J. (1978). Energy expenditure during simulated rowing. *Journal of Applied Physiology*, 45(1), 87-93.

- Hagerman, F. C., Fielding, R. A., Fiatarone, M. A., Gault, J. A., Kirkendall, D. T., Ragg, K. E., & Evans, W. J. (1996). A 20-year longitudinal study of Olympic oarsmen. *Medicine and Science in Sports and Exercise*, 28(9), 1150-1156.
- Hagg, G. M. (1992). Interpretation of EMG spectral alterations and alteration indexes at sustained contraction. *Journal of Applied Physiology*, 73(4), 1211-1217.
- Hahn, A., Ryan, R., Bellenger, S., Lawton, E., & Kearns, A. (1995a). *Study of the effectiveness of sodium bicarbonate as an ergogenic aid*. Tasmania Rowing Science Camp (Handout).
- Hahn, A., Ryan, R., Lawton, E., Kearns, A., & Bellenger, S. (1995b). *Are the physiological demands of a 2000 meter ergometer test similar to those of racing?* Tasmania Rowing Science Camp (Handout).
- Hahn, A., Ryan, R., Lawton, E., Kearns, A. & Bellenger, S. (1996). *Prediction of Race performance from physiological and anthropometric variables*. Tasmania Rowing Science Camp (Handout).
- Hahn, A. G., Tumity, D. M., Shakespear, P., Rowe, P., & Telford, R. D. (1988). Physiological testing of oarswomen on Gjessing and Concept II rowing Ergometers. *EXCEL*, 5(1), 19-23.
- Hakkinen, K. (1994). Neuromuscular fatigue in males and females during strenuous heavy resistance loading. *Electromyography and Clinical Neurophysiology*, 34, 205-214.
- Hartmann, U., Mader, A., Wasser, K., & Klauer, I. (1993). Peak Force, Velocity and Power During Five and Ten Maximal Rowing Ergometer Strokes by World Class Female and Male Rowers. *International Journal of Sports Medicine*, 14, S42-S45.

- Helal, J. N., Guezennec, C. Y., & Goubel, F. (1987). The aerobic - anaerobic transition: Re-examination of the threshold concept including an electromyographic approach. *European Journal of Applied Physiology*, 56, 643-649.
- Henneman, E., Somjen, G., & Carpenter, D. O. (1965). Functional significance of cell size in spinal motoneurons. *Journal of Neurophysiology*, 28, 560-580.
- Henry, J. C., Clark, R. R., McCabe, R. P., & Vanderby, R. (1995). An evaluation of instrumented tank rowing for objective assessment of rowing performance. *Journal of Sports Sciences*, 13, 199-206.
- Horita, T., & Ishiko, T. (1987). Relationships between muscle lactate accumulation and surface EMG activities during isokinetic contraction in man. *European Journal of Applied Physiology and Occupational Physiology*, 56(1), 18-23.
- Horstmann, T. (1998). *EMG Reliability*. Paper presented at the American Sports Medicine Association, Orlando, Florida.
- Housh, T. J., De Vries, H. A., Johnson, G. O., Evans, S. A., Housh, D. J., Stout, J. R., Bradway, R. M., & Evetovich, T. K. (1996). Neuromuscular Fatigue Thresholds of the Vastus Lateralis, Vastus Medialis and Rectus Femoris Muscles. *Electromyography and Clinical Neurophysiology*, 36, 247-255.
- Ishiko, T. (1971). Biomechanics of Rowing. *Medicine and Sport*, 6, 249-252.
- Jansen, R., Ament, W., Verkerke, G. J., & Hof, L. (1997). Median power frequency of the surface electromyogram and blood lactate concentration in incremental cycle ergometry. *European Journal of Applied Physiology and Occupational Physiology (Berlin)*, 75(2), 102-108.
- Jensen, R. L., Freedson, P. S., & Hamill, J. (1996). The prediction of power and efficiency during near-maximal rowing. *European Journal of Applied Physiology*, 73, 98-104.

