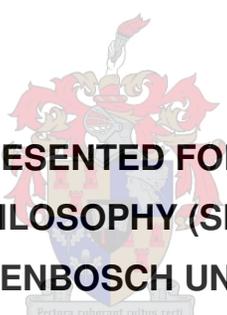


**THE VALIDITY AND RELIABILITY OF A BIOFEEDBACK
SYSTEM DURING SEGMENTAL STABILISATION IN PATIENTS
WITH LOW BACK PAIN**

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**DISSERTATION PRESENTED FOR THE DEGREE OF
DOCTOR OF PHILOSOPHY (SPORT SCIENCE)
AT STELLENBOSCH UNIVERSITY**

The crest of Stellenbosch University is centered behind the text. It features a shield with a blue and red design, topped with a crown and a banner. The Latin motto "Perfata celebrant cultus recti" is inscribed on a scroll at the base of the crest.

PROMOTER: PROF. J.G. BARNARD

DECEMBER 2009

DECLARATION

By submitting this dissertation electronically, I declare that the entirety of the work contained therein is my own, original work, that I am the owner of the copyright thereof (unless to the extent explicitly otherwise stated) and that I have not previously in its entirety or in part submitted it for obtaining any qualification.

December 2009

ABSTRACT

Worldwide epidemiological findings strongly indicate low back pain as a growing epidemic despite the latest diagnostic and treatment methods used (Jellema *et al.*, 2001:377; Woolf & Pfleger, 2003:646; Kopec *et al.*, 2004:70; Frost & Sullivan, 2006; Dagenais *et al.*, 2008:9). From this clinical problem, a need arose to quantify lumbar muscle performance for the safe monitoring of rehabilitation programmes and assessments.

The quantification of muscular strength is especially important from a kinematic viewpoint, because activities of daily living are dependent on muscular strength (Nobori & Maruyama, 2007:9). Furthermore, it is of utmost importance to combine the complex muscular system with the complexity of motor control (Richardson *et al.*, 2005) and biomechanics (McGill, 2007), to specifically address the problem of low back pain.

Scientists have developed a better understanding of muscle function or dysfunction pertaining to low back pain and highlighted the clinical importance of quantitative muscle testing of the lumbar spine. Various clinical assessment devices and methods such as ultrasound, kinesiological electromyography, isokinetic dynamometry and the Biering-Sorensen test are being used to record muscle atrophy and dysfunction of the *m. lumbar multifidi* at L5. However, some of these assessments are very costly and some are clinically unsafe, and therefore the need for reliable and valid low back tests still exists. In the study presented, a system called the pressure air biofeedback (PAB) device was developed, to scientifically assess *m. lumbar multifidus*' isometric contraction in asymptomatic (n=24) and low back pain (n=18) subjects.

A closed chain test model with a neutral spine posture in zero degrees upright sitting was used. This study compared results of *m. lumbar multifidus*' isometric contraction between tests of pressure air biofeedback, electromyography and real-time ultrasound. Pressure air biofeedback values in millibar (mb), root-mean-square (RMS) values of electromyography in microvolts (μ V) and real-

time ultrasound muscle cross-sectional area in centimeters square (cm²) of *m. lumbar multifidus*' isometric contraction were recorded during lumbar extension at maximal voluntary isometric contraction (MVIC), as well as at 50% and 80% MVIC. Trial-to-trial and day-to-day comparisons of measurements were made to estimate the reproducibility of the device. Also, day-to-day comparisons of PAB calibration measurements with calibrated kilogram (kg) weight increments of 2.5 kg (between 2.5 kg and 160 kg) were made to estimate the validity of the device.

Pearson correlation calculations were significant between PAB force (mb) and EMG activity (μ V) for the asymptomatic group at all effort levels over the two days [from $r=0.63$ ($p<0.01$) to $r=0.76$ ($p<0.01$)]. A non-significant correlation between PAB and EMG [from $r=-0.02$ ($p<0.94$) to $r=0.26$ ($p<0.29$)] for low back pain subjects at all effort levels over the two days was calculated. No correlation was found between US and PAB for all effort levels tested on both days [from $r=0.27$ ($p<0.17$) to $r=0.47$ ($p<0.01$)]. The lumbar extension strength (at MVIC) of asymptomatic subjects was significantly more by 52.2 kg force (kgf) or 515.02 Newton ($p<0.01$) than the low back pain subjects. Significant low back strength differences ($p<0.01$) were also calculated for all sub-MVICs.

Calibration results of PAB force output (mb) and applied external force comparisons (calibrated weights in kg) demonstrated high agreement or validity ($r=0.997$, $p<0.01$) while the ICC calculation of 0.997 (SEM=1.55) of PAB force calibration measurements between day one and two indicated a significant correlation and very good reliability of the PAB device.

The testing effect of the PAB device, specific to the L5 lumbar muscles recruited was exemplified by the different modes of isometric contractions and showed a high level of reliability and validity in asymptomatic as well as low back pain subjects. The findings of the PAB test further indicated that the method used, in this case an upright seated (neutral spine), closed chain loaded lumbar extension test, may be critical to determine the exact muscles measured in the lumbar spine.

The PAB test findings may also define an upright sitting posture most suitable for the closed chain, lumbar extension PAB test. Finally, this research study has indicated that air pressure, as applied in the closed chain, upright sitting PAB test, may be used as an applicable, reliable and alternative testing method to the Biering-Sorensen and/or isokinetic tests in quantifying muscle strength of the L5 m. lumbar multifidus.

Keywords: Low back pain, *m. lumbar multifidus*, quantifying muscle strength, reliability and validity, PAB test

SAMEVATTING

Wêreldwye epidemiologiese bevindings toon aan dat lae ruggpyn 'n groeiende epidemie is, ongeag die nuutste diagnostiese en behandelings metodes (Jellema *et al.*, 2001:377; Woolf & Pfleger, 2003:646; Kopec *et al.*, 2004:70; Frost & Sullivan, 2006; Dagenais *et al.*, 2008:9). Hierdie kliniese probleem het gelei tot 'n behoefte aan die kwantifisering van lumbale rug spierkrag, sodat rehabilitasie programme en toetse veilig gemonitor kan word.

Die kwantifisering van spierkrag is veral belangrik vanuit 'n kinematiese oogpunt, omdat meeste daaglikse aktiwiteite van spierkrag afhanklik is (Nobori & Maruyama, 2007:9). Verder, is dit van uiterste belang om die komplekse muskulêre sisteem met die kompleksiteit van motoriese spierbeheer (Richardson *et al.*, 2005) en biomeganika (McGill, 2007) te kombineer om spesifiek die probleem van lae ruggpyn aan te spreek.

Wetenskaplikes het gedurende die laaste paar jaar 'n beter begrip van spierfunksie of –disfunksie, wat met lae ruggpyn verband hou, ontwikkel. Dit het die kliniese belangrikheid van kwantitatiewe spier-assessering van die lae rug benadruk. Verskeie kliniese toetsinstrumente en –metodes, soos bv. ultra-klank, kinesiologiese elektromiografie, isokinetiese dinamometrie en die Biering-Sorensen toets word gebruik om spieratrofie en disfunksie van die *m. lumbale multifidus* by die vyfde lumbale vlak (L5) te meet. Verskeie van hierdie toetse is egter baie duur en klinies onveilig en daarom bestaan daar steeds 'n behoefte vir 'n betroubare en geldige toets vir lae rug spierkrag.

In hierdie navorsingstudie het die navorser 'n lugdruk bioterugvoerings-instrument, genaamd die PAB apparaat, ontwerp. Die PAB apparaat is gebruik om isometriese spierkontraksies van die *m. lumbale multifidus* wetenskaplik te toets in asimptomatiese (n=24) en lae ruggpyn (n=18) proefpersone. 'n Geslote ketting toetsmetode, met 'n neutrale rugpostuur in 'n regop-sit posisie van nul grade, is gebruik. Tydens hierdie studie is die *m. lumbale multifidus* se

isometriese spierkontraksie met drie verskillende toetsmetodes, naamlik lugdruk bio-terugvoering, elektromiografie en ultra-klank, vergelyk.

Lumbale multifidus spierkontraksie is gemeet in die volgende eenhede; lugdruk bio-terugvoeringsdata in millibar (mb), wortel van gemiddelde kwadraat data van elektromiografie in mikrovolt (μV) en ultra-klank spieroppervlakte in vierkante sentimeter (cm^2). Hierdie metings van maksimale isometriese kontraksie (MIK) is tydens lumbale ekstensie gemeet.

Dieselfde data is ook gemeet tydens onderskeidelik 50 % en 80 % MIK. Toets-hertoets en dag-tot-dag vergelykings van metings is geneem om die betroubaarheid van die instrument te bepaal. Daar is ook dag-tot-dag vergelykings van kalibrasie metings gedoen met gekalibreerde gewigte van 2.5 kg toenames tussen 2.5 kg en 160 kg om die geldigheid van die instrument te bepaal.

Pearson korrelasie berekenings tussen PAB kraguitset (mb) en elektromiografie (μV) was beduidend vir asimptomatiese proefpersone by alle kraguitset vlakke oor twee dae [$r=0.63$ ($p<0.01$) tot $r=0.76$ ($p<0.01$)]. 'n Nie-beduidende korrelasie tussen PAB en elektromiografie [$r=-0.02$ ($p<0.94$) tot $r=0.26$ ($p<0.29$)] vir lae rugpyn proefpersone by alle kraguitset vlakke oor twee dae is bereken. Geen korrelasie is tussen ultra-klank en PAB gevind vir alle isometriese kraguitset vlakke oor die twee dae gemeet [$r=0.27$ ($p<0.17$) tot $r=0.47$ ($p<0.01$)]. Lumbale ekstensie krag van asimptomatiese proefpersone was beduidend meer teen 52.5 kg of 515.02 N ($p<0.01$) teenoor die lae rugpyn proefpersone. Beduidende lae rugkrag verskille ($p<0.01$) is ook vir alle sub-MIK's bereken.

Kalibrasie uitslae van PAB kraguitset (mb) en toegepaste eksterne kragte (gekalibreerde gewigte in kg) het 'n hoë geldigheidswaarde getoets ($r=0.997$, $p<0.01$). Die IKK berekening van 0.997 (SEM=1.55) vir PAB kalibrasie metings tussen dag een en twee het op 'n beduidende korrelasie en betroubaarheid van die PAB apparaat gedui.

Die toetseffek van die PAB apparaat, met spesifieke verwysing na die aktivering van die L5 lumbale spiere, is duidelik bevestig deur die verskillende vlakke van isometriese kontraksies en het 'n hoë vlak van betroubaarheid en geldigheid in a-simptomatiese, sowel as lae rugpyn proefpersone aangetoon. Die bevindings van die PAB toets het verder aangetoon dat die metode wat gebruik is, in hierdie geval 'n geslote ketting toetsmetode, met 'n neutrale rugpostuur in 'n regop-sit posisie tydens lumbale ekstensie, kritiek mag wees in die bepaling van die presiese spiere wat gemeet word in die lumbale rug.

Die PAB toetsresultate het die neutrale rugpostuur in regop-sit geïdentifiseer as die mees geskikte postuur vir die geslote ketting, lumbale ekstensie PAB toets. Laastens het die navorsingstudie aangetoon dat lugdruk, soos toegepas in die PAB toets, aangewend kan word as 'n toepaslike, betroubare en alternatiewe toetsmetode vir die Biering-Sorensen en/of isokinetiese toetse in die kwantifisering van spierkrag van die L5 *m.* lumbale multifidus.

Sleutelwoorde: Lae rugpyn, *m.* lumbale multifidus, kwantifisering van spierkrag, betroubaarheid en geldigheid, PAB toets

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CONTENTS

CHAPTER ONE		p
	THE LOW BACK PAIN PROBLEM	
1.1	INTRODUCTION	2
1.2	PROBLEM-STATEMENT	4
1.3	EMERGING OPPORTUNITIES IN MUSCLE ASSESSMENT TECHNOLOGY	7
1.4	PRESSURE AIR BIOFEEDBACK: A NEW DIRECTION	7
1.5	OBJECTIVE	10
CHAPTER TWO		
	EPIDEMIOLOGY OF LOW BACK PAIN	
2.1	INTRODUCTION	14
2.2	EPIDEMIOLOGY OF LOW BACK PAIN WORLDWIDE	16
2.2.1	Australia	17
2.2.2	Japan	18
2.2.3	The Netherlands	19
2.2.4	South Africa	20
2.2.5	Sweden	20
2.2.5.1	Direct and indirect costs of low back pain in Sweden	21
2.2.5.2	Total cost of chronic low back pain in Sweden	21
2.2.6	The United Kingdom	22
2.2.6.1	Annual prevalence of low back pain in the United Kingdom	24
2.2.6.2	Cost of general practice in the United Kingdom	24
2.2.6.3	Cost of healthcare professionals in the United Kingdom	24
2.2.7	The United States of America	26
2.2.7.1	Epidemiology of lumbar surgery in the United States of America	27
2.2.7.2	Cost analysis of low back pain in the United States of America	29
2.3	COMPENSATION CLAIMS	31
2.4	EPIDEMIOLOGY OF LUMBAR REVISION SURGERY	32

2.5	REVIEW OF TREATMENT MODALITIES IN LOW BACK PAIN	34
2.6	LUMBAR SUPPORT BELTS	37
2.7	EPIDEMIOLOGY OF LOW BACK PAIN IN SPORTSMEN AND WOMEN	40
2.7.1	Athletics	40
2.7.2	Ball Sports	41
2.7.3	Gymnastics and Dance	43
2.7.4	Strength Sports	44
2.7.5	Water sports	44
2.8	CHILDREN AND LOW BACK PAIN	45
2.9	SUMMARY	48

CHAPTER THREE

PATHOPHYSIOLOGY OF LOW BACK DISORDERS

3.1	INTRODUCTION	52
3.2	BIOMECHANICS AND INJURY MECHANISM OF LOW BACK DISORDERS	53
3.2.1	Injury mechanism during lumbar flexion	57
3.2.2	Injury mechanism during lumbar extension	62
3.2.3	Injury mechanism during sitting	63
3.3	THE LUMBAR MULTIFIDUS MUSCLE MECHANISM IN LOW BACK DISORDERS	65
3.3.1	Introduction	65
3.3.2	Mechanical anatomy of the <i>m. lumbar multifidus</i>	67
3.3.3	Morphology of the <i>m. lumbar multifidus</i>	74
3.3.4	Histology of the <i>m. lumbar multifidus</i>	77
3.3.5	Neutral spine posture and muscle activation of <i>m. lumbar multifidus</i>	80
3.4	SUMMARY	85

CHAPTER FOUR**NEW TECHNOLOGY DEVELOPMENT OF A PRESSURE AIR
BIOFEEDBACK DEVICE**

4.1	INTRODUCTION	88
4.2	MARKET OPPORTUNITIES AND EMERGING TECHNOLOGIES	89
4.3	HISTORIC DEVELOPMENT OF MODERN DYNAMOMETRY	91
4.3.1	Development of the pneumatic dynamometer	93
4.4	DESIGN AND VALIDATION OF THE PRESSURE AIR BIOFEEDBACK DEVICE	94
4.4.1	The pressure air biofeedback device	95
4.4.2	Pressure air biofeedback device calibration	98
4.4.3	Calibration data of the pressure air biofeedback device versus calibrated weights	101
4.5	SUMMARY	105

CHAPTER FIVE**METHODOLOGY**

5.1	INTRODUCTION	108
5.2	OBJECTIVE	109
5.3	RESEARCH DESIGN	109
5.4	SUBJECTS	109
5.4.1	Exclusion criteria for asymptomatic subjects	110
5.4.2	Exclusion criteria for low back pain subjects	111
5.5	SECTION ONE: THE OSWESTRY DISABILITY QUESTIONNAIRE	112
5.5.1	Development	112
5.5.2	Modifications and translations	112
5.5.3	Construct validity	113
5.5.4	Internal consistency	113
5.5.5	Reproducibility	113
5.5.6	Scoring the Oswestry Disability Index	114
5.6	SECTION TWO: MORPHOLOGICAL MEASUREMENTS	114
5.6.1	The measurement protocol	114

5.6.2	Age	115
5.6.3	Anthropometrical measurements	115
5.6.3.1	Body mass	115
5.6.3.2	Standing body height	115
5.6.3.3	Body mass index	116
5.6.3.4	Waist-to-hip ratio	116
5.6.3.5	Standing lumbar posture	117
5.7	SECTION THREE: PRESSURE AIR BIOFEEDBACK TESTING PROTOCOL – APPLICATION OF SCIENTIFIC EVIDENCE	118
5.7.1	Parallelograms of force vectors in the pressure air biofeedback and Biering-Sorensen tests	125
5.8	SECTION FOUR: GENERAL SET-UP AND TESTING PROCEDURE	128
5.8.1	Surface electromyography set-up procedure	130
5.8.2	The pressure air biofeedback and electromyography trial	131
5.8.3	The pressure air biofeedback and ultrasound trial	134

CHAPTER SIX

RESULTS

6.1	INTRODUCTION	138
6.2	STATISTICAL ANALYSIS	139
6.3	RESULTS OF DESCRIPTIVE STATISTICS	140
6.3.1	Descriptive statistics for age and body mass index	140
6.3.2	Descriptive statistics for waist-to-hip ratio	141
6.3.3	Descriptive statistics for standing lumbar posture	141
6.3.4	Descriptive statistics for the Oswestry Disability Index	142
6.4	PRESSURE AIR BIOFEEDBACK DEVICE CALIBRATION	142
6.5	REAL TIME BIOFEEDBACK GRAPHS FOR PRESSURE AIR BIOFEEDBACK, ELECTROMYOGRAPHY AND ULTRASOUND	142
6.5.1	First case – an asymptomatic 42 year old male	143
6.5.2	Second case – a 49 year old low back pain woman	144
6.5.3	Third case – an 80 year old asymptomatic male	146
6.5.4	Fourth case – a 66 year old asymptomatic male	147

6.6	INTRACLASS CORRELATION COEFFICIENT AGREEMENT RESULTS OF ELECTROMYOGRAPHY, PRESSURE AIR BIOFEEDBACK AND ULTRASOUND VARIABLES	148
6.7	ADJUSTED TESTING PROTOCOL	151
6.8	RESULTS OF DAY-TO-DAY REPEATABILITY ANALYSIS	152
6.8.1	Pressure air biofeedback day one versus day two during electromyography testing	152
6.8.2	Pressure air biofeedback day one versus day two during ultrasound testing	154
6.8.3	Pressure air biofeedback during the trial-to-trial tests of electromyography and ultrasound on day one and two	155
6.8.4	Electromyography root mean square values day one versus day two	157
6.8.5	Ultrasound day one versus ultrasound day two	157
6.8.6	Ultrasound day one and two versus pressure air biofeedback (ultrasound testing) day one and two	157
6.8.7	Electromyography versus pressure air biofeedback (electromyography testing) day one and two for the whole group	159
6.8.8	Electromyography versus pressure air biofeedback day one and two for two subgroups	162
6.9	LOW BACK STRENGTH IN ASYMPTOMATIC AND LOW BACK PAIN SUBJECTS	165

CHAPTER SEVEN

DISCUSSION

7.1	RESULTS OF MORPHOLOGY AND THE OSWESTRY DISABILITY INDEX	170
7.2	ADJUSTED TESTING PROTOCOL	171
7.3	PRESSURE AIR BIOFEEDBACK RELIABILITY AND VALIDITY TESTS	172
7.4	ULTRASOUND AND ELECTROMYOGRAPHY RELIABILITY	174
7.5	ULTRASOUND VERSUS THE PRESSURE AIR BIOFEEDBACK TEST	175

7.6	ELECTROMYOGRAPHY VERSUS THE PRESSURE AIR BIOFEEDBACK TEST	179
7.7	LOW BACK STRENGTH - ASYMPTOMATIC VERSUS LOW BACK PAIN SUBJECTS	182
7.8	THE PRESSURE AIR BIOFEEDBACK TEST VERSUS THE BIERING-SORENSEN TEST	183
7.9	CONCLUSION	189

CHAPTER EIGHT

RECOMMENDATIONS AND LIMITATIONS

8.1	SUMMARY	191
8.2	RECOMMENDATIONS	192
8.3	LIMITATIONS	194

REFERENCES	195
-------------------	-----

APPENDIX A	236
-------------------	-----

APPENDIX B	238
-------------------	-----

APPENDIX C	240
-------------------	-----

APPENDIX D	241
-------------------	-----

APPENDIX E	243
-------------------	-----

APPENDIX F	245
-------------------	-----

LIST OF FIGURES

- Figure 2.1 Economic burden of selected diseases in the UK
(adapted from Maniadakis & Gray, 2000:102). 23
- Figure 3.1 The oblique angle of the lumbar portions of *m. iliocostalis*
and *m. longissimus*. (a) neutral spine position. (b) oblique
angle of these muscles as viewed with US. (c) lumbar spine
flexion and (d) loss of the oblique angle with spine flexion so
that anterior shear forces cannot be counteracted (adapted
from McGill, 2007:53). 58
- Figure 3.2 Biomechanics of lifting. (a) a flexed spine imposing
anterior shear force and (b) a neutral spine posture
aligns the fibres to support the shear forces. (adapted
from McGill, 2007:102). 61
- Figure 3.3 The vertebra-to-vertebra attachments of *m. lumbar*
multifidus within the lumbar and between lumbar
and sacral vertebrae. (a) the laminar fibres at every level;
(b-f) the longer fascicles from the caudal edge and
tubercles of the spinous processes at levels L1-L5.
(reproduced with permission from Bogduk, 1997:106). 69
- Figure 3.4 Oblique orientation of fascicles of *m. lumbar multifidus*
into major vertical and minor horizontal vectors
(reproduced with permission from Macintosh &
Bogduk, 1986:205-213). 70
- Figure 3.5 Average perpendicular actions of *m. lumbar multifidus*
on the lumbar spinous processes. (reproduced with
permission from Macintosh & Bogduk, 1986:205-213). 70

Figure 4.1	The Regnier dynamometer testing low back strength (left) and the dynamometer itself (right) (adapted from Horne & Talbot, 2002).	92
Figure 4.2	The PAB device set-up during a seated, closed chain back extension test in a neutral spine posture (pre-test set-up).	97
Figure 4.3	Schema of technique used to determine pressure-force characteristic of a ball device (adapted from Axen <i>et al.</i> , 1992:7).	99
Figure 4.4	Schema of calibration technique used to determine the pull-compression force characteristic of the PAB device.	100
Figure 4.5	A significant correlation or linear relationship between PAB (mb) and calibrated weights (kg) was found on day one.	102
Figure 4.6	Day two again showed a significant correlation or linear relationship between PAB (mb) and calibrated weights (kg).	102
Figure 4.7	A significant correlation or linear relationship between the average PAB (mb) values vs calibrated weights, calculated over the two days, was indicated.	103
Figure 4.8	A significant correlation or linear relationship between PAB vs PAB (mb) values on two different days was indicated.	103

Figure 4.9	Day one, an example of a real time PAB force graph (mb) for five different calibrated weight loads between 140kg and 160kg.	104
Figure 4.10	Day two, an example of a real time PAB force graph (mb) for the same five calibrated weight loads between 140kg and 160kg.	105
Figure 5.1	Posture assessment with the inclinometer. (A) represents the lumbosacral angle (L-S joint) and (B) represents the thoracolumbar angle (T-L joint) (adapted from Saunders, 1998:7).	117
Figure 5.2	Closed chain loading in (a) a leg press exercise and (b) in a closed chain loaded posture as in squatting (adapted from Richardson, 2005:94, 96)	123.
Figure 5.3	Directing the pulling force vector through the low back in Figure a (adapted from McGill, 2007:142). Similar pulling force vector in Figure b, as executed in this study.	124
Figure 5.4	Graphical drawing of force vectors with respect to the subscapula (a), feet attachment (c) and hip joint (b) in the PAB test. Intersection (i) at L4.	125
Figure 5.5	Graphical drawing of force vectors with respect to the <i>m. teres</i> major alignment (a), gravity (c) and the hip joint (b) in the Biering-Sorensen test. Intersection (i) at T11.	127
Figure 6.1a	PAB force real-time graph (mb) in an asymptomatic 42 year old male, illustrating high air pressure output values for MVIC, 80% and 50% MVIC.	143

- Figure 6.1b Electromyographic RMS values (μV) of the same 42 year old male at different effort levels. Left (L) *m. lumbar multifidus* on the top is shown in relation to right (R) *m. lumbar multifidus* at the bottom. 144
- Figure 6.2a PAB force real-time graph (mb) in a 49 year old low back pain woman, illustrating weak low back extension strength according to the air pressure values shown, at three different effort levels. 145
- Figure 6.2b RMS values of EMG (μV) of different strength contractions for left (L) and right (R) *m. lumbar multifidus* of the same 49 year old, low back pain woman. Note the EMG imbalance between left (L) and right (R) sides, demonstrating muscle denervation of right *m. lumbar multifidus*. 145
- Figure 6.3a Example of weak low back strength as shown in the low air pressure output levels of the PAB force graph (mb) of an 80 year old asymptomatic male. 146
- Figure 6.3b RMS values of EMG (μV) of left (L) and right (R) *m. lumbar multifidus* of the same 80 year old asymptomatic male. EMG difference between L and R demonstrates muscle denervation of the right side. 146
- Figure 6.4a PAB force real-time graph (mb) in a 66 year old asymptomatic male, illustrating high air pressure output values for MVIC, 80% and 50% MVIC. 147

Figure 6.4b	Axial US image of <i>m. lumbar multifidus</i> (CSA in cm ²) of resting and MVIC (stress) of the same 66 year old asymptomatic male. Left (L) and right (R) <i>m. lumbar multifidus</i> are shown at the fifth lumbar vertebrae (L5). At rest, the L side measured 7.29 cm ² and the R side, 8.27 cm ² . At MVIC (stress) the L side measured 8.28 cm ² and the R side, 8.83 cm ² .	148
Figure 6.5	PAB (mb) day one versus PAB (mb) day two, reflecting a MVIC during EMG testing (n=43).	152
Figure 6.6	PAB (mb) day one versus PAB (mb) day two, reflecting an 80% MVIC during EMG testing (n=43).	153
Figure 6.7	PAB (mb) day one versus PAB (mb) day two, reflecting a 50% MVIC during EMG testing (n=43).	153
Figure 6.8	PAB (mb) day one versus PAB (mb) day two, reflecting a MVIC during US testing (n=28).	154
Figure 6.9	PAB (mb) day one versus PAB (mb) day two at 80% MVIC (n=28).	154
Figure 6.10	PAB (mb) day one versus PAB (mb) day two at 50% MVIC (n=28).	155
Figure 6.11	PAB (mb) versus PAB (mb) day one, reflecting a MVIC during EMG and US tests on the same day (n=28).	155
Figure 6.12	PAB (mb) versus PAB (mb) day one, reflecting an 80% MVIC during EMG and US tests on the same day (n=28).	156
Figure 6.13	PAB (mb) versus PAB (mb) day one, reflecting a 50% MVIC during EMG and US tests on the same day (n=28).	156

Figure 6.14	PAB (mb) versus US (CSA in cm ²) day two, reflecting a non-linear relationship at MVIC during US testing (n=28). See paragraph 7.5 for the explanation of the discrepancy between subjects 20 and 23.	158
Figure 6.15	Day one, EMG (μ V) versus PAB (mb) at MVIC (n=42).	160
Figure 6.16	Day one, EMG (μ V) versus PAB (mb) at 80% MVIC (n=42).	160
Figure 6.17	Day one, EMG (μ V) versus PAB (mb) at 50% MVIC (n=42).	160
Figure 6.18	Day two, EMG (μ V) versus PAB (mb) at MVIC (n=42).	161
Figure 6.19	Day two, EMG (μ V) versus PAB (mb) at 80% MVIC (n=42).	161
Figure 6.20	Day two, EMG (μ V) versus PAB (mb) at 50% MVIC (n=42).	161
Figure 6.21	A non-significant relationship ($r=26$, $p<0.29$) between EMG (μ V) and PAB (mb) at MVIC for low back pain subjects (n=18) on day one, is illustrated.	163
Figure 6.22	A highly, non-significant relationship ($r= -0.11$; $p<0.68$) between EMG (μ V) and PAB (mb) at MVIC for 18 low back subjects on day two is illustrated, again.	163
Figure 6.23	A very good linear relationship is demonstrated ($r=0.75$, $p<0.01$) between EMG (μ V) and PAB (mb) on day one, for MVIC (n=24).	164

- Figure 6.24 A very good linear relationship is also demonstrated ($r=0.73$, $p<0.01$) on day two, between EMG (μV) and PAB (mb) for MVIC (n=24). 165
- Figure 6.25 Illustrated, is a PAB force difference of 42.09 mb ($p<0.01$) between asymptomatic (n=24) and low back pain subjects (n=18) at MVIC. 166
- Figure 6.26 PAB force difference of 35.8 mb ($p<0.01$) is illustrated between asymptomatic (n=24) and low back pain subjects (n=18) at 80% MVIC. 167
- Figure 6.27 Illustrated, is a PAB force difference of 22.46 mb ($p<0.01$) between asymptomatic (n=24) and low back pain subjects (n=18) at 50% MVIC. 167

LIST OF TABLES

Table 2.1	Comparison of lumbar fusion surgery with other musculoskeletal procedures (adapted from Deyo <i>et al.</i> , 2005:1444).	28
Table 2.2	Five year re-operation percentage rates for degenerative lumbar surgery (adapted from Diwan <i>et al.</i> , 2003:310).	33
Table 2.3	Results of low back pain interventions from systematic reviews (adapted from Furlan <i>et al.</i> , 2001:E160).	35
Table 2.4	Outcome of non-invasive treatment modalities on low back pain (adapted from van Tulder <i>et al.</i> , 2006:S74).	36
Table 2.5	The most common sites of injury in amateur and professional golfers as reported in surveys (adapted from Metz, 1999:27).	42
Table 3.1	The <i>m. lumbar multifidus</i> CSA at L4 and L5 in normal subjects (adapted with permission from Richardson <i>et al.</i> , 2005:66).	76
Table 5.1	Example of four test trials over two separate days for subject A.	129
Table 5.2	Example of four test trials over two separate days for subject B.	129

Table 6.1	ICC results of RMS values of EMG during four effort levels measured on day one and two between left (L) and right (R) <i>m. lumbar multifidus</i> over a three second period (n=43).	149
Table 6.2	ICC of PAB values at three effort levels (during EMG testing) measured over two days, taken per second (s) over a three second period (n=43).	150
Table 6.3	ICC of US values during four effort levels measured on Day one and two between left (L) and right (R) <i>m. lumbar multifidus</i> (n=28).	151
Table 6.4	Pearson correlation calculations between ultrasound and pressure air biofeedback for different effort levels on day one and day two.	158
Table 6.5	Pearson correlation calculations between PAB force (mb) and RMS values of EMG (μ V) for different effort levels on day one and day two for the whole group (n=42).	159
Table 6.6	Pearson correlation calculations between PAB force (mb) and RMS values of EMG (μ V) for different effort levels on day one and day two for the low back pain group (n=18).	162
Table 6.7	Pearson correlation calculations between pressure air biofeedback and EMG (RMS) values for different effort levels on day one and day two for the asymptomatic group (n=24).	164

LIST OF ABBREVIATIONS AND DEFINITIONS

- ACSM – American College of Sports Medicine
- A.D. – after Christ
- AHCPR – Agency for Health Care Policy and Research
- BMI – body mass index
- CPG – Chronic Pain Grade Questionnaire
- cm – centimeter
- cm² - square centimeter
- CSA – cross sectional area
- CSAG – Clinical Standards Advisory Group
- CT – computed tomography
- df – degrees of freedom
- Dr. – doctor
- EMG – electromyography
- *et al.* – and others
- FT – fast twitch fibers
- GP – general practitioner
- HFE – Human Factors Engineering
- Hz – Herz
- i – intersection
- ICC – intraclass correlation coefficient
- ISAK – International Society for the Advancement of Kinanthropometry
- ISSLS – International Society for the Study of the Lumbar Spine
- *in vitro* – term used for observing a biological phenomenon outside the human body e.g. in a test-tube
- *in vivo* – term used for observing a biological phenomenon inside the human body
- kg – kilogram
- kgf – kilogram force
- kg/m² - kilograms by standing height in meters squared
- kN – kilonewton
- L – left

- L1 – lumbar vertebral level one
- L2 – lumbar vertebral level two
- L3 – lumbar vertebral level three
- L4 – lumbar vertebral level four
- L5 – lumbar vertebral level five
- LRS – lumbosacral radicular syndrome
- L-S angle – lumbosacral angle
- *m.* – musculus
- *ma* – moment arm
- *mb* – millibar
- MHz – megahertz
- MRC – Medical Research Council
- MRI – magnetic resonance imaging
- ms – milliseconds
- MVIC – maximal voluntary isometric contraction
- *n* – number of test subjects
- *N* – Newton
- NIOSH – National Institute of Occupational Safety and Health
- NSAID's – nonsteroidal anti-inflammatory drugs
- Nm – Newton-meter
- ODI – Oswestry Disability Index
- OPCS – Office of Population Censuses and Surveys
- PAB – pressure air biofeedback
- PC – personal computer
- PM – paraspinal muscle mapping
- PRA – postural restraint apparatus
- PSI – pounds per square inch
- *r* – Pearson correlation coefficient
- *R* – right
- RCT – randomized controlled trial
- *rf* – resultant force vector
- RMS – root mean square
- *s* – second
- SD – standard deviation

- SEM – standard error of measurement
- ST – slow twitch fibers
- S1 – the first fused vertebrae (from five) of the sacrum
- TENS – transcutaneous electrical nerve stimulation
- T-L angle – thoracolumbar angle
- tvl – transmissible vector
- type I – slow oxidative muscle fibre
- type II – fast glycolytic muscle fibre
- T1 – thoracic vertebral level one
- T8 – thoracic vertebral level eight
- T12 – thoracic vertebral level 12
- UK – The United Kingdom
- US – ultrasound
- USA – United States of America
- WCPT – World Confederation of Physical Therapy
- WHO – World Health Organisation
- WHR – waist-to-hip ratio
- \$ - American dollar
- £ - British pound
- € - European monetary value, the euro
- μV – micro Volt
- % - percentage
- $^{\circ}$ - degree

CHAPTER ONE

THE LOW BACK PAIN PROBLEM

- 1.1 INTRODUCTION
- 1.2 PROBLEM-STATEMENT
- 1.3 EMERGING OPPORTUNITIES IN MUSCLE ASSESSMENT
TECHNOLOGY
- 1.4 PRESSURE AIR BIOFEEDBACK: A NEW DIRECTION
- 1.5 OBJECTIVE

1.1 INTRODUCTION

Dysfunction caused by low back pain may not be a life-threatening condition, but it is a major public health problem worldwide and has grown to epidemic proportions (Cassidy *et al.*, 1998:1860; Deyo, 1998:49; Maniadakis & Gray, 2000:95; Nourbakhsh & Arab, 2002:447; Tandon *et al.*, 2002:165; van Tulder *et al.*, 2003:30; Pai & Sundaram, 2004:1; Walker *et al.*, 2004:238; Cassidy *et al.*, 2005:2817; Davis & Kotowski, 2005:453; Ekman *et al.*, 2005:1777; Humphrey *et al.*, 2005:175; Richardson, 2005:3; Demoulin *et al.*, 2006:43; van Tulder *et al.*, 2006:S64). Because the lifetime incidence of low back pain is exceptionally high, people who suffer from recurrent or chronic low back pain are those who also incur the majority of the cost, both personally and financially (Bouchard *et al.*, 1990:22; Nachemson, 1990:533; Maniadakis & Gray, 2000:95; Richardson, 2005:3).

Low back pain, as one of the major musculoskeletal disorders, such as osteoporosis, osteoarthritis, rheumatoid arthritis, spinal disorders and fractures, may cause severe long-term pain and physical disability and affects millions of people worldwide. This global burden has been recognised by the United Nations and the World Health Organization (WHO) with their endorsement of the Bone and Joint Decade 2000-2010 (Woolf & Pfleger, 2003:646). This burden of major musculoskeletal disorders has also led to the proclamation of the National Bone and Joint Decade 2002-2011, in the United States of America (USA) by President George W. Bush (Bush, 2002).

The disease of low back pain affects both men and women equally, with prevalence increasing steadily with age (Pai & Sundaram, 2004:1). It is therefore important to reflect on the WHO's description of health. According to Nachemson (1990:537), it has been accepted that health, according to the WHO, has more than one dimension and that health is not only the absence of sickness or a handicap, but also a state of complete physical, psychological and social well-being. To achieve complete health, as described by the WHO, the World Confederation of Physical Therapy (WCPT), at their 13th WCPT General Meeting in 1995, defined physical therapy as the assessment and treatment of

the neuromuscular and cardio-respiratory systems by physical or mechanical means with special reference to the rehabilitation of movement dysfunction and specifically dysfunction of the neuromuscular system of the patient (World Confederation of Physical Therapy, 1995). It is clear from present knowledge that this is also definitely true with regard to low back pain and its prevention and treatment.