- Jones, D. A., & Bigland-Ritchie, B. (1986). Electrical and Contractile Changes in Muscle Fatigue, *Biochemistry of Exercise. International series of Sports Science* (Vol. 16, pp. 377-392).
- Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., Gainey, J., Gorton, G., & Cochran, G. V. B. (1989). Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *Journal of Orthopaedic Research*, 7, 849-860.
- Kollmitzer, J., Ebenbichler, G. R., & Kopf, A. (1999). Reliability of surface electromyographic measurements. *Clinical neurophysiology*, 110(4), 725-734.
- Komi, P. V., & Tesch, P. (1979). EMG Frequency Spectrum, Muscle Structure and Fatigue During Dynamic Contractions in Man. *European Journal of Applied Physiology*, 42, 41-50.
- Kostka, C. E., & Cafarelli, E. (1982). Effect of pH on sensation and vastus lateralis electromyogram during cycling exercise. *Journal of Applied Physiology: Respirat. Environ. Exercise Physiology*, 52(5), 1181-1185.
- Kupa, E. J., Roy, S. H., Kandarian, S. C., & De Luca, C. J. (1995). Effects of muscle fiber type and size on EMG median frequency and conduction velocity. *Journal of Applied Physiology*, 78(1), 23-32.
- Kyrolainen, H., & Smith, R. (1999). Mechanical power output and muscle activities during maximal rowing with different stroke rates. *Journal of Human Movement Studies*, 36 (2), 75-94.
- Lee, G., Minamitani, H., Ju, K., Wakano, K., Onishi, S., & Yamazaki, H. (1996). Fatigue evaluation of lumbar muscles during repeated dynamic trunk exercise. *Electromyography and Clinical Neurophysiology*, 36, 237-245.
- Lee, J. (1994). Roots of Fatigue. *Inside Science*, 71, 1-4.

- Lindstrom, L., Magnusson, R., & Petersen, I. (1970). Muscular fatigue and action potential conduction velocity changes studied with frequency analysis of EMG signals. *Electromyography*, 4, 341-356.
- Loeb, G. E., & Gans, C. (1986). *Electromyography for Experimentalists*. Chicago: The University of Chicago Press.
- Lormes, W., Buckwitz, R., Rehbein, H., & Steinacker, J. M. (1993). Performance and Blood lactate on Gjessing and Concept II Rowing Ergometers. *International Journal of Sports Medicine*, 14, S29-S31.
- Lutoslawska, G., Klusiewicz, A., Sitkowski, D., & Krawczyk, B. (1996). The effect of simulated 2-km laboratory rowing on blood lactate, plasma, inorganic phosphate and ammonia in male and female junior rowers. *Biology of Sport (Warsaw)*, 13(1), 31-38.
- MacFarlane, D. J., Edmond, I. M., & Walmsley, A. (1997). Instrumentation of an ergometer to monitor the reliability of rowing performance. *Journal of Sport Science*, 15, 167-173.
- Mahler, D. A., Andrea, B. E., & Andresen, D. C. (1984). Comparison of 6 min "all out" and incremental exercise tests in elite oarsmen. *Medicine and Science in Sports*, 16(6), 567-571.
- Mannion, A. F., & Dolan, P. (1994). Electromyographic Median Frequency Changes During Isometric Contraction of the Back Extensors to Fatigue. *Spine*, 19(11), 1223-1229.
- Mannion, A. F., & Dolan, P. (1996a). The Effects of Muscle Length and Force Output on the EMG Power Spectrum of the Erector Spinae. *Journal of Electromyography and Kinesiology*, 6(3), 159-168.

- Mannion, A. F., & Dolan, P. (1996b). Relationship between myoelectric and mechanical manifestations of fatigue in the quadriceps femoris muscle group. *European Journal of Applied Physiology*, 74, 411-419.
- Mannion, A. F., Dumas, G. A., Stevenson, J. M., & Cooper, R. G. (1998). The influence of Muscle Fiber Size and Type Distribution on Electromyographic Measures of Back Muscle Fatigability. *Spine*, 23(5), 576-584.
- Marr, A., & Stafford, P. (1982). A kinematic and electromyographical comparison of a junior and novice rower. *Pelops*, 4, 1-6.
- Marriott, H. E., & Lamb, K. L. (1996). The use of ratings of perceived exertion for regulating exercise levels in rowing ergometry. *European Journal of Applied Physiology and Occupational Physiology*, 72(3), 267-271.
- Martindale, W. O., & Robertson, D. G. E. (1984). Mechanical energy in sculling and in rowing an ergometer. *Canadian Journal of Applied Science*, 9(3), 153-163.
- Masuda, T., Sadoyama, T., & Shiraishi, M. (1996). Dependence of Average Muscle Fibre Conduction Velocity on Voluntary Contraction Force. *Journal of Electromyography and Kinesiology*, 6(4), 267-276.
- McBride, M. E., & Elliott, B. C. (1999). Use of real-time telemetry to monitor instantaneous seat and boat velocity in pair oared rowing. In R. H. Sanders & B. J. Gibson (Eds.), *Scientific Proceedings of the XVII International Symposium on Biomechanics in Sports* (pp. 93-96). Perth: Edith Cowan University.
- McLellan, T. M., & Cheung, K. S. Y. (1992). A comparative evaluation of the individual anaerobic threshold and the critical power. *Medicine and Science in Sports and Exercise*, 24(5), 543-550.

- Merletti, R., & Roy, S. (1996). Myoelectric and Mechanical Manifestations of Muscle Fatigue in Voluntary Contractions. *The Journal of Orthopaedic and Sports Physiotherapy*, 24(6), 342-353.
- Messonnier, L., Freund, H., Bourdin, M., Belli, A., & Lacour, J.R. (1997). Lactate exchange and removal abilities in rowing performance. *Medicine and Science in Sports and Exercise*, 29(3), 396-401.
- Miller, R. G., Gianninni, D., Milner-Brown, H. S., Layzer, R. B., Koretsky, A. P., Hooper, D., & Weiner, M. W. (1987). Effects of fatiguing exercise on high energy phosphates, force and EMG: evidence of three phases of recovery. *Muscle and Nerve*, 10, 810-821.
- Mills, K. R., & Edwards, R. H. T. (1984). Muscle fatigue in myophosphorylase deficiency: power spectral analysis of the electromyogram. *Journal of Electroencephalography and Clinical Neurophysiology*, 57, 330-335.
- Millward, A. (1987). A study of the forces exerted by an oarsman and the effect on boat speed. *Sport Sciences*, 5, 93-103.
- Milner-Brown, H. S., & Miller, R. G. (1986). Muscle membrane excitation and impulse propagation velocity are reduced during muscle fatigue. *Muscle and Nerve*, 9, 367-374.
- Moritani, T., Muro, M., & Nagata, A. (1986). Intramuscular and surface electromyogram changes during muscle fatigue. *Journal of Applied Physiology*, 60(4), 1179-1185.
- Moritani, T., Takaishi, T., & Matsumoto, T. (1993). Determination of maximal power output at neuromuscular fatigue threshold. *Journal of Applied Physiology*, 74(4), 1729-1734.

- Morton, A. R., Lawrence, S. R., Blanksby, B. A., & Bloomfield. (1984). A report of profiles of elite Australian oarsmen. *Australian Rowing*, 7(3), 21-23.
- Muro, M., Nagata, N., & Moritani, T. (1983). Analysis of myoelectric signals during dynamic and isometric contractions. In H. Matsui & K. Kobayashi (Eds.), *Biomechanics* (Vol. V111-A, pp. 432-439). Illinois: Champaign.
- Murphy, A. J., & Wilson, G. J. (1996). Poor Correlations between isometric tests and dynamic performance: relationship to muscle activation. *European Journal of Applied Physiology*, 73, 353-357.
- Nelson, W. N., & Widule, C. J. (1983). Biomechanics: New frontier of rowing. *Rowing U.S.A.* 15 (2), 23-27.
- Oda, S., & Moritani, T. (1995). Cross-correlation of bilateral differences in fatigue during sustained maximal voluntary contraction. *European Journal of Applied Physiology and Occupational Physiology*, 70(4), 305-310.
- Peltonen, J. E., Rantamaki, J., Niittymaki, S. R. T., Sweins, K., Rusko, H. K., & Viitasalo, J. T. (1994). Effects of $F_1 O_2$ on Force Production and EMG Activity during All-out Ergometer Rowing. In J. Vitasalo & U. Kujala (Eds.), *The Way to Win* (pp. 291-294). Helsinki: International Congress on Applied Research in Sports.
- Peltonen, J., & Rusko, H. (1993). Interrelations between power, force production and energy metabolism in maximal leg work using a modified rowing ergometer. *Journal of Sports Sciences*, 11, 233-240.
- Petrofsky, J. A., & Lind, A. R. (1980). Frequency analysis of the surface electromyogram during sustained isometric contractions. *European Journal of Physiology*, 43, 173-182.

- Petrofsky, J. S. (1979). Frequency and amplitude analysis of the EMG during exercise on the cycle ergometer. *European Journal of Applied Physiology*, 41, 1-15.
- Potvin, J. R. (1997). Effects of muscle kinematics on surface EMG amplitude and frequency during fatiguing dynamic contractions. *Journal of Applied Physiology*, 82(1), 144-151.
- Potvin, J. R., & Bent, L. R. (1997). A Validation of Techniques Using Surface EMG Signals from Dynamic Contractions to Quantify Muscle Fatigue during Repetitive Tasks. *Journal of Electromyography and Kinesiology*, 7(2), 131-139.
- Powers, S. K., & Howley, E. T. (1994). *Exercise Physiology. Theory and Application to Fitness and Performance*. (2nd ed.). Dubuque, USA: Brown & Benchmark.
- Psek, J. A., & Cafarelli, E. (1993). Behaviour of co-active muscles during fatigue. *Journal of Applied Physiology*, 74, 170-175.
- Robertson, J. (1982). Central signals of perceived exertion. *Medicine and Science in Sport and Exercise*, 14(5), 390-396.
- Rodriguez, R. J., Rodriguez, R. P., Cook, S. D., & Sandborn, P. M. (1990). Electromyographic analysis of rowing stroke biomechanics. *Journal of Sports Medicine and Physical Fitness*, 30(1), 103-108.
- Roeleveld, K., Stegeman, D. F., Vingerhoets, H. M., & Van Oosterom, A. (1997). Motor unit potential contribution to surface electromyography. *Acta Physiologica Scandinavica*, 160, 175-183.
- Sadoyama, T., Masuda, T., & Miyano, H. (1983). Relationship between muscle fiber conduction velocity and frequency parameters of surface EMG during sustained contraction. *European Journal of Applied Physiology*, 51, 247-256.