According to Borenstein (2000:225), knowledge of disorders of the lumbar spine has matured since the 1990s into the 21st century with advances being made in biochemistry, genetics, imaging technology and clinical therapeutics. This has resulted in improvements both in the outlook for individuals with these conditions, and in our understanding of these conditions. However, present diagnostic and treatment methods have not been able to slow down the worldwide epidemic of low back problems (Cairns *et al.*, 2000:127; Maniadakis & Gray, 2000:101; Jellema *et al.*, 2001:377; Kopec *et al.*, 2004:70). Despite the wide variety of health care professionals and treatment strategies that are used in the management of low back pain, we still do not fully understand the biomechanical or underlying physical risk factors that lead to clinical spinal instability (O'Sullivan *et al.*, 1997:2959; Radebold *et al.*, 2000:947; Comerford & Mottram, 2001b:23-24; Woolf & Pfleger, 2003:652; Richardson, 2005:4).

The simplistic model of thinking that if something does not move well enough it is tight and needs stretching, or is weak and needs strengthening, no longer holds the answer to mechanical dysfunction, especially in the case of low back pain (Comerford & Mottram, 2001a:3). This is why, over the last two decades, a significant body of academic and clinical research has developed a more detailed understanding of human movement function and in particular movement dysfunction (Comerford & Mottram, 2001b:15; Richardson *et al.*, 2005:vii).

This greater understanding of muscle function or dysfunction has become especially applicable to the low back pain patient and highlights the clinical importance of quantitative muscle testing and the corrective prescription and monitoring of exercise therapy (Hasue *et al.*, 1980:143; Helewa *et al.*, 1990:965;

Gallagher, 1997:1864; Owen, 1999:888; Keller *et al.*, 2004:3; Lygren *et al.*, 2005:1070; Mayer *et al.*, 2005:2556). However, exercise programmes alone will not give a global solution to this problem, but a cohesive effort of scientists, physicians, politicians and employers is necessary to solve one of the most expensive medical problems of modern industrialised societies (Nachemson, 1990:537).

1.2 PROBLEM-STATEMENT

Musculoskeletal stability is essential to the lumbopelvic region to avoid harmful strain and injury to its structures (Wilke *et al.*, 1995:192; Cholewicki & McGill, 1996:1; Vasseljen *et al.*, 2006:905; Kiesel *et al.*, 2007:161). Keeping this in mind, Elfving *et al.* (2003:628), Hides (2005:159) and MacDonald *et al.* (2006:260) have highlighted the impairment found in the deep abdominal and paraspinal muscles which are essential in the local segmental control of the lumbar spine and pelvis. Among the physical deficiencies that might cause long-term low back trouble, lumbar muscle function, with specific reference to the lumbar multifidus muscle or *m. lumbar multifidus*, is considered to be an especially important component of back health (Elfving *et al.*, 2003:620,628; Richardson *et al.*, 2005:149).

Patients with low back problems often have reduced isometric back muscle endurance (Suter & Lindsay, 2001:E361; Hakkinen *et al.*, 2003:1070; Keller *et al.*, 2004:7) and reduced back muscle strength (Crossman *et al.*, 2004:631; Keller *et al.*, 2004:7; Kim *et al.*, 2004:127), probably as a consequence of disuse and pain-related deconditioning. This is why several researchers have produced numerous scientific studies and discussions regarding the *m. lumbar multifidus*, as it pertains to low back pain, biomechanics, rehabilitation, manual therapy, clinical anatomy, and surgery over the past several decades. These include Morris *et al.* (1962:509), Pauly (1966:223), Jonsson (1970:5), Mattila *et al.* (1986:732), Lehto *et al.* (1989:3070), Rantanen *et al.* (1993:568), Ng *et al.* (1997:954), Roy *et al.* (1997:409), Weber *et al.* (1997:1765), Kader *et al.* (2000:145), Danneels *et al.* (2001:186), Hermann and Barnes (2001:971), Kay (2000:102), Kay (2001:17), Elfving *et al.* (2003:628), Hakkinen *et al.*

(2003:1071), Crossman *et al.* (2004:630, 632), Hides (2005:149), Bradl *et al.* (2005:275), Choi *et al.* (2005:768), Johnson & McCormick (2005:772), Hodges *et al.* (2006:2926), MacDonald *et al.* (2006:254) and Kiesel *et al.* (2007:161).

Because of lumbar muscle deconditioning, it is important that new and more efficient clinical evaluation methods are developed to specifically address the problem and mechanism of joint protection and level of impairment (Richardson, 2005:7; Mayer *et al.*, 2005:2556; McGill, 2007:189). However, the requirement for more efficient evaluation and treatment methods should be a cross-scientific approach. In order to master the use of motion and loading in low back treatment, a unique combination of medical- and exercise sciences must be utilised to help evaluate, identify and treat the problematic condition of clinical instability (Richardson *et al.*, 1999:1-2).

Patients are commonly informed that they have improved because they are able to successfully perform more advanced exercises. Precisely how clinicians are making such judgments regarding the effects of rehabilitation exercise programmes is not clear (Hagins *et al.*, 1999:547). Therefore, it would be fair to say that objective, quantitative evaluation assessments would be essential to judge functional recovery of patients, however, the problem is that evaluation equipment has become too large to use (Gubler-Hanna *et al.*, 2007:920). This is why several researchers have developed or tested different evaluation apparatus for measuring muscle performance in a clinical office setting (Helewa *et al.*, 1981:353; Hyde *et al.*, 1983:420; Hyde *et al.*, 1983:424; Giles, 1984:36; Helewa *et al.*, 1986:1044; Helewa *et al.*, 1990:965; Goldman *et al.*, 2003:95; Stokes *et al.*, 2005:116; Costa *et al.*, 2006:48; Martin *et al.*, 2006:154; Descarreaux *et al.*, 2007:92).

However, apart from these evaluation apparatus being too large to use (Gubler-Hanna *et al.*, 2007:920), Choi *et al.* (2005:770) have also mentioned the unnecessary utilization of expensive specialized equipment (e.g. the MedX Lumbar Extension System) in postoperative lumbar rehabilitation. Secondly, apart from the problem with large and expensive apparatus, certain methods of assessing low back muscle performance are widely criticised and debated.

Arguably, the most widely known and used test is the Biering-Sorensen test. Although the Biering-Sorensen test is the most widely used test in published studies, it has several major drawbacks (Demoulin *et al.*, 2006:47):

- The contribution of the hip extensor muscles apart from the paraspinal muscles (most notably *m. lumbar multifidus*) is unknown.
- The better position-holding performance of females compared to males remains partly unexplained.
- The different modifications of the test in research studies have led to the absence of a single standardized test protocol which is an impediment to comparative studies.
- Individual factors such as motivation and pain tolerance can lead patients to stop the test.
- The lumbar compression load of 4 000 N is above the recommended value of the National Institute of Occupational Security and Health (NIOSH) which may classify the test as high risk.
- The impossibility of the test to quantify the relative muscle strength developed by an individual is one of the most important shortcomings.

Despite these shortcomings, the Biering-Sorensen test has become the tool or standard of reference for assessing muscle performance in low back pain patients, especially pre- and post rehabilitation (Demoulin *et al.*, 2006:47; McGill, 2007:211). Apart from the unsolved problems with low back evaluation apparatus and testing methods, it is also important to be aware of the shifting trends, regarding technology development in the global orthopaedic industry. For instance, different opportunities have emerged regarding new assessment technology in the spinal and medical device markets worldwide, which are driven by various factors.

1.3 EMERGING OPPORTUNITIES IN MUSCLE ASSESSMENT TECHNOLOGY

The global orthopaedic market is estimated to have been worth approximately 29 billion USA dollars (\$) in 2006, with the spinal market becoming the largest sector during 2006, surpassing the knee implant sector for the first time (Taylor, 2007). One of the major factors or trends that are driving these global markets is the world's ageing population (Taylor, 2007; Driscoll & Watson, 2008). It can be argued that the strong growth in these markets, and specifically the spinal market, may stimulate extra growth in the rehabilitation medicine market, with specific reference to post-operative back rehabilitation.

This may lead to the emergence of new rehabilitation technology that is reflected in the design and development of various portable and inexpensive muscle strength evaluation devices as recently tested by various researchers in a clinical office setting (Rezasoltani *et al.*, 2003:7; Li *et al.*, 2006:411; Martin *et al.*, 2006:154; Nobori & Maruyama, 2007:9). These portable muscle evaluation devices form part of the growing, global market sector for rapid testing devices, estimated to be already worth \$4 billion during 2002. Diagnostic, rapid testing devices are also proving themselves to be increasingly accurate and reliable and are also becoming smaller and more portable (Clinica Reports, 2002; Ward & Clarkson, 2004:2).

Taken the mentioned difficulties with cumbersome apparatus, the shortcomings of the Biering-Sorensen test, plus ongoing medical technology development in a fast changing and growing global spinal market, it was decided to develop a new portable muscle assessment device for low back pain and to use air pressure as the preferred medium of resistance.

1.4 PRESSURE AIR BIOFEEDBACK: A NEW DIRECTION

A common problem faced by rehabilitation specialists is the fact that objective strength assessments of patients in a typical office setting have been limited because of the availability of mainly heavy, large-scale, expensive equipment

and the lack of portable, inexpensive, clinical office-testing devices. This has led to a shortage of standardised or validated methods for strength assessments in a clinical office setting (Rainville *et al.*, 2003:2466; Martin *et al.*, 2006:154). Accurate estimates of muscular strength and function are essential to guide clinicians in analysing and evaluating treatment effects of rehabilitation programmes (Helewa *et al.*, 1981:353; Giles, 1984:36). This is why researchers like Helewa *et al.* (1981:353), Giles (1984:36), Helewa *et al.* (1986:1044), Helewa *et al.* (1990:966), Axen *et al.* (1992:2) and Richardson *et al.* (1999:122) have developed and used air pressure biofeedback devices (e.g. a modified sphygmomanometer and a compressible ball) to objectively measure muscle strength and function of patients in a clinical office setting.

In the 1970s, the isokinetic dynamometer was widely promoted as a muscle exercise and strength testing device by measuring opposing muscle groups by controlling the speed and movement of the joint (Thistle & Hislop, 1967:279; Moffroid & Whipple, 1969:739). According to Helewa *et al.* (1981:353) and Helewa *et al.* (1986:1044), this dynamometer has been tested to determine its accuracy at different speeds with various weights, but its large size and high cost limits its use in clinical practice, which is why Helewa *et al.* (1986:1044), Helewa *et al.* (1990:966), Axen *et al.* (1992:2) as well as Richardson *et al.* (1999:122) developed and tested air pressure devices for objective, “simple to use” clinical office testing.

According to Helewa *et al.* (1981:353) and Giles (1984:37), the modified sphygmomanometer fulfilled the criteria for validity as stipulated by them. These are: that any evaluation device used should be quantitative so that numbers may be produced; be objective so that observer impressions do not affect the measurement values; be sensitive to changes in muscle strength; be reliable in that the instrument is free from defect and does not require frequent maintenance; be reproducible in the hands of different observers; and be portable so that it may be used in different settings. Further to this, it also should be fast, safe, comfortable, simple to apply and inexpensive. According to these researchers, devices with these properties would be considered valid.

One of the most telling factors resulting from objective feedback provided by the use of the pressure air biofeedback unit is the tendency for patients to be trained at loads far in excess of that which they could control, thereby causing injury to the lumbar spine (Richardson *et al.*, 1999:146-147). Persons must be trained to cope with the mechanical loads inherent in their daily activities, work practices or sport. What must be ensured is that the deep muscles are trained to supply the inner support so that forces absorbed by the global muscles in high or rapid loading events can be transferred safely and efficiently to the passive structures of the spinal column (Richardson & Jull, 1995:2; O'Sullivan *et al.*, 1998:114; Cholewicki *et al.*, 2000:1377-1378; Radebold *et al.*, 2000:953; Comerford & Mottram, 2001a:11-13). This is why Richardson *et al.* (1999:122) and Hides *et al.* (2005:218) advocated the use of the stabilizer, an air pressure biofeedback device that objectively identifies how well the *m. transversus abdominis* can be contracted without using a posterior pelvic tilt.

The research of these scientists has stimulated the current research study to develop and test a pressure air biofeedback device (PAB) that allows non-invasive identification of *m. lumbar multifidus* contraction and rehabilitation training of this most important paraspinal muscle in low back patients. Research to discover whether certain clinical tests reflect automatic deep muscular function or control in functional activities, or indeed research to develop certain measures capable of demonstrating whether problems do exist in the deep muscles during such activities, presents the ultimate challenge to the treatment of motor control problems in the deep muscles of low back pain patients (Richardson *et al.*, 1999:155; Comerford & Mottram, 2001a:11).

Of particular importance would be the development of a clinical office assessment device that is reliable and valid and according to Li *et al.* (2006:411), may represent a technical advance in portable muscle strength devices that provide comparable information to those obtained by isokinetic dynamometers at a fraction of the cost and size. With respect to cost, the Cybex isokinetic dynamometer for example, costs anything between \$38 000 or 290 000 South-African rand (R) for a used one, to \$64 000 (R490 000) for a new one, while it also covers a large floor space of seven square metres (2.34m x

3.02m) (Hitech Therapy, 2009). These values have been calculated at an exchange rate of R 7.69 to the dollar, as traded on 22 July 2009. On the other hand, the PAB device is a small, portable device, which will sell at two to four percent of the price of a new Cybex isokinetic dynamometer (Pienaar & Webster, 2009).

For these reasons, an office exercise-testing instrument, called the PAB device, has been developed to try and quantify the deep stabilising contraction of the *m. lumbar multifidus* in a functional body position such as upright sitting. Also, over-the-counter diagnostic products have proved themselves to be increasingly accurate and reliable, while clinicians and patients are moving from the hospital set-up to the convenience of clinical office and even home testing (Clinica Reports, 2002). Furthermore, the prevalent and pervasive impact of musculoskeletal conditions globally, led to the endorsement of the Bone and Joint Decade 2000-2010 by the United Nations and WHO. This is the reason why a portable, simple to apply, inexpensive and reliable office muscle-testing device, specifically addressing the low back problem, is more than ever relevant.

1.5 OBJECTIVE

This study will investigate the reliability and validity of a newly developed pressure air biofeedback device (PAB) in measuring the muscle strength and postural contraction of the *m. lumbar multifidus* described by Hides (2005:67-71) and Richardson (2005:4), that are based on the impairment found in these specific paraspinal muscles of low back pain patients (Hides, 2005:158-161). Therefore, the main purpose of the study is three-fold:

- To develop a PAB device that reliably demonstrates the *m. lumbar multifidus* strength contraction in a closed chain loaded, upright sitting, back extension test in asymptomatic subjects and low back pain patients.
- To determine if a correlation exists between PAB pressure output in milliBar (mb) and EMG activity (μV) of *m. lumbar multifidus* contraction in

a closed chain loaded, upright sitting, back extension test in asymptomatic subjects and low back pain patients.

- To determine if a correlation exists between PAB pressure output (mb) and muscle cross-sectional area change (CSA in cm^2) of *m. lumbar multifidus* contraction in a closed chain loaded, upright sitting, back extension test in asymptomatic subjects and low back pain patients.

Apart from these research objectives, it is also important to formulate the null and research hypotheses for this research study as follows:

Null hypotheses: The PAB device does not measure the *m. lumbar multifidus* strength contraction in a closed chain loaded, upright sitting back extension test.

Alternative hypotheses: The PAB device measures the *m. lumbar multifidus* strength contraction in a closed chain loaded, upright sitting back extension test.

It is important to note, that the effects of the PAB device were tested against an alpha level of $p < 0.05$ for statistical significance. This decision was taken because the PAB device was developed as a low risk, lumbar spine extension testing device, that would not incur injury to low back pain patients or asymptomatic subjects.

With respect to the global epidemic of low back pain and its influence and impact on micro- and macroeconomic and social systems worldwide, the epidemiology of this much debated and researched musculoskeletal condition, will be discussed in Chapter Two. In Chapter Three, the patho-physiology of low back disorders, with specific reference to biomechanics, injury mechanisms and the lumbar multifidus muscle mechanism, will be discussed. In Chapter Four, the development of the PAB device is explained, with respect to emerging technologies, development of modern dynamometry, and the design and validation of the PAB device. Chapter Five will give a detailed explanation of the methodology used to scientifically assess subjects with respect to anthropometry, PAB device versus EMG testing, and PAB device versus US

testing. In Chapter Six, the results will be reported and statistically analyzed in detail. In Chapter Seven, a detailed discussion of the study is done, while in Chapter Eight, conclusions, recommendations and limitations of the study are given.

CHAPTER TWO

EPIDEMIOLOGY OF LOW BACK PAIN

- 2.1 INTRODUCTION
- 2.2 EPIDEMIOLOGY OF LOW BACK PAIN WORLDWIDE
 - 2.2.1 Australia
 - 2.2.2 Japan
 - 2.2.3 The Netherlands
 - 2.2.4 South Africa
 - 2.2.5 Sweden
 - 2.2.5.1 Direct and indirect costs of low back pain in Sweden
 - 2.2.5.2 Total cost of chronic low back pain in Sweden
 - 2.2.6 The United Kingdom
 - 2.2.6.1 Annual prevalence of low back pain in the United Kingdom
 - 2.2.6.2 Cost of general practice in the United Kingdom
 - 2.2.6.3 Cost of healthcare professionals in the United Kingdom
 - 2.2.7 The United States of America
 - 2.2.7.1 Epidemiology of lumbar surgery in the United States of America
 - 2.2.7.2 Cost analysis of low back pain in the United States of America
- 2.3 COMPENSATION CLAIMS
- 2.4 EPIDEMIOLOGY OF LUMBAR REVISION SURGERY
- 2.5 REVIEW OF TREATMENT MODALITIES IN LOW BACK PAIN
- 2.6 LUMBAR SUPPORT BELTS
- 2.7 EPIDEMIOLOGY OF LOW BACK PAIN IN SPORTSMEN AND WOMEN
 - 2.7.1 Athletics
 - 2.7.2 Ball Sports
 - 2.7.3 Gymnastics and Dance
 - 2.7.4 Strength Sports
 - 2.7.5 Water sports
- 2.8 CHILDREN AND LOW BACK PAIN
- 2.9 SUMMARY

2.1 INTRODUCTION

Pain, by definition, is a subjective phenomenon and thus, any measure of pain that is not based on a person's self-report is likely to be inaccurate. As to the validity of self-reported low back pain data, the basic problem concerns the underlying pathology and mechanism causing low back pain and various factors influencing a person's response to symptoms (Salminen *et al.*, 1992:1038). With this in mind, it can be said that low back pain is frequently diagnosed and most commonly treated in primary healthcare settings (Ekman *et al.*, 2005:1777; van Tulder *et al.*, 2006:S64). Further to this, it appears that low back pain is especially frequent during the most productive years of a person's life, causing major societal, industrial and personal problems resulting in substantial annual healthcare costs, lost productivity and disability (Hides *et al.*, 1995:54; Cassidy *et al.*, 2005:2817; Hansson & Hansson, 2005:337; Dagenais *et al.*, 2008:8).

Epidemiological studies show that low back pain affects 50%-90% of us during our lifetime, 15%-30% of us at any given time while the annual incidence rate is calculated at five percent (Skovron *et al.*, 1994:129; Andersson, 1998:28; Maniadakis & Gray, 2000:95; Pai & Sundaram, 2004:1; Cassidy *et al.*, 2005:2817; Descarreaux *et al.*, 2005:185). With reference to the different sexes, males and females are equally affected with the peak incidence at approximately 40 years of age (Frymoyer, 1988:291; Helliuvaara, 1989:257; Herring & Weinstein, 1995:1172). Also, 12%-26% of children and adolescents are reported to experience lumbosacral pain (Fairbank *et al.*, 1984:461; Salminen, 1984:1).

However, the natural history of lumbosacral pain appears to be favourable because up to 90% of cases of low back pain apparently resolve without medical attention in 6-12 weeks while 40%-80% are symptom free within one to two weeks (Berquist-Ullman & Larsson, 1977:1; Coste *et al.*, 1994:578; Hansson & Hansson, 2005:337). It has also been found that 75% of patients with sciatica reported relief of pain after six months (Weber, 1983:131). However, these statistics, which may appear reassuring, should be tempered by the knowledge that as many as 70%-90% of patients have recurrent episodes of

low back pain (Herring & Weinstein, 1995:1172; van den Hoogen *et al.*, 1997:1515; Hides *et al.*, 2001: E243; Rasmussen-Bar *et al.*, 2003: 233).

Given the widespread occurrence of low back pain in the global general population, it has also been found that excessive physical activity that is representative of competition sports, can lead to acute dynamic overload and/or chronic repetitive exertion of the mechanical lower back (Herring & Weinstein, 1995:1172). Thus, it may be universally present in the physically active as well as physically competitive population, which will be discussed under Paragraph 2.7.

According to the above-mentioned information, as well as the National Institute for Occupational Safety and Health (NIOSH) in the USA (NIOSH, 1997; Waddell, 1996:2820; Davis & Kotowski, 2005:453), it can be said that low back pain represents the most common and costly musculoskeletal disorder. In more calculated terms, low back pain accounts for disability in three to six percent of the population per annum. Despite widespread opinion, research studies show that 78% of patients who experience a first episode of low back pain still have pain after a six month period. Twenty-six percent (26%) of these patients experience significant disability (Wahlgren *et al.*, 1997:213; Quittan, 2002:423). Although this represents only a small percentage, the cost of hospitalization and invalidity are considerable. Research data further reveals that within one year, one to two percent of patients do not return to work because of their low back pain (Frymoyer & Cats-Baril, 1991:263; Quittan, 2002:423).

This substantial epidemiological and economic impact of low back pain on society is also expected to increase in time (Maniadakis & Gray, 2000:95). This can already be seen in countries like the USA, the United Kingdom (UK), Sweden, the Netherlands and Australia, to name just a few. According to Borenstein (1999:131), Luo *et al.* (2004:79), Kopec *et al.* (2004:70), Carragee and Hannibal (2004:7) and Demoulin *et al.* (2006:43), the lifetime incidence of back pain in the USA and other countries is between 60% and 80%. Norris (1995:61) found that the same tendency has occurred in the UK where as much as 80% of the population has experienced back problems at one stage or

another. Also, Foster *et al.* (1999:1332), Jellema *et al.* (2001:377) and Oksuz (2006:E968) reported that low back pain is also a very common health problem in developing as well as developed countries and as much as 70% of all back problems are specifically related to the lumbar region. This also has been the experience of other researchers like Lanes *et al.* (1995:801), Andersson (1999:581) and Nourbaskhsh and Arab (2002:447).

Of great concern however is the fact that worldwide, lumbar back injuries increased dramatically over the last few decades. Nachemson already reported that over a period of 20 years, low back injuries increased by an estimated 2 500% in North America and by 4 000% over a 30 year period in Sweden (Nachemson, 1990:533). These increases also reflect the low back epidemic in other countries like the UK, the Netherlands, Australia and the USA.

Apart from the epidemiology of low back pain, it is also essential to broaden one's knowledge of the anatomy and biomechanics of the lumbar spine to better understand the pathophysiology of low back pain caused by micro- or macrotrauma and to continually improve low back rehabilitation and treatment programmes. Therefore, the pathophysiology of low back pain with specific reference to the biomechanics, musculoskeletal injury mechanisms and the *m.* lumbar multifidus mechanism, will be discussed in Chapter 3.

2.2 EPIDEMIOLOGY OF LOW BACK PAIN WORLDWIDE

Since low back pain is a common and often disabling condition during the most productive years of a person's life, the burden of illness to society is high in terms of medical costs, lost workdays and production losses (Waddell, 1991:719; Guo *et al.*, 1995:591; van Tulder *et al.*, 1995:233; Andersson, 1999:581; Walker, 2000:205; Cassidy *et al.*, 2005:2817; Ekman *et al.*, 2005:1778; Hansson & Hansson, 2005:337; Volinn *et al.*, 2005:697; Dagenais *et al.*, 2008:8). Following is a detailed description of the global burden of low back pain.

2.2.1 Australia

There is a high prevalence of low back pain in the Australian adult population demonstrating the magnitude of this particular health problem in this country (Walker *et al.*, 2004:242). The findings of this study clearly reflect a sample that is generally representative of the Australian adult population. The prevalence estimates and Chronic Pain Grade Questionnaire (CPG) grades (I-IV) are very close to those reported by Cassidy *et al.* (1998:1860) in Canada. However, the prevalence estimates are moderately higher than those reported in Belgium (Skovron *et al.*, 1994:129), Denmark (Harreby *et al.*, 1996:312) and the UK (Walsh *et al.*, 1992:227; Hillman *et al.*, 1996:347).

Even though 64% of the entire sample had had low back pain in the six months prior to the survey date, two thirds of these had a grade I CPG score, meaning low-intensity pain and low disability from it (Walker *et al.*, 2004:241), arguably the reason why the majority (55.5%) did not seek medical care (Walker *et al.*, 2004:327). In the last six months, 1 235 of 1 913 respondents (64.6%) reported at least one episode of low back pain, where 64.1% of these had grade I pain, 16.7% had grade II pain and 16.0% had grade III-IV pain. Of those adults with low back pain, 44.5% (95% confidence interval, 41.8-47.3) did seek medical care. Significantly more subjects visited a general practitioner (11.8%) or a chiropractor (14.3%) compared to a massage therapist (6.2%). Similarly, significantly more subjects visited a chiropractor rather than a physiotherapist (8%). Also, women sought care for low back pain more often than men (Walker *et al.*, 2004:333).

Although 42.6% of the adult population had experienced low intensity pain and disability from low back pain, the high level of significant disability associated with low back pain should be a cause for concern. Over ten percent (10.5%) had been significantly disabled by low back pain in the past six months, resulting in considerable time off from their usual activities (Walker *et al.*, 2004:243).

2.2.2 Japan

Musculoskeletal pain, as an occupational health problem, is one of the most common disorders, especially in construction workers. In 1996, the Japanese Ministry of Labour reported that low back pain was the primary cause (about 60%) of occupational sick leave for a minimum of four days, while in 1994 they reported that the incidence of low back pain in construction workers was next highest to transportation workers (Yamamoto, 1997:173; Ueno *et al.*, 1999:449; Kaneda *et al.*, 2001:316). Compared to other countries, a nationwide health survey in the USA showed that construction workers are the highest risk group for work-related low back pain (Guo *et al.*, 1995:591).

According to another report published by the Labour Ministry of Japan, the number of workers with occupational disease, reported in 1979, was 13 807, while 11 564 of them had low back pain, which is a remarkable 83.8%. In 1997, the rate of occupational illnesses decreased to 6 034, while the rate of low back pain cases still remained as high as 83.5% (5 041) (Miyamoto *et al.*, 2000:186; Kaneda *et al.*, 2001:310). Also, in 1990, five percent of all indemnity claims under Workers' Accident Compensation in Japan reflected disabling "back injuries", while 25% were claimed under workers' compensation in the USA (Hadler, 1994:1113).

The study of Matsui *et al.* (1997:1245) showed that the highest prevalence of low back pain in Japan was in the age range of 40-49 years for Japanese men (73.1%) and 50-59 years for Japanese women (61.1%). The perceived cause found to be most associated with low back pain was lifting, especially in workers with jobs requiring moderate to heavy work. Interestingly, obesity was not a risk indicator for low back pain (Matsui *et al.*, 1997:1242). Similar patho-ergonomical factors were found to play a role in low back pain among Japanese hospital nurses (Ando *et al.*, 2000:211).

With respect to the vibration syndrome (found in Japanese long distance truck drivers) as an occupational health hazard, Miyashita *et al.* (1992:349) reported that low back pain (38%-50%) was the second most frequent complaint behind

stiff shoulder (43.5%-56.8%) among power shovel, bulldozer and forklift workers in Japan. Miyashita *et al.* (1992:350) also reported that back pain complaints in 211 young tractor drivers increased from 20% in 1961 to 57% in 1971 (from 106 drivers) with some significant radiological changes.

2.2.3 The Netherlands

In the Netherlands, it is estimated that there are between 60 000 and 75 000 new cases of lumbosacral radicular syndrome (LRS) each year. The LRS is based on a lumbar disc prolapse, characterised by irradiating pain over an area of the buttocks or legs, innervated by one or more spinal nerve roots of the lumbar vertebrae or sacrum, combined with the phenomena associated with nerve root tension or neurological deficit (Ostelo *et al.*, 2003:209). According to these researchers, in the Netherlands, with a population of about 16 million people, 10 000 to 11 000 operations are performed each year because of the lumbar disc prolapse.

The study of van Doorn (1995:62) reported that in a self-employed subset of Dutch dentists, veterinarians, physicians and physical therapists, 23% of compensation claims lasted longer than six months or were deemed chronic. According to them, the risk of chronic back pain increased with older age. It is estimated that the total indirect costs of back pain for the entire labour force of the Netherlands in 1991 were 4.6 billion USA dollars (\$), of which the breakdown was calculated at \$3.1 billion for absenteeism and \$1.5 billion for disablement. It was further calculated that the total direct medical costs of back pain in the Netherlands were \$367.6 million (Jellema *et al.*, 2001:377; van Tulder *et al.*, 1995:238) and consisted of \$200 million for hospital care (56.5%), \$139 million for paramedical care (36%), \$22 million for general practice care (6%) and \$6.6 million for medical specialist care (1.8%).

Because of the unavailability of certain information, such as pharmaceutical care, artificial devices, nursing homes, home nursing, management and administration, as well as limited disclosure of medical insurance figures, van Tulder *et al.* (1995:238) have substantially underestimated the direct medical

costs of back pain. These researchers agree that direct medical costs of back pain in 1991 in the Netherlands might have been as high as \$550 million.

2.2.4 South Africa

South African statistics have shown that lower back pain is one of the most commonly treated conditions by physical therapists and that 80% of the South African population will experience back pain at some time, due to problems arising from low back disorders (Belot, 2005; Health 24, 2009). According to van Vuuren *et al.* (2007), 30 000 South Africans suffer daily from back and neck problems, while 10% of them will become chronic cases. It is estimated that workers compensation for low back pain has cost the South African economy two billion rand per year since 2000 and is escalating on a yearly basis (Belot, 2005; van Vuuren *et al.*, 2007; Health 24, 2009).

In a review of 27 epidemiological studies of low back pain in Africa (Louw *et al.*, 2007:105), it was found that 63% of the studies were conducted in South Africa and Nigeria. Of these 63%, 37% were done in South Africa and the remaining 26% in Nigeria. The most common population group involved workers (48%), while 15% of the population comprised of scholars. Analyzing the low back pain prevalence of these specific groups, it was found that the average one year prevalence of low back pain was 33% among adolescents and 50% among adults. The average lifetime prevalence of low back pain calculated to 36% among adolescents and 62% among adults. The findings of this study support the global burden of low back pain and highlight the rise in the prevalence of low back pain in Africans.

2.2.5 Sweden

Ekman *et al.* (2005:1781) and Hansson and Hansson (2005:344) calculated the direct, indirect and total costs of low back pain in Sweden. In general, low back pain data from Sweden indicated that 11%-19% of all annual sickness days since 1961 can be attributed to low back pain (Andersson, 1999:581; Ekman *et al.*, 2005:1778). According to the National Social Insurance Board of Sweden,

the 2001 expenditures for back pain represented 11% of the total costs for short-term sickness absence in Sweden and 13% of all early retirement pensions were granted for back problems (Ekman *et al.*, 2005:1778). Hansson and Hansson (2005:337) further reported that annual costs for those sick-listed in Sweden due to low back or neck problems have been estimated at 3.5 billion euros (€) corresponding to nearly one-third of the nation's total health-care costs of 1995.

With respect to the difference in the rate of surgery between Sweden and several other countries, Hansson and Hansson (2000:3055) reported that the rate of surgery in Sweden was five times less than the USA and three times less than the Netherlands. However, the burden of low back pain in terms of costs was still high as specifically described below.

2.2.5.1 Direct and indirect costs of low back pain in Sweden

Based on a sample of 302 patients of Ekman *et al.* (2005:1780-1781), the most commonly prescribed medications for low back pain were analgesics, nonsteroidal anti-inflammatory drugs (NSAIDs), muscle relaxants and antidepressants. The total yearly pharmaceutical cost per patient was €183. Low back pain patients consulted general practitioners (GPs) and specialists on average three times during the previous 6 months and physiotherapists and chiropractors an average of 8.2 times. Radiograph examinations represented 71% of all radiologic examinations and diagnostic tests. Taking all the above into consideration, the total average annual direct cost per patient amounted to almost €3 090. Concerning indirect costs of low back pain, the average yearly cost per patient for early retirement was estimated at €2 774. The largest indirect cost item was sickness absence from work, which calculated to an average yearly cost per patient of €9 563 (Ekman *et al.*, 2005:1781).

2.2.5.2 Total cost of chronic low back pain in Sweden

The total annual direct costs per patient were estimated at €3 090 (in 2002 prices) or 15% of the total annual low back pain costs. The indirect costs

(mainly productivity loss and reduced work capacity) were estimated at €17 600 per patient or 85% of the total low back pain costs. There was a significant difference in total costs between men and woman, €17 800 and €23 300 respectively. Woman incurred both larger direct and larger indirect costs (Ekman *et al.*, 2005:1781). According to Hansson and Hansson (2005:344), the total direct costs (health-service costs) and indirect costs (production losses) for a group of 1 822 Swedish patients who were sick-listed for more than 28 days over a two year period, totalled almost €47 million.

The indirect costs for absenteeism and early retirement are substantially higher than the direct costs for pharmaceuticals, medical visits, physiotherapy and hospitalizations for Swedish low back pain patients. Therefore, there is a great need for treatment therapies that have the potential to reduce the high indirect costs (Ekman *et al.*, 2005:1784). Also, the results of this study confirm what most other research studies have concluded, which is that the small percentage of patients with chronic low back pain account for the largest percentage of costs where indirect costs for sick leave and early retirements represent 92% of the total costs (Maetzel, 2002:23). Similar figures have been obtained in a Dutch and British study where indirect costs contributed 93% and 87% of the total costs respectively (Van Tulder *et al.*, 1995:238; Maniadakis & Gray, 2000:95).

2.2.6 The United Kingdom

In the research study of Rudy *et al.* (1995:2547), it was found that 11.7 million British patients had back problems, of which 2.6 million patients were temporarily disabled and 2.6 million permanently disabled for any type of job. In the United Kingdom (UK) during the decade 1983-1993, outpatient attendances for back pain increased fivefold (Palmer *et al.*, 2000:1577). The same researchers also found that over a 10 year interval between two surveys (1987-88 and 1997-98), the one year prevalence of low back pain rose from 36.4%-49.1% respectively (Palmer *et al.*, 2000:1578). In terms of hospitalisation, McCombe *et al.* (1989:908) mentioned that 23% of all patients that had been admitted to the Royal Orthopaedic Hospital had lumbar back problems. In 1993,

hospital costs for the National Healthcare services of Britain amounted to 480 million pounds (£) (Foster *et al.*, 1999:1332).

In one of the most comprehensive cost analysis studies of low back pain in the UK, Maniadakis and Gray (2000:96) were able to make use of more precise and recent data in their research study. They calculated that 35% of the total cost of low back pain related to private sector services and are most likely paid for directly by patients and their families. They estimated the direct health care cost of low back pain in 1998 to be £1 632 million. With respect to the distribution of costs across different health care providers, 37% related to physiotherapy and allied specialists care, 31% related to hospital care, 14% related to primary care, seven percent (7%) to medication, six percent (6%) to community care and five percent (5%) to radiology, imaging and investigation purposes.

Overall, back pain imposes a greater economic burden than any other disease for which economic analysis has been carried out in the UK (Maniadakis & Gray, 2000:101). In particular, as shown in Figure 2.1, back pain is more costly than coronary heart disease, as well as any other selected diseases in the UK.