- Sale, D. G. (1988). Neural adaptation to resistance training. *Medicine and Science in Sports and Exercise*, 20(5), S135-S145.
- Sale, D. G., (1991). Testing strength and power. In MacDougall, J. D., Wenger, H. A., Green, H. J. (Eds.). *Physiological testing of high performance athletes* pp.21-106.
- Schneider, E., Angst, F., & Brandt, J. D. (1975). Biomechanics of Rowing. pp.115-119
- Secher, N. H. (1993). Physiological and Biomechanical aspects of Rowing. *Sports Medicine*, 15(1), 24-42.
- Seiler, S. (1996). Physiology of the elite rower. *Coaching Science Abstracts*, 1(6).
- Sherwood, A. M., Priebe, M. M., & Graves, D. E. (1997). Consistency of Multi-channel Surface EMG Recordings: Application in Spinal Cord Injured Subjects. *Journal of Electromyography and Kinesiology*, 7(2), 97-111.
- Solomonow, M., Barrata, R., Zhou, B.H., Shosi, H., Bosew, A., Beck, C., & D'Ambrosia, R. (1987). The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *American Journal of Sports Medicine*, 15 207-213
- Smith, R. M. & Milburn, P. (1996). Differences in patterns of power production between club and national level female rowers during maximal ergometer rowing. *Australian Conference in Science and Medicine in Sports*, 326-327
- Smith, R. M., & Spinks, W. L. (1995). Discriminant analysis of biomechanical differences between novice, good and the elite rowers. *Journal of Sport Sciences*, 13, 377-385.
- Smith, R. M., Spinks, W. L., & Moncrieff, J. (1992). ROWSYS: An "On Water" Biomechanical analysis system for rowing.

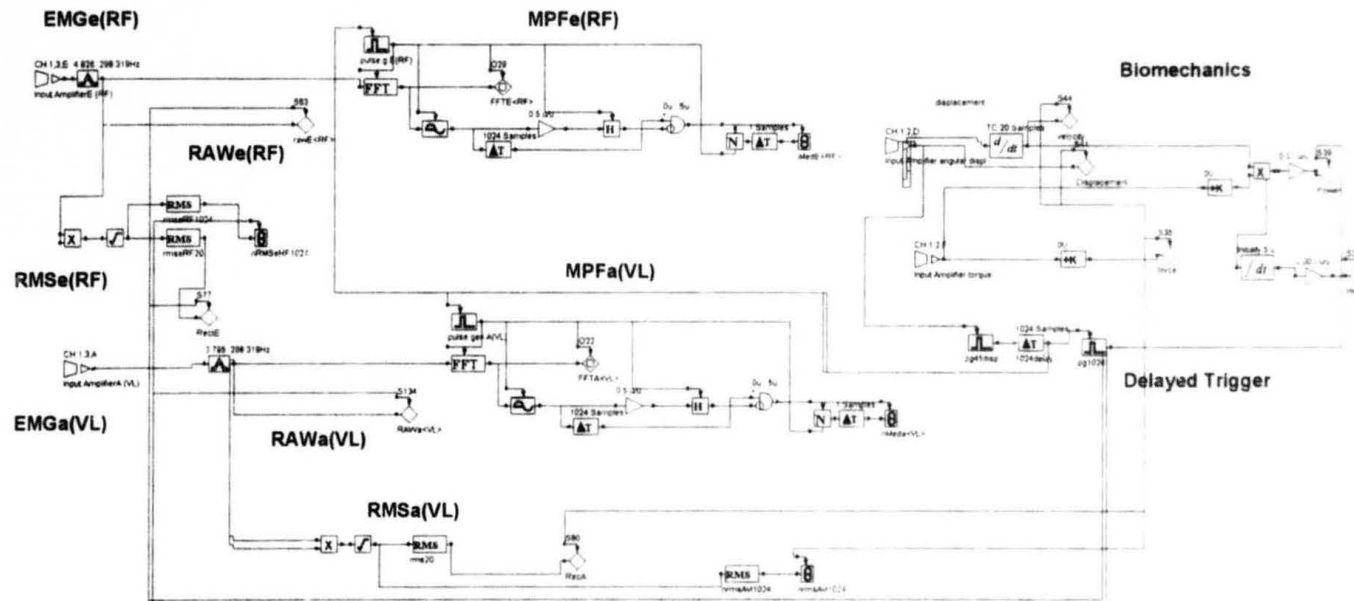
- Staniak, Z., Nosarzewski, Z., & Karpilowski, B. (1994). Computerized measuring set for rowing ergometry. *Biology of Sport*, 11(4), 271-282.
- Steinacker, J. M. (1993). Physiological aspects of training in rowing. *International Journal of Sports Medicine*, 14, S3-S10.
- Steinacker, J. M., Laske, R., Hetzel, W. D., Lormes, W., Liu, Y., & Stauch, M. (1993). Metabolic and hormonal reactions during training in junior oarsmen. *International Journal of Sports Medicine*, 14, S24-S28.
- Takaishi, T., Yasuda, Y., Ono, T., & Moritani, T. (1996). Optimal pedaling rate estimated from neuromuscular fatigue for cyclist. *Medicine and Science in Sports and Exercise*, 28(12), 1492-1497.
- Taylor, A. D., & Bronks, R. (1994). Electromyographic correlates of the transition from aerobic to anaerobic metabolism in treadmill running. *European Journal of Applied Physiology and Occupational Physiology*, 69(6), 508-517.
- Taylor, A. D., Bronks, R., Smith, P., & Humphries, B. (1997). Myoelectric evidence of peripheral muscle fatigue during exercise in severe hypoxia: some references to muscle vastus lateralis myosin heavy chain composition. *European Journal of Applied Physiology and Occupational Physiology*, 75(2), 151-159.
- Tesch, P. A., Dudley, G. A., Duvoisin, M., Hather, B., & Harris, R. T. (1990). Force and EMG signal patterns during repeated bouts of concentric and eccentric muscle actions. *Acta Physiologica Scandinavica*, 138, 263-271.
- Tesch, P. A., Komi, P. V., Jacobs, I., Karlsson, J., & Viitasalo, J. T. (1983). Influence of lactate accumulation of EMG frequency spectrum during repeated concentric contractions. *Acta Physiologica Scandinavica*, 119, 61-67.
- Vaughan, C. L., (1989). *Biomechanics of Sport*. Florida: C.R.C. Press.