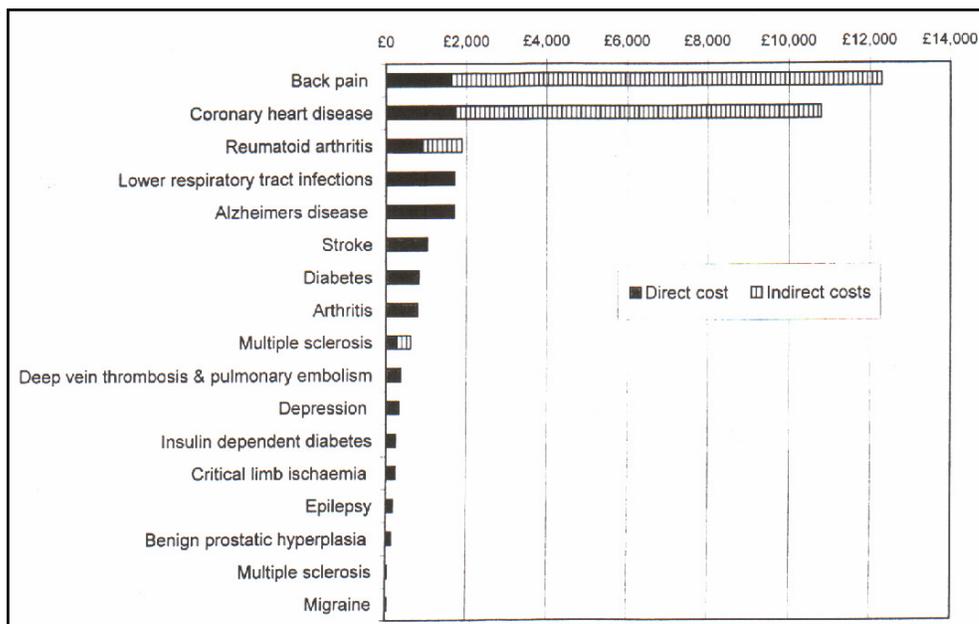


Figure 2.1: Economic burden of selected diseases in the UK (adapted from Maniadakis & Gray, 2000:102).

2.2.6.1 Annual prevalence of low back pain in the United Kingdom

According to Maniadakis and Gray (2000:96-97), the Office of Population Censuses and Surveys (OPCS) in 1997 reported the annual prevalence rate of low back pain in the UK as 37%. For 20% of low back sufferers, the pain had started within the previous 12 months, while 19% had suffered throughout the whole year. Applying these OPCS figures to the UK adult population of 47.7 million (Office of National Statistics, 1998), they suggest that the annual incident population experiencing low back pain is about 3.5 million and the prevalent population experiencing low back pain is 17.3 million. They further determined that adults suffering from low back pain during the entire year, calculated to 3.1 million.

2.2.6.2 Cost of general practice in the United Kingdom

Maniadakis and Gray (2000:97) further reported that population surveys (OPCS, 1997) show that 12% to 16% of all British adults visit their GPs yearly due to low back pain. However, studies based on GP medical records suggested lower consultation rates. According to a survey of 500 GP records (Intercontinental Medical Statistics during 1993), 9.4% of adults over 15 years of age consulted with low back pain. Further to this, a well validated and widely used survey in the UK, the fourth national study on Morbidity Statistics from General Practice (OPCS, 1996) reported that each year 8.4% of adults consult a general practitioner due to low back pain. Applied to the UK population of 1998, these data suggest that 5 million adults consult with a medical practitioner on a yearly basis. Thus, the total cost of primary care related to back pain in 1998 is estimated at £140.6 million.

2.2.6.3 Cost of healthcare professionals in the United Kingdom

A low back pain consultation and cost analysis of a few medical professions in the UK will be stipulated and discussed in short.

- According to Maniadakis and Grey (2000:95-97), three percent of those suffering from low back pain visit a private consultant, which calculates to 0.52 million (520 000) adult persons per year (OPCS, 1997). These figures yield a cost of £72.7 million for private consultations.
- Looking at physiotherapy figures, 9% of those suffering from low back pain visit a physiotherapist (OPCS, 1997). This implies that 1.6 million adults receive physiotherapy per annum in the UK. Assuming seven visits per patient gives an estimate of 10.9 million sessions of physiotherapy (Maniadakis & Grey, 2000:97). According to Foster *et al.* (1999:1332), in Britain, physiotherapists as part of the allied medical profession play a key role in the management of low back pain, treating approximately 1.3 million patients each year at a cost of \$98 million per annum. Maniadakis and Grey (2000:97) calculated the cost for private physiotherapist visits at £100.5 million per annum.
- Five percent of low back sufferers visit an osteopath, which calculates to 0.86 million (860 000) adults consulting per annum (OPCS, 1997) and 4.3 million sessions. This calculates to a total cost of £172.8 million per annum (Maniadakis & Grey, 2000:96).
- Two percent of those experiencing low back pain visit a chiropractor (OPCS, 1997). This implies 0.35 million (350 000) adults consulting and 1.7 million sessions spent at a chiropractor which is in line with the estimate of the Working Party for Chiropractors (1993) which estimated a total of 3.9 million consultations in the UK of which 50% are due to back pain. At an estimated cost of £40 per visit, the total cost for chiropractic treatment calculates to £69.1 million per annum (Maniadakis & Grey, 2000:97).

A consultation and cost analysis of only a few medical professions in the UK has been highlighted above. It excludes an analysis of costs in hospital care, community care, medication and radiology. However, the direct cost of back pain is insignificant compared to the cost of informal care and the production

losses related to it, which total £10 668 million. According to Foster *et al.* (2003:239), a cost-of-illness study of low back pain in the UK estimated that the overall costs varied between £6.6 billion and £12.3 billion depending on the cost method used. Overall, low back pain is one of the most costly conditions for which an economic analysis has been carried out in the UK and this is in line with findings in other countries (Maniadakis & Gray, 2000:95).

2.2.7 The United States of America

Low back pain is a major public health problem in the USA and is reaching epidemic proportions (Deyo, 1998:48; Pai & Sundaram, 2004:1). According to Bratton (1999:2299), Lively (2002:643), Carragee and Hannibal (2004:7) and Pai and Sundaram (2004:1), low back pain is the most common cause of disability in patients younger than 45 years of age and is also the main reason for which people under 45 years of age limit their physical activities. It is the second leading symptomatic cause for physician visits (Hart *et al.*, 1995:11; Pai & Sundaram, 2004:1), the third most common cause for surgical procedures and the fifth most common reason for hospitalizations in the USA (Taylor *et al.*, 1994:1207; Hart *et al.*, 1995:11; Wolsko *et al.*, 2003:292; and Pai & Sundaram, 2004:1).

Further to this, more than seven million American patients were treated for chronic lumbar back pain in 1994 and the numbers increase by approximately two million a year (Lahad *et al.*, 1994:1286). The research of Hart *et al.* (1995:11), as well as the data of Andersson (1999:583), indicated that there were almost 15 million physician office visits for low back pain in the USA in 1990, ranking this specific problem one of the top five medical reasons for all physician visits in that country. If looking at chiropractic services in the USA and Canada from 1985 through to 1991, the chiropractic visit rate is twice that reported 15-20 years ago and mirrors the significant increase in chiropractor practices (Hurwitz *et al.*, 1998:775).

Wolsko *et al.*, (2003:295) analysed the amount of visits made to complementary professionals in the USA in 1997, specifically for the treatment of back or neck

pain. Because of the high prevalence of back and neck pain and because of the frequency of visits made to complimentary providers for the treatment, they estimated that >200 million visits were made to complementary professionals in 1997. More than 88 million of these visits were made to chiropractors, while >10 million visits were made to providers of massage, energy healing, relaxation techniques and yoga. All these estimates were based on the 1997 USA adult population estimate of 198 million (Wolska *et al.*, 2003:296). According to the researcher's calculation, it suggests that an estimated 10.41 million back patients visited chiropractors and 9.12 million back patients visited providers of massage, energy healing, relaxation techniques and yoga in the USA in 1997.

2.2.7.1 Epidemiology of lumbar surgery in the United States of America

Investigating low back surgery in the USA, Davis (1994:1117) found that lumbar surgery in both males and females increased by 33% while lumbar fusions in both genders increased by 60%. Also, lumbar intervertebral disc surgery increased by 40% in males and by 21% in females. An international comparison further showed that the rate of back surgery in the USA was at least 40% higher than in any other country and was more than five times those in England and Scotland and more than double the rate for Australia (Cherkin *et al.*, 1994:1203; Andersson, 1999:584; Bozic *et al.*, 2004:1311; Walker *et al.*, 2004:328).

According to the USA National Hospital Discharge Survey, Andersson (1999:584) reported that in 1990, 46 500 lumbar fusions as well as 232 500 low back operations without fusions were done. Volinn *et al.* (1994:64) also examined the USA National Hospital Discharge Survey for the time period 1979-1987, while Taylor *et al.* (1994:1207) extended the time span to include data up to 1990. During these 11 years surveyed, surgery among adults for low back pain increased by 55%, from 147 500 patients in 1979 to 279 000 patients in 1990. According to Andersson (1999:584), this rise corresponds to an increase from 102-158 patients per 100 000 adults. This rise also represents a significant increase of 100% in lumbar fusions, from 13-26 patients per 100 000 adults.

With specific reference to lumbar fusion surgery after 1996, when intervertebral fusion cages were approved in the USA, Deyo *et al.* (2005:1442) reported the following fusion rates in 2001. Of the 356 638 hospitalizations that met their criteria for “definite lumbar” spinal surgery, 122 316 cases of lumbar spinal fusions for degenerative conditions were done, compared to 32 701 operations in 1990. These procedures correspond to age- and sex-adjusted rates of 61.1 operations per 100 000 adults in 2001 compared to just 19.1 operations per 100 000 in 1990. This reflects a two-fold increase of lumbar fusion surgery for degenerative conditions from 1990-2001, or a relative increase of 220%. During the 1980s, lumbar fusions increased by 100% according to Taylor *et al.* (1994:1207), which means that the pace of the increase in fusion surgery has accelerated dramatically since 1996.

To put the accelerated rate of fusion surgery into perspective with other musculoskeletal procedures, Deyo *et al.* (2005:1444) focused on changes in procedures from 1996-2001 (Table 2.1). In 2001, there were almost three times more knee arthroplasties and hip replacements than lumbar fusions for degenerative disease. However, the knee and hip procedures increased only 13%-15% during the years 1996-2001, compared with a 113% increase for lumbar fusions.

Table 2.1: Comparison of lumbar fusion surgery with other musculoskeletal procedures (adapted from Deyo *et al.*, 2005:1444).

	Lumbar Fusions for Degenerative Conditions	Inpatient Hip Replacement	Inpatient Knee Arthroplasty
Procedures, 2001	122 469	329 900	363 536
Increase (%), 1996-2001	113%	13%	15%
Mean hospital days, 2001	4.5	5.4	4.5
Median hospital days	4	4	4
Mean total hosp. charges, 2001	\$39 906	\$28 234	\$25 309
Median total hosp. charges	\$33 119	\$24 017	\$22 335
*National Hospital Bill, 2001	\$4.8 billion	\$9.3 billion	\$9.2 billion
*(mean charges times (x) no. of hospitalizations)			

2.2.7.2 Cost analysis of low back pain in the United States of America

According to Deyo & Bass (1989:501) and Carey *et al.* (1995:787), lumbar spinal injuries in the USA are the single most expensive injury in terms of production loss. Further to this, lumbar spinal injuries are also considered the most expensive orthopaedic problem in industrialised countries in the world (Nachemson, 1990:533; Lahad *et al.*, 1994:1286). In 1990, the financial cost of lumbar spinal injuries to the USA was estimated at \$24 billion (Lahad *et al.*, 1994:1286).

The reason for this enormous financial burden on the USA economy is that this particular injury occurs in 80% of all adult persons in the USA and is the most common musculoskeletal reason for medical care. According to Hazard *et al.* (1996:945) and Cooper *et al.* (1996:2329), these costs have risen to \$50 billion per year in the USA. Apart from the above-mentioned, the prevalence of musculoskeletal disorders has increased as people enter the job market. In this regard, the review of NIOSH has reported that by the age of 35, most people have had their first episode of back pain which, as part of musculoskeletal impairments, is among the most prevalent and symptomatic health problems of middle and old age. In the USA, age groups with the highest rates of compensable back pain and strains are the 20-24 age group for men and the 30-34 age group for women (NIOSH, 1997). Low back pain accounted for 23% (\$8.8 billion) of total workers' compensation payments in 1995 (Murphy & Volinn, 1999:691).

In an attempt to determine the proportion of costs for certain components of back care, Williams *et al.* (1998:2329) researched and reported the following health and back care costs data. Twenty percent (20%) of claimants with back pain for four months or more accounted for 60% of health care costs. The most costly services for low back problems were diagnostic procedures (25%), surgery (21%) and physical therapy (20%). Physician evaluation was 15% of the total cost, whereas 2% was attributed to medication. These data confirmed information reported by Federspiel *et al.* (1989:919) that a small proportion of

individuals with chronic disability account for a disproportionate amount of resources used to treat back problems.

Grazier *et al.* (1984:79) conducted the most comprehensive cost analysis in 1984, estimating the USA back pain costs using three national survey data sets. Frymoyer & Durett (1997:149) later adjusted the Grazier *et al.* estimates to the value of the USA dollar as experienced in the years 1990 and 1994. These adjusted estimates came to \$24.3 billion in 1990 and \$33.6 billion in 1994. More recently Luo *et al.* (2004:79) and Hampton (2004:415) used data from the 1998 Medical Expenditure Panel Survey, a national survey on health care utilisation and expenditures, to calculate the total health care expenditure incurred by individuals with back pain in the USA. According to these researchers, this healthcare cost figure reached \$90.7 billion in 1998.

To quantify the back epidemic in the USA, the scientific research study of Luo *et al.* (2004:81) reported the following:

- A total of 25.9 million adults reported back pain sometime in 1998.
- Of this back pain population, 55% were female and ± 61% were married.
- The average age was 48 years and the majority were white (88.3%).
- The most prevalent back diagnosis was: unspecified back disorders including spinal stenosis, lumbago and sciatica (59.5%).
- Other diagnosis included back sprains and strains (16.2%), disc disorders (14.2%) and other disorders of the cervical region (9.6%).

It is clear from these studies that low back pain has imposed huge burdens on the USA health care system and has a devastating impact on the USA population. Since low back pain is one of the most costly diseases, it is important to identify the determinants of health care expenditures for this

specific patient population. Such information can help with the development of optimal intervention strategies and appropriate health care policies for low back pain (Luo *et al.*, 2004:84).

2.3 COMPENSATION CLAIMS

The burden of illness in terms of compensation claims has been costly to the labour force of several countries. In a study of van Tulder (1995:237), disablement benefit payouts for back pain in the Netherlands in 1991 were estimated at \$1.5 billion. Musculoskeletal diseases accounted for 31% of the total costs of disablement while back pain accounted for 48% of these disability costs of musculoskeletal problems. By way of comparison, diseases of the circulatory system accounted for only 10% of total disability costs. During 2001 in Sweden, 13% of all early retirement pension payouts were granted for back problems (Ekman *et al.*, 2005:1778). In the USA, low back pain accounted for 23% (\$8.8 billion) of total workers' compensation payments in 1995 (Murphy & Volinn, 1999:691).

However, Murphy & Volinn (1999:691) and Borenstein (2000:225) also reported positive news regarding a decline in the frequency of occupational low back claims reported over a nine year period. Workers compensation data from Liberty Mutual Insurance Company (1987-1995), the Washington State Department of Labor and Industry (1991-1995) and the Bureau of Labor Statistics (1992-1995) were reviewed for frequency of low back claims from industrial settings. According to these researchers, the USA estimates of annual low back pain claims decreased by 34% between 1987 and 1995. However, because the rate of filing remained 1.8 per 100 workers, the estimated cost of low back pain claims for 1995 was \$8.8 billion, as mentioned previously.

Occupational back pain is common among workers in both Japan and the USA, which economically are ranked seventh and eighth in the "high income" grouping of countries, and is the reason why wage compensation for time off work is substantial in both countries. Accordingly, back pain claim rates in Japan seemingly would be of the same order of magnitude as rates in the USA.

However, in 1999, about one worker out of 10 000 filed a back pain claim in Japan, while about 58 out of 10 000 workers filed a back pain claim in the USA. To convert these rates back into raw numbers, the Workers' Accident Compensation Insurance in Japan covered 48 492 908 workers in 1999 who filed 4 632 claims for back pain. In contrast, the Washington state fund covered the equivalent of 1 455 893 workers in 1999 but 8 441 back pain claims were filed (Volinn *et al.*, 2005:700). This means that the rate of back pain claims in 1999 was 60 times higher in the USA than in Japan. The question of why the back pain claim rate is so much higher in the USA than in Japan remains essentially unsolved according to these researchers.

2.4 EPIDEMIOLOGY OF LUMBAR REVISION SURGERY

The lumbar region or lower back is the region of the spine, according to Diwan *et al.* (2003:309), that is most commonly operated on, while the most common cause for lower back surgery is that of a degenerative disorder, such as degenerative disc disease, herniated nucleus pulposus, lumbar canal stenosis and degenerative instability. Further to this, nearly 15% of patients with operations on the lumbar spine for degenerative causes or disorders may end up requiring another operation of the lower back (Malter *et al.*, 1998:814; Diwan *et al.*, 2003:309). The most frequent surgical procedure performed for treating degenerative disorders of the lower back is spinal decompression, followed by a combination of decompression and fusion surgery. Fusion of vertebral bodies as an isolated procedure is the least frequently performed operation for degenerative lumbar spine conditions (Hu *et al.*, 1997:2265; Malter *et al.*, 1998:814; Diwan *et al.*, 2003:309-310).

According to Hu *et al.* (1997:2265), it appeared that individuals younger than 45 years of age are more likely to undergo discectomy whereas fusion is commonly performed as the first choice operation for patients in the 45-65 age group. In persons older than 65 years of age, a laminectomy is more likely to be performed. Their data also indicated that discectomy was associated with fewer complications than laminectomy, but elderly patients are 25% less likely to undergo revision surgery. However, over a period of three years the re-

operation rate for all three groups was approximately 10%. Several good results after revision surgery have also been reported to occur in approximately 25%-80% of cases (Kim & Michelssen, 1992:957; Bernard, 1993:2196; Herron, 1994:161; Stewart & Sachs, 1996:706; Ozgen *et al.*, 1999: 287).

In the Washington study of Malter *et al.* (1998:814) involving 6 376 patients, the re-operation rate for decompression was 14.6%, compared to an 18.2% re-operation rate for fusion alone and decompression with fusion. Revision surgery in 4 722 patients in the Ontario study of Hu *et al.* (1997:2265) came to 9.5% for decompression, was 10.2% for decompression with fusion and 9.2% for fusion alone. According to Table 2.2 the difference in re-operation rates for different procedures for degenerative lumbar surgery over a five year period can be seen.

Table 2.2: Five year re-operation percentage rates for degenerative lumbar surgery (adapted from Diwan *et al.*, 2003:310).

<u>Washington Study</u> (n=6 376)	Decompression	Decompression with Fusion	Fusion alone
Annual surgical rate	83.7% (n=5 337)	10.9% (n=695)	5.4% (n=344)
Re-operation rate	14.6%	18.2% (including fusion alone)	
<u>Ontario Study</u> (n=4 722)			
Annual surgical rate	78.3% (n=3 697)	13.5% (n=638)	8.1% (n=383)
Re-operation rate	9.5%	10.2%	9.2%

However, the reported success rates for lumbar spine surgery vary significantly from series to series, but overall revision surgery is reported to have poorer outcome results than primary surgery. Additionally, the outcome results vary according to the procedure being performed, with better results reported with revision decompressive procedures including discectomy than with revision fusion cases. Nasca (1987:809) has reported better results after lumbar spinal stenosis surgery when done primarily than when done as part of revision

surgery (79.2% versus 59.4%). According to Diwan *et al.* (2003:323), significant back and leg symptoms developed in approximately 10%-15% of patients who had undergone a spinal decompression procedure and in approximately 15%-20% of patients who had a spinal fusion procedure for degenerative disease of the lumbar spine.

In analysing the safety and efficacy of revision lumbar spinal surgery, Diwan *et al.* (2003:319) found that revision surgery on the lumbar spine is typically associated with a higher level of complexity than with primary surgery. This is mostly explainable by the presence of fibrous scar tissue, dural adhesions, altered anatomy and poor tissue quality. Therefore, revision surgery can be expected, according to these researchers, to predispose lumbar spinal patients to higher rates of complications.

2.5 REVIEW OF TREATMENT MODALITIES IN LOW BACK PAIN

The diagnosis of chronic low back pain is a problematic area in the sense that, in most cases, secondary causes are responsible for the pain that is experienced (Jayson, 1994:681). According to Teasell and Harth (1996:844) and Koes *et al.* (2001:2504), for the majority of patients with chronic back pain, there is no definite pathologic-anatomical or pathologic-physiological diagnosis that can be made and diagnostic management seems to vary between back care specialists. The problem of making a definite diagnosis complicates the treatment of low back pain patients. This is why recent reviews on the clinical outcomes of different treatment methods of low back pain show conflicting results, as well as no scientific evidence for most interventions (Waddell, 1998:263-272; Furlan *et al.*, 2001:E160; van Tulder *et al.*, 2006:S64; van Tulder *et al.*, 2006:S82).

Furlan *et al.* (2001:E155) conducted searches of Medline, Embase, Psychinfo and The Cochrane Database of Systematic Reviews. They yielded 1 102 specific abstracts, while 109 articles were retrieved for detailed reading. A review of these articles excluded the majority of them, leaving only 36 articles which fulfilled the original inclusion criteria for randomised controlled trial

studies. Furlan *et al.* (2001:E160) and Waddell (1998:265-272) critically appraised or reviewed the methodology of systematic reviews of different treatment therapies for low back pain, as shown in Table 2.3.

Table 2.3: Results of low back pain interventions from systematic reviews (adapted from Furlan *et al.*, 2001:E160).

<u>Intervention</u>	<u>Reviews</u>	<u>Patients</u>	<u>Conclusions</u>
Analgesics	1	29	negative
Antidepressants	6	408	conflicting
Injections (epidural, facet)	4	898	conflicting
Muscle relaxants	1	50	positive
NSAIDs	2	1 126	conflicting
Opioids	1	38	positive
Back schools	7	>2 575	conflicting
Bed rest	1	>203	uncertain
EMG biofeedback	1	176	negative
Cognitive/behaviour	3	>999	conflicting
Couple therapy	1	56	negative
Multidisciplinary teams	3	>561	positive
Acupuncture	6	>645	conflicting
Exercises	6	1 980	conflicting
Laser	2	20	negative
Orthoses	3	806	conflicting
Spinal manipulation	9	>3 050	conflicting
TENS	4	>397	conflicting
Traction	3	>108	conflicting

Waddell (1998:265) rated the scientific evidence on the treatment of low back pain on a three-star system where three stars represents a generally consistent finding in a majority of multiple, acceptable, randomized controlled trial (RCT) studies, where one star, for instance, represents limited scientific evidence which does not meet all the criteria of acceptable RCT studies. Waddell (1998:263-264) built their evidence on the USA clinical guidelines provided by the Agency for Health Care Policy and Research (AHCPR, 1994), which was

the most comprehensive review of back treatment evidence ever undertaken. The AHCPR spent over two years and close to \$1 million reviewing and evaluating more than 10 000 articles.

Most recently, van Tulder *et al.* (2006:S64) and van Tulder *et al.* (2006:S82) also summarized the best available evidence from systematic reviews conducted within the framework of the Cochrane Back Review Group on non-invasive, as well as invasive treatments for low back pain. However, within the context of this study a summary of the effect of non-invasive treatments on low back pain (van Tulder *et al.*, 2006:S74) is given in Table 2.4. However, there is no evidence that any of all these interventions provide long-term solutions for pain and function.

Table 2.4: Outcome of non-invasive treatment modalities on low back pain (adapted from van Tulder *et al.*, 2006:S74).

<p><u>Pain Relief</u></p> <p>Acute low back pain</p> <ul style="list-style-type: none"> • Traditional non-steroidal anti-inflammatory drugs, muscle relaxants and advice to stay active are effective for short-term pain relief in acute low back pain. <p>Chronic low back pain</p> <ul style="list-style-type: none"> • Antidepressants, COX2 inhibitors, back schools, progressive relaxation, cognitive respondent treatment, exercise therapy and intensive multidisciplinary treatment are effective for short-term pain relief in chronic low back pain. <p><u>Function</u></p> <p>Acute low back pain</p> <ul style="list-style-type: none"> • Advice to stay active is also effective for long-term improvement of function in acute low back pain. <p>Chronic low back pain</p> <ul style="list-style-type: none"> • COX2 inhibitors, back schools, progressive relaxation, exercise therapy and multidisciplinary treatment are also effective for short-term improvement of function in chronic low back pain.

While there seems to be no real long-term effective method in treating low back pain patients, it may lead to the recurrence of low back pain. According to Andersson (1999:583), the recurrence rate of low back pain is so high that it seems to be part of its natural history. Valkenburg and Haanen (1982:9) reported lifetime recurrences of up to 85%, whereas in Sweden, the one year recurrence of sick-listing for low back pain was 44% in 1987. Rossignol *et al.* (1988:502) and Abenhaim *et al.* (1988:829) showed low back pain recurrence rates of 20% in one year and as much as 36% over three years in Canada. Men had a higher risk of recurrence than women and people between the ages of 25-44 years had the highest rate of recurrence of low back pain (Andersson, 1999:583).

However, with the progression in scientific rehabilitation techniques over the past few years, Hides (2005:157) reported that in a 2-3 year reanalysis of first-episode low back pain patients, the subjects in the specific exercise group had a 5.9 times less chance of suffering from recurrences of low back pain than the control group who did not receive any treatment. This may be a positive step towards the reduction of the recurrence problem of low back pain.

2.6 LUMBAR SUPPORT BELTS

Because the deep local muscle support system of the lumbopelvic region is the focus of this research study and mirrors the biomechanical effect of external back support belts, it has been deemed necessary to look at the epidemiology of external lumbar support belts or braces.

In response to the increasing human and economic costs of low back injury, employers have attempted preventive measures, specifically the widespread use of back support belts, approximately four million of which were purchased during 1995 in the USA (Labar, 1996:33). However, controversy exists over the effectiveness of back support belts for the prevention of occupational low back pain, which many workers use as a safety measure during lifting (Ammendolia *et al.*, 2005:128, 133; McGill, 2007:163).

In the largest study of its kind ever conducted, the Centres for Disease Control and Prevention's National Institute for Occupational Safety and Health (NIOSH) found no evidence that back belts reduce low back injury or back pain for retail workers who lift or move merchandise (Wassell *et al.*, 2000:2727; NIOSH, 2008). Also, Waddell (1998:271), in their review of randomized control trials of effective treatment for low back pain, found no evidence that lumbar corsets and supports are effective for acute low back problems.

The growing popularity of lumbar support belts, according to Jellema *et al.* (2001:377:385), has led to several studies, as well as reviews investigating the preventive and therapeutic effects of these belts. Because of the controversial research results on the effectiveness of lumbar support belts and with the emergence of biomechanical science, different research studies on different back support belts have been undertaken to such an extent that there are now more than 70 types of lumbar supports for prevention and more than 30 types for treatment of spinal disorders, worldwide.

One such example is the Serola sacroiliac belt which has been designed by Dr. Serola, a chiropractor by profession, to compress and support the sacroiliac joints. By normalizing the mechanics of the joint, a person normalizes its physiology, including muscle strength and proprioception. According to the Serola biomechanical theory, the key is normalization rather than simply stabilization (Serola Biomechanics Inc, 2008). However, research results from recent studies strongly suggest that specific exercise of the local muscle system stabilizes the lumbopelvic area and plays a vital role in lumbopelvic joint control and support, with subsequent pain reduction and improved functionality (Richardson *et al.*, 1999:41; Richardson *et al.*, 2002:404; Richardson *et al.*, 2005:4-7).

The review by Jellema *et al.* (2001:377) distinguishes itself from other reviews by evaluating the literature systematically, using up-to-date methodology recommended by the Cochrane Collaboration Back Review Group (Van Tulder *et al.*, 1997:2323-2330), by including the most recent literature and by reviewing lumbar back supports in the context of both treatment and prevention. Results

from the review of Jellema *et al.* (2001:385) showed that there is conflicting evidence on the effectiveness of lumbar back supports in the treatment of low back pain. Lumbar back supports are not recommended for primary prevention, as well as treatment of low back pain.

According to the available international guidelines for the management of low back pain in primary care, a lumbar back support should not be prescribed for patients with low back pain (Jellema *et al.*, 2001:385), although it would be interesting to know the specific effects of different types of lumbar back supports (Jellema *et al.*, 2001:385). However, none of the studies reviewed by Jellema *et al.* (2001:385) evaluated the effectiveness of lumbar back supports in the secondary prevention of low back pain and therefore they recommended that future studies, if any, should focus on this topic.

Given the available literature on back belts, McGill (2007:163) reported that the universal prescription of back belts is not in the best interest of reducing both the risk of injury and compensation costs, globally. McGill (2007:163-164) further highlighted certain mandatory conditions for back belt prescription, which are summarised as follows:

- Candidates for back belt wearing should be screened for cardiovascular risk, given the risk of elevated blood pressure.
- Belt wearers must receive education on lifting mechanics and tissue injury.
- A full ergonomic assessment of a person's job should examine and attempt to correct the cause of the musculoskeletal overload and should only prescribe back belts as a supplement for a few individuals.
- Back belts should not be considered for long-term use.

2.7 EPIDEMIOLOGY OF LOW BACK PAIN IN SPORTSMEN AND WOMEN

The era of professionalism in sport has brought with it injuries due to the increased intensity of playing (and training) sport at the highest level (Ranawat *et al.*, 2003:915). Published rates of low back pain in sportsmen and women range from 1%-30% and are influenced by sport type, gender, training intensity, training frequency and technique (Bono, 2004:382). Although not directly related to competitive sportsmen, the much quoted studies of Cady *et al.* (1979:269) and Cady *et al.* (1985:111) showed that a significant correlation between physical fitness and reported low back injury existed over a 10 year period in a select group of fire fighters in Los Angeles. The least-fit group reported 10 times more low back pain accidents than the best-fit group.

Although physical fitness may maintain the health of the lumbar spine, excessive physical activity that is representative of competition sports can lead to acute dynamic overload and/or chronic repetitive exertion of the mechanical lower back (Herring & Weinstein, 1995:1172). However, with conflicting reports it is not clear whether sportsmen or women are at higher risk for low back pain (Bono, 2004:382), although numerous studies have reported on the negative effect of low back pain in the world of competitive sport. Most sporting activities can result in injury to the lumbosacral spine, but low back pain is most frequently reported in cricket, golf, gymnastics, football, weightlifting, wrestling, dance and rowing (Jackson, 1979:364; Hoshina, 1980:75; Stanitski, 1982:77; Micheli, 1983:473; Micheli, 1985:85; Grimshaw *et al.*, 2002:655; Ranawat *et al.*, 2003:915). Following, are specific sport types that researchers have investigated as high risk for the development of low back pain. As a matter of differentiation, the specific sporting codes have been categorized as athletics, ball sports, gymnastics and dance, strength sports, and water sports.

2.7.1 Athletics

Kujala *et al.* (1996:165) documented that 30 of 65 (46%) adolescent athletes reported low back pain, compared with 6 of 33 (18%) non-athletes. In contrast to these studies, Videman *et al.* (1995:699) reported that low back pain was

less common in former elite athletes (275 out of 937 or 29.3%) than it was in non-athletes (273 out of 620 or 44%).

2.7.2 Ball Sports

Because low back pain is a common reason for lost playing time by competitive sportsmen/women, Hainline (1995:241) found that 38% of professional tennis players he studied reported low back pain as the reason for missing at least one tournament, while Lundin *et al.* (2001:103) documented that 32% of tennis players they studied during 1996 and 1999, developed severe low back pain. In a recent MRI study of the lumbar spine in 33 asymptomatic elite adolescent tennis players, Alyas *et al.* (2007:836) documented that 84.8% (28 of 33 players) had abnormal magnetic resonance imaging (MRI) findings. Although asymptomatic, pars injuries and facet arthroses were common, predominantly in the lower (L4/5 and L5/S1) lumbar spine.

With respect to cricket, schoolboy fast bowlers have been shown to be at most risk (47.4%) of injuring themselves (Stretch, 1995:1182), while A-grade or provincial cricketers have a 42% risk of injury (Stretch, 1992:339). According to these researchers the most common site of injury in these players is the back, with incidences of 33.3% for schoolboys and 17% for provincial players. Although these data represent a South African population, it supports unpublished injury statistics from Australian state teams (Elliott, 2000:983). Furthermore, the problem of lumbar spondylolysis in cricketers was already diagnosed a few years ago. A five year epidemiological study of an A-grade cricket team by Payne *et al.* (1987:17) showed that 50% of fast bowlers were diagnosed with a stress fracture of a lumbar vertebra.

According to Hardcastle (1993:398), the fast bowling action places immense stresses on the spine, taking place as many as 300-500 times per week. Professional cricketers are also involved in up to 99 days out of 149 days of the English season, excluding days of practice. Because of some of these factors, low back pain has now become common and many young fast bowlers are being lost to the game of cricket (Ranawat *et al.*, 2003:915). Apart from being

lost to the game of cricket, Ranson *et al.* (2005:1111) reported that low back injuries account for the greatest loss of playing time for professional fast bowlers. Apart from cricket injuries to the lower back, Gregory *et al.* (2004:737) also investigated the occurrence of spondylolysis in soccer players. He found that 82.1% (23 of 28 soccer players) studied, showed signs of complete or incomplete stress fractures of the lumbar spine. Furthermore, the study of Lundin *et al.* (2001:103) documented a 37% injury rate of severe low back pain in soccer players.

In the world of golf, with around 57 million golfers in 119 countries (Golf Research Group, 2008), low back pain is becoming an epidemic (Jakubowski, 2004:28). With more than 26 million golfers in the USA, which reflects an increase of 33% between 1986 and 1997, low back pain has become the most common musculoskeletal complaint experienced by both amateur and professional golf players (Grimshaw & Burden, 2000:1667; Lindsay & Horton, 2002:599; Parziale, 2002:499). Looking at injury patterns of golfers (Table 2.5), the prevalence of injury however, varies depending on a number of factors, including gender and professional versus amateur status (Metz, 1999:27).

Table 2.5: The most common sites of injury in amateur and professional golfers as reported in surveys (adapted from Metz, 1999:27).

<u>Amateur: % of Players Injured at Site</u>		
<u>Site</u>	<u>Men</u>	<u>Women</u>
Lower back	36.0	27.4
Elbows	32.5	35.5
Hands & wrists	21.2	14.5
Shoulders	11.0	16.1
<u>Professionals: % of Total Injuries</u>		
<u>Site</u>	<u>Men</u>	<u>Women</u>
Hands & wrists	29.6	44.8
Lower back	25.0	22.4
Shoulder	11.4	7.5
Elbow	7.3	6.0

It can be seen in Table 2.5 that the most commonly injured sites in amateur golfers overall (by percentage of players injured at each site) were the lower back (34.5%), while other injuries, for example elbows (33.1%), hands and wrists (20.1%) and shoulders (11.7%) followed. Bulbulian *et al.* (2001:570) agree with these findings of Metz (1999:27). Furthermore, men had virtually the same percentages as the group overall, but lower back injuries in amateur women were second only to elbow injuries in that gender group. In contrast, professional male and female golf injuries to the lower back were second only to hand and wrist injuries (Metz, 1999:27).

Parziale (2002:499-500) treated a total of 145 individual golfers between 1994 and 1997, which included 109 male amateurs, seven male golf professionals and 29 female amateur golfers. The most common injury in both men and women was low back pain, followed by shoulder and elbow pain. Sixty-five (65) of the 145 injuries seen were of the lower back, which included 57 male golfers (53 amateurs, four professionals) and eight female golfers. The percentage of male golfers with low back pain complaints was 49% vs. 28% for female golf-related injuries. Grimshaw *et al.* (2002:657) also identified the profile for golf-related injuries to amateur and professional golfers. In contrast to Metz (1999:27), injury data by Grimshaw *et al.* (2002:657) indicated that for professional golfers, injury is more common in the back/lower back (30%) and wrist (18%), while in amateur golfers the profile is also more prevalent in the back/lower back (27%) and then elbow (25%).

2.7.3 Gymnastics and Dance

Sward *et al.* (1991:437) reported a significantly higher rate of low back symptoms in elite gymnasts (79% or 19 of 24 subjects) than in a control group (38% or 6 of 16 subjects). In comparison with other athletes, gymnasts appear to be among the most likely to report severe low back pain (Sward *et al.*, 1990:124). It was found that six of seven elite rhythmic gymnasts reported low back pain over a seven week period (Hutchinson, 1999:1686). The study of Goldstein *et al.* (1991:463) showed that injuries in gymnastics frequently involve the lumbosacral spine. Injuries of the pars interarticularis are mostly recognized

and have approximately a 10% incidence in female gymnasts, almost four times the frequency of the general population (Herring & Weinstein, 1995:1172).

In the dance population, ballet dancers appear to be experiencing lumbosacral injuries at a growing rate. Herring and Weinstein (1995:1172) reported that during the 1990-1991 dance season a San Francisco ballet injury survey revealed that 24% of all injuries were to the spine. This made the spine the leading site of injury in ballet dancers, while one half of all these spine injuries were to the lumbosacral region.

2.7.4 Strength Sports

In weightlifting, low back pain, including pars interarticularis injuries, is the most frequent injury (Stith, 1990:259). Calhoun & Fry (1999:232) studied injuries to competitive USA weightlifters over a 6-year period and documented that the lower back is the most commonly injured area of the body in weightlifting. They found that injuries to the back consisted primarily of strains (74.6%). Low back pain is also more common in some sportsmen/women than in others. In this respect, Lundin *et al.* (2001:103) found that wrestlers had the highest rate of severe low back pain (54%) compared to tennis and soccer players.

2.7.5 Water sports

A retrospective study of injuries to elite rowers at the Australian Institute of Sport, which occurred over a 10 year period (1985-1994), was documented by Hickey *et al.* (1997:1567). These competitive male and female rowers had a 15% and 25% prevalence of low back pain over this 10-year period. In a similar fashion, swimming may also target the lumbosacral spine. An activity like swimming that is often prescribed as therapy for lumbosacral problems, is sometimes associated with lumbosacral problems when overused. This may not be surprising when one realizes that competitive swimmers undergo approximately 600 000 lumbar rotatory movements per year (Herring & Weinstein, 1995:1172).

2.8 CHILDREN AND LOW BACK PAIN

It has been generally believed that lumbar pain is uncommon among children and adolescents before the age of 20 (Kelsey *et al.*, 1990:699), while traditional paediatric orthopaedic teaching “classified” a child with back pain as having a tumour or infection until proven otherwise (Combs & Caskey, 1997:789). In a study by Balagué and Nordin (1992:575), subjects younger than 20 years represented one percent of a whole population of patients operated on for disc herniation. However, the National Health and Nutrition Examination Survey in the USA reported onset of low back pain before the age of 20 in 11% of the general population (Deyo & Tsui-Wu, 1987:264). Studies like these were arguably the reason why interest in low back pain and school-aged children has increased exponentially, according to Balagué *et al.* (2003:1403). A PubMed search done by these scientists, retrieved only four references over a four year period from January 1970 to December 1973 compared with 337 for 1998-2001.

Since the 1980s, epidemiological studies started concentrating more on the adolescent lumbar spine (Salminen, 1984:1) and have shown that the lifetime prevalence of low back pain in children is high, reaching that of adults by the end of their growth spurt (Taimela *et al.*, 1997:1132; Leboeuf-Yde & Ohm Kyvik, 1998:228; Gunzburg *et al.*, 1999:440; Kovacs *et al.*, 2003:259; Bejia *et al.*, 2005:331; Trevelyan & Legg, 2006:45). Because a potential link between the adolescent growth spurt and the increased prevalence of adult low back pain was found, several researchers have suggested that the study of low back pain in young people might provide insight into the origins of adult low back pain (Fairbank *et al.*, 1984:461; Goodman & McGrath, 1991:247; Terti *et al.*, 1991:503; Kujala *et al.*, 1992:627; Olsen *et al.*, 1992:606; Salminen *et al.*, 1992b:405; Duggleby & Kumar, 1997:505).

Investigation of the adolescent growth spurt as part of the Harpenden Growth Study indicated that the average age of onset of the adolescent growth spurt in males was 12.5 (± 2 years), with completion of the growth spurt ranging from 13.5-17.5 years. Similar results were obtained for adolescent females, on average 2 years prior to males (Tanner *et al.*, 1976:109). Studies by Fairbank *et*

al. (1984:461) and Olsen *et al.* (1992:606) showed that the onset of low back pain in adolescents corresponded approximately to the same age ranges as reported in the Harpenden Growth Study. In their review of studies on juvenile low back pain, Duggleby and Kumar (1997:508) found that chronological age had a significant positive correlation with low back pain. The cumulative prevalence of low back pain at age 10 was 9.9% and increased by 10% each year to age 12, with a dramatic increase to 52.8% at age 13. In a study by Balagué *et al.* (1988:175) they reported that by age 15, a cumulative prevalence of 71.3% existed, while most other studies found a cumulative prevalence of 31.2% and 36% at that age (Duggleby & Kumar, 1997:508).

In a study of Kovacs *et al.* (2003:261-266), 7 048 adolescents (from 44 schools) between the ages of 13-15, as well as 9 309 parents from a Spanish community, participated. Significant ($p < 0.001$) differences between boys and girls were found for lifetime prevalence of low back pain (50.9% in boys vs. 69.3% in girls), as well as regard to lifestyle activities, such as smoking (63.5% of girls vs. 47.9% of boys), leisure sitting of more than two hours (46.8% of girls vs. 41.3% of boys) and sports participation (more than one sport; boys 88.4%, girls 80.6%; team sports: boys 83.4%, girls 20.4%). There was also a significant difference between boys and girls having been more diagnosed with scoliosis (19.1% vs. 28.1%), as well as having a leg length discrepancy (9.9% vs. 12.8%). No association between low back pain in the child and the biological/non-biological parent was found, nor did the risk of scoliosis in an adolescent increase when both biological parents had it. Of importance was that pain in bed, or upon rising, was found to be the most highly relevant and significant association with low back pain in both adolescents, parents and both sexes.

Danish school children between the ages of 13-16 years showed a cumulative lifetime prevalence of low back pain of 58.9% and an increase in low back pain prevalence of 6.4% from 14-15 years, independent of gender (Harreby *et al.*, 1999:444). In a previous study by Burton *et al.* (1996:2323) it was found that the annual incidence of low back pain increased from 12% to 21.5% between the ages 12-15 years, with the greatest increase (7%) occurring in the final year.

These studies again confirm the trend of low back pain to worsen in the maturing adolescent. Statistical analysis of independent variables in the study of Harreby *et al.* (1999:448) showed that female gender, daily smoking and heavy activities during leisure time are important factors associated with severe low back pain in the Danish adolescent. An explanation for the correlation between some of these factors and low back pain is the fact that smoking is allowed in the majority of Danish schools and is not as taboo as in other countries. Also, more than half the Danish schoolchildren of 14 years and older have a job in their leisure time, involving heavy loads on the lumbar spine, such as cleaning jobs, work in a supermarket and the distribution of newspapers (Harreby *et al.*, 1999:445, 448).

In a prediction of future low back pain in British schoolchildren, Jones *et al.* (2003:827) demonstrated that a strong association exists between high levels of negative psychosocial dimensions (e.g. conduct problems and hyperactivity) and an increased probability of new low back pain 12 months subsequently, and further propose that the origin of the adult back pain “career” may begin as early as adolescence. Apart from this, they also found no evidence of an increase in risk associated with a daily mechanical load (schoolbag weight) carried by children to school.

Another study of psychosocial factors affecting low back pain in Norwegian adolescents (14-16 years of age), indicated a strong association between poor self-perceived fitness and low back pain (Sjölle, 2002:582). Szpalski *et al.* (2002:459) also found that poor self-perception of health was a significant variable behind the reporting of low back pain in Belgium children, 9-12 years of age and further reported that new low back pain was observed more frequently (significant at $p < 0.0001$) in children who do not walk to school.

Low back pain cumulative lifetime prevalence among Tunisian schoolchildren and adolescents was 28.4% and chronic low back pain prevalence was eight percent. The sitting position (school chair height and comfort) was the only factor associated with both low back pain and chronic low back pain in 67% of these school children (Bejia *et al.*, 2005:335). In a Finnish study, Salminen *et al.*

(1992a:1039) reported that most children who had trouble with sitting at school thought that low back pain was due to an unsuitable school desk. Balaqué *et al.* (1988:178) found that in Swiss children who experienced low back pain, sitting was found to be the most troublesome situation in connection with low back pain.

In a study of nine year old Belgium children, Gunzburg *et al.* (1999:439) reported that 36% suffered at least one episode of low back pain while children who played video games for more than two hours per day reported significantly more low back pain than children watching television only. There was also a significant correlation between self-reported low back pain and indicators of general well being (tiredness, less happy and sleeping badly), as well as between self-reported low back pain and low back pain in these children's parents.

As can be seen, the risk for the development of low back pain in adolescents is multifactorial and therefore complex. Several factors such as gender, anthropometry, hamstring muscle tightness, hypermobility, sitting position, psychological and social factors, sports activities, smoking status and TV watching have been associated with nonspecific low back pain in young people (Salminen *et al.*, 1995:2101; Duggleby & Kumar, 1997:507; Harreby *et al.*, 1999:444; Sjölie, 2002:582; Szpalski *et al.*, 2002:459; Bejia *et al.*, 2005:335).

2.9 SUMMARY

Among populations in western industrialized countries, low back disorders are a major health problem and have increased with epidemic proportions over the last few decades (Waddell, 1996:2820; Deyo, 1998:50; van Tulder *et al.*, 2003; Woolf & Pfleger, 2003:652; van Tulder *et al.*, 2006:S64). Therefore, a committee called the Clinical Standards Advisory Group (CSAG) was formed during 1992 by the British Ministry of Health to investigate the problem of lumbar back injuries in Britain (Foster *et al.*, 1999:1332). According to CSAG, there was an increase of 208.5% in back injuries in the period 1978/79-1991/92. In the same time period, other orthopaedic injuries increased by 136.4%, while

cases of vascular disease increased only by 91% in comparison. Other researchers, such as Hart *et al.* (1995:11) and Andersson (1999:581), also studied the growing tendency of lumbar back injuries in the USA and came to the realisation of the enormous proportion of this specific epidemic, as well as the financial liability it has on society. By observing the epidemiology of low back pain worldwide, it can be said that although jobs involving heavy lifting and physical labour - many of the factors traditionally believed to “cause” low back pain - have decreased steadily, the number of people suffering from low back pain has not decreased (Deyo, 1998:48).

Low back pain is therefore a condition that is bound to have a significant economic impact on healthcare expenditures of countries. The economics of low back pain are of further interest because, although there is limited evidence that the incidence of low back pain has increased during the last few years, disabilities resulting from this condition have grown at a rate that exceeds population growth and the rate of growth of virtually all other health problems (Frymoyer & Cats-Baril, 1991:263; Pai & Sundaram, 2004:1).

Taking all the above into consideration, it should be remembered that most of the figures reported in international literature are estimates only (van Tulder *et al.*, 1995:239). Comparison of these figures is hampered by the fact that they have been derived from various sources, for example different countries, industrial settings, different health care systems and socio-economic patterns. Methodological differences due to various classification systems or survey techniques can also lead to discrepancies among studies (Dagenais *et al.*, 2008:19).

To counter these differences and discrepancies and to increase consistency in the management of low back pain across countries, the European Commission for example, has approved and funded a project called “COSTB13”. The main objectives of the COSTB13 action are: “developing European guidelines for the prevention, diagnosis and treatment of non-specific low back pain, ensuring an evidence-based approach through the use of systematic reviews and existing clinical guidelines, enabling a multidisciplinary approach, and stimulating

collaboration between primary health care providers and promoting consistency across providers and countries in Europe” (van Tulder *et al.*, 2006:S75).

From available literature however, it may still be concluded that the financial impact of low back pain on industry and society is enormous (van Tulder *et al.*, 1995:239; Maniadakis & Gray, 2000:101; Dagenais *et al.*, 2008:8). Also, Maniadakis and Gray (2000:95), as well as van Tulder *et al.* (1995:239), agree that the epidemic increase of low back pain is becoming a threat to social welfare and thereby not only a medical but also an economical problem. They further warn that it may even become a political problem. According to Dagenais *et al.* (2008:19): “further studies are required to estimate the total costs of low back pain and inform decision making for this complex and challenging condition”.

CHAPTER THREE

PATHOPHYSIOLOGY OF LOW BACK DISORDERS

- 3.1 INTRODUCTION
- 3.2 BIOMECHANICS AND INJURY MECHANISM OF LOW BACK DISORDERS
 - 3.2.1 Injury mechanism during lumbar flexion
 - 3.2.2 Injury mechanism during lumbar extension
 - 3.2.3 Injury mechanism during sitting
- 3.3 THE LUMBAR MULTIFIDUS MUSCLE MECHANISM IN LOW BACK DISORDERS
 - 3.3.1 Introduction
 - 3.3.2 Mechanical anatomy of the *m. lumbar multifidus*
 - 3.3.3 Morphology of the *m. lumbar multifidus*
 - 3.3.4 Histology of the *m. lumbar multifidus*
 - 3.3.5 Neutral spine posture and muscle activation of *m. lumbar multifidus*
- 3.4 SUMMARY

3.1 INTRODUCTION

To deliver comprehensive low back rehabilitation to the general population at large or to competitive sportsmen and women, one needs to understand the anatomic structures, biomechanical and functional deficits or injury mechanisms that develop with injuries of the lumbar spine (Herring & Weinstein, 1995:1172; McGill, 1997a:448-449; McGill, 2007:72). Also, it is important to understand the interplay between the lumbar spine and the specific biomechanical demands related to activities of daily living and with specific reference to this research study, the physical testing of the low back pain patient.

Before looking at the patho-mechanics of low back disorders, it is necessary to understand the definition of low back pain. Anderson (1986:91) and Woolf and Pfleger (2003:652) defined low back pain as follows; pain localized below the line of the twelfth rib and above the inferior gluteal folds, with or without leg pain. It can further be classified as “specific”, indicating a suspected pathological cause or as “non-specific”, as in about 90% of cases. Duration of low back pain lasting less than six weeks is defined as acute; between six weeks and three months as sub-acute; and if it lasts longer than three months, as chronic. Frequent episodes are described as recurrent low back pain (Frymoyer, 1988:291, Woolf & Pfleger, 2003:652).

It is often very difficult to determine whether an injury originates in the lumbar spine, sacro-iliac joints or hip joints despite technological advances (Fritz *et al.*, 2005:743). Therefore, Waddell (1998:263) points out that in only 15% of cases can a definitive diagnosis as to the pathology of low back pain be made. According to Nachemson (1990:534), only in about 2%-30% of these patients can an established diagnosis be made, while known patho-anatomic causes for low back pain and sciatica include disc herniation, spinal stenosis, fracture, spondylolisthesis, infection, rheumatic diseases and different types of demonstrable instability. However, looking at the natural history of low back pain, it seems likely that most cases of low back pain are caused by small ruptures or swelling of the myotendinous structures (Nachemson, 1990:534; McGill, 1997a:449). Of interest, although not directly related to patho-

physiology, Bigos *et al.* (1986:246) and Hadler (2001:1309) mentioned that low back pain was more strongly predicted by psychosocial factors and discarded biomechanical evidence. McGill (2007:33) strongly objected to such unscientific conclusions.

With specific reference to “non-specific”, Haldeman (1999:1) reported that the Agency for Health Care Policy and Research (AHCPR) Guidelines on Low Back Problems and similar guidelines in other countries have given up on attempts to clarify or define the origin of this condition and have resorted to differentiating between so-called Red Flag pathologic conditions and patients with non-specific low back pain, with or without leg symptoms. However, Bogduk *et al.* (1996:313) and McGill (2007:27) stated that clinicians and scientists must question the statement that 85% of low back disorder cases are idiopathic or have no definitive patho-anatomical cause. According to McGill (2007:28), improved tissue-based diagnosis, provocative testing and functional diagnosis with a better understanding of the interaction between patho-anatomical and psychological variables, will help build the foundation for better prevention and rehabilitation techniques in the future.

For the purpose of this study, the researcher will review and analyze scientific evidence related to the most important injury mechanisms of low back pain and, in light of that, set up a sound biomechanical and clinical test for the low back pain patient in the office setting with the PAB.

3.2 BIOMECHANICS AND INJURY MECHANISM OF LOW BACK DISORDERS

Overloading of low back anatomical structures by means of peak or cumulative compressive or shear forces is considered to be one of the strongest factors affecting low back disorders (McGill, 1997a:449; Richardson, 2005:105; McGill, 2007:29). Epidemiological studies (Aggrawall *et al.*, 1979:58; Kornberg, 1988:934; Videman *et al.*, 1990:728; Burnett *et al.*, 1996:305; McGill, 1997b:467; Yingling & McGill, 1999a:1882; Callaghan & McGill, 2001a:28; Gunning *et al.*, 2001:471) have established that excessive loading on tissue in

the lumbar spine will result in injury. According to McGill (2007:6) it is important for clinicians and therapists to be aware of the mechanics of spine function before low back rehabilitation can start. To understand the biomechanics of injury, several tissue damage scenarios will be described in brief:

- Spinal ligaments seem to avulse at lower load rates but tear in their midsubstance at higher load rates (Noyes *et al.*, 1994:236) and can lead to prolonged disability given the loading forces (Troup *et al.*, 1981:61). McGill (2007:99) stated that this is consistent with the prolonged length of time it takes for ligamentous tissue to regain structural integrity when compared with other tissues.
- The classic disc herniation injury (damage to the disc annulus) appears to be associated with repeated flexion motion with only moderate compressive loading required (Callaghan & McGill, 2001a:28), while full flexion with lateral bending and twisting leads to the same injury (Gordon *et al.*, 1991:450).
- With reference to the disc nucleus, McGill (2007:98) stated that the tissue injury mechanism to the disc nucleus still remains obscure. However, Lotz and Chin (2000:1477) documented that cell death (apoptosis) within the nucleus increases under excessive compressive load but is generally not detectable *in vivo*.
- End plate injury has been shown to happen first under excessive compressive loading of spinal units in the laboratory (Callaghan & McGill, 2001a:28), while end plate avulsion has been revealed under excessive anterior-posterior shear loading (McGill, 2007:98). Kornberg (1988:934) documented, by means of magnetic resonance imaging (MRI), traumatic Schmorl's node formation in a patient following forced lumbar flexion.
- Compressive loading also damages vertebral cancellous bone and often accompanies disc herniation and annular delamination (Gunning *et al.*, 2001:471).

- Looking at the posterior bony elements (neural arch), repeated stress-strain reversals associated with full flexion and extension movements are thought to cause spondylitic fractures (Burnett *et al.*, 1996:305), while excessive shear forces can fracture parts of the arch (Cripton *et al.*, 1995:111; Yingling & McGill, 1999a:1882).

The above-mentioned injury mechanisms, are caused by high compressive loading and are known to be accelerated with repetitive loading (Richardson *et al.*, 2005:105). A few scientists reported that large compressive forces are produced in the lumbar intervertebral disc during various sport and physical activities. For example, a golfer executing a golf swing produces between 6 100 and 7 500 Newton (N) of compressive forces across the L3–L4 disc in amateur and professional players, respectively (Bono, 2004:386). Fast bowling during cricket can also place large forces on the lumbar spine which may be lessened with proper technique (Elliot & Khangure, 2002:1714). These cricket researchers also found that small-group coaching aimed at reducing the level of shoulder counter-rotation during cricket bowling decreased the prevalence and progression of lumbar disc degeneration as measured with MRI.

Gatt *et al.* (1997:317) recorded forces in the L4-L5 motion segment during blocking manoeuvres in five football linesmen, with an average peak compressive load of >8 600 N and with an average peak sagittal shear force of 3 300 N. The magnitude of these forces exceeded the reported in vitro forces necessary to cause fatigue failure of the lumbar intervertebral disc. According to these data it is suggested that football linesmen are at risk for routine repetitive lumbar disc microtrauma.

With respect to rowing, Reid and McNair (2000: 321) reported on the factors that may influence the onset of low back pain in rowers. They reported compressive loads of 3 919 N for men and 3 330 N for women, while anterior shear forces were calculated at 848 N and 717 N for men and women respectively. Peak compressive loads during the stroke were 6 066 N for men and 5 031 N for women. Furthermore, for 70% of the stroke cycle, rowers are in a flexed posture that contributes to these compressive forces.

With reference to weight lifters, compressive forces in the lumbar spine are far greater than were recorded in any other sport or physical activity. Cholewicki *et al.* (1991:1179) recorded forces in the lumbar L4-L5 motion segment in 57 competitive weight lifters, with an average compressive load of more than 17 000 N. Other studies (Cappozzo *et al.*, 1985:613) found that when a person performed half-squat exercises with weights approximately 1.6 times body weight, the L3-L4 motion segment was absorbing compressive loads approximately ten times body weight calculating to about 7 000 N for an average 70 kg person. However, above researchers found that increasing lumbar flexion was the most influential factor affecting compressive loads or forces in the lumbar spine.

Further to this McGill (2007:46, 99) stated that researchers have overemphasized a single variable to the lumbar injury mechanism, namely, acute or once off maximum exposure to lumbar compression; yet some studies have shown that higher rates of low back disorders occur with lower levels of lumbar compression. In the study of Callaghan and McGill (2001a:35), it was shown that the numbers of flexion/extension cycles were more important than the actual magnitude of compressive load. In an *in vitro* study, these researchers could produce no herniations with 260 N compressive load and 85 000 flexion/extension cycles; however herniations were produced by 867 N load and 22 000 to 28 000 cycles and with 1 472 N and only 5000 to 9 500 cycles. From this evidence it can be documented with more clarity that disc herniations are a function of repeated flexion/extension motion, with only a modest level of accompanying compressive load.

The clinical implication of this finding is that prescription of classical flexion/extension exercises under simultaneous compressive loading, such as the seated flexion/extension machines, may lead to disc herniation. Further to this, flexion stretches and sit-ups or daily activities like prolonged sitting which are characterized by a flexed spine must be reconsidered. Avoiding these motions can spare the posterior annulus portion of the disc (McGill, 2007:47).

The antithesis to flexion strained, disc damage is the McKenzie therapy approach, which is based on the back extension posture. The theory is that extension of the lumbar spine will drive the nucleus forward or anteriorly within the disc (McKenzie, 1979:22). In 2005, Scannell and McGill found that with flexion-strained disc pathology, the herniation process begins from failure in the innermost annulus rings, filling with nucleus material and progresses radially outward. With the McKenzie technique, they have found that the extended posture may drive the nucleus material that is in the delaminated pockets of the posterior nucleus back (anteriorly) toward the central part of the disc (McGill, 2007:47). This brings us to the flexion injury mechanism of the lumbar spine.

3.2.1 Injury mechanism during lumbar flexion

Various mechanical variables contribute by increasing or lowering the risk of low back disorders. To illustrate, Holm and Nachemson (1983:866) showed the benefit of increased levels of movement in providing nutrition to the structures of the intervertebral disc, while McGill (2007:100) stated that their research has demonstrated that too many movement cycles to full flexion resulted in intervertebral disc herniation. However, too little motion from sedentary work also resulted in intervertebral disc injury (Videman *et al.*, 1990:728). Understanding the muscle forces, their components of compression and shear force and their role in supporting the lumbar spine is very useful (McGill, 2007:82).

One of the most common movement functions of daily living is the standing-to-full-flexion manoeuvre where, in full flexion, the lumbar extensors shut down their neural drive by reflex while the passive tissues absorb the load as they strain under full flexion (Schultz *et al.*, 1985:195; Richardson, 2005:111; Shin *et al.*, 2004:486; McGill, 2007:76; Konrad, 2008). This is called the “flexion-relaxation” syndrome. The “flexion-relaxation” syndrome can be further explained in that the lumbar extensors undergo eccentric contraction as the spine approaches full flexion, while the passive tissues take over moment production, relieving the extensor muscles of this role and accounting for their myoelectric silence. Furthermore, during flexion of the lumbar spine, a forward

or anterior shear is exerted on the intervertebral joint, especially at the lower lumbar levels where these forces are greatest (Hides, 2005:69). McGill (2007:52, 76) warns that shear loading is substantial when lumbar muscles lose their line of action (Figure 3.1, a-d) and re-orientate (to be parallel) to the compressive axis of the spine with lumbar flexion causing the flexed spine to be unable to resist damaging shear forces.

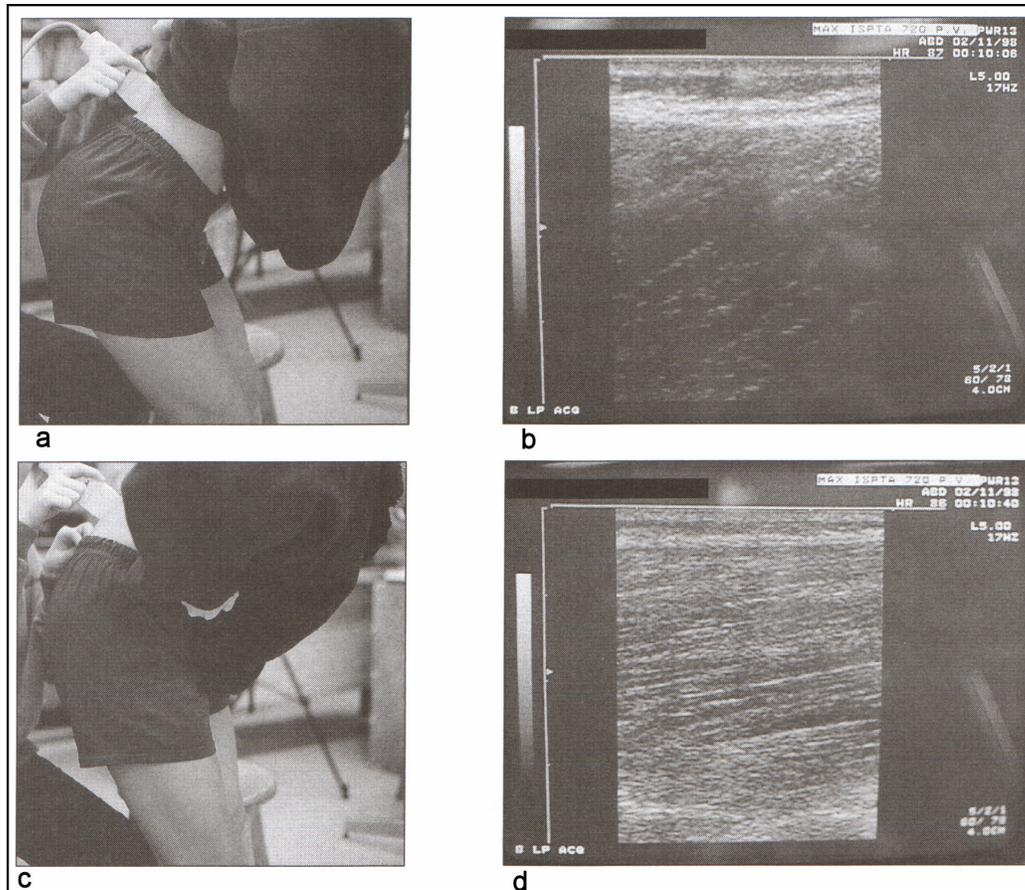


Figure 3.1: The oblique angle of the lumbar portions of *m. iliocostalis* and *m. longissimus*. (a) neutral spine position. (b) oblique angle of these muscles as viewed with US. (c) lumbar spine flexion and (d) loss of the oblique angle with spine flexion so that anterior shear forces cannot be counteracted (adapted from McGill, 2007:53).

According to Hides (2005:69), when *m. lumbar longissimus* and *m. lumbar iliocostalis* contract bilaterally, they draw their vertebrae of origin posteriorly

owing to their posterior and caudal direction and therefore oppose anterior shear. Contraction of the *m. lumbar multifidus* fascicles produces posterior sagittal rotation of the vertebrae of origin rather than posterior translation.

It may be true that during activities of forward bending and lifting the induced forces are controlled by the lumbar erector spinae muscles and the *m. lumbar multifidus* together, as one of the known roles of the *m. lumbar multifidus* is adjustment and control of lumbar lordosis. It is therefore important to know that the ability of the lumbar paraspinal muscles to protect the spine against anterior shear forces is a function of spine curvature. Figure 3.1 (a-d) shows the pathomechanics of adopting a neutral spine during trunk flexion tasks.

McGill (2007:80) documented muscle and passive tissue forces and moments during full flexion where the flexion reaction moment was 171 Nm (muscle=38 Nm; ligaments=113 Nm; passive tissues like disc, skin=20 Nm) causing the joint to compress at 3 145 N, while shear force was 1 026 N. It is clear that patients with spondylolithesis or shear instabilities would be contraindicated for lumbar flexion exercises, for example, straight-leg toe touch or sit-ups. However, prescription of classical repetition exercises for example, bent knee sit-ups to increase back health has been shown to be contraindicated when biomechanically analysed.

Axler and McGill (1997:804) showed that each sit-up, with full lumbar flexion as well as disc annulus compression, causes low back compression levels close to the action limit set by the National Institute for Occupational Safety and Health (NIOSH). During 1981, NIOSH set the compression limit to 3 300 N (325 kg) (McGill, 2007:88). Axler and McGill, (1997:804) have shown that the traditional sit-up causes approximately 3 350 N of compression (causing tissue damage) on the spine with each repetition of the bent leg sit-up.

This exercise prescription was based on one theory, for example that *m. psoas* is shortened on the length-tension relationship so that compressive loading is reduced. However, Juker *et al.* (1998:308) showed that while *m. psoas*, with its specific anatomical attachments, is shortened with hip flexion; its activation

contraction is higher during bent-knee sit-ups because the shortened *m. psoas* must contract to higher levels (causing increased lumbar compression) to compensate for its compromised length. Biomechanical studies like these clearly indicate that bent-knee sit-ups are contra-indicated for persons with low back disorders (McGill, 2007:9, 89).

One of the other most common movement functions of daily living, apart from forward bending, is simultaneous bend and lift where the muscle and ligament forces that are required to support the posture and control movement impose excessive loads on the lumbar spine. In determining the biomechanical load on the L4-L5 joint, when an average man lifts 27 kg using a squat lift style, the complex three-dimensional, biologically driven model of McGill was used to predict extensor reaction moment and loading force.

According to this model, an extensor reaction moment of 459 Nm (223 m/kg) was produced, causing a compressive load of over 7 000 N (700 kg) on the lumbar spine (McGill, 2007:16, 82). This amount of compression can already damage very weak spines, although the lumbar spine tolerance in an average healthy man can approach 12-15 kN (1 200–1 500 kg) (Adams & Dolan, 1995:3). In extreme situations, Cholewicki *et al.* (1991:1179) have documented compressive spinal loads (risk free) of over 20 kN (2 000 kg) in competitive weightlifters.

As mentioned before, it is important to know that the ability of the paraspinal muscles to protect the spine against anterior shear forces is a function of spine curvature, especially when lifting stoop or squat style. In this regard, Adams *et al.* (1994:5), McGill (1997b:465) and McGill (2007:102) stressed the importance that a neutral lumbar posture in lifting (squat style), while allowing the lumbar spine to flex, will cause the posterior ligaments to strain (Figure 3.2 a-b). This lifting technique of spine flexion has significant effects on shear loading or resultant injury risk of the intervertebral spine.

The dominant line of action of the pars lumborum fibers of *m. longissimus* and *m. iliocostalis* when in neutral lumbar lordosis, caused these muscles to

produce a posterior shear force on the superior vertebra, with the interspinous ligament (obliquity to resist posterior shear of superior vertebrae) controlling posterior sliding of the facet joint.

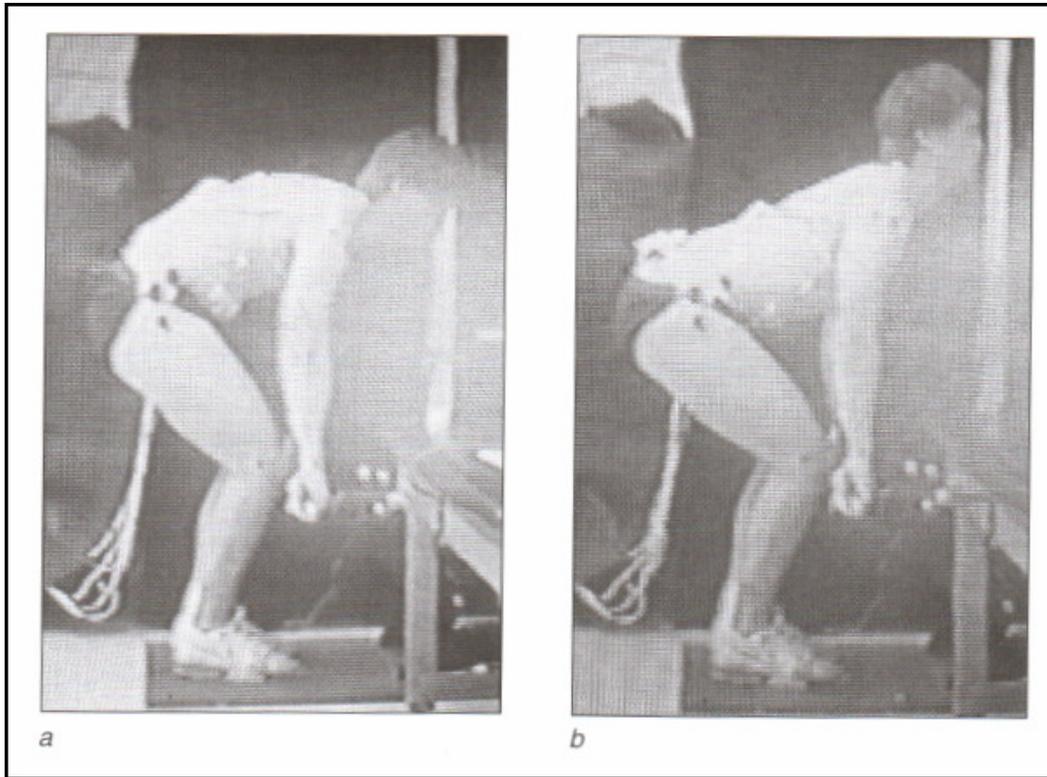


Figure 3.2: Biomechanics of lifting. (a) a flexed spine imposing anterior shear force and (b) a neutral spine posture aligns the fibres to support the shear forces. (adapted from McGill, 2007:102).

In contrast, during full flexion, the interspinous ligament (with opposite obliquity) imposes anterior shear force on the superior vertebra. Anterior shear forces that result from flexing the lumbar spine in lifting (Figure 3.2, a-b) are likely to exceed 1 000 N (100 kg), with recruited ligaments contributing to this shear motion (McGill, 2007:100).

The research of Cripton *et al.* (1995:111) quantified the risk of injury with respect to shear tolerance of the lumbar spine to be between 2 000 and 2 800 N in adult cadavers for one-time loading. McGill (1997b:473) illustrated a fully

flexed and neutral spine during lifting (Figure 3.2, *a-b*) that reflects the flexion injury mechanism of the lumbar spine, that is, myoelectric silence in the back extensor muscles, strained posterior passive tissues, as well as high shearing forces from both reaction shear on the upper body and interspinous ligament strain. The neutral spine posture recruits the pars lumborum muscle groups and aligns the fibres to support the anterior shear action of gravity on the upper body and hand held load. The shear load created in posture (a), resulted in 1 900 N load on the lumbar spine, while posture (b) reduced the shear load to about 200 N. This example illustrates that anterior-posterior shear load is the real risk compared to compressive load, since the spine can safely tolerate 10 kN in compression but 1 000 N of shear causes injury simply because the spine is fully flexed (McGill, 1997b:473; McGill, 2007:102).

Another mechanism to consider in the interpretation of injury risk to the lumbar spine is the evidence that the ability of the spine to bear load is a function of the curvature of the spine *in vivo* (McGill, 2007:102). Gunning *et al.* (2001:471) recently showed that a fully flexed porcine spine model is 20%-40% weaker than if it was in a neutral spine posture. The clinical importance of these findings is that clinicians and therapists must consider the anterior shear force of the superior vertebra during full flexion and resultant shear forces imposed on the joint by interspinous ligament strain and thereby avoid prescribing flexion stretches for patients with shear pathology, such as spondylolithesis (McGill, 2007:65).

3.2.2 Injury mechanism during lumbar extension

The injury mechanism of extension injuries to the lumbar spine can be better understood if we are more aware of the muscle activation levels and the resultant spinal load created by this movement. Callaghan *et al.* (1998:16) and McGill (2007:91) reported the compression load of performing an upper body extension exercise with legs fixed and the cantilevered upper body extending over the bench or roman chair. This extension movement activates the thoracic and lumbar portions of *m. longissimus* and *iliocostalis* (four extensors), which impose over 4 000 N (397 kg) on the lumbar spine. In their review of the

Biering-Sorensen test Demoulin *et al.* (2006:46) also reported this lumbar compression load to be unsafe. These calculations were based on the virtual spine model of the McGill group (McGill & Norman, 1985:883; McGill & Norman, 1986:666; Cholewicki & McGill, 1996:13; McGill, 2007:16-21) and precaution should be taken when prescribing this type of test to patients. Creating even more unwanted pressure is the prone lying position, with both legs extended and both arms outstretched which loads the L4-L5 lumbar spine to 6 000 N (580kg).

Epidemiological studies in athletes, such as gymnasts and cricket bowlers, have shown that damage to posterior elements of the vertebrae is associated with the cyclic flexion/extension motion in these sports and that excessive shear forces play a major role (Hardcastle *et al.*, 1992:421; Ranawat *et al.*, 2003:915). Repetitive extension movements (eg. in gymnasts) can specifically cause posterior shear of the superior vertebrae leading to ligamentous damage, end plate avulsion from the vertebral body or fatigue crack in the pars, eventually resulting in a fracture (Yingling & McGill, 1999b:501, McGill, 2007:42). Higher load rates of posterior shear forces produced wedge fractures and facet damage to the vertebral body (McGill, 2007:42).