- Van Dieen, J. H., Boke, B., Oosterhuis, W., & Toussaint, H. M. (1996). The Influence of torque and velocity on erector spinae muscle fatigue and its relationship to changes of electromyogram spectrum density. *European Journal of Applied Physiology and Occupational Physiology (Berlin)*, 72(4), 310-315.
- Vandervoort, A. A., Sale, D. G., & Moroz, J. (1984). Comparison of motor unit activation during unilateral and bilateral leg extensions. *Journal of Applied Physiology*, 56(1), 46-51.
- Vestergaard-Poulsen, P., Thomsen, C., Sinkjaer, T., & Henriksen, O. (1995). Simultaneous ³¹P-NMR spectroscopy and EMG in exercising and recovering human skeletal muscle: A correlation study. *Journal of Applied Physiology*, 79(5), 1469-1478.
- Vollestad, N. K. (1995). Metabolic correlates of fatigue from different types of exercise in man. *Advances in Experimental Medicine and Biology*, 384, 185-194.
- Weir, J. P., Wagner, L. L., & Housh, T. J. (1992). Linearity and reliability of the IEMG v torque relationship for the forearm flexors and leg extensors. *American Journal of Physical Medicine and rehabilitation*, 71, 283-287.
- Westerblad, H., Lee, J. A., Lannergren, J., & Allen, D. G. (1991). Cellular mechanisms of fatigue in skeletal muscle. *American Journal of Physiology*, 261, C195-C209.
- Wilson, J. M. J., Robertson, D. G. E., & Stothart, J. P. (1988). Analysis of Lower Limb Muscle Function in Ergometer Rowing. *International Journal of Sport Biomechanics*, 4, 315-325.
- Wing, A. M., & Woodburn, C. (1995). The co-ordination and consistency of rowers in a racing eight. *Journal of Sports Sciences*, 13, 187-197.