Apart from flexion and extension induced injuries to the lumbar spine, sitting as a hypokinetic activity may also lead to biomechanical strain and injury to the lumbar spine.

3.2.3 Injury mechanism during sitting

Hypokinesia is the greatest clinical symptom caused by the chair. Most of the time, people sit at home watching television or relaxing, sit in their cars whilst driving and sit at work whilst working (Pienaar, 2004). People generally sit with the lumbar spine in flexion which minimizes muscle activity while sitting more upright requires higher activation of the *m. psoas* and the hip extensor group (Juker *et al.*, 1998:301). Already in the 1960s, Nachemson (1966:107) documented higher loads on the discs in various sitting postures compared to the standing posture, using intradiscal pressure measurements, while Videman

et al. (1990:728) documented the increased risk of disc herniation in people who perform sedentary jobs characterized by sitting.

In terms of trunk muscle activity during sitting, O'Sullivan *et al.* (2006:E710) more recently reported that the local (*m. lumbar multifidus* and *m. internal oblique*) and global (*m. thoracic erector spinae* and *m. external oblique*) muscles of the lumbopelvic area can be preferentially activated or deactivated in three different unsupported sitting postures. Thoraco-lumbar and lumbopelvic sitting postures were compared to slump sitting. They found that lumbopelvic sitting appeared to be the optimal posture because it avoids end range postures which minimize connective tissue strain as reported by Macintosh *et al.* (1993a:884).

Also, it resulted in preferential activation of the local lumbar stabilizing muscles known to be fatigue resistant (Gardner-Morse and Stokes, 1998:86) and capable of providing a local stabilizing effect on the lumbopelvic region without placing a high compressive load on the lumbar spine, as reported by Kavcic *et al.* (2004:1254). Furthermore, O'Sullivan *et al.* (2006:E711) found that thoracic upright sitting in contrast, resulted in high levels of coactivation of *m. external oblique* and *m. thoracic erector spinae* which exert high compressive loads on the spine. Slump sitting resulted in flexion-relaxation of the lumbar stabilizing muscles which led to increased intervertebral disc and connective tissue loading (O'Sullivan *et al.*, 2002:1238; O'Sullivan *et al.*, 2006:E711).

In a study of 20 minutes prolonged sitting in a slouched or flexed posture, McGill and Brown (1992:43) found that creep developed in the posterior passive tissues and subjects could only regain half of their intervertebral joint stiffness after two minutes recovery from 20 minutes of spinal flexion. Even after 30 minutes recovery, some residual joint laxity remained. According to these research results, a time factor seems to be associated with the resetting of posterior tissue stiffness and it would be unwise to lift something immediately following prolonged sitting in flexion where posterior movement of the nucleus has taken place. Furthermore, with the McKenzie approach in mind (McKenzie, 1979:22), anterior movement of the nucleus would decrease pressure on the

posterior portion of the annulus, which is the most problematic site of herniation. Repositioning of the nucleus after a postural change takes time because of the viscosity of the nuclear tissue. McGill (2007:96) stated that if compressive forces were applied to a disc in which nuclear tissue was still posterior, for example lifting immediately after a prolonged period of flexion (like sitting), a concentration of stress load would occur on the posterior annulus. He explains that these two scenarios would demonstrate the concept of spinal memory where the function of the spine is modulated by certain previous activity or loading history. Loading history of the spine determines disc hydration (the size of disc space and disc geometry), which in turn affects ligament length, joint mobility, stiffness, and load distribution.

The functional and clinical relevance of spinal memory in the sporting world would apply to “bench-sitting sports” where those with sensitive backs would do well to avoid sitting with a flexed lumbar spine while waiting to perform (Green *et al.*, 2002:1076). Callaghan and McGill (2001b:373) suggested that sitting with variable postures is recommended as a strategy to minimize the risk of tissue overload and that no single, ideal sitting posture exists.

With a better understanding of the biomechanical and injury mechanism of low back disorders, it is appropriate to focus on the muscle mechanism of *m. lumbar multifidus* in low back pain.

3.3 THE LUMBAR MULTIFIDUS MUSCLE MECHANISM IN LOW BACK DISORDERS

3.3.1 Introduction

A review of medical textbooks, according to Biedermann *et al.* (1991:1179), suggests that most disorders of the lower back have commonly been thought to be caused by mechanical problems of the spine. However, the role of muscles in the biomechanics of the lumbar spine has been examined for many years (Goel *et al.*, 1993:1531). During 1961, Lucas and Bresler provided evidence that muscles provide the majority of the resistance to external load in stabilizing

the spine (Goel *et al.*, 1993:1531). Since then more and more studies (De Vries, 1968:175; Mayer *et al.*, 1985:765; Seidel *et al.*, 1987:592; Biedermann *et al.*, 1991:1182; Bogduk *et al.*, 1992:904; Panjabi, 1992:383; Cholewicki *et al.*, 1997:2210; Richardson *et al.*, 1999:68-76; Comerford & Mottram, 2001b:15; Danneels *et al.*, 2001:186; Huble-Kozey & Vezina, 2002:1100; van Dieën *et al.*, 2003:834; Richardson *et al.*, 2005:4; McGill, 2007:21,49), to name a few, have focussed on the significant role muscles play during increased stresses on passive structures of the lumbar spine.

However, the antithesis to this is that spinal tissue failure results in joint injury, pain and changes in muscle function, including muscle size (inhibition) and motor control changes (Richardson *et al.*, 2005:106). According to these researchers, biomechanical deloading or reduction in weightbearing load leads to muscle dysfunction, specifically in the weight-bearing or local muscles, which have a major role in protecting the joints from injury. For example, studies on *m. vastus medialis oblique* in knee pathology and in microgravity conditions have shown this specific muscle to change the most of all the quadriceps muscles with deloading (Pevsner *et al.*, 1979:869; Musacchia *et al.*, 1992:44S; Richardson *et al.*, 2005:112-114). Richardson (1987:400) documented that *m. vastus medialis oblique* had a predominantly tonic recruitment EMG pattern in healthy matched controls, while the muscle displayed phasic patterns of recruitment in patients with chronic patellofemoral pain syndrome. According to Richardson *et al.* (2005:113), "this would concur with the predicted changes resulting from deloading of the weightbearing muscles".

With respect to low back disorders, Richardson *et al.* (2005:107) stated that research is beginning to produce evidence that changes in weightbearing and non-weightbearing muscle occur as a result of deloading and that the intrinsic lumbar muscles or local muscles show the greatest loss of mass in reduced weight bearing situations. More specifically, several researchers have shown that dysfunction of *m. lumbar multifidus* has been demonstrated in low back patients with reference to muscle activation, fatigability, muscle composition (histology) and size (pathological gross anatomy) (Biedermann *et al.*, 1991:1179; Sihvonen *et al.*, 1991:1080; Rantanen *et al.*, 1993:568; Kay,

2000:102; Kay, 2001:17; Yoshihara *et al.*, 2003:494; Fryer *et al.*, 2004:354; Hides, 2005:149-155; McGill, 2007:109). However, McGill (2007:110) warns against the clinical emphasis that has been placed on the multifidus complex and suggests that clinicians should consider all muscles and not just those that have been chosen for study.

On the other hand, Richardson *et al.* (1999) and Richardson *et al.* (2005) in their books documented a comprehensive review of literature on motor control problems in the deep abdominal and paraspinal mechanisms, specifically *m. transversus abdominis* and *m. lumbar multifidus* as a basis of exercise prescription for lumbar segmental stabilisation. This new paradigm of exercise rehabilitation addressed the motor control problems experienced in these muscles, focussing on improving mechanical support of the lumbar vertebral segments through specific deep-muscle contraction exercises.

Because the past few decades have produced a vast amount of clinical and research studies regarding the *m. lumbar multifidus*, it is important to understand the mechanical anatomy and patho-mechanics of the *m. lumbar multifidus* in developing the optimum evaluation and treatment strategies for patients with low back pain and lumbar segmental instability (Kay, 2000:102; Kay, 2001:17; Richardson *et al.*, 2005: vii). From an evaluation perspective, Ng and Richardson (1994:115) suggested that an additional or different testing model needs to be developed to selectively recruit muscles with a greater stability role, such as the *m. lumbar multifidus*.

3.3.2 Mechanical anatomy of the *m. lumbar multifidus*

Although the global and local muscles of the abdominal mechanism, as well as the torque producing paraspinal muscles play a role in lumbar spinal support, this discussion will mainly focus on the mechanical anatomy of the *m. lumbar multifidus* muscle. The *m. lumbar multifidus* is the largest, deepest and most medial of the major paraspinal muscles that span the lumbosacral junction (Macintosh *et al.*, 1986:196; Bogduk, 1994:116; Kay, 2000:104). Because the multifidi's muscle bulk increases on progression caudally from L2-S1 (Hides *et*

al., 1995:54; Hides, 2005:64,66), it suggests that it is the muscle most capable (as segmental stabilizer) of providing support at this level. In contrast, the cross-sectional area of the *m. lumbar longissimus* and *m. iliocostalis* decreases on progression caudally and are primarily extensors of the spine when acting bilaterally, while they can also assist in lateral flexion when acting unilaterally (Hides, 2005:62, 66).

The clinical importance of the *m. lumbar multifidus* cannot be underestimated because it is the L4-L5 and L5-S1 segments that have the highest incidence of pathology in low back disorders. This is arguably the reason why a large amount of scientific research has focussed on the patho-physiological changes and anato-mechanical function of the segmental *m. lumbar multifidus* (Kay, 2000:102; Kay, 2001:17; Hides, 2005:66, 71).

Anatomical data on the exact attachment sites of the *m. lumbar multifidus* revealed a tight clustering of fascicle attachments with little variation in the 21 subjects studied by Macintosh and Bogduk (1986:210). The *m. lumbar multifidus* has five separate bands, each consisting of a series of fascicles that stem from the caudal one-fifth of the spinous processes and laminae of the lumbar vertebrae and insert into the mamillary processes of the vertebrae two, three, four or five levels inferiorly (Hides, 2005:60). According to Macintosh and Bogduk (1994:189), there is a common tendon from the spinous process with five overlapping fascicles from each level. These fascicles diverged caudally to attach separately to mammillary processes, the iliac crest and sacrum (Figure 3.3).

The tendon from the base of the L1 spinous process attaches to the L4 mamillary process, the mamillary processes of L5-S1 and to the posterior superior iliac spine. The tendon from the base of the L2 spinous process attaches caudally to the mamillary process of L5. The common tendon coming from the base of the spinous process of L3 attaches to the mamillary process of the sacrum. The multifidi from the L4 spinous process attaches more medially to the L3 insertion area but still lateral to dorsal sacral foramina. The multifidi from the L5 level travels medial to the L4 muscles and inserts medial to dorsal sacral

foramina. The deeper fibres attach to the facet capsule allowing the multifidi to control the capsule from being pinched during extension movements. Therefore, the *m. lumbar multifidus* is arranged for individual control of the vertebrae and this is supported by its neural innervation.

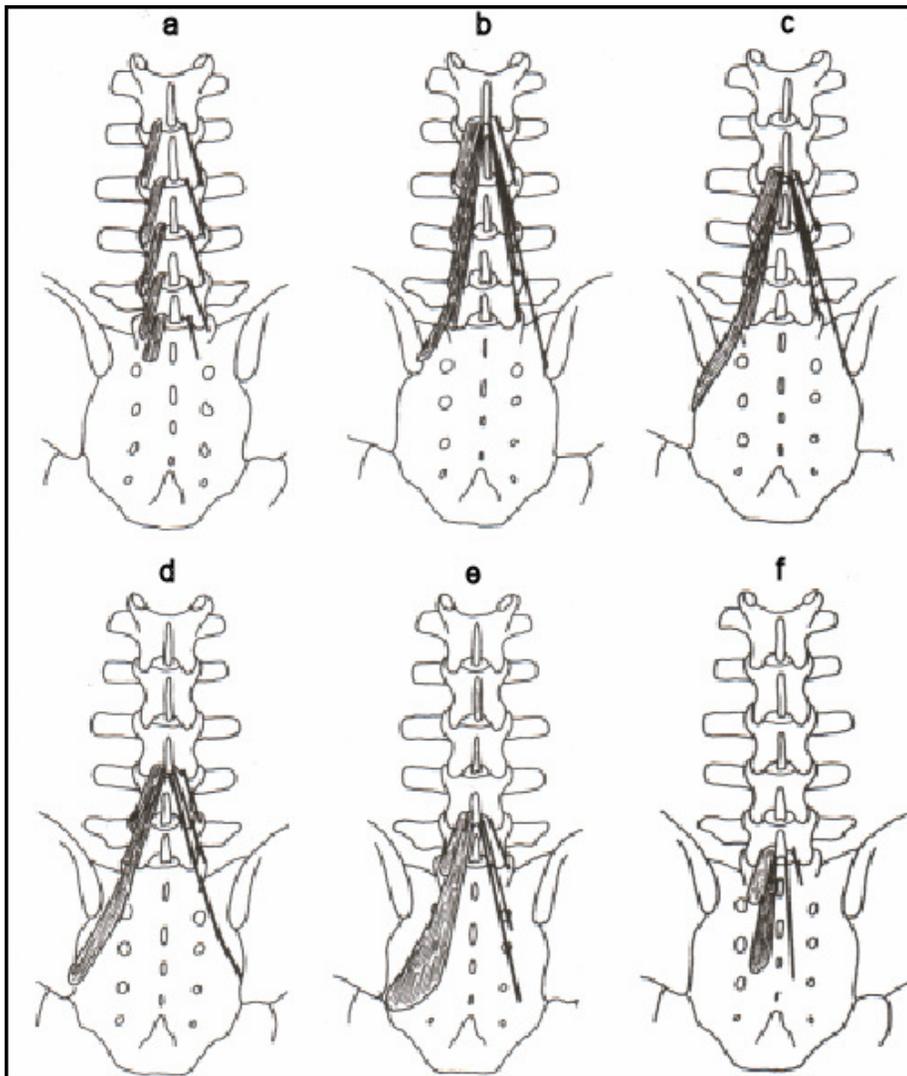


Figure 3.3: The vertebra-to-vertebra attachments of *m. lumbar multifidus* within the lumbar and between lumbar and sacral vertebrae. (a) the laminar fibres at every level; (b-f) the longer fascicles from the caudal edge and tubercles of the spinous processes at levels L1-L5. (reproduced with permission from Bogduk, 1997:106).

In terms of the insertions of the *m. lumbar multifidus*, a vertical and horizontal vector was indicated (Figures 3.4 and 3.5). A posterior view showed an increasing obliquity from L1 to L3, while below L3 the obliquity gradually narrowed to attach more medially; this represented a major vertical vector and a minor horizontal vector (Figure 3.4). Viewed laterally, all *m. lumbar multifidus* fascicles are oriented at approximately 90 degrees to the long axis of the vertebral pedicle (Figure 3.5). From this it can be seen that these fascicles are oriented to provide primarily posterior sagittal rotation of their vertebra of origin, while the length of the spinous processes endowed them with considerable mechanical advantage. However, they were not well oriented to resist shearing (Kay, 2001:23).

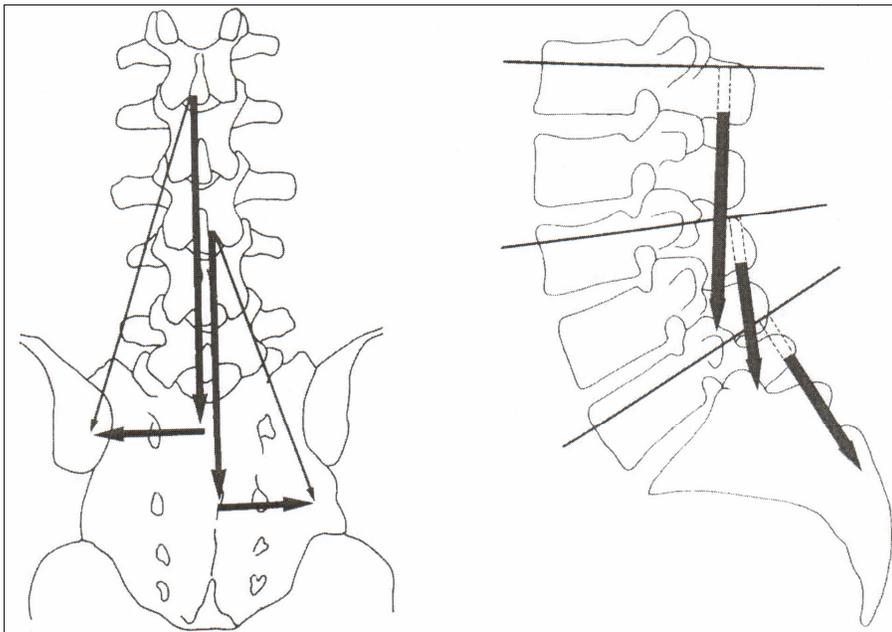


Figure 3.4

Oblique orientation of fascicles of *m. lumbar multifidus* into major vertical and minor horizontal vectors.

(reproduced with permission from Macintosh & Bogduk, 1986:205-213).

Figure 3.5

Average perpendicular actions of *m. lumbar multifidus* on the lumbar spinous processes.

Considering the neural innervation of segmental *m. lumbar multifidus*, Bogduk *et al.* (1982:388) reported that pain from the facet joint is due to innervation from

the medial branch of the L1-4 dorsal rami and that this should be the focus of denervation for pain relief. After dissecting the lumbar dorsal rami in six cadavers, it was found that the L1-4 dorsal rami form three branches: a medial branch innervating the multifidi, an intermediate branch for the *m. iliocostalis* and a lateral branch for the *m. longissimus*. The medial branch also innervates the interspinous ligament and muscle, as well as the adjacent facet joint.

Aspden (1992:384), Macintosh and Bogduk (1994:189) and Hides (2005:63-64) described the separate innervation of the *m. lumbar multifidus*, attached to the vertebra, as an essential part of enabling separate control of individual vertebra. Because each vertebra can be controlled independently it means that applied loads on the lumbar spine can be adjusted more precisely. According to Hides (2005:64), no specific segmental nerve-muscle relationship exists with the *m. lumbar longissimus* and *m. iliocostalis*, suggesting a more general relationship to the spinal segments.

According to Macintosh *et al.* (1993b:205), Kay (2000:105) and Kay (2001:23-24), the muscles best suited for trunk rotation are the oblique abdominal group, while McGill (1991a:91) also added the *m. latissimus dorsi*. Because of the flexion moment associated with the oblique abdominals, the paraspinal muscles might be acting as an antagonist to trunk flexion. In a similar function, the *m. lumbar multifidus* is thought to work as an anti-flexor to balance the anterior sagittal rotation produced by the trunk flexors. In trunk flexion, the *m. lumbar multifidus* and *m. lumbar longissimus* and *m. iliocostalis* control the anterior rotation and anterior translation of the spine (Hides, 2005:63). However, an important biomechanical characteristic, known as the “flexion-relaxation” phenomenon or “critical point” in trunk flexion, where the myoelectricity of the trunk extensors diminishes, needs to be considered (Hides, 2005:71; Shin *et al.*, 2004:486).

On return to upright, the *m. lumbar multifidus* induces posterior sagittal rotation, assisted by the lumbar erector spinae which also control the posterior sagittal translation. Bogduk *et al.* (1992:897) found that the *m. lumbar multifidus* contributes only 20% to the total extensor moment calculated at the L4-5

vertebral level, the lumbar erector spinae contributes 30%, while the thoracic component of the erector spinae contributes 50%. This finding suggests that the *m. lumbar multifidus* is at a mechanical disadvantage to produce extension of the thoracic cage on the pelvis from a flexion position, even though it is the largest muscle at the lumbosacral junction. Also, because of its insertions, there is a large vertical vector indicating it as a stabiliser of sagittal rotation rather than a torque producing extensor muscle of the back (Kay, 2000:105).

When the horizontal force vector for axial rotation was calculated, it did not exceed 29% of the total force generated by the *m. lumbar multifidus* and was even less for most of the fascicles. Macintosh *et al.* (1993b:205) reported that the fascicles of the *m. lumbar multifidus* contributed only 1.5 Nm (1.5%) axial torque for upright rotation at L3, while 1.3 Nm and 0.5 Nm were exerted at L4 and L5 respectively. In an earlier study, Parnianpour *et al.* (1989:409) measured the total torque exerted in trunk rotation to be 100 Nm, while only 5% came from the back muscles.

With respect to the axial attachments of the *m. lumbar multifidus*, it is poorly suited to contribute to the motion of lateral side bending (Kay, 2001:23). Zetterberg *et al.* (1987:1038) reported that in maximal lateral-bending, the ipsilateral *m. rectus abdominus* was active at 50%, the *m. longissimus* at 70% and the L3 multifidi at about 40%. Contralateral muscle activity was 20% in the *m. rectus abdominus* and almost seven percent with the *m. lumbar multifidus* and *m. longissimus*. This coincides with the biomechanical study of Crisco and Panjabi (1991:796) where they found that the *m. lumbar multifidus* is capable of controlling spinal segmental stiffness in the sagittal plane with its deeper, intersegmental fibres, but it needs the assistance of the *m. lumbar longissimus* and *m. iliocostalis* to control neutral zone motion in the frontal plane. Various studies have investigated the lumbar muscles' ability to increase spinal segmental stiffness and specifically the neutral zone motion.

Lewin *et al.* (1962:299) and Hides (2005:62, 66) reported that the *m. lumbar multifidus* has a close relationship to the zygapophyseal or facet joints; it controls the sliding motion in the cranio-caudal direction and therefore stabilises

the stresses within the vertebral triad (the two zygapophyseal joints and the cartilaginous joints between the vertebral bodies) preventing overloading of the facet joints. According to these researchers, “the deep *m. lumbar multifidus* is the only muscle which functions primarily as a protective mechanism for this vertebral triad”. This lumbar segmental protective mechanism is possible because of the unique arrangement of predominantly vertebra-to-vertebra attachments within the lumbar and between the lumbar and sacral vertebrae (Hides, 2005:60, 66).

The biomechanical study of Panjabi *et al.* (1989:198) specifically simulated the effect of intersegmental muscle forces on spinal instability without the influence of *m. lumbar longissimus* and *m. iliocostalis*. They concluded that the intersegmental nature of the deep multifidi fibres gave a significant advantage to the neuromuscular system to control the stability of the lumbar segment.

A few years later, the biomechanical study of Wilke *et al.* (1995:196) investigated the influence of different muscle groups on the L4-L5 motion segment in human spines. The muscles examined were the *m. lumbar multifidus*, *m. lumbar longissimus*, *m. lumbar iliocostalis* and *m. psoas major*. The total neutral zone motion was decreased by 83% in flexion and extension, decreased by 76% in lateral flexion and no significant change happened in axial rotation. The strongest influence was created by the *m. lumbar multifidus*, which was responsible for more than two thirds of the increase in segmental stiffness. Also, the animal studies of Indahl *et al.* (1995:2652) and Indahl *et al.* (1997:2834) made a clear connection between the irritation (by means of electrical stimulation and saline injection) of a lumbar facet joint or the annular fibres of a disc and a protective contraction of the *m. lumbar multifidus*.

According to Hides (2005:70), it is possible for the different fascicles of the *m. lumbar multifidus* to have different primary functions, for instance, the longer fascicles which originate from the spinous processes have a mechanical advantage over the shorter, deeper fibres. The longer fascicles may contribute more to extensor torque production compared to the shorter, deeper fibres with little leverage. The shorter fibres may be more involved in a tonic stabilising role

during the maintenance of upright postures. This has been found in various EMG and biomechanical studies (Valencia & Munro, 1985:205; McGill, 1991b:809; Wilke *et al.*, 1995:192; Keifer *et al.*, 1998:471; Moseley, *et al.*, 2002:E29).

3.3.3 Morphology of the *m. lumbar multifidus*

Over the past few years, dysfunction of the lumbar muscles in patients with low back pain has been demonstrated using imaging modalities that allow assessment of muscle size or cross-sectional area (CSA) and muscle consistency. With respect to the cross-sectional anatomy of the lumbar spine, MRI, computed tomography (CT) and real-time ultrasound imaging (US) have been used fairly extensively to show dysfunction of the *m. lumbar multifidus* in terms of muscle size or CSA, as well as muscle consistency, for example the degree of fatty infiltration in the multifidi (Hides, 2005:151).

Various studies have reported on paraspinal muscle atrophy in patients with chronic low back pain but in most instances it has been ascribed primarily to disuse and deconditioning (Cooper *et al.*, 1992:389; Hultman *et al.*, 1993:114; Parkkola *et al.*, 1993:830). In post-operative patients, paraspinal muscle differences between sides, as well as vertebral levels, have been reported (Laasonen, 1984:9; Alaranta *et al.*, 1993:137). In unilateral cases, paraspinal muscles were shown to be 10%-30% smaller on the affected side than on the unaffected side. A positive relationship between the fat content of the paraspinal muscles at the lumbosacral junction and disability index was demonstrated in post-operative patients and those with chronic low back pain.

Hides *et al.*, (1994:165) used US to measure the CSA of the *m. lumbar multifidus* in patients with acute and sub-acute low back pain, as well as normal subjects. The scans revealed marked asymmetry of the *m. lumbar multifidus* in each patient and were isolated to one vertebral level. In patients, the mean percentage between side differences were 31% ($p < .001$) while normal subjects were only three percent ($p < .001$), indicating symmetrical *m. lumbar multifidus* CSA between sides. Earlier, Stokes *et al.* (1992:280P) used US to compare *m.*

lumbar multifidus wasting in 26 patients with acute low back pain (less than three weeks of pain) to 10 normal subjects. Significant differences ($p < 0.05$) between the sides in CSA at the level of symptoms were reported.

The effects of low back pain on the *m. lumbar multifidus* size have been investigated by CT scan in postoperative patients (Sihvonen *et al.*, 1993:578). The findings demonstrated that, in some cases, lumbar surgery for spinal stenosis and/or disc herniation could lead to severe changes in the *m. lumbar multifidus* at L5–S1. Two groups of subjects were studied: those with poor outcomes and those with good outcomes from surgery. A variable, which related to poor outcomes, was *m. lumbar multifidus* atrophy and this was more prevalent in the poor postoperative group. They speculated over possible iatrogenic trauma to the dorsal nerve root and subsequent denervation of the *m. lumbar multifidus* following laminectomy.

Apart from investigating histological changes in the *m. lumbar multifidus* in first-time and repeat-operation patients, Kawaguchi *et al.* (1994:2600) also found multifidi atrophy in the repeat surgery group. This finding indicated the long lasting effect of severe degenerative changes of the low back muscles after posterior lumbar surgery and it was recommended that the retraction time should be shortened during posterior lumbar surgery.

The research group of Kelly used US to find a segmental decrease in multifidi CSA area in patients with chronic low back pain with disc protrusion and radicular leg pain prior to surgery (Hides, 2005:152). The L4-L5 vertebral level showed the greatest difference. Subjects with more low back pain showed greater wasting of the *m. lumbar multifidus*. At the one year post-operative follow-up, there was still a significant correlation between leg pain and *m. lumbar multifidus* atrophy. Similar findings were observed in the MRI study of Kader *et al.* (2000:145), where muscle consistency was graded as mild (fatty or fibrous tissue replacement less than 10%), moderate (replacement less than 50%) and severe (greater than 50%) in 78 chronic low back pain patients. Lumbar multifidus muscle degeneration was present in 80% of the subjects and was seen most commonly at the L4-L5 and L5-S1 vertebral levels. Danneels *et*

al. (2000:266) also reported selective atrophy of the *m. lumbar multifidus* muscle in chronic low back pain patients and it was specific to the L5 vertebral level.

Hides and Associates also measured and documented the *m. lumbar multifidus* CSA at five vertebral levels in 40 normal subjects (13 males and 27 females) (Hides, 2005:66). Because the *m. lumbar multifidus* at the fifth vertebral level (L5) is the focus of this study, only the *m. lumbar multifidus* CSA at L4 and L5 will be mentioned (Table 6.1).

Table 3.1: The *m. lumbar multifidus* CSA at L4 and L5 in normal subjects (adapted with permission from Richardson *et al.*, 2005:66).

Vertebral Level	CSA (mean cm ² (SE))	Confidence Interval
Female		
L4	4.78 (0.2)	4.37- 5.18
L5	6.38 (0.18)	6.01- 6.75
Male		
L4	6.27 (0.29)	5.68- 6.85
L5	6.79 (0.27)	6.25- 7.32

With respect to acute low back pain or injury, Hides *et al.* (1996:2763) documented the rapid and persistent effect of reflex inhibition that was seen in the multifidi muscle at vertebral level L5 in these patients. Despite having been pain free at four weeks and resuming normal work, sport and leisure activities for the next six weeks, multifidi muscle size was not restored at 10 weeks in patients who did not receive specific intervention to reactivate the *m. lumbar multifidus*. Long-term results showed that the specific exercise group experienced fewer recurrences of low back pain (30%) compared to the control group with 84% ($p < 0.001$) (Hides *et al.*, 2001:E246). This study highlighted the segmental multifidi muscle inhibition, still evident in the control group at 10 weeks, which exposed the injured segment (L5) to decreased muscle support

and a predisposition to further injury. These results further supported other research studies (Stokes & Young, 1984:7; Wise *et al.*, 1984:95; Hurley & Newham, 1993:127; Hides *et al.*, 1994:165) in favour of the argument that the most likely cause of decreased multifidi CSA in patients with acute low back pain is reflex inhibition.

According to Hides (2005:158,160), there is a significant body of evidence that illustrates *m. lumbar multifidus* dysfunction. By using imaging techniques like MRI and CT imaging and/or US, the impairment can be shown to the patient and it also offers an incentive to comply with a rehabilitation programme.

3.3.4 Histology of the *m. lumbar multifidus*

Apart from the studies that investigated *m. lumbar multifidus* atrophy, as well as fatty infiltration of this muscle, several studies have examined the histological or fibre characteristics of the *m. lumbar multifidus*. Jorgensen *et al.* (1993:1439) took muscle biopsies from the L3 multifidi, *m. lumbar longissimus* and *m. lumbar iliocostalis* within 24 hours of death from six young male cadavers. The fibres were classified as slow twitch (ST) and fast twitch (FT), or as Häggmark and Thorstensson (1979:319) classified them, type I (slow oxidative) and type II (fast glycolytic). The *m. lumbar longissimus* muscle had an average of 70.5% type I fibres compared to the 54% of the *m. lumbar multifidus* and 55% in *m. lumbar iliocostalis* ($p < .001$). According to the fibre characteristics, the investigators concluded that the trunk extensors have prolonged endurance capabilities compared to other skeletal muscle. This suggests that the lumbar paraspinal muscle group is fatigue resistant and therefore good for postural function.

However, Thorstensson and Carlson (1987:198) found no statistically significant differences between the L3 *m. lumbar longissimus* and *m. lumbar multifidus* in regard to type I or type II fibres in 16 healthy subjects. The overall mean values for all subjects for type I fibres were 62% in *m. lumbar longissimus* and 57% in the *m. lumbar multifidus*. According to these researchers, they found no

evidence of functional specialization in the muscles of the lumbar spine with respect to their fibre types.

In two previous studies from the same research group, muscle biopsies were taken from 17 back surgery patients, three cadavers within 24 hours of death and 10 patients with idiopathic scoliosis (Jowett *et al.*, 1975:158; Fiddler *et al.*, 1975:220). The cadaver biopsies, taken from the L5 *m. lumbar multifidus* (assumed to be normal backs) resulted in type I fibres ranging from 58% to 69% (Fiddler *et al.*, 1975:220). Further to this, muscle biopsy analysis from the back surgery and scoliosis patients led to various findings. There was an increase in type I fibres ($p < 0.01$) with advancing age, which was due to a smaller size of type II fibres. Patients with nerve root signs showed a significant reduction in the number of type II fibres ($p < 0.05$), leading to muscle atrophy. From these findings it appears that the relative percentage of type I fibres increases both with age or pathology. This indicated a more postural (tonic) role and less of an ability to create phasic activity (Jowett *et al.*, 1975:158).

In the study of Mattila *et al.* (1986:732) the most significant pathological findings were the differences in the number of core-targetoid and moth-eaten fibres in surgery patients compared to the control group (cadavers). Lumbar multifidus muscle biopsies were taken from the L4-5 or L5-S1 levels. A very large individual variation in the ratio of type I to type II fibres were found, but with no significant correlation in terms of relative numbers. Type II fibres were significantly smaller than type I fibres. In the end, the findings were non-specific and multiple factors like denervation, excessive physical strain, etc. were indicated as a possible cause for pathology. Sirca and Kostevc (1985:135) investigated, among others, the fibre distribution of the L2 multifidi and *m. longissimus* in male cadavers and 17 patients who underwent surgery for a herniated disc. There was nearly an equal distribution of type I and type II fibre types in the lumbar spine. However, the *m. lumbar multifidus* had more type I fibres than the *m. longissimus*. Also, a higher percentage of type I fibres, in the range of 8%-13%, have been reported in the *m. lumbar multifidus* compared to the *m. lumbar longissimus* (Verbout *et al.*, 1989:120).

In a later study, Rantanen *et al.*, (1993:572) investigated the histological profile of the *m. lumbar multifidus* five years after surgery in two groups of subjects (positive outcome or negative outcome). Biopsies collected at operation from all subjects showed evidence of type II muscle fibre atrophy and type I fibre internal structural changes. Histochemical studies of Zhao *et al.* (2000:2197), Yoshihara *et al.* (2001:625) and Yoshihara *et al.* (2003:494) also found significant decreases in the size of type I and type II fibres, along with structural change or atrophy at the L5 *m. lumbar multifidus*. Rantanen *et al.*, (1993:572) found that there were no significant changes in atrophy between the two patient groups at follow-up. However, changes in the internal structure of type I muscle fibres were markedly different between the groups. In the positive-outcome group, the presence of moth-eaten and core-targetoid fibres decreased. In contrast, the negative outcome-group showed a marked increase in the frequency of these abnormalities, especially the percentage increase in moth-eaten fibres from 2.7%-16.7%. According to Hides (2005:151), these results highlight the potential clinical importance of dysfunction in this muscle.

In summary, the importance of the supporting function of the three lumbar muscles may be reflected in the distribution of muscle fibre type. Postmortem studies have revealed that the *m. lumbar multifidus* and the lumbar and thoracic components of the erector spinae muscles have a high percentage of type I fibres. The presence of both a higher percentage of type I fibres, as well as a larger type I fibre size, compared with the more fatigable type II fibres, arguably supports the tonic role of these muscles (Hides, 2005:63). Further to this, the concentration of oxidative enzymes in all the lumbar muscles is large and the endurance capacity high. This histochemical composition, as well as the high composition of type II fibres, indicates the tonic holding and thus supportive function of these muscles (Jorgensen *et al.*, 1993:1439).

When the tonic holding function of *m. lumbar multifidus* gets disturbed by injury, the effect of reflex inhibition is known to be rapid (Hides, 2005:123). This is evident in the segmental decrease in the *m. lumbar multifidus* CSA and is localized to the side of painful symptoms in patients with unilateral low back pain (Hides *et al.*, 1994:165). Further evidence regarding muscle response to

reflex inhibition comes from biopsy studies to analyze the relative effects of reflex inhibition on different muscle fibre types. Häggmark *et al.* (1981:15) showed that type I fibre atrophy occurred within the first week following knee joint injury. Also, a drastic fall from 81%-57% in type I fibres of the quadriceps muscle was reported in a cross-country skier who was treated surgically. It took several months for the quadriceps to return to its pre-injury fibre distribution, despite early active rehabilitation.

Finally, the impairment in *m. lumbar multifidus*, which has similar muscular characteristics as *m. vastus medialis* (Hides, 2005:125), may be explained as follows: firstly, muscle atrophy has been demonstrated in immobilization studies and has been explained in terms of the greater amount of type I fibres in *m. vastus medialis* that are the most vulnerable to immobilization-induced atrophy of the quadriceps group (Appell, 1990:56; Musacchia *et al.*, 1992:48S). However, Zhao *et al.* (2000:2197) and Yoshihara *et al.* (2001:625) found significant decreases in both type I and type II fibres of *m. lumbar multifidus*. Secondly, Richardson (2005:107) and Hides (2005:124) suggested that extensor muscles undergo more severe atrophy than flexors in reflex inhibition. Thirdly, muscles that function as anti-gravity muscles, otherwise called the local muscles of the weightbearing group, are mostly affected.

3.3.5 Neutral spine posture and muscle activation of *m. lumbar multifidus*

Physiological posture, or the many-curved configuration of the spine adopted during manoeuvres, is important when assessing spinal function. When spinal curves are maintained, it is the most energy-efficient position for the body to stay upright against the forces of gravity, as well as other forces that are applied to the spine (Aspden, 1992:372; Hides, 2005:68). For this, a complex organisation of the paraspinal muscle group is required to control the curvature and compressive force along the spine to ensure mechanical stability. For the lumbar spine, it has been proposed that *m. lumbar multifidus* is one of the major contributors to the stability of the lower back via control of the physiological lordosis (Aspden, 1992:385; Hides *et al.*, 2005:205).

The biomechanical studies of Keifer *et al.* during 1997 and 1998 investigated the relationship between local and global muscles to maintain neutral spinal curves during axial compression. The addition of local and global muscles into their model increased the ability of the spine to withstand compressive forces without buckling. Pelvic rotation or anterior tilt by only two degrees allowed the spine to carry axial compression of up to 400 N with minimal anterior displacement of T1 and highlighted the importance of integration of the two muscle systems (Keifer *et al.*, 1997:45). In the other study, Keifer *et al.* (1998:471) showed that the global muscles and passive structures are sufficient to stabilise the spine in neutral for very small displacements. However, the system is far more efficient with inclusion of the local muscles. Considering the contribution of the local muscles, 80% was provided by the *m. lumbar multifidus*, with some contribution by *m. iliocostalis*.

Quantification of the muscle activity of *m. lumbar multifidus* in dynamic posture function would be helpful in determining effective intervention and back assessment strategies. The gold standard measurement tool for assessing muscle activation is kinesiological electromyography (EMG) (Kiesel *et al.*, 2007:162). Furthermore, EMG has been extensively used during the last decades in studying muscle activation and fatigue patterns (Elfving & Dederling, 2007:28). In recent years, EMG studies on the multifidi have increased dramatically (Haig *et al.*, 1995:715; Ng *et al.*, 1997:956; Elfving *et al.*, 2000:117; Arnall *et al.*, 2002:769; Moseley *et al.*, 2002:E30; Sung, 2003:1314; Kramer *et al.*, 2004:531; Vasseljen *et al.*, 2006:906; Kiesel *et al.*, 2007:162).

With respect to the assessment of muscle activity in maintaining upright postures, the researchers Asmussen and Klausen used surface EMG electrodes on *m. rectus abdominis* and the lumbar erector spinae muscles during a static standing posture (Kay, 2001:19). Seventy eight percent (78%) of the subjects showed activity in the low back muscles (36 of 46 normal subjects studied) while the remaining 22% of the subjects displayed abdominal activity during static upright standing. These EMG findings led the researchers to speculate that the line of gravity must be anterior to the axis for sagittal lumbar movement, the reason for more activity in the low back muscles to counter

anterior gravity pull. According to lateral X-rays and the dropping of a line vertically from the external auditory meatus, these researchers calculated that the line of gravity was located, on average, one centimetre anterior to the centre of the L4 vertebral body.

More studies confirmed greater EMG activity in the paraspinal muscles in upright postures. Kippers and Parker (1985:103) reviewed EMG literature pertaining to the erector spinae (including *m. lumbar multifidus*) in different postures. However, they did not distinguish the *m. lumbar multifidus* from the *m. iliocostalis lumborum* and the *m. longissimus thoracis*. Standing was described as intermittent erector *spinae* activity, but three times greater than abdominal activity. In the study by Jonsson (1970:19), bilateral fine-wire EMG was used to assess the function of the *m. lumbar multifidus*, *m. lumbar longissimus* and *m. iliocostalis* at different lumbar levels (T12-L5) in normal subjects. Various tests were done in the prone, standing and sitting positions. Looking only at muscle activity in the three static postures, no activity was found in the three muscles in the resting prone position. However, the *m. lumbar multifidus* was found to be the most active of the three muscles when the subjects were standing at ease, as well as sitting in an erect posture. This action would probably indicate its antigravity support to the spine with continuous activity (Hides, 2005:70).

Donisch and Basmajian (1972:34), Valencia and Munro (1985:219) and Hides (2005:70) reported that slight to moderate activity of the *m. lumbar multifidus* has been demonstrated in the standing posture, again stressing its tonic postural role. According to Hides (2005:70), maintaining a neutral spine position may be the one element that is crucial to maintain this tonic activation. However, some studies performed in the upright sitting position have had varied results. Valencia and Munro (1985:219) documented inactivity in the *m. lumbar multifidus*, while Donisch and Basmajian (1972:34) reported an active *m. lumbar multifidus* in straight, unsupported sitting, again stressing its tonic antigravity function.

In contrast, if a person's body gave in under the force of gravity, the body would gradually progress into a flexed "relaxed" posture, transferring stresses to the

passive joint structures that may cause the trunk and hip extensors to “turn off”, leading to possible imbalances between trunk and hip flexors and extensors. This may lead to low back injury and can be explained in terms of changes in tissue creep with lumbar flexion and the resulting reduced passive joint protection mechanism (Richardson, 2005:111; McGill, 2007:102).

Also, in tasks that involve static trunk flexion postures, an important biomechanical characteristic known as the flexion-relaxation response, needs to be considered. As a full flexion posture is approached, the passive tissues (ligaments, disc, skin, etc) rapidly take over moment production, relieving the back extensor muscles of this function and accounting for their myoelectric silence. The clinical implication would be to avoid static flexion postures of the lumbar spine that causes the local multifidi to deactivate or reach a period of myoelectric silence (Schultz *et al.*, 1985:195; Shin *et al.*, 2004:486; Richardson, 2005:111; McGill, 2007:76; Konrad, 2008). The flexion-relaxation phenomenon is described in more detail in Paragraph 3.2.1.

Haig *et al.* (1993:482) and Haig *et al.* (1995:719) used EMG to do paraspinal muscle mapping (PM), including the *m. lumbar multifidus*, in normal subjects (1995-study) and low back pain patients (1993-study). The mapping technique involved inserting fine-wire EMG electrodes in search of fibrillation or denervation in the paraspinal muscles and the *m. lumbar multifidus*. Greater numbers of abnormalities were documented at the lower vertebral levels in the spine ($p < 0.0001$). The mean PM score was 1.11 for the normal subjects (1995 study) compared to a PM score of 12.18 for low back pain patients (1993 study). Frequent mild denervation of the L5 *m. lumbar multifidus* was documented, which led the authors to suggest that pathology in this segmentally stabilizing muscle might predispose a normal subject to segmental instability and eventual disc degeneration and possible herniation.

Ng and Richardson (1994:119) investigated the muscle activity of the right erector spinae at L1 and the right *m. lumbar multifidus* muscle at L5 in neutral and extended postures in normal subjects. Surface electrodes were used. The testing procedure included a prone arch test (used as reference because of

maximum *m. lumbar multifidus* contraction), horizontal trunk-holding and leg-holding tests of five seconds each. Significant differences were found between the normalised EMG data for trunk and leg-holding for the erector spinae ($p < 0.05$) and *m. lumbar multifidus* ($p < 0.005$). This indicated greater muscle activity for trunk-holding compared to leg-holding for both muscles, but more significantly for the *m. lumbar multifidus*. It was suggested that the horizontal trunk and leg-holding positions be used for low back strengthening because it allows one to train in a neutral position which places less stress on the lumbar discs. The authors speculated that although these tests showed high muscle activity, an additional exercise model might be necessary to specifically recruit lumbar muscles with a greater stability role, such as the *m. lumbar multifidus*.

In another EMG study on normal subjects, Ng *et al.* (1997:958) used surface electrodes to assess the muscle activity and fatigue patterns of *m. iliocostalis lumborum* at the L2 level and *m. lumbar multifidus* at the L5 level during a trunk-holding test. Greater activity was recorded in the *m. lumbar multifidus*, 78% compared to the 68% for the *m. iliocostalis lumborum* ($p < 0.005$). There was also a greater fatigue rate in the *m. lumbar multifidus* compared to the *m. iliocostalis lumborum* and there was increased activity in both muscles with fatigue. It was suspected that the increased muscle activity with fatigue might have been due to greater recruitment of other motor units to maintain the required force output to meet the demands of the test. The higher fatigue rate of the *m. lumbar multifidus* compared to the *m. iliocostalis lumborum* was thought to be due to the greater activity levels.

In the study of O'Sullivan *et al.* (2006:E711), three different upright sitting postures resulted in altered *m. lumbar multifidus* activation patterns. Two of the three postures, namely, the slump sitting- and thoracic upright sitting postures showed no real difference in *m. lumbar multifidus* activity. However, the lumbopelvic upright sitting- or neutral spine posture, showed significant greater muscle activity ($p < 0.001$) of *m. lumbar multifidus*, compared to the other two postures. The results of this study highlighted the importance of specific postural training when the aim is to activate the anti-gravity lumbar extensor, *m. lumbar multifidus*.

3.4 SUMMARY

The clinical importance of the *m. lumbar multifidus* cannot be underestimated because it is the L4-5 and L5-S1 segments that have the highest incidence of pathology in low back disorders (Hides, 2005:66), arguably affecting *m. lumbar multifidus* the most of all paraspinal muscles. Furthermore, the clinical importance of *m. lumbar multifidus* is reflected in the segmental arrangement of the multifidi fascicles, the large size of *m. lumbar multifidus* at the lumbosacral junction, the segmental innervation and control function of *m. lumbar multifidus* and the close relationship between the *m. lumbar multifidus* and the zygapophyseal joints (Kay, 2000:104).

With respect to the morphology of *m. lumbar multifidus* in acute low back pain or injury, as well as chronic low back pain patients, a significant body of evidence has documented *m. lumbar multifidus* dysfunction that was seen at the 5th vertebral level (L5). The most likely cause of decreased multifidi CSA in patients with low back pain is reflex inhibition of the neuromuscular system and deloading of the musculoskeletal system (Richardson, 2005:107; Hides, 2005:125).

The multifidi has a high proportion of type I muscle fibres and is well suited to a tonic, holding function or isometric muscle contraction. EMG has provided further evidence of this continuous activation of the deep fibres of the *m. lumbar multifidus* (Hides, 2005:72). It has been suggested that *m. lumbar multifidus* primarily controls the neutral zone and increases segmental stiffness because the deep fibres of multifidi lie close to the spinal motion segment or vertebral triad and secondly, the length of the deep muscle fibres remains unchanged in any physiological posture.

With respect to the lumbar flexion injury mechanism, it is important to take note of the flexion-relaxation response. When a standing, full flexion posture is approached, the passive tissues (ligaments, disc, skin, etc) take over moment production, relieving the back extensor muscles of this function and accounting for their myoelectric silence. This causes a forward or anterior shear on the

intervertebral joint, especially at the lower lumbar levels where these forces are greatest (Hides, 2005:69). The clinical implication would be to avoid daily, flexion postures of the lumbar spine that cause the local multifidi to deactivate or reach a period of myoelectric silence (Schultz *et al.*, 1985:195; Richardson, 2005:111; Shin *et al.*, 2004:486; McGill, 2007:76; Konrad, 2008).

With respect to dynamic postural control and maintaining the normal lordosis during lifting, it is important that the spine be adjusted to match the applied mechanical forces (McGill, 2007:100-103). Activation of the local muscles (80% contribution from the *m. lumbar multifidus*) plays a major role, while integration of the weightbearing (one-joint) and non-weightbearing (multi-joint) muscles must be undertaken (Hides, 2005:73; Richardson, 2005:94).

Furthermore, the exercise that may be effective in the management of low back pain involves teaching patients to form a dynamic muscle “corset” involving co-activation of the *m. transversus abdominis* and *m. lumbar multifidus*. However, clinical evidence of the effectiveness of retraining the corset action in low back pain patients has demonstrated that strength training (with added load) is actually required to achieve significant improvement in muscle function of the *m. lumbar multifidus*, as assessed by its CSA (Danneels *et al.*, 2000:270; Richardson & Hides, 2005:88).

These studies lend support to the view of Richardson and Hides (2005:88) that stabilization and protection of the low back in patients with low back disorders involves higher level strengthening exercise. These researchers suggest that this strengthening exercise should focus on the anti-gravity system, which controls high-load weightbearing function. In the light of these scientific findings, the researcher has designed a new muscle testing instrument, the pressure air biofeedback (PAB) device and developed a new biomechanical test for the low back pain patient in the clinical office setting. With this in mind, a detailed discussion of the PAB device development will follow in Chapter Four.

CHAPTER FOUR
NEW TECHNOLOGY DEVELOPMENT OF A PRESSURE AIR
BIOFEEDBACK DEVICE

- 4.1 INTRODUCTION
- 4.2 MARKET OPPORTUNITIES AND EMERGING TECHNOLOGIES
- 4.3 HISTORIC DEVELOPMENT OF MODERN DYNAMOMETRY
 - 4.3.1 Development of the pneumatic dynamometer
- 4.4 DESIGN AND VALIDATION OF THE PRESSURE AIR BIOFEEDBACK DEVICE
 - 4.4.1 The pressure air biofeedback device
 - 4.4.2 Pressure air biofeedback device calibration
 - 4.4.3 Calibration data of the pressure air biofeedback device versus calibrated weights
- 4.5 SUMMARY

4.1 INTRODUCTION

In the 21st century, different trends or factors are driving the need to develop new and effective measuring devices in the rehabilitation medicine market, specifically with reference to the quantification of muscle strength. This is especially important from the viewpoint of kinematic assessment because activities of daily living are dependent on muscular strength (Nobori & Maruyama, 2007:9). According to Martin *et al.* (2006:154), there has been a growing interest in the development of suitable portable methodology to measure muscle strength in clinical and epidemiological settings, but has been limited because the equipment is too costly. More specifically, muscle strength assessment devices in rehabilitation medicine are too large in size to operate in a clinical office setting, difficult to transfer and too cumbersome to operate (Gubler-Hanna *et al.*, 2007:920). Choi *et al.* (2005:770) have also mentioned the unnecessary utilization of expensive rehabilitation and testing equipment, for example, the MedX Lumbar Extension System.

Thus, any new muscle strength measuring device should aim to be the opposite, i.e. small, simple to operate, portable, inexpensive, valid and reliable. Furthermore, the continuing use of poorly validated and unreliable measurement devices, combined with the lack of sufficient instrumentation available for clinical and epidemiological use, highlights the need for new muscle assessment technology. It also highlights the need to properly evaluate measuring devices that have been developed before incorporating them into clinical practice (Alexander & Clarkson, 2000:53; Tousignant *et al.*, 2001:235).

This specific need for portable, valid muscle testing instruments in the rehabilitation medicine market is reflected in the growing market sector for rapid testing devices, estimated to be worth four billion US dollars (\$) worldwide. Diagnostic technology, in the form of rapid testing devices, is moving out of the hospital set-up into health-care professionals' offices and into patients' homes. Diagnostic, rapid testing devices are also proving themselves to be increasingly accurate and reliable and are also becoming smaller and more portable (Clinica Reports, 2002; Ward & Clarkson, 2004:2). Healthcare technology developers

should take cognizance of this specific growing market seeing that there is an increasing need for portable, reliable and valid muscle strength assessment devices in rehabilitation medicine.

Apart from the growing market sector for rapid testing devices, it is essential to also be aware of the growing orthopaedic and spinal markets and the factors and trends that affect emerging technologies in these specific markets.

4.2 MARKET OPPORTUNITIES AND EMERGING TECHNOLOGIES

The global orthopaedic market is estimated to have been worth approximately \$29.1 billion in 2006, with the spinal market becoming the largest sector during 2006, surpassing the knee implant sector for the first time (Taylor, 2007). According to Driscoll & Watson (2008), the spine surgery market has driven dramatic change in clinical practice, leading to a global market that will grow to over \$26 billion over the next 10 years. Furthermore, the worldwide spine surgery market will be characterized by the introduction of new technologies, as well as solid market growth in double-digit terms for the same period (Frost & Sullivan, 2006).

The USA holds the biggest share of the world spine market, accounting for 66% of spine products sales in the global market. The European market is next with Germany and France having the biggest share. Japan is the third largest spine market while Asian countries, such as China and Korea, have small, but fast growing spine markets (Clinica Medical Technology News, 2004). The spine surgery sector therefore represents an unusually strong market opportunity in the healthcare technology industry. The striking number of new spine surgery companies founded in the last several years is testimony enough to the potential in this market (Driscoll & Watson, 2008).

With the dramatic changes happening in the worldwide spinal and the rapid-testing device markets, it is important to look at the various trends or factors that drive emerging technologies and market opportunities. They are as follows:

- The world's ageing population is one of the most important factors driving the significant growth in the global orthopaedic market (Woolf & Pfleger, 2003:646; Chantler, 2004:87; Frost & Sullivan, 2006; Taylor, 2007; Driscoll & Watson, 2008). With 77 million baby boomers entering their 60s and 70s in the USA, health care technology companies are expected to develop inventive technologies, since this demographic group is comfortable in using technology (Frost & Sullivan, 2006).
- Low back disorders cause patients to commonly consult doctors and is obviously the main factor driving the global spine sector (Woolf & Pfleger, 2003:652; Clinica Medical Technology News, 2004; van Tulder *et al.*, 2006:S64; Driscoll & Watson, 2008).
- Healthcare technology developers who can produce devices that offer greater clinical benefits and ease-of-use, are expected to succeed, according to Frost & Sullivan (2006).
- Healthcare technology companies who can focus on designing devices that stand out due to their inventiveness, cost-effectiveness and functionality, are also expected to succeed (Romero *et al.*, 2000; Frost & Sullivan, 2006).
- The well documented limitations of manual muscle testing (Frese *et al.*, 1987:1072; Hayes & Falconer, 1992:145; Bohannon, 2005:662; Martin *et al.*, 2006:158; Cuthbert & Goodheart, 2007:2) have also created a need for developing an objective alternative to this subjective method (Fung, 2003:1; Nobori & Maruyama, 2007:9).
- The continuing use of poorly validated and unreliable measurement devices, combined with the lack of sufficient instrumentation available for quantitative muscle testing (Escolar *et al.*, 2001:787; Tousignant *et al.*, 2001:235), further highlight the need for new muscle assessment technology.

Taking the above trends and factors into consideration, it can be argued that healthcare technology companies may strategically position themselves to benefit from these factors. Furthermore, the growth in the spine surgery market may create more growth in the spine rehabilitation market, highlighting the need for clinically effective and cost effective care, as suggested by Chantler (2004:87). This may lead healthcare technology companies to aggressively enter the spine rehabilitation market in providing new technology devices for the quantitative assessment of low back strength. This may already reflect in the design and development of various portable and inexpensive muscle strength testing devices for the orthopaedic market, as recently tested by various researchers in a clinical office setting (Rezasoltani *et al.*, 2003:7; Li *et al.*, 2006:411; Martin *et al.*, 2006:154; Nobori & Maruyama, 2007:9).

While clinical assessment instruments are getting smaller and smarter, it is important to know that modern, 21st century clinical dynamometry has evolved over hundreds of years, from primitive measurements to the highly technological assessment devices of today. Therefore, it would seem appropriate to give a short review of the evolution of modern dynamometry up to the development of the pressure air biofeedback (PAB) device.

4.3 HISTORIC DEVELOPMENT OF MODERN DYNAMOMETRY

According to Pearn (1978b:133), the evolution of modern clinical dynamometry, as far as can be determined, began in 1699 A.D. when the first scientific study on the assessment of muscle strength was produced. Upper limb strength, for example, was measured by the ability to lift or move loads of known weights. This led to a need (in the early 18th century) for an assessment method that would measure muscle strength along a continuum that would allow easy standardization and reproducibility.

During 1763, the Graham-Desaguliers dynamometer was developed in London to measure human muscular force, in such a way that synergistic muscles could not impart a false mechanical advantage to the test. John Desaguliers was the first to establish a standard position for muscle testing. He made quantitative

dynamometry possible and he established that there were differences in the strength of individual muscles from person to person, even though their physical appearance was comparable (Pearn, 1978b:127).

Later, in 1798, the French anthropologists Buffon and Gueneau of Montbelliard commissioned Regnier to invent a multi-purpose dynamometer after they tried the Graham-Desaguliers device. They argued that the Graham-Desaguliers dynamometer “was too large and too heavy to be carried” (Pearn, 1978b:127). The portable Regnier dynamometer was therefore invented (Figure 4.1) to measure a large range of human muscle forces (such as grip strength, the strength of the lower back, etc.) against the pulling strength of artillery-horses (Pearn, 1978a:167; Pearn, 1978b:127; Horne & Talbot, 2002:4).

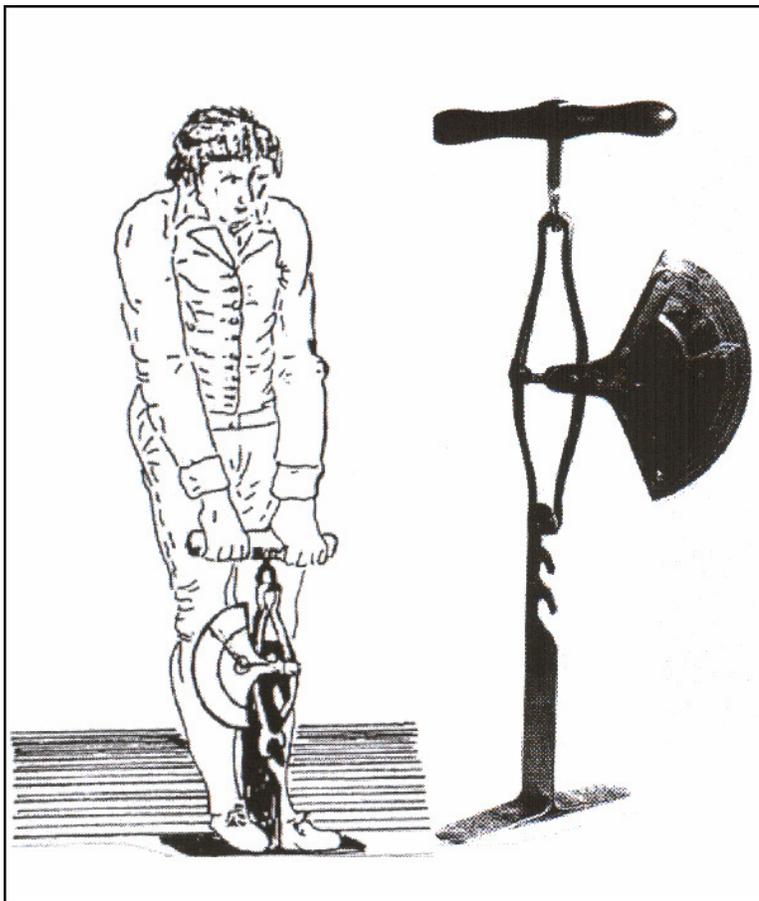


Figure 4.1: The Regnier dynamometer testing low back strength (left) and the dynamometer itself (right) (adapted from Horne & Talbot, 2002).

The argument of Buffon and Gueneau reflected a similar problem that 21st century clinicians experience in that muscle strength assessment and exercise devices are too large in size and too expensive (Helewa *et al.*, 1986:1044; Rozman *et al.*, 2001:236; Li *et al.*, 2006:411; Nobori & Maruyama, 2007:10).

4.3.1 Development of the pneumatic dynamometer

While cost and portability have been repeatedly cited as noteworthy factors for the rejection or redesigning of dynamometers in the 1800s, Allan Hamilton, a North American physician, invented the first pneumatic (water-filled) dynamometer in 1875 to conform to these two factors (Hunsicker & Donnelly, 1955:408; Solgaard *et al.*, 1984:569). After this, in 1939, Geckler published a brief report about his slightly more sophisticated pneumatic (air-filled) device. It consisted of a rubber bulb connected by means of a short tube to an air compressor gauge. The amount of force created as the bulb was squeezed was transmitted to the air gauge and the reading was taken directly from the dial (Clerke, 2006).

This led to the development of the Martin Vigorimeter (air-pressure device for grip strength) that is most commonly used in British occupational therapy departments (Fraser & Bente, 1983:296), which is still compared to other grip strength measuring tools (Desrosiers *et al.*, 1995:137) and is still in use today (Sheldon, 2003:154; Molenaar *et al.*, 2008:1053).

Over the decades, other pneumatic measuring devices have also been promoted by various researchers like Helewa *et al.* (1981:353), Giles (1984:36), Helewa *et al.* (1986:1044), Helewa *et al.* (1990:966), Axen *et al.* (1992:2) and Richardson *et al.* (1992:105). These scientists have developed and used different air pressure biofeedback devices like the modified sphygmomanometer, a compressible ball and a pressure biofeedback unit to quantify the strength, as well as the motor control function, of various muscle groups of patients in a clinical office setting.

The commercially available pressure biofeedback unit (an inelastic, three-section air-filled bag) of Richardson *et al.* (1992:105) is the latest pneumatic device that has been evaluated for the reliable assessment of the motor control contraction of *m. transversus abdominis* (Cairns *et al.*, 2000:127; Storheim *et al.*, 2002:239; von Garnier *et al.*, 2009:8). However, no other research developments on air pressure measuring devices that test and monitor muscle strength have taken place apart from these studies.

4.4 DESIGN AND VALIDATION OF THE PRESSURE AIR BIOFEEDBACK DEVICE

Medical devices are ubiquitous in the diagnosis and treatment of medical conditions at home, the clinical office setting and in hospitals and are the trade of healthcare professionals (Ward & Clarkson, 2004:2). Furthermore, the dramatic growth in the spinal and rapid testing device markets, combined with the factors that drive these markets, was reason enough to design and develop the PAB device.

In designing and developing the PAB device, some basic principles of Human Factors Engineering (HFE), or sometimes referred to as usability engineering, were incorporated. The capture of device requirements, device development, device design validation and device use are HFE principles that have been followed in the design and development process of the PAB (Ward & Clarkson, 2004:13-18). Also, for the PAB device to be used in clinical and epidemiological settings, it has to conform to specific criteria described by Helewa *et al.* (1981:353), Helewa *et al.* (1986:1044), Sapega (1990:1562) and Matheson *et al.* (1992:109), namely:

- It should be quantitative so that numbers may be produced.
- It should be sensitive to changes in muscle strength.
- It should be reliable in that the instrument is free from defect and does not require frequent maintenance.

- It should be reproducible in the hands of different observers.
- It should be adaptable to different muscle groups.
- It should be portable so that it may be used in different settings.
- It should be fast and safe to apply.
- It should be comfortable and simple to apply.
- It should be inexpensive.

Instruments or devices with these properties would be considered valid according to these researchers.

Furthermore, because the PAB device was tested against the Toshiba 6000C US imaging scanner and the Noraxon Myotrace 400 EMG device, it was important that these two clinical instruments be reliable and valid. This was reflected in the Toshiba and Noraxon companies' trackrecord of research and development and their compliance with international accreditation requirements for technology development (Toshiba, 2009; Noraxon, 2008).

Taking the mentioned HFE principles and nine criteria into consideration, the PAB was designed and developed with the cooperation and assistance of an electronic engineer.

4.4.1 The pressure air biofeedback device

The prototype PAB device can be described as an isometric/isotonic exercise-testing device which consists of a self-inflated, air-filled elastic ball, about 22 centimetres (cm) in diameter and which is inflated to a predetermined internal pressure. The inflatable ball is spherical when inflated and may be of rubber or a like material and as such, may be slightly elastic. The PAB ball is partially enclosed within two rigid (fibreglass) shell capsules that fit on the inflatable body

on opposite sides thereof to cover the opposite outer surface segments of the ball. The shell capsules are separable from the PAB ball and may be formed of a fibreglass material or any like suitable rigid material.

Each shell capsule defines an internal profile complementary to the outer surface profile of the outer surface segment of the inflatable ball. The rigid shell capsules are joined together by a primary strap, a nine (9) cm non-elastic, nylon Velcro strap that passes securely over the shell capsules to surround the inflatable ball. In its operative configuration, the primary strap forms a closed loop that is size adjustable with the aid of a suitable complementary Velcro attachment and fits tightly around the inflatable body.

Another nine (9) cm adaptable, non-elastic, nylon Velcro strap, which acts as a handle or torso strap, attaches and extends from the primary strap on one side, through which the torso can be placed. On the other side, a light chain and steel pipe attach and extend from the primary strap for both feet to push against. By pulling the chain and torso strap attachments away from one another, the primary strap pulls the shell capsules towards one another to thereby increase the internal air pressure in the inflatable ball (Figure 4.2).

The pressure monitoring system of the PAB device includes a suitable pressure sensor called the Druck PMP 1400 (see Appendix F) that is connected in communication with the internal space of the elastic ball via Festo tubing (Festo AG & Co.KG, Esslingen) that ensures reliable compressed air supply to the system. An air hand pump is connected in line with the Festo tubing to adjust the baseline internal pressure (mb) of the PAB ball. The pressure transducer comprises of electronically operable features that connect to a personal computer to provide a visual (biofeedback) and/or printed record of pressure changes within the inflatable ball during the performance of muscle strength testing on subjects (Figure 4.2).



Figure 4.2: The PAB device set-up during a seated, closed chain back extension test in a neutral spine posture (pre-test set-up).

More specifically, movement against the PAB ball (direct compression) or away from the PAB ball (Figure 4.2) by pulling against the torso nylon strap attached to the PAB ball and simultaneously pushing against the steel pipe, chain attachment with both feet (pull-compression force) causes volume changes in the PAB ball. These volume changes are then registered by the pressure transducer and communicated to and reflected on the personal computer (PC) screen. The relationship between the external force applied by the subject through the fibreglass shells to the elastic ball and the internal pressure developed within the system is determined by the area of the apposition between the rigid shells and the PAB ball.

The pressure transducer has a range from 0-250 mb with 0.1 mb intervals on the scale, while the accuracy of the pressure transducer device is around three (± 3) mb according to the electronic software developer (Webster, 2008). A

conversion programme may convert mb to pounds per square inch (PSI), which is the international unit most widely recognised. However, the mb unit has been chosen, firstly, because it is more sensitive by measuring air pressure as raw data, secondly, from a practical point of view, it gives better visual feedback to subjects and thirdly, calibration is more precise in smaller units than working with bigger units.

The PAB device was set up in a seated, closed chain, isometric test with the spine in an upright neutral position which reflected a zero degree (0°) test angle for the trunk (Elfving *et al.*, 2000:118; Elfving *et al.*, 2003:621; O'Sullivan *et al.*, 2006:E710). Angles for the hips were measured at approximately 90° - 95° flexion, while the knees were flexed at 40° - 45° , using a goniometer. These measured angles at the knees and hips determined the length of the chain for each subject. The postural restraint apparatus (PRA) was fitted (with non-stretchable elastic Velcro straps) around the thorax, just below the inferior angles of the scapulae (dorsal view) and just below the mamilla line (frontal view).

This postural restraint position needs to be maintained (strap position not to shift on body) to maximise the L5-S1 extension moment in the test. The PRA was attached to the PAB, while the PAB was again connected with a light steel chain to a steel pipe for both feet to push against. By pulling away with the PRA and simultaneously pushing with both feet against the steel pipe, a pull-compression force is diverted through the two rigid capsules on the PAB ball. A display unit (PC screen) in front of the subject continuously shows the air pressure (real-time) in mb visual feedback (see Figure 4.2).

4.4.2 Pressure air biofeedback device calibration

Comparisons of strength values between pneumatic manometers and isokinetic or isometric dynamometers have been difficult to compare. Pneumatic manometers do not measure the same parameter with the same units of measurement as an isokinetic or isometric dynamometer and thus comparisons between these instruments have been problematic (Solgaard *et al.*, 1984:569).

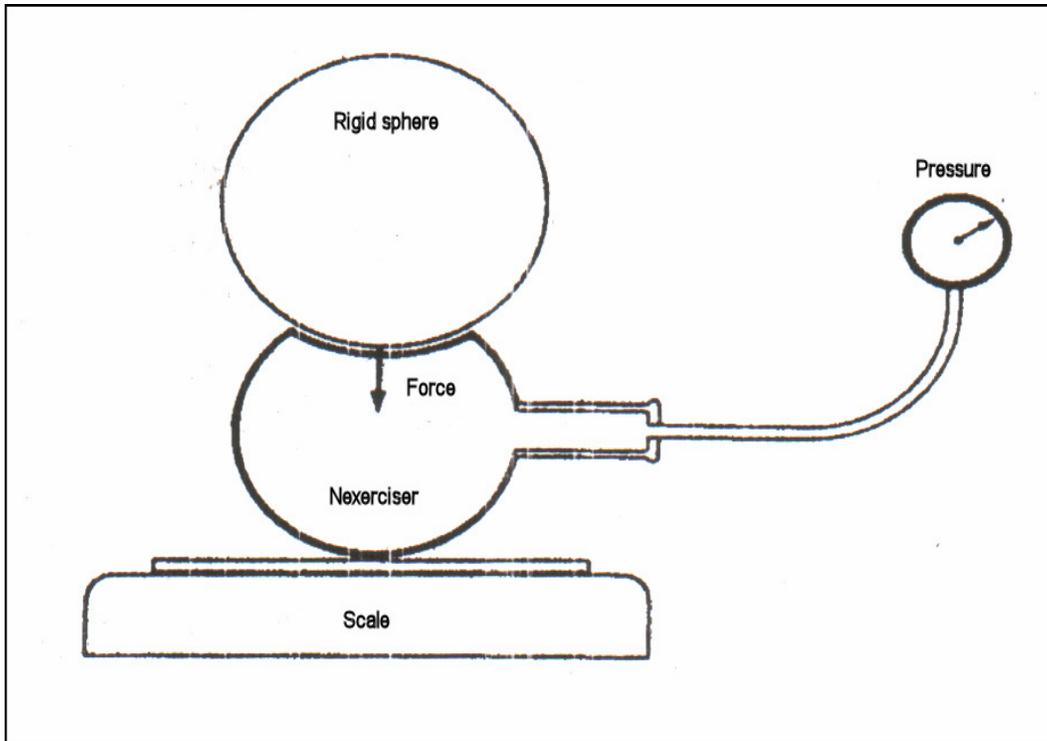


Figure 4.3 Schema of technique used to determine pressure-force characteristic of a ball device (adapted from Axen *et al.*, 1992:7).

However, this technical problem was solved in the studies of Helewa *et al.* (1981:356) and Axen *et al.* (1992:2) by using incremental calibrated weights in kilograms (kg) to compress the sphygmomanometer bag and/or elastic ball, as used in their studies (Figure 4.3). This was done to determine if a linear relationship existed between the applied external force (kg) and internal pressure (mb) of the ball and/or modified sphygmomanometer.

In the case of the PAB device, a different calibration protocol was followed compared to the protocols of Helewa *et al.* (1981:356) and Axen *et al.* (1992:2). Firstly, the applied calibrated weights to the PAB ball eventually totalled 160 kg compared to much smaller total weights applied in the other studies. Secondly, the area of apposition between the applied external forces and the PAB ball was exactly the same in the calibration test as in the seated back extension test done with the research subjects. Thirdly, in quantifying the relationship between an external force and internal pressure of the PAB ball, the values of mb

pressure were calibrated with respect to kilogram weights by simulating the pull-compression action on the PAB ball.

This meant that calibrated weights (accurate within one (1) gram as weighed on a calibrated physician's scale) were hanged from the PAB ball attachments by an inflexible rope, as suggested by Coetzee (2009) which compressed the PAB ball from both sides by the rigid fibreglass capsules surrounding the inflated elastic ball (Figure 4.4).

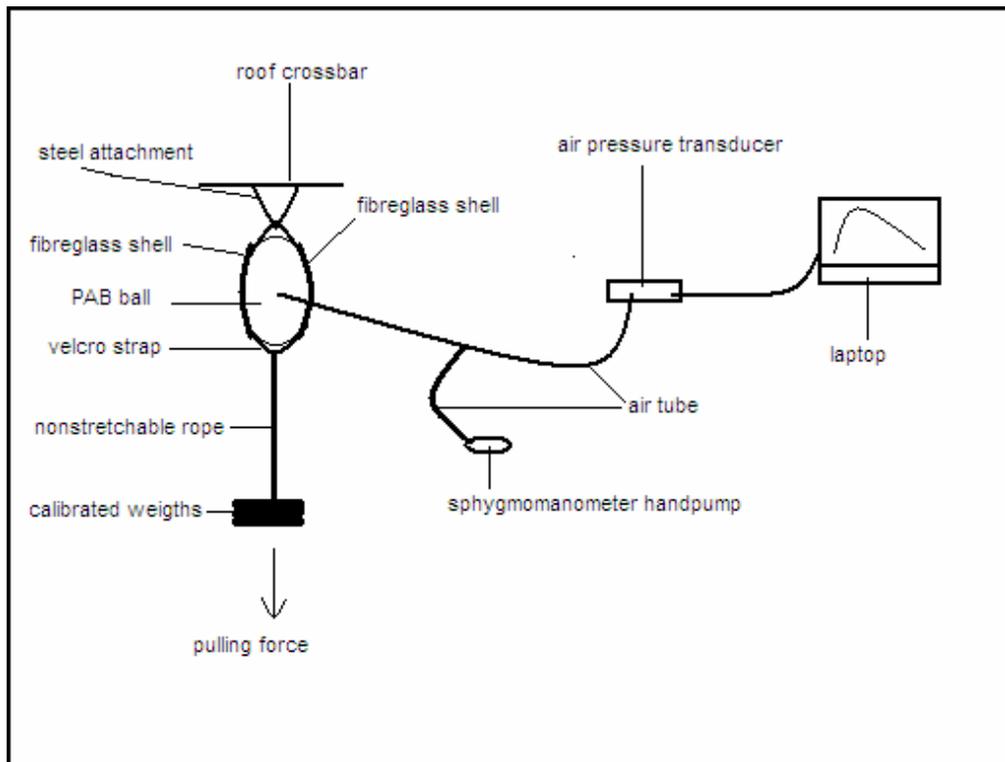


Figure 4.4: Schema of calibration technique used to determine the pull-compression force characteristic of the PAB device.