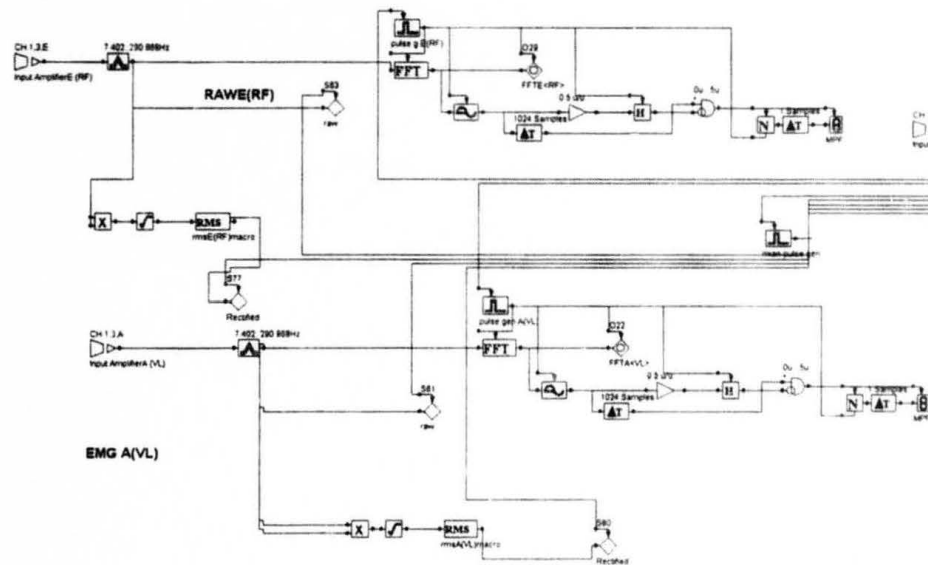
- Winkel, J., & Jorgensen, K. (1991). Significance of skin temperature changes in surface electromyography. *European Journal of Applied Physiology and Occupational Physiology (Berlin)*, 63(5), 345-349.
- Womack, C. J., Davis, S. E., Wood, C. M., Sauer, K., Alvarez, J., Weltman, A., & Gaesser, G. A. (1996). Effects of Training on Physiological Correlates of Rowing Ergometry Performance. *Journal of Strength and Conditioning Research*, 10(4), 234-238.
- Wretling, M. L., & Henriksson-Larsen, K. (1998). Mechanical Output and Electromyographic Parameters in Males and Females During Fatiguing Knee-Extensions. *International Journal of Sports Medicine*, 19, 401-407.
- Yang, J. F., & Winter, D. A. (1983). Electromyography reliability in maximal and submaximal isometric contractions. *Arch Phys Med Rehab*, 64, 417-420.
- Zatsiorsky, V. M., & Yakunin, N. (1991). Mechanics and biomechanics of rowing: A review. *International Journal of Sport Biomechanics*, 7, 229-281.

APPENDIX A

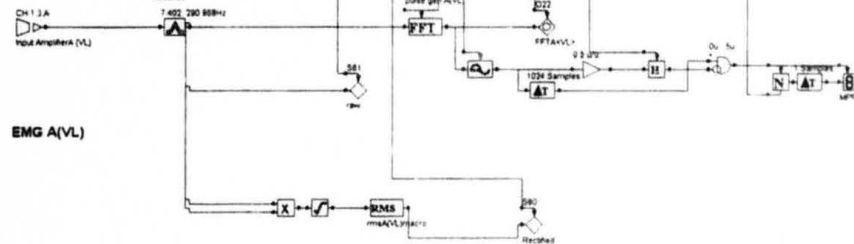
Amlab Schematic Projects



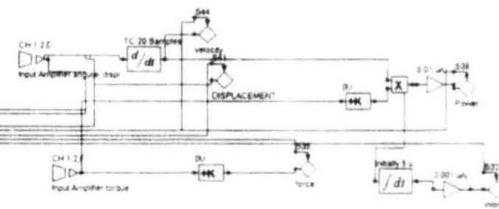
EMG E(RF)



EMG A(VL)