The calibration procedure was done in increments of 2.5 kg weights. The relationship between the external force applied by the subject through the fibreglass shells to the elastic PAB ball and the internal pressure developed within the system is determined by the area of the apposition between the rigid shell capsules and PAB ball. Because accurate methods of measuring apposition area are lacking, the PAB device was designed with fibreglass shell

capsules to specifically solve this shortcoming. PAB ball pressure measurements obtained in mb for applied external forces of 2.5 kg increments between 2.5 kg and 160 kg were then plotted against the corresponding forces applied.

4.4.3 Calibration data of the pressure air biofeedback device versus calibrated weights

For the purpose of the present study, real-time air pressure data is reflected on a personal computer, which is connected in parallel with a Druck PMP 1400 pressure transducer (see Appendix F) to obtain permanent accurate pressure recordings. Validity of the PAB device was tested by comparing two sets of pressure measurements (obtained while applying known forces of 2.5 kg weight increments between 2.5 kg and 160 kg) collected on two different days. Results of PAB force values (mb) versus calibrated weights (kg) tested over these two days are shown in Figures 4.5 - 4.7. Pearson correlation coefficients (r) were calculated in order to examine the correlation between PAB force (mb) and an applied external force (calibrated weights in kg). An alpha level of 0.05 was selected for statistical significance. Also, the intraclass correlation coefficient (ICC) agreement result of the PAB versus PAB force values of the two calibration tests was calculated (figure 4.8).

According to Figures 4.5 and 4.6, a significant linear relationship emerged between air pressure output (PAB force in mb) and the whole range of applied external forces (calibrated kg weights) in the two calibration tests done on day one ($r=0.995$, $p<0.01$) and day two ($r=0.998$, $p<0.01$). Also, Figure 4.7 showed a similar significant correlation ($r=0.997$, $p<0.01$) between average PAB force data (mb), calculated over the two days in relation to calibrated weights. Therefore, the results of PAB force (mb) and applied external force comparisons (calibrated weights in kg) demonstrated high agreement or validity between measures (calibrated weights in kg) and the associated criterion (PAB force in mb).

Furthermore, the ICC calculation of 0.997 (SEM=1.55) between the two sets of PAB force calibration measurements between day one and two (Figure 4.8), indicated a significant correlation and therefore excellent reliability of the PAB device.

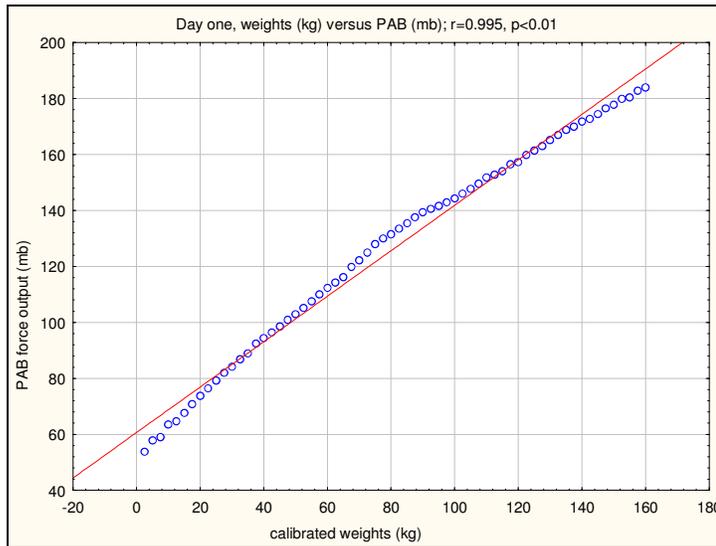


Figure 4.5 A significant correlation or linear relationship between PAB (mb) and calibrated weights (kg) was found on day one.

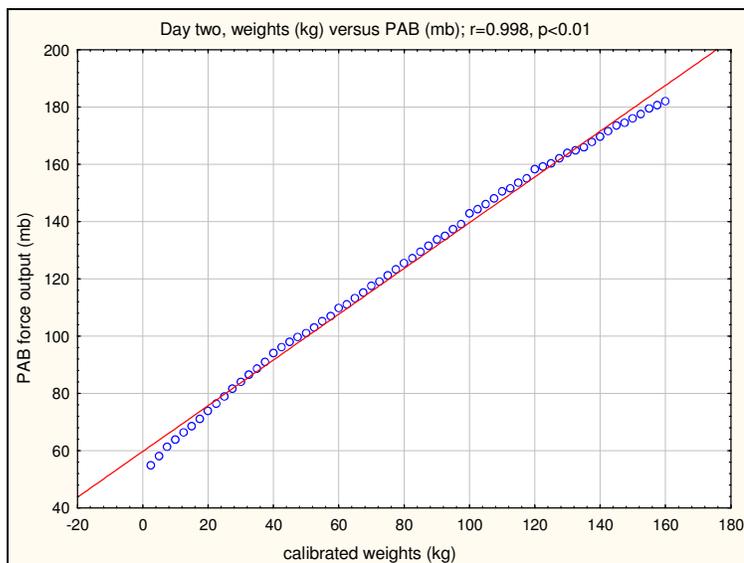


Figure 4.6 Day two again showed a significant correlation or linear relationship between PAB (mb) and calibrated weights (kg).

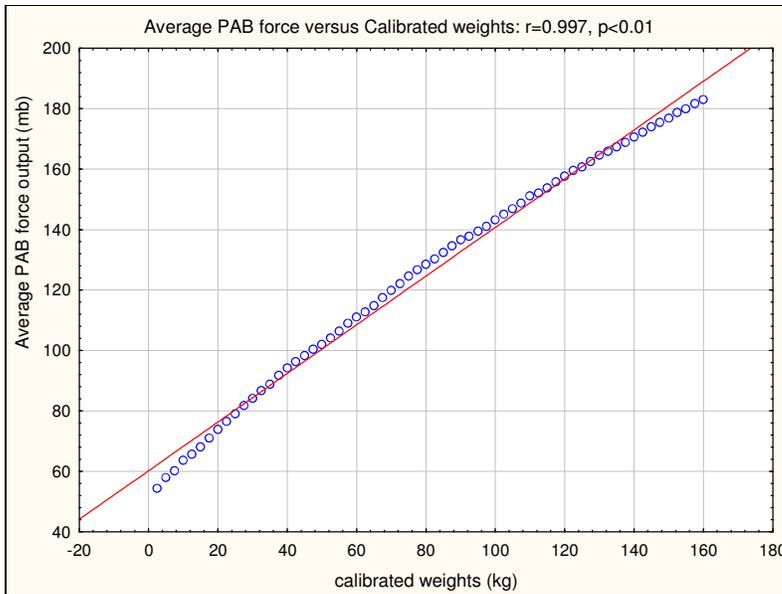


Figure 4.7 A significant correlation or linear relationship between the average PAB (mb) values vs calibrated weights, calculated over the two days, was indicated.

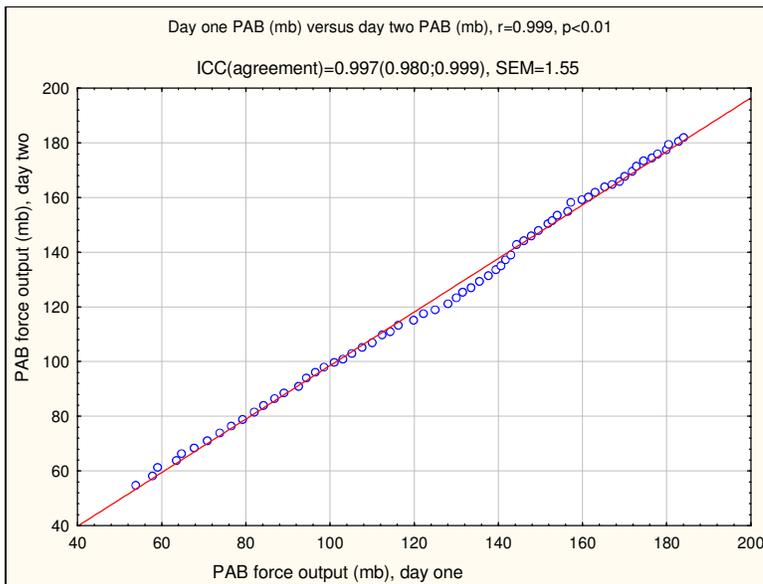


Figure 4.8 A significant correlation or linear relationship between PAB vs PAB (mb) values on two different days was indicated.

The implication of this linear force pressure characteristic is that any given increase in mb pressure signifies the same increase in kg external force applied. Therefore, under these experimental and calibrated conditions, a measured increase in maximum or submaximal PAB force values signifies a proportional increase in peak or submaximal external force (kg) or corresponding muscle strength.

Also, on the earth's surface a mass of 1kg exerts a down force of 9.81 Newton (N) or 1 kg force (kgf) (Newton, 2009). Because the approximation of 1 kg corresponding to 9.81 N is used as a rule in everyday life and in engineering, as well as the linear relationship that exists between the calibrated weights (kg) and PAB force (mb), the kg unit has also been converted [Onlineconversion, 2009a; Onlineconversion, 2009b] to the SI metric unit of force, the N. Therefore, in terms of the SI metric conversion, the air pressure output of the PAB may also be expressed as PAB force in kgf or N (Appendix E), especially when muscle strength is expressed.

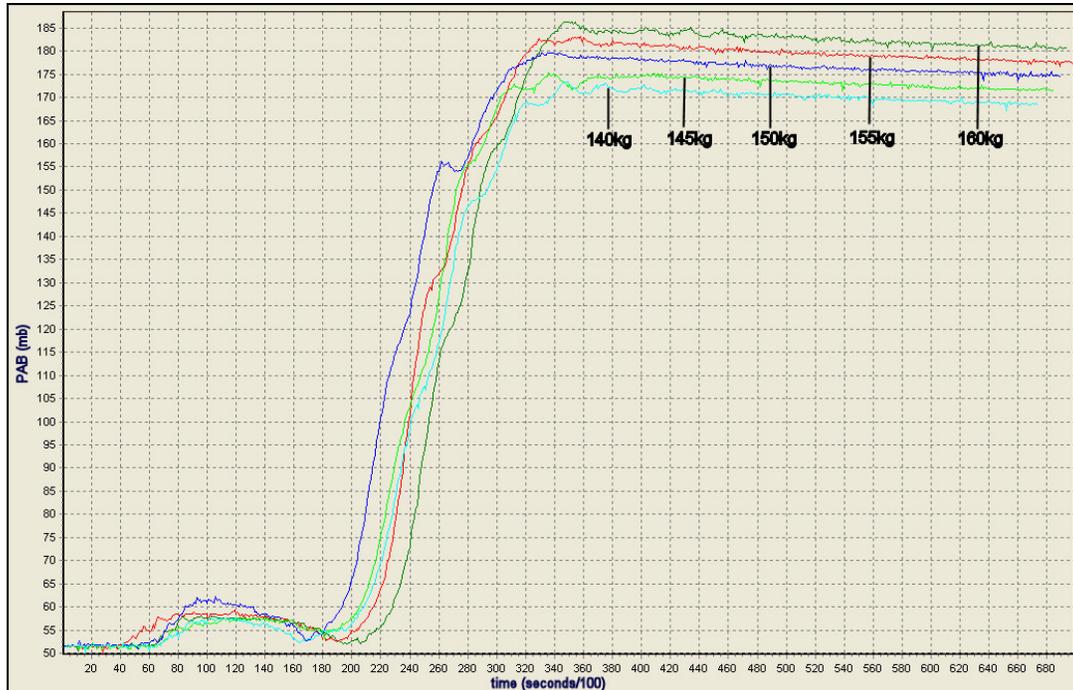


Figure 4.9 Day one, an example of a real time PAB force graph (mb) for five different calibrated weight loads between 140kg and 160kg.

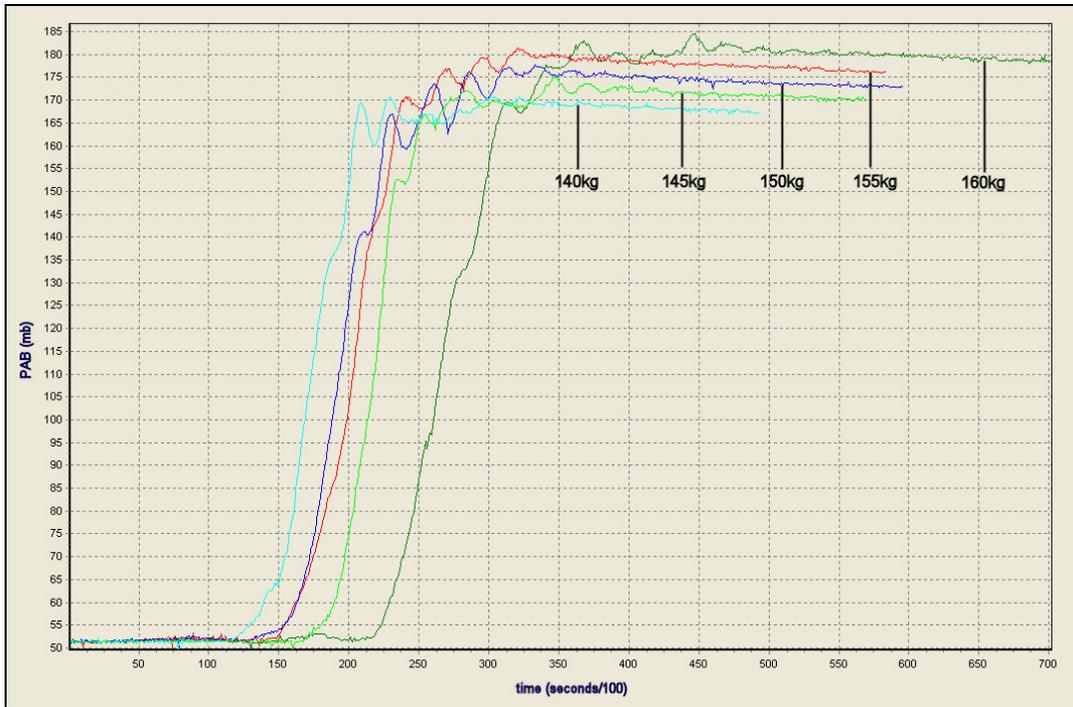


Figure 4.10 Day two, an example of a real time PAB force graph (mb) for the same five calibrated weight loads between 140kg and 160kg.

With further reference to the calibration procedure, Figures 4.9 and 4.10 illustrate how PAB pressure data was collected from a real time pressure data graph in response to incremental calibrated weights. Calibrated PAB force data was recorded one second after the peak pressure value for each loaded weight had been reached. This calibration procedure was strictly followed on both days for each incremental weight load of 2.5kg between 2.5kg-160kg.

It has been acknowledged that the PAB device is an inflatable, elastic ball, which is not definitively calibrated because of its elasticity. This has been controlled as far as possible with two rigid hemispheric shells on opposite outer surface segments of the PAB ball.

4.5 SUMMARY

This chapter highlighted the growing need for portable and valid muscle testing devices in a clinical office setting with reference to the emerging rapid testing

device market. However, this growing need may turn to an “essential must” if one looks at the global expansion of the growing orthopaedic and especially, the spinal market. It may be more than ever necessary to develop valid, portable and “easy to use” technology and testing protocols for the strength assessment of low back pain. Furthermore, this chapter also highlighted the technology development and valid assessment of the portable PAB device for testing low back muscle strength. It was found that the PAB device is a valid and reliable device in relation to the kg force exerted by calibrated weights. Therefore, it was decided to test the PAB device against electromyography and ultrasound instruments during the assessment of a seated low back extension test in asymptomatic and low back pain subjects in a clinical office setting.

CHAPTER FIVE

METHODOLOGY

- 5.1 INTRODUCTION
- 5.2 OBJECTIVE
- 5.3 RESEARCH DESIGN
- 5.4 SUBJECTS
 - 5.4.1 Exclusion criteria for asymptomatic subjects
 - 5.4.2 Exclusion criteria for low back pain subjects
- 5.5 SECTION ONE: THE OSWESTRY DISABILITY QUESTIONNAIRE
 - 5.5.1 Development
 - 5.5.2 Modifications and translations
 - 5.5.3 Construct validity
 - 5.5.4 Internal consistency
 - 5.5.5 Reproducibility
 - 5.5.6 Scoring the Oswestry Disability Index
- 5.6 SECTION TWO: MORPHOLOGICAL MEASUREMENTS
 - 5.6.1 The measurement protocol
 - 5.6.2 Age
 - 5.6.3 Anthropometrical measurements
 - 5.6.3.1 Body mass
 - 5.6.3.2 Standing body height
 - 5.6.3.3 Body mass index
 - 5.6.3.4 Waist-to-hip ratio
 - 5.6.3.5 Standing lumbar posture
- 5.7 SECTION THREE: PRESSURE AIR BIOFEEDBACK TESTING
PROTOCOL – APPLICATION OF SCIENTIFIC EVIDENCE
 - 5.7.1 Parallelograms of force vectors in the pressure air biofeedback and Biering-Sorensen tests
- 5.8 SECTION FOUR: GENERAL SET-UP AND TESTING PROCEDURE
 - 5.8.1 Surface electromyography set-up procedure
 - 5.8.2 The pressure air biofeedback and electromyography trial
 - 5.8.3 The pressure air biofeedback and ultrasound trial

5.1 INTRODUCTION

In clinical practice, there is an increasing need for the objective and accurate assessment of muscle function or dysfunction (Hyde *et al.*, 1983:420; Giles, 1984:36; Helewa *et al.*, 1986:1044; Cairns *et al.*, 2000:127; Danneels *et al.*, 2001:146). The purpose of scientific testing in this study is to assess the reliability and validity of the PAB device in measuring the *m. lumbar multifidus* contraction in upright seated, isometric back extension, thereby developing a quantifiable functional test for the strength contraction of the *m. lumbar multifidus*. In the evaluation of experimental subjects, Matheson *et al.* (1993:66) describe five components that are important to adhere to when testing subjects. The PAB testing protocol specifically attempted to address these five components. These components were also highlighted in the research of Helewa *et al.* (1981:353) and Helewa *et al.* (1986:1044):

- Safety: the evaluation should be completed without any risk of injury to the subject.
- Reliability: the apparatus as well as the examiner must be able to repeat the test with consistent results.
- Validity: each test needs to be specific in what it is testing and the results must be able to exactly measure that.
- Practical: is it cost-effective in terms of time efficiency?
- Versatility: how applicable are the results of the evaluation?

Furthermore, the clinical evaluation instruments namely, the Toshiba 6000C Ultrasound (US) imaging scanner and the Noraxon Myotrace 400 electromyography (EMG) device used in this study, were tested against the PAB device. Therefore the methods and testing procedures used in this study were strictly executed. This was done, to be able to scientifically compare the PAB device to the EMG device and the US scanner.

5.2 OBJECTIVE

The aim of this study was to assess the reliability and validity of the newly developed PAB device (Patent number, P42817ZP00 MR/mjm) in assessing low back strength with specific reference to *m. lumbar multifidus* during an upright seated, closed chain isometric back extension test.

5.3 RESEARCH DESIGN

A prospective validation study was conducted by means of an experimental research design. An occasional random test was done in the selection of low back pain and asymptomatic subjects and they were assessed by a test-retest method (Thomas & Nelson, 1985:227, 263; Kidd, 2009).

5.4 SUBJECTS

In this study, subjects participated of their own free will and were randomly selected from asymptomatic persons as well as low back pain persons. Twenty five (25) asymptomatic subjects (13 females and 12 males) as well as 18 low back pain subjects (9 males and 9 females) between the ages of 18-80 years, were recruited from the clientele of a health club, as well as a biokinetic practice situated in the city of Pietermaritzburg, KwaZulu-Natal. Furthermore, with respect to the low back pain group (n=18), seven subjects had low back surgery with instrumentation, one patient was diagnosed with tuberculosis at the L3-4 level, one patient had low back pain due to cancer at the L4 level (diagnosed only after testing had been completed), one patient was diagnosed with a unilateral pars defect at L5, one patient was diagnosed with a grade 1-2 anterolithesis, one patient was diagnosed with multi-level degenerative disc disease while six patients were diagnosed with mechanical low back pain. All 43 subjects gave their written, voluntary consent (Appendix A). Also, for the evaluation device to be used in clinical practice and to be clinically relevant for its purpose, Weaver *et al.* (2001:5) and Haynes and Edmondston (2002:585) suggested that symptomatic individuals be included in validation studies of newly developed medical instruments. Therefore, it was decided to include

asymptomatic, as well as symptomatic individuals to assess the clinical usefulness of the PAB device.

The study was approved by the Stellenbosch University Ethics Committee. Inclusion criteria for both the asymptomatic and low back pain subjects for the study were willingness to participate in the study and the ability to correctly maintain a neutral lumbar lordosis during an upright seated, closed chain, back extension test. However, exclusion criteria were different for each group and will be described in brief.

5.4.1 Exclusion criteria for asymptomatic subjects

- If the subject had episodes of low back pain during the six months before testing started.
- If the subject had abdominal or spinal surgery.
- If the subject was pregnant – in the case of female subjects.
- If the subject was menstruating on the testing day – female subjects.
- If a subject presented with marked spinal deformity, e.g. scoliosis.
- If the subject had a neuromuscular disorder.
- General poor health, that would impair compliance of assessment.
- If the subject had a history of alcohol or substance abuse (which may affect compliance with the research protocol).
- An inability to perform the test in a controlled way, as well as to maintain the correct neutral spine posture during the test.

5.4.2 Exclusion criteria for low back pain subjects

- If the subject was pregnant – in the case of female subjects.
- If the subject was menstruating on the testing day - female subjects.
- If a subject presented with marked spinal deformity, e.g. scoliosis.
- General poor health, apart from low back pain, that would impair compliance or assessment.
- If the subject had a history of alcohol or substance abuse (which may affect compliance).
- Clinical signs of any known neurological disorder or muscular degenerative conditions, such as muscular dystrophy.
- Having received a spinal epidural or having undergone abdominal or spinal surgery within three months prior to testing day.
- An inability to perform the test in a controlled way as well as to maintain the correct neutral spine posture during the test.

Exclusion criteria were chosen in order to avoid factors that might influence the ability to correctly contract the *m. lumbar multifidus*. This was specifically done to stay within the framework of a randomised controlled trial, which is the scientific standard for conducting proper research (Waddell, 1998:263). The evaluation protocol consisted of four sections. The first section is the completion of the Oswestry Disability Index (ODI) (Appendix B) by either a symptomatic or asymptomatic subject. The second section is the morphological assessment of each individual. The third section describes the application of scientific evidence to the pressure air biofeedback testing protocol. The fourth section comprises of the general set up and testing procedure. A detailed description of the four sections now follows:

5.5 SECTION ONE: THE OSWESTRY DISABILITY QUESTIONNAIRE

The ODI is a condition-specific measuring instrument to assess a low back pain patient's status and progress in routine clinical practice. Evidence of the ODI's validity and reliability is also described by Roland and Fairbank (2000:3115). This is the reason for using the questionnaire in this study. When used, as reproduced in Appendix B, no permission is required from the authors (Roland & Fairbank, 2000:3115) to use the ODI in this study.

5.5.1 Development

The development of the ODI was initiated by John O'Brien in 1976 when large numbers of chronic low back pain patients were seen at a specialist referral clinic. Low back pain patients were interviewed by an orthopaedic surgeon, an occupational therapist and a physiotherapist to identify the disturbance of activities of daily living through chronic back pain. After various drafts were tried, the first version (Version 1.0) of the index was published in 1980 and widely disseminated after the 1981 meeting of the International Society for the Study of the Lumbar Spine (ISSLS) in Paris (Roland & Fairbank, 2000:3117).

5.5.2 Modifications and translations

In a study by a Medical Research Council (MRC) group in the United Kingdom, the ODI was validated and improved to Version 2.0 and has been recommended for general use (Baker *et al.*, 1989:174; Meade *et al.*, 1995:350). The ODI has stood the test of time and has been used in a wide variety of clinical situations in the United Kingdom, the USA and various other countries. The ODI has been translated into at least nine languages and therefore permits comparison between studies performed in numerous countries. The questionnaire can be completed in less than five minutes and scored in less than one minute (Roland & Fairbank, 2000:3117, 3119).

5.5.3 Construct validity

The wording of the ODI was designed on the basis of a patient's self-report and symptoms of chronic low back pain. Concerning pain measures, Gronblad *et al.* (1993:194) indicated that the ODI shows moderate correlation with a visual analog scale ($n=94$, $r=0.62$). The ODI has also been used to validate the Pain Disability Index, the Low Back Outcome Score, the Manniche Scale, the Aberdeen Score, a new German language scale, the Curtin Scale and a functional capacity evaluation (Roland & Fairbank, 2000:3118). Also, two different mechanical methods of lumbar spine assessment for return-to-work status were found to be less effective in successfully predicting this than the ODI (Nordin *et al.*, 1997:25; Loisel *et al.*, 1998:1592). Further to this, the ODI predicted isokinetic performance (Kaivanto *et al.*, 1995:97), isometric endurance (Kuukkaken & Malkia, 1996:119) and performance with sitting and standing (but not with lifting) in a study involving secret observations (Fisher & Johnson, 1997:78). Physical tests also correlated with the ODI (Gronblad *et al.*, 1994:201).

5.5.4 Internal consistency

Using Version 2.0, Fisher and Johnson (1997:78) found Cronbach's α to be 0.76, while Kopec *et al.* (1996:159) found 0.87. All these investigations indicated an acceptable degree of internal consistency.

5.5.5 Reproducibility

In an original study by Fairbank *et al.* (1980:272), patients with chronic low back pain, who were tested twice at a 24-hour interval, showed a high correlation ($n=22$, $r=0.99$). This may include a memory effect. If the test-retest interval is extended to four days, the correlation of scores decreases to $n=22$, $r=0.91$ (Kopec *et al.*, 1996:159) and if retested after a week, the correlation decreases to $n=22$, $r=0.83$ (Gronblad *et al.*, 1993:194). The disadvantage of increasing the time interval is that natural symptoms may fluctuate, which can influence the strength of the results.

5.5.6 Scoring the Oswestry Disability Index

The ODI consists of 10 sections, with six statements per section (see Appendix B). For each section of six statements the total score is five; if the first statement is marked, the score is zero; if the last statement is marked, it is five. Intervening statements are scored according to rank. If more than one box is marked in each section, the highest score is taken. If all 10 sections are completed the score is calculated as follows: $\frac{16 \text{ (total scored)}}{50 \text{ (total possible score)}} \times 100 = 32\%$. If one section is missed (or not applicable) the score is calculated as follows: $\frac{16 \text{ (total scored)}}{45 \text{ (total possible score)}} \times 100 = 35.5\%$. Therefore, the final score may be summarised as: total score (5 x number of questions answered) x 100%. Roland and Fairbank (2000:3123) also suggest rounding the percentage to a whole number for convenience. According to Sung (2003:1314) interpretation of disability scores is as follows: 0%-20%, minimal disability; 20%-40%, moderate disability; 40%-60%, severe disability; 60%-80%, crippled; 80%-100%, bed bound or the patient is greatly exaggerating his/her symptoms. However, a limitation of this classification system is that certain percentage scores may indicate two disability classifications e.g. if a subject scores 40%, he/she can be classified as moderate or severe disability. The scores should have been classified as 20%-39% and 40%-59%, so that the end of one disability count does not overlap the start of the following disability count. This will avoid confusion, regarding classification of border cases.

5.6 SECTION TWO: MORPHOLOGICAL MEASUREMENTS

5.6.1 The measurement protocol

Anthropometry, as a science, depends upon adherence to the specific rules of measurement as determined by national and international standards bodies, such as the International Society for the Advancement of Kinanthropometry (ISAK) (Ross & Marfell-Jones, 1991:223; De Ridder, 1999:24; Marfell-Jones *et al.*, 2006). The measurement protocol (Appendix C) in this study was followed as prescribed by ISAK.

5.6.2 Age

The subject's age was determined from date of birth until date of testing.

5.6.3 Anthropometrical measurements

5.6.3.1 Body mass

Equipment: A calibrated Healthometer beam balance scale.

Method: Body weight, representing body mass, is a critical measure. The body mass was accurately measured to 0.1 kg. Each 100 gram was measured accurately to a maximum of 160 kg (maximum weighing capacity of scale). Before weighing, the subjects emptied their bladders and took off unnecessary clothes and shoes. The subject stood on the centre of the scale without support and with the subject's weight distributed evenly on both feet. The subject had to stand still, eyes looking ahead and with arms relaxed next to the body (Bekker, 1996:219; De Ridder, 1999:34). In quantifying the relationship between an external force and the measured scale output, the scale was calibrated with respect to kilogram weights by placing calibrated weights (accurate within 10 grams) on top of the centre of the scale.

5.6.3.2 Standing body height

Equipment: A Healthometer stadiometer.

Method: There are three general techniques for measuring stature: free standing, stretch and recumbent. The stretch stature method was used in this study and required the subject to stand with the feet together and the heels, buttocks and upper part of the back touching the vertical part of the scale. The head when placed in the Frankfort plane need not be touching the vertical part of the scale. The Frankfort plane was achieved when the orbital (lower edge of the eye socket) was in the same horizontal plane as the tragion (the notch superior to the tragus of the ear). When these two anatomical landmarks were

aligned, the vertex was the highest point on the skull. The maximum body height was measured from the soles of the feet to the vertex of the skull. The subject was instructed to take and hold a deep breath while keeping the head in the Frankfort plane. The examiner placed the head plate firmly down on the vertex, crushing the hair as much as possible. Measurement was taken at the end of a deep inward breath and to the last millimetre (Marfell-Jones *et al.*, 2006:85).

5.6.3.3 Body mass index

The BMI was used to assess body mass relative to body height and was calculated by dividing body mass into kilograms by standing height into metres square (kg/m^2). The American College of Sports Medicine (ACSM) classified a BMI of under 18.5 kg/m^2 as underweight, 18.5 to 24.9 kg/m^2 as normal, 25.0 to 29.9 kg/m^2 as overweight and a BMI of equal or greater than 30.0 kg/m^2 as obese (ACSM, 2006:58).

5.6.3.4 Waist-to-hip ratio

The distribution of body fat is recognised as an important predictor of health risk (e.g. hypertension, coronary artery disease) in humans. More specifically, the waist-to-hip ratio (WHR) is the circumference of the waist divided by the circumference of the buttocks and hips and has been used as a simple method for determining body fat distribution (health risk). The subject stood in a relaxed position with stomach exposed and dressed in underclothes. A calibrated cloth tape measure was used to measure each site and it also reduced skin compression and improved consistency of measurement (ACSM, 2006:59; Marfell-Jones *et al.*, 2006:87). To measure the waist, the subject stands in front of the researcher, allowing the tape to be passed around the narrow part of the abdomen. The base of the tape is held in the right hand, while the researcher uses the left hand to adjust the level of the tape at the narrowest point. Using the cross-hand technique, the researcher positions the tape in front at the target level. The subject should breathe normally and the measurement is taken at the end of a normal expiration (Marfell-Jones *et al.*, 2006:87). To measure the hips,

the researcher passes the tape around the hips from the side. The base of the tape is held in the right hand, while the researcher uses the left hand to adjust the level of the tape at the greatest posterior protuberance of the buttocks. Using the cross-hand technique, the researcher positions the tape at the side in a horizontal plane and at the target level, before taking the measurement (Marfell-Jones *et al.*, 2006:88).

5.6.3.5 Standing lumbar posture

Saunders and Ryan made the following remark on the importance of using an inclinometer in assessing posture angles and range of motion: “The American Medical Association’s Guides to the evaluation of permanent impairment state that standard goniometric techniques for measuring spinal movement can be highly inaccurate and that measurement techniques using inclinometers are necessary to obtain reliable spinal mobility measurements.” (Saunders & Ryan, 2004:46). Standing lumbar posture was evaluated with the subject in an upright standing position (Figure 5.1), viewing the subject from the side (lateral view) (Saunders, 1998:5, 7).

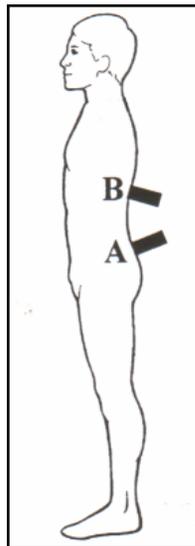


Figure 5.1 Posture assessment with the inclinometer. (A) represents the lumbosacral angle (L-S joint) and (B) represents the thoracolumbar angle (T-L joint) (adapted from Saunders, 1998:7).

The standing postural assessment only measured two angles in this study, that is the lumbosacral angle (L-S joint) and the thoracolumbar angle (T-L joint) and was taken with the certified Saunders digital inclinometer. With the subject standing erect, the sacral midpoint was located and marked to measure the lumbosacral angle. According to Saunders (1998:5), the sacral midpoint is on a line midway between the inferior aspects of the posterior superior iliac spines (PSIS). The L-S joint is approximately three centimetres superior to this line and can usually be palpated as the segmental space that is relatively easy to locate and has very limited movement. It is also helpful to have the subject bend forward a little to find the L-S joint and to palpate the spinous processes easier.

To measure the thoracolumbar angle, the T12-L1 interspace was located and marked (T-L joint). According to Saunders (1998:5), the T-L joint can be found by starting with the L-S joint as number one and count up six interspinous spaces to find this specific joint. The inclinometer was verified by reading 0° when the long base was horizontal. The inclinometer was placed at the L-S joint and recorded the reading as the lumbosacral angle. Keeping the inclinometer at the L-S joint, it was then zeroed. The inclinometer was then placed at the T-L joint and recorded the reading as the thoracolumbar angle. The normal values for the lumbosacral angle are between 15° to 30° and for the thoracolumbar angle, between 30° to 40° lordosis (Saunders, 1998:5).

5.7 SECTION THREE: PRESSURE AIR BIOFEEDBACK TESTING PROTOCOL – APPLICATION OF SCIENTIFIC EVIDENCE

In developing a new low back assessment test and device, it was important to incorporate scientific principles in the design of such a device and test. Therefore, the methodology of the PAB testing protocol was developed according to scientific biomechanical and physiological evidence, with specific reference to the lumbar spine. Mentioned below are a few scientific findings of various researchers, that have been regarded and/or calculated in the set-up design of the PAB testing protocol. The PAB testing protocol is described in short, after each mentioned scientific finding:

- EMG activity and back extension torque have shown a linear relationship in most sagittal spine flexion/extension angles, except in greater trunk flexion angles where back extensor muscles show myoelectric silence (Konrad, 2005:49). This “flexion-relaxation” syndrome can be further explained in that the lumbar extensors undergo eccentric contraction as the spine approaches full flexion, while the passive tissues take over moment production, relieving the extensor muscles of this role and accounting for their myoelectric silence (Hides, 2005:69). Thus, in lumbar hyperflexion the flexion-relaxation response becomes evident at the L4-S1 level (Shin *et al.*, 2004:485; Konrad, 2005:46; McGill, 2007:76). This syndrome may accelerate disc herniation which is a function of repeated flexion motion of the lumbar spine (McGill, 2007:46).

According to the PAB testing protocol, no sagittal flexion of the lumbar spine is allowed because of the flexion-relaxation response. Therefore, an isometric back extension test is done with surface EMG electrodes at the fifth lumbar level (L5) (see Paragraph 5.8.1 for a detailed discussion). The subject is set-up in a seated, upright (trunk angle of 0°) and neutral spine posture and should maintain this specific posture, while executing the PAB test.

- McGill (1997b:470) and McGill (2007:102, 142) reported that back extension in a neutral spine posture when lifting a dead weight is biomechanically sound. This biomechanical “test posture” reduces high shearing forces, avoids interspinous ligament strain and aligns fibres to support the shear forces. Keifer *et al.* (1997:45) and O’Sullivan *et al.* (2006:E711) also stressed the importance of the lumbar neutral or lumbopelvic posture in standing or sitting to reduce spinal compressive load.

According to the PAB testing protocol, a seated back extension test is done in neutral lordosis, or lumbopelvic posture, at a trunk angle of 0° and with hips flexed to approximately 90° and knees flexed to approximately 45°. This is to minimize the shear and compressive forces

on the lumbar spine as found in the above-mentioned studies.