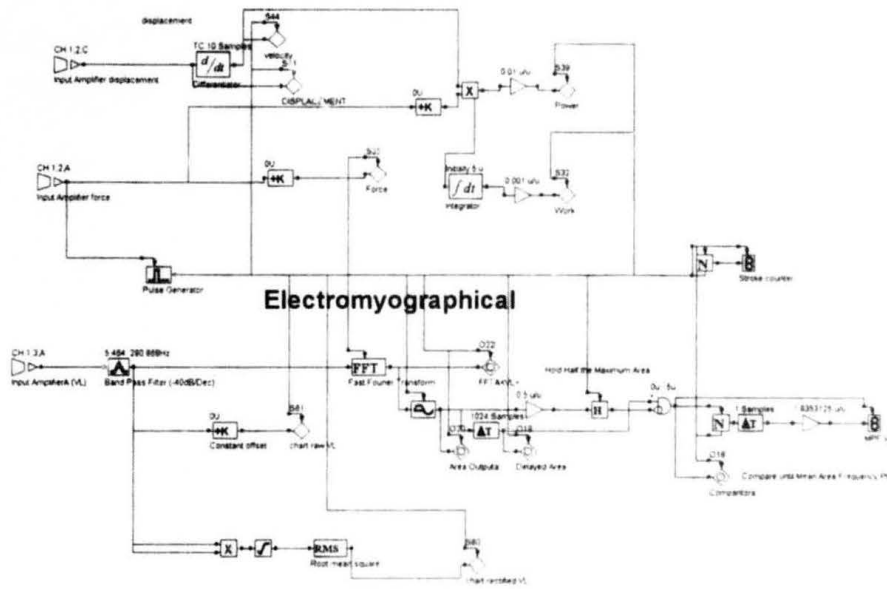


BIOMECHANICS

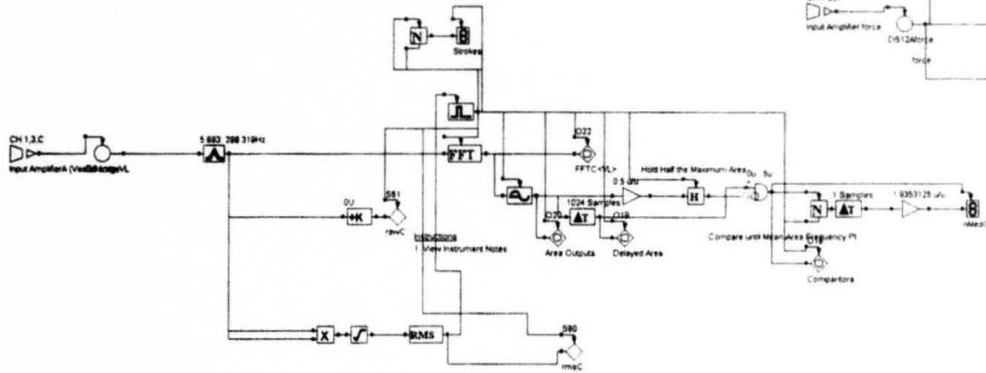




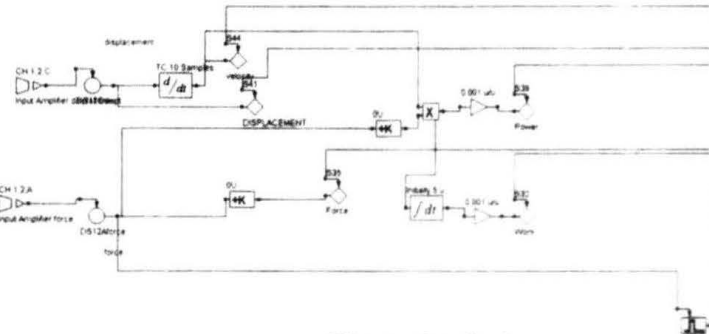
Biomechanical



Electromyographical



Biomechanical



APPENDIX B

Statement of disclosure

For potential participants:

During maximal rowing, fatigue alters physiological parameters that subsequently have a detrimental effect on the optimal and efficient application of skill. This study proposes to monitor, record and correlate force, fatigue and muscle activation parameters during maximal rowing ergometer performance.

Subjects will perform 6-minute maximal efforts on a rowing ergometer or an isokinetic dynamometer whilst force output and surface electromyographic (EMG) measurements are recorded. Capillary blood will be analyzed for lactate levels using finger-prick samples (50 μ l) pre- and post-exercise.

The subjects and rowing crew to which they belong will gain knowledge on their technical skills during fatigue and optimal pacing strategies that will enhance efficiency and performance outcomes. Knowledge of the changing parameters during maximal efforts will enable insight into the fatigue processes that affect skilful performance. A better understanding of local and central mechanisms of fatigue will advance optimal outcomes during athletic performance.

Participants will be required to attend Edith Cowan University for eight-ten sessions of approximately one-hour duration arranged at convenient times to suit the participants and the researcher. Any subjects withdrawing from the study will suffer no prejudice regarding further care, selection, or involvement in other rowing matters.

Any questions concerning this project entitled **“An investigation into force and fatigue characteristics during maximal rowing”** should be directed to Darryl Turner of the Human Movement Department ECU on 400 5868 or 307 6973.

INFORMED CONSENT

I _____ (the participant) have read the information above
and have been informed about all aspects of the above research project.
Any questions I have asked have been answered to my satisfaction.
I agree to participate in this activity, realizing I may withdraw at any time.
I agree that the research data gathered for this study may be published
provided that I am not identifiable (or, "understanding that I may be identified").

Participant or authorized representative

Date

Investigator

Date

APPENDIX C

University Ethical Approval



EDITH COWAN
UNIVERSITY

PERTH WESTERN AUSTRALIA
CHURCHLANDS CAMPUS

Pearson Street, Churchlands
Western Australia 6018
Telephone (09) 273 8333
Facsimile (09) 387 7095

12 March 1997

Committee for the Conduct of Ethical Research

Mr Darryl Turner
Department of Human Movement
Science, Technology & Engineering
Joondalup Campus

Dear Mr Turner

Re: **Ethics Approval**

Code: 97-18

Project Title: *An investigation into force and fatigue characteristics during maximal rowing performance.*

This project was reviewed by the Committee for the Conduct of Ethical Research at its meeting on 7 March, 1997.

I am pleased to advise that the project complies with the provisions contained in the University's policy for the conduct of ethical research, and has been cleared for implementation.

Period of approval is from 10 March 1997 to 28 February 2000.

With best wishes for success in your work.

Yours sincerely

ROD CROTHERS
Executive Officer

cc. Ms Colleen Ledwith, Secretary, HDC
Dr Paul Sacco, Supervisor
Mrs A Johnsen, Secretary, Doctoral Studies Committee

Please note: Students conducting approved research are required to submit an ethics report as an addendum to that which they submit to their Faculty's Higher Degrees Committee.

APPENDIX D

Warm-Up Protocol

a) 5 min at a pace of 2 min (2:00) per 500 m followed immediately by 12 maximal strokes, b) 6 min recovery, c) 3 min consisting of 1 min at a pace of 2:00 per 500 m, 1 min at 1:50 pace, 1 min at 2:00 pace, d) 2 min recovery.

(Hahn et al., 1995a)

APPENDIX E

TABLES

Table E1

Mean (SD) Effect of Ergometer Type and Muscle on Isometric
Force/torque-rmsEMG Relationships for the Same Subjects

		Ergometer type					
		T _{cyb}		T _{adap}		Total	
<u>Muscle</u>		<u>m</u>	<u>SD</u>	<u>m</u>	<u>SD</u>	<u>m</u>	<u>SD</u>
(RF)		.93	(.05)	.91	(.08)	.92	(.01)
(VL)		.94	(.04)	.94	(.02)	.94	(.01)
Total		.93	(.01)	.93	(.01)		

Note. Values are Linear Regression Coefficient "least squares (r^2) scores

Total = mean of ergometer types and mean of muscles. T_{cyb} = Cybex ergometer.

T_{adap} = adapted leg-only ergometer, RF = rectus femoris, VL = vastus lateralis.

N = 9.

Table E2

Mean (SD) Effect of Ergometer Type and Muscle on Isokinetic and Dynamic
Force/torque-rmsEMG Relationships for the Same Subjects

Ergometer type									
		T _{cyb}		T _{adap}		T _{stan}		Total	
<u>Muscle</u>		<u>m</u>	<u>SD</u>	<u>m</u>	<u>SD</u>	<u>m</u>	<u>SD</u>	<u>m</u>	<u>SD</u>
(RF)		.87	(.05)	.70	(.19)	.74	(.08)	.77	(.02)
(VL)		.87	(.05)	.67	(.13)	.81	(.11)	.78	(.01)
Total		.87	(.01)	.69	(.05)	.78	(.03)		

Note. Values are Linear Regression Coefficient "least squares" (r^2) scores.

Total = mean of ergometer types and mean of muscles. T_{cyb} = Cybex ergometer,

T_{adap} = adapted leg-only ergometer, RF = rectus femoris, VL = vastus lateralis

N = 8.

Table E3

Mean (SD) Effect of Ergometer Type and Muscle on
Force/torque-rmsEMG Relationships (r^2) for the Same Subjects During Fatigue

	Ergometer type							
	T _{cyb}		T _{adap}		T _{stan}		Total	
Muscle	m	SD	m	SD	m	SD	m	SD
(RF)	.49	(.31)	.13	(.16)	.25	(.25)	.29	(.06)
(VL)	.54	(.28)	.24	(.36)	.39	(.32)	.39	(.06)
Total	.52	(.09)	.19	(.11)	.32	(.11)		

Note. Values are Linear Regression Coefficient "least squares (r^2) scores. Total = mean of ergometer types and mean of muscles. T_{cyb} = Cybex ergometer. T_{adap} = adapted leg-only ergometer. RF = rectus femoris. VL = vastus lateralis.

N = 5.

Table E4

Effect of Ergometer Type and Muscle on Mean (SD)

Force/torque-MPF Relationships (r^2) for the Same Subjects During Fatigue

	Ergometer type			
	T _{cyb}	T _{adap}	T _{stan}	Total
Muscle	m (SD)	m (SD)	m (SD)	m (SD)
RF	.60 (.23)	.24 (.22)	.23 (.30)	.36 (.07)
VL	.34 (.12)	.28 (.28)	.42 (.25)	.35 (.05)
Total	.47 (.05)	.26 (.09)	.33 (.07)	

Note. Values are Linear Regression Coefficient "least squares (r^2)" scores. Total = mean of ergometer types and mean of muscles. T_{cyb} = Cybex ergometer. T_{adap} = adapted leg-only ergometer. RF = rectus femoris. VL = vastus lateralis

N = 5.