- O'Sullivan *et al.* (1997:2964) and Hides (2005:68-70) found that the *m. lumbar multifidus* is active in the upright posture (reflects its tonic postural role), it contributes significantly to controlling of the lumbar lordosis, as well as controlling stability of the lumbar segment (Panjabi *et al.*, 1989:194; Kay, 2001:33). In a recent study, it was found that the lumbopelvic upright sitting position, compared to slump sitting, was associated with significantly greater muscle activity of superficial *m. lumbar multifidus* in controlling lumbar posture and stability (O'Sullivan *et al.*, 2006:E711).

According to the PAB testing protocol, a lumbopelvic upright sitting posture used in the seated, isometric back extension test may lead to greater muscle activity of *m. lumbar multifidus* as found in the above-mentioned research findings. Therefore, the PAB testing protocol may place the lower lumbar spine at a lower risk, because of the stability contraction of *m. lumbar multifidus* in this position.

- Clinically, significant muscle atrophy is commonly observed in quadriceps depth (CSA), mainly the one-joint, anti-gravity *m. vastus medialis*, with a greater amount of type I fibres that are most vulnerable to atrophy during immobilization (Jorgensen *et al.*, 1993:1439 and Hides, 2005:124-125, 153). It may explain the finding of atrophy of specifically the L5 *m. lumbar multifidus*, caused by reflex inhibition, which has similar muscular characteristics. However, Zhao *et al.* (2000:2197), Yoshihara *et al.* (2001:625) and Yoshihara *et al.* (2003:494) found significant decreases in both type I and type II fibres of the L5 lumbar multifidus muscle. Also, anti-gravity extensors, such as *m. vastus medialis* and *m. lumbar multifidus* do undergo more atrophy than flexors in reflex inhibition.

According to the PAB testing protocol, the *m. lumbar multifidus* at the L5 vertebral level is specifically tested as an anti-gravity extensor muscle.

Clinically, it is the lumbar muscle most affected from the lumbar paraspinal muscle group, the reason for assessing at the L5 level.

- Kay (2000:104) reported that the *m. lumbar multifidus*' anatomical position is the most medial of the major paraspinal muscles and the largest to directly cross the lumbosacral junction, and therefore back extension tests have been specifically focussing on testing the L5/S1 extension moment (Gallagher, 1997:1865; Lariviere *et al.*, 2003:307; Shin *et al.*, 2004:488; Granata *et al.*, 2005:1031).

According to the PAB testing protocol, the upright seated, closed chain back extension test may transfer the pulling force vector specifically through the lower lumbar levels, which may lead to greater EMG activation of *m. lumbar multifidus*. Also, *m. lumbar multifidus* increases in muscle size caudally from L1 to S1, and therefore the biggest cross sectional area (CSA) of *m. lumbar multifidus* at vertebral level five (L5) is tested.

- The studies of Bogduk *et al.* (1992:897), Choi *et al.* (2005:768) and Lee *et al.* (2006:2258), found that maximum force, exerted by a muscle, is proportional to its size, including CSA. Therefore, US imaging is more extensively used in the reliable assessment of *m. lumbar multifidus* thickness and CSA (Hides *et al.*, 1992:19; Hides *et al.*, 1995:54; Rezasoltani, 2003:35; Hides, 2005:154-155; Stokes *et al.*, 2005:125; Lee *et al.*, 2006:2261; Vasseljen *et al.*, 2006:911; Kiesel *et al.*, 2007:164). However, the assessments in these studies were done in the prone position, which does not reflect the functional and physiological contraction of *m. lumbar multifidus*. Also, these researchers measured either muscle thickness or CSA on US which reflects different opinions on measuring muscle size.

According to the PAB testing protocol, CSA of the *m. lumbar multifidus* on US imaging has been selected as the preferred choice of muscle size measurement. The subject is seated in an upright, lumbopelvic posture

during the isometric back extension test, which does reflect the functional and physiological contraction of *m. lumbar multifidus*.

- The research of Kiesel *et al.* (2007:164) revealed that a high correlation ($r=.79$, $p<.001$) exists between EMG activity and US thickness change of the *m. lumbar multifidus* in typical contractions in a prone position. However, apart from testing in a prone position, these researchers measured muscle thickness and not CSA.

According to the PAB testing protocol, the correlation between EMG activity and specifically the US imaging CSA of the *m. lumbar multifidus* is tested in relation to the PAB force measurements. However, the PAB testing protocol has been done in an upright seated, lumbopelvic posture during the isometric back extension test and not in prone, while the CSA of *m. lumbar multifidus* has been specifically measured.

- Isokinetic testing has shown disadvantages in that isokinetic movement seldom occurs in actual human performance tasks. Also, being normally an isolated joint movement (open chain loading), isokinetic testing can produce large loads on the involved joints and may even be dangerous for the healing of tissues (Kannus, 1994:S11).

According to the PAB testing protocol, a seated closed chain isometric back extension test simulates a closed chain squat posture as shown in Figure 5.2b, but without cyclic, sagittal flexion and extension of the lumbar spine. According to Richardson (2005:97), Tesch found that the shoulder loaded, squat or lunge exercise has the same muscle recruitment patterns as the closed chain leg press (Figure 5.2a). Arguably, the specific physical set-up of the PAB device may simulate an improved closed chain loaded test compared to a functional closed chain squat, because the subject is loaded through the feet (steel pipe and chain), to the subscapula attachment of the Velcro strap and not the shoulders as in the squat posture.

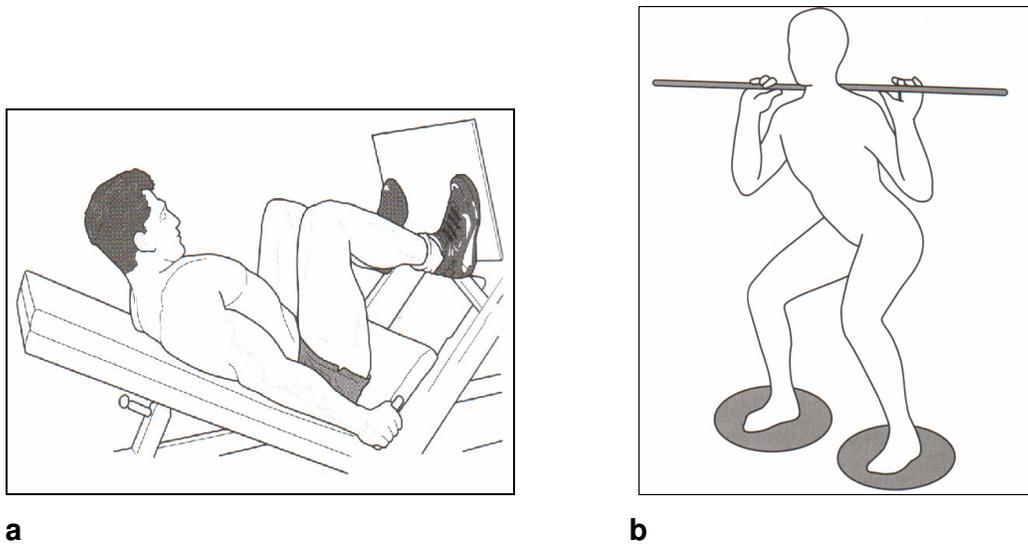
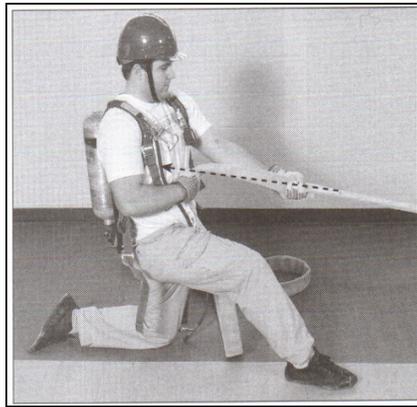


Figure 5.2 Closed chain loading in (a) a leg press exercise and (b) in a closed chain loaded posture as in squatting (adapted from Richardson, 2005:94, 96).

- Richardson (2005:96) reported that closed chain exercise loading involves proximal and distal segments moving together to compress a joint (weightbearing) and to load longitudinally through the body including through the feet as in the leg press exercise (Figure 5.2a). Little research has focused on muscle force production of the antigravity extensors in an erect loaded posture or closed chain loading (see Figure 5.2b). However, the high use of antigravity muscles in closed chain loading was reported by Richardson (2005:96-97).

According to the PAB testing protocol, a closed loop nylon strap is fixed around the torso (below the mamilla line) in an upright sitting posture and is attached to the PAB. The PAB is then further attached to the feet by a light chain and steel pipe for the feet to push against. This specifically simulates closed chain loading (Figure 5.3b). This may be seen as a low risk test for assessing low back muscle performance, while an open chain test may be classified as high risk, because it may stimulate the wrong muscle recruitment patterns that may be contraindicated for testing low back pain patients (Richardson 2005:95).

- Because the PAB-test simulates daily pulling forces (opening a door, pulling on a rope, etc) it is important to direct the pulling force vector through the lower back by pulling through the hands at the lumbopelvic level. By directing the transmissible vector through the lumbar spine (Figure 5.3a), the moment arm (and moment), the muscle forces and lumbar spine compression is reduced (McGill, 2007:139, 142). To further reduce the reaction force on L4-L5, spine posture (neutral) and whole-body posture are important variables for injury avoidance (McGill, 2007:16-17, 142).



a



b

Figure 5.3 Directing the pulling force vector through the low back in Figure a (adapted from McGill, 2007:142). Similar pulling force vector in Figure b as executed in this study.

According to the PAB testing protocol, an upright seated (neutral spine), closed chain test, with a torso-strap below the scapula and with feet pushing against a steel pipe, is done. This may direct the pulling force vector through the lower lumbar spine. Figures 5.3a and b illustrate a similar pulling force vector as McGill (2007:142) has suggested and as executed in the PAB test. Arguably, the PAB test may be well suited to the recruitment patterns of the anti-gravity muscle system that needs to be addressed first, in low back rehabilitation (Richardson 2005:102).

5.7.1 Parallelograms of force vectors in the pressure air biofeedback and Biering-Sorensen tests

With respect to the use of force vectors in developing the PAB testing protocol, it was decided to construct a force vector parallelogram in relation to the lower back, with respect to applied mechanical forces (Figure 5.4).

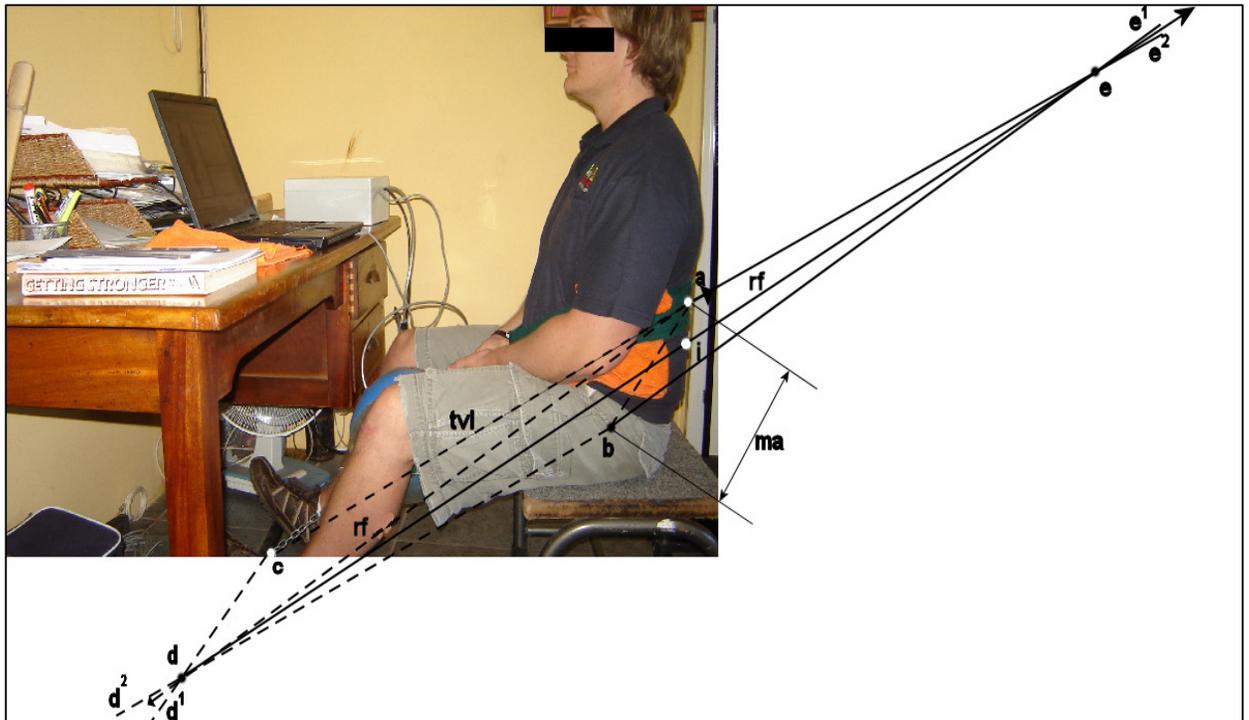


Figure 5.4 Graphical drawing of force vectors with respect to the subscapula (a), feet attachment (c) and hip joint (b) in the PAB test. Intersection (i) at L4.

This was based on the basic fundamentals of applied mechanics, as described in Gowitzke and Milner (1984:62-63) and McGill (2007:137-142). Furthermore, because the Biering-Sorensen test has become the standard of reference in published studies for assessing low back muscle performance (Demoulin *et al.*, 2006:47), parallelograms were also drawn for this test. By drawing parallelograms for each test (Figures 5.4 and 5.5), it was possible to see differences between the two tests, in terms of the resultant force vectors.

Parallelogram one (broken lines) of the PAB test (Figure 5.4) was drawn as follows:

- Line ab = vector from subscapula attachment of PAB (a) to hip (b).
- Line ac = transmissible vector (tv) from subscapula (a) to feet (c).
- Line cd¹ = drawn in parallel to vector ab.
- Line bd² = drawn in parallel to vector ac.
- Line ad = resultant force vector (rf) is projected from point a to d.

Parallelogram two (solid lines) in Figure 5.4 was drawn from the resultant force vector (ad) from parallelogram one:

- Line ad = resultant force (rf) vector from parallelogram one.
- Line db = vector from resultant force vector point (d) to hip joint (b).
- Line be¹ = drawn in parallel to the resultant force vector (ad).
- Line ae² = drawn in parallel to vector db.
- Line de = resultant force vector (rf) of parallelogram two.
- i = intersection of resultant force vector (de) at lumbar level four (L4).
- ma = moment arm from subscapula attachment of PAB (a) to hip (b).

With respect to the Biering-Sorensen test (Figure 5.5), parallelogram one (broken lines) was drawn, based on the calculation of the line of gravity by the segmental method (Gowitzke & Milner, 1984:140-143; Kriel 1999:122-126):

- Line ab = vector from *m. teres major* (a) to hip (b).
- Line ac = transmissible vector (tv) from *m. teres major* (a) to earth (c) (gravity).
- Line cd¹ = drawn in parallel to vector ab.
- Line bd² = drawn in parallel to vector ac.
- Line ad = resultant force (rf) vector is projected through point d.

Parallelogram two (solid lines) in Figure 5.5 was constructed from the resultant force vector (ad) from parallelogram one:

In summary, the intersection (i) of the resultant force vector line (de) in the PAB test, projects through the L4 level, while it runs through the T11 level in the Biering-Sorensen test. Furthermore, the moment arm (ma) from the subscapula attachment to the hip joint in the PAB test is shorter than the ma (*m. teres major* to hip) in the Biering-Sorensen test. These differences in the i and ma of the two tests indicate that the thoracic and lumbar portions of *m. longissimus* and *m. iliocostalis* are activated in the Biering-Sorensen test, imposing over 4 000 N of compression on the lumbar spine (McGill 2007:91). The closed chain neutral spine PAB test may only activate the lumbar portions of *m. longissimus* and *m. iliocostalis* while *m. lumbar multifidus* is maximally activated. This may reduce unnecessary loading on the lumbar spine.

5.8 SECTION FOUR: GENERAL SET-UP AND TESTING PROCEDURE

It is important to note that the testing-protocol was set up in such a way to assess the hypotheses that the PAB device measures the *m. lumbar multifidus* strength contraction in a closed chain loaded, upright sitting back extension test in low back pain as well as asymptomatic subjects.

Before the actual testing started, a particular set-up procedure was followed to prepare the subject for the PAB test. It is important to note that each subject had been informed and understood the testing procedure before the actual assessment started. Each subject signed informed consent forms (Appendix A). The same set-up procedure was followed for each subject, and involved completion of the ODI, the anthropometrical assessment, the EMG set-up procedure and the PAB set-up procedure.

Subjects had been randomly tested over the two days. Four isometric test trials were done over two separate days. The first trial consisted of four tests, which included one extra MVIC test for standardisation. The other three trials consisted of three tests per trial, calculating to 13 tests over two separate days (see Tables 5.1 and 5.2). Because subjects had been randomly selected, one subject's testing protocol on day one started with PAB and EMG testing and finished with PAB and US testing, while another subject started with PAB and

US and finished with PAB and EMG testing. More specifically, trial one on day one for subject A started with PAB and EMG testing, which was done at the biokinetic rooms. Trial two on day one ended with PAB and US testing, which was done at the radiology department at St. Annes' Hospital. Day two started with trial three, testing PAB and EMG (biokinetic rooms) and finished with trial four at the hospital, testing PAB and US (Table 5.1).

Another subject (B) started the same as subject A, that is, trial one on day one started with PAB versus EMG testing at the biokinetic rooms, while trial two on day one ended with PAB versus US testing, at the hospital. However, subject B started trial three, day two with PAB versus US testing done at the hospital. The last test, trial four on day two was done at the biokinetic rooms, testing PAB versus EMG (see Table 5.2). It is important to note that a five second resting test (baseline measurements for EMG and US) was done before each trial was started. The four resting tests are not included in tables 5.1. and 5.2.

Table 5.1 Example of four test trials over two separate days for subject A.

<u>Subject A</u>	
Day one:	Trial 1 = PAB + EMG (biokinetic rooms) = 4 tests ¹
	Trial 2 = PAB + US (hospital) = 3 tests
Day two:	Trial 3 = PAB + EMG (biokinetic rooms) = 3 tests
	Trial 4 = PAB + US (hospital) = 3 tests

Table 5.2 Example of four test trials over two separate days for subject B.

<u>Subject B</u>	
Day one:	Trial 1 = PAB + EMG (biokinetic rooms) = 4 tests ¹
	Trial 2 = PAB + US (hospital) = 3 tests
Day two:	Trial 3 = PAB + US (hospital) = 3 tests
	Trial 4 = PAB + EMG (biokinetic rooms) = 3 tests

¹ included one extra MVIC test for standardisation.

For the purpose of this discussion, the PAB and EMG testing protocol, trial one on day one will be explained in detail in the following two paragraphs (5.8.1 and 5.8.2)

5.8.1 Surface electromyography set-up procedure

Preparation for EMG testing started after the ODI and the anthropometrical assessments of the subject had been completed (see Section One and Two of this chapter for detailed descriptions). The subject was requested to lie prone on the examination bed while the lumbosacral area was exposed. After anatomical landmark-screening was done (by means of palpation) for the correct placement of the electrodes on the subject, the specific skin areas were properly cleaned with surgical alcohol. The spinous process of L5 was palpated and the location confirmed by palpating L4/5 interspinous space and L4 spinous process from the position of the iliac crests. The skin over the L5 spinous process was marked for reference (Kiesel *et al.*, 2008:134) with a permanent pencil.

Two pairs of disposable, self-adhesive surface electrodes (Bluetrode, GP 00-50/D) were attached bilaterally to the marked L5 spinous process on the skin overlying the muscles of interest. A reference electrode was placed on the left lateral iliac crest. The muscle sampled was the *m. lumbar multifidus* at the fifth lumbar vertebrae (L5) on the right and the left side over the greatest convexity of the specific muscle to be tested (Elfving *et al.*, 2000:118; Dederling *et al.*, 2002:172; Elfving *et al.*, 2003:621; Larivière *et al.*, 2003:307). After the preparation for EMG testing was finished, the subject was asked to sit on a bench in an upright, neutral spine position for the PAB device and its attachments (Velcro strap, chain and steel pipe) to be affixed to the subject's body.

With respect to collecting EMG data, the raw EMG signal was recorded from the two electrode locations [Myotrace 400 (MT400), Noraxon, USA] and stored in a personal computer using Myoresearch XP software. The sampling rate for the MT400 EMG device is 1000 samples per second (s). The band was set

between 10-500 Herz (Hz) for the preamplifier. The low pass cutoff was 500 Hz and the highpass cutoff, 20 Hz. The root mean square (RMS) on the EMG, MT 400 display was smoothed with a window of 300 milliseconds (ms) (Homann, 2007).

The EMG variable, RMS, was reported in this study because it is also a way of calculating power or the amount of electrical muscle potential, measured in micro Volts (μV) (Humphrey *et al.*, 2005:181). These researchers found that RMS was significantly different between back pain subjects and normal controls. They have further reported that RMS was highly correlated with MVIC.

The EMG variable, median frequency slope (MF slope), was not investigated, because it is a technique used for the assessment of back muscle fatigue (Mannion & Dolan, 1994:1223; Roy *et al.*, 1989:992), which was not investigated in this study. The EMG variable, initial median frequency (IMF) was also not investigated, because it is a variable that is relatively independent of load (Nargol *et al.*, 1999:883) however, Mannion and Dolan (1996:159) have shown that it is load-dependent at lower loads of force output. This study measured relatively high force output levels (MVIC and 80% MVIC), except for 50% MVIC. The IMF is also a variable used in back muscle fatigue studies (Humphrey *et al.*, 2005:182), however, the current study did not investigate back muscle fatigue.

5.8.2 The pressure air biofeedback and electromyography trial

An upright seated, closed chain isometric back extension test was done with the lumbar spine in neutral (Elfving *et al.*, 2000:118; Elfving *et al.*, 2003:621; O'Sullivan *et al.*, 2006:E707; Creamer, 2008). A neutral posture for the lumbar spine or lumbopelvic posture was always maintained during the testing protocol. Angles for the hips were measured at approximately 90° flexion, while the knees were flexed close to 45°, using a goniometer. A postural restraint apparatus (PRA) was fitted (with non-stretchable elastic Velcro straps) around the thorax, just below the inferior angles of the scapulae (dorsal view) and just below the mamilla line (frontal view). This postural restraint position needs to be

maintained (strap position not to shift on body) to maximise the L5-S1 extension moment in the test.

The PRA was attached to the PAB, while the PAB was again connected with a light steel chain to a steel pipe for both feet to push against. A display unit (PC screen) in front of the subject continuously showed the air pressure (real-time) in mb visual feedback. The pre-test, set-up procedure was done in the following way with all subjects:

- The subject was instructed to sit upright and maintain the neutral spine posture in the seated position.
- After the subject had been properly set up with the PAB device, he/she performed two submaximal contractions in the seated neutral spine posture to become familiar with the PAB device and the real-time feedback graph as shown on the computer screen in front of the subject.

After the two submaximal contractions, a two minute break was given before the research testing was started. The internal air pressure of the PAB ball was always checked and if necessary calibrated at 50 mb before and during the PAB testing protocol.

The subject was instructed to sit relaxed in an upright neutral spine position while resting EMG was recorded over a five second period. The subject was then instructed to do two maximal voluntary isometric contractions (MVIC) for standardisation, with the hips flexed to approximately 90°-95° and knees flexed to approximately 40°-45°. With both hands holding the front attachment, the subject produced two maximum isometric low back extensions without pulling with the arms on the front attachment.

The subject started with a two second build up to a maximum isometric extension effort and then held for three seconds at maximum, giving a total of five seconds. A one minute rest break was given between the two MVIC tests, while a three minute break was given between the MVIC and sub MVIC tests.

After the two MVIC tests, the best PAB force output of the two tests was used as the standard maximum PAB force. Fifty percent (50%) and 80% of the maximum PAB force were then calculated for the sub maximum tests to be done. The calculation of the sub maximal value of 50% was done according to the following example:

A subject's highest air pressure output was 187 mb on average, as read over a three second period from the real time PAB force graph. Fifty percent (50%) from 187 mb, was calculated by subtracting the internal pressure of the PAB ball (constant of 50 mb) from 187 mb ($187-50=137$ mb) which gave the real air pressure increase from the starting pressure of 50 mb. The 50% were then calculated from 137 mb which gave a figure of 68.5 mb. But, to calculate 50% from 187 mb the 68.5 mb was added to 50 mb (constant PAB ball pressure) which calculated to 118.5 mb or 50% of 187 mb. The 80% value was calculated in the same way.

After the subject was informed of the 50% and 80% values to be tested in terms of his/her MVIC achieved, one 50% and one 80% MVICs were then tested over a five second period each. With both hands holding the front attachment, the subject started with a two second build-up to the calculated sub-maximal pressure and then held for three seconds, for a total of five seconds. The subject produced an isometric back extension at 50% and 80% of his/her PAB force maximum in this fashion. The subject could see and control the sub maximum effort value as depicted on the PC screen in front of him/her. A two minute break was given between the 50% and 80% MVIC tests. The PAB and EMG data collected after each test was saved for analysis. The saved PAB and EMG graphs over the three second interval were analysed. That is, three PAB and three EMG recordings each were taken of the left and right *m. lumbar multifidus* and then averaged.

After finishing the PAB and EMG trial, the subject was transported to the radiology department of the St Anne's Private Hospital for US testing. There was a thirty (30) minute break between the PAB and EMG trial done at the biokinetic practice and the PAB and US trial done at the hospital.

5.8.3 The pressure air biofeedback and ultrasound trial

For US testing, the PAB device set-up procedure was exactly the same as for the PAB and EMG trial. Ultrasound images in brightness (B) mode of bilateral *m. lumbar multifidus* were obtained using a Toshiba 6000C diagnostic ultrasound instrument with a 7.5 MHz curvi-linear transducer for optimum penetration and resolution (Toshiba eub555; Toshiba Medical Corp, Tokyo, Japan). The US scanning and measurements were done by a specialist diagnostic radiologist for reliable measurements. Gel was interposed between the transducer and the skin area over the L5 spinous process. The researcher was mainly involved with the correct mechanical set-up procedure of the subject in the testing position, the calibration of the PAB between tests and recording of all the pressure air biofeedback results. Great care was taken to maintain the correct neutral lumbar posture of each subject during all the tests, as well as to maintain the ultrasound scanner head in the correct position at all times. The radiologist also adjusted the angle of the transducer to optimize visualization of the image.

The spinous process of L5 was palpated and the location confirmed by palpating L4-5 interspinous space and L4 spinous process from the position of the iliac crests. The skin over the L5 spinous process was already marked from the PAB/EMG trial previously done. The site for L5 *m. lumbar multifidus* was confirmed *in vivo* on the US image (Coldron *et al.*, 2003:162; Stokes *et al.*, 2005:117). The transducer was placed longitudinally over the skin marking of the L5 spinous process to orientate the radiologist and also to confirm the anatomical skin marking. The transducer was then rotated by 90° to obtain a transaxial view, showing the L5 spinous process and bilateral echogenic laminae. The transducer was angled to allow the beam to be perpendicular to the lamina (i.e. to record brightest image of lamina) as this provided a consistent landmark.

A US image was taken at the L5 landmark while the subject was in a relaxed upright seated neutral spine posture. This was saved and measured as a resting image. Following the resting measurement, the subject did one MVIC,

one 50% MVIC and one 80% MVIC of five seconds each as was done in the PAB and EMG trial. It was important for the subject to hold the five second isometric contractions within 10% of the calculated target air pressure.

The same rest breaks as in the PAB and EMG trial were taken (see Appendix D for testing protocol). One axial or cross-sectional image of each of the four different effort levels (rest, MVIC, 50% MVIC and 80% MVIC) was taken at the level of L5 of both the left and right *m. lumbar multifidus* and measured. Visual feedback on the PC of the PAB force efforts was given to subjects. The ultrasound images were collected at the fourth second of the five second period and then saved on the US system.

The lateral borders of the left and right *m. lumbar multifidus* were identified on the frozen US scan and then measured using the on-screen calipers to trace the perimeter of the muscle and to provide a measure of area in cm² (Hides *et al.*, 1995:56; Coldron *et al.*, 2003:162; Kristjansson, 2004:85). One cross-sectional image of *m. lumbar multifidus* at rest in upright sitting, at MVIC, 50% MVIC and 80% MVIC was recorded. Therefore, a total of four cross-sectional images of the left and right *m. lumbar multifidus* per subject were obtained. The average for the left and right CSA of *m. lumbar multifidus* were calculated for each effort level.

The methodology, as explained in this chapter, was strictly followed for all subjects. Forty three subjects (n=43) were assessed in total. In Chapter Six, the results of all the 43 subjects tested, will be reported and statistically analyzed in detail.

CHAPTER SIX

RESULTS

- 6.1 INTRODUCTION
- 6.2 STATISTICAL ANALYSIS
- 6.3 RESULTS OF DESCRIPTIVE STATISTICS
 - 6.3.1 Descriptive statistics for age and body mass index
 - 6.3.2 Descriptive statistics for waist-to-hip ratio
 - 6.3.3 Descriptive statistics for standing lumbar posture
 - 6.3.4 Descriptive statistics for the Oswestry Disability Index
- 6.4 PRESSURE AIR BIOFEEDBACK DEVICE CALIBRATION
- 6.5 REAL TIME BIOFEEDBACK GRAPHS FOR PRESSURE AIR BIOFEEDBACK, ELECTROMYOGRAPHY AND ULTRASOUND
 - 6.5.1 First case – an asymptomatic 42 year old male
 - 6.5.2 Second case – a 49 year old low back pain woman
 - 6.5.3 Third case – an 80 year old asymptomatic male
 - 6.5.4 Fourth case – a 66 year old asymptomatic male
- 6.6 INTRACLASS CORRELATION COEFFICIENT AGREEMENT RESULTS OF ELECTROMYOGRAPHY, PRESSURE AIR BIOFEEDBACK AND ULTRASOUND VARIABLES
- 6.7 ADJUSTED TESTING PROTOCOL
- 6.8 RESULTS OF DAY-TO-DAY REPEATABILITY ANALYSIS
 - 6.8.1 Pressure air biofeedback day one versus day two during electromyography testing
 - 6.8.2 Pressure air biofeedback day one versus day two during ultrasound testing
 - 6.8.3 Pressure air biofeedback during the trial-to-trial tests of electromyography and ultrasound on day one and two
 - 6.8.4 Electromyography root mean square values day one versus day two
 - 6.8.5 Ultrasound day one versus ultrasound day two
 - 6.8.6 Ultrasound day one and two versus pressure air biofeedback (ultrasound testing) day one and two

- 6.8.7 Electromyography versus pressure air biofeedback
(electromyography testing) day one and two for the whole group
- 6.8.8 Electromyography versus pressure air biofeedback day one and two
for two subgroups
- 6.9 LOW BACK STRENGTH IN ASYMPTOMATIC AND LOW BACK
PAIN SUBJECTS

6.1 INTRODUCTION

Measurement of muscular strength is very important, especially from the viewpoint of kinematic assessment, because activities of daily living are dependent on muscular strength. Therefore, assessment is essential, not only in the field of orthopaedic surgery, but also in most fields of medicine (Nobori & Maruyama, 2007: 9). Various force measuring devices have been developed and improved (Nobori & Maruyama, 2007: 9) and have been used, for example, to determine and monitor the effects of training programmes on back extensor muscles (Rezasoltani *et al.*, 2003: 7). In this study, a pressure air biofeedback device (PAB) was tested during an upright seated, closed chain back extension test to measure *m. lumbar multifidus* strength.

To establish the measurement precision of the new testing apparatus and to determine possible sources of error in the results, a series of tests were performed to test the reliability and validity of the PAB device. The PAB device was validated against calibrated weights in kilogram (kg), and validity was further assessed by comparing maximum and sub-maximum air pressure output (PAB force) in millibar (mb) to a criterion, the Noraxon electromyography (EMG) device (microvolt in μV) and the Toshiba ultrasound scanner [cross-sectional area (CSA) in cm^2]. Furthermore, trial-to-trial and day-to-day comparisons of PAB measurements were made to estimate the reliability of the new instrument. Descriptive statistics were also reported for age, body mass index (BMI), waist-to-hip ratio (WHR), standing lumbar posture and the Oswestry Disability Index (ODI). Statistical tests and analysis were performed to test the following:

- To determine the correlation between PAB force (mb) and calibrated weights (kg).
- To determine the correlation between trial-to-trial and day-to-day PAB force (mb) measurements, thereby estimating the reliability of the PAB device.

- To determine the correlation between PAB force (mb) and EMG root mean square (RMS) data points (μV), that is to determine if the different load tasks or maximum voluntary isometric contractions (MVIC) and sub-maximum voluntary isometric contractions (80% MVIC and 50% MVIC) adequately increased *m. lumbar multifidus* activation and PAB force output, thereby assessing validity of the PAB device.
- To determine the correlation between PAB (mb) and US (CSA in cm^2) data points during MVIC, 80% MVIC and 50% MVIC.
- To determine the difference between low back strength, expressed in PAB force (mb) of asymptomatic and low back pain subjects during the seated low back extension test.

6.2 STATISTICAL ANALYSIS

Repeated measures ANOVA were performed to determine the effects of age, BMI, WHR, standing lumbopelvic posture and ODI on the dependent variables of group (low back pain and asymptomatic) and gender (male and female). Also, repeated measures ANOVA were performed to determine the difference in low back strength, expressed in PAB force (mb), between the low back pain and asymptomatic group during MVIC, 80% MVIC and 50% MVIC (during EMG testing). Effects were considered significant at $p < 0.05$.

Intraclass correlation coefficient (ICC) agreement results of root mean square (RMS) values of EMG between left (L) and right (R) *m. lumbar multifidus* at four different effort levels (resting, MVIC, 80% MVIC and 50% MVIC) on day one and day two were calculated. Also, ICC agreement results of three PAB effort levels (MVIC, 80% MVIC and 50% MVIC) during EMG testing on day one and day two were reported. Furthermore, ICC agreement results of US between left and right *m. lumbar multifidus* at four different effort levels (resting, MVIC, 80% MVIC and 50% MVIC) on day one and day two were calculated.

Furthermore, the Pearson correlation coefficient (r) was calculated in order to examine the correlation between PAB force (mb), EMG (μV) and US (CSA cm^2) data points. An alpha level of 0.05 was selected for statistical significance in this study, because the PAB test was developed as a low risk, low back extension test, that would not put low back patients' or asymptomatic subjects' lumbar spines at risk for injury. The Statistica software program at Stellenbosch University was used for data analysis.

6.3 RESULTS OF DESCRIPTIVE STATISTICS

Descriptive statistics of the morphological characteristics of the random selected 43 subjects (21 males and 22 females) were calculated. From these 43 subjects, 25 were classified as asymptomatic, while 18 were referred to a biokinetic practice for low back rehabilitation. From these 18 low back pain subjects, seven had lumbar surgery, six were diagnosed with mechanical low back pain, while the rest (five) were diagnosed with different clinical lumbar pathologies.

6.3.1 Descriptive statistics for age and body mass index

The average age of the 43 subjects was 47.58 years ($\text{SD} = \pm 18.58$). Back pain subjects were on average 48.89 years ($\text{SD} = \pm 14.22$) of age, while the asymptomatic subjects were 46.64 years ($\text{SD} = \pm 21.43$) old. With respect to gender, females average age was 47.68 ($\text{SD} = \pm 17.08$), while male subjects were 47.48 years ($\text{SD} = \pm 20.47$) of age. No significant difference in age between the asymptomatic and low back pain groups ($p < 0.70$), and between the two genders ($p < 0.97$) were reported. Male subjects mean body mass was 80.51 kg ($\text{SD} = \pm 11.50$), they were 1.76 m ($\text{SD} = \pm 0.08$) tall and had a mean body mass index (BMI) of 25.99 kg/m^2 ($\text{SD} = \pm 2.77$). The female subjects weighed 65.1 kg ($\text{SD} = \pm 13.39$) on average while they were 1.64 m ($\text{SD} = \pm 0.06$) tall. Their mean BMI was 23.96 kg/m^2 ($\text{SD} = \pm 4.21$). According to the American College of Sports Medicine (ACSM, 2006:58), the male subjects mean BMI was classified as overweight (25.0-29.9 kg/m^2), while the female subjects fell within the normal

BMI of 18.5-24.9 kg/m². However, there was no significant difference in the BMI profile ($p>0.05$) between the low back pain group and the asymptomatic group.

6.3.2 Descriptive statistics for waist-to-hip ratio

In terms of waist-to-hip ratio (WHR), the male subjects had a mean waist girth of 91.01 cm (SD= ± 8.64) and gluteal hip girth of 98.16 cm (SD= ± 6.16), which calculated to an average WHR 0.93 (SD= ± 0.07). With respect to the female subjects, mean waist girth was 80.05 cm (SD= ± 13.26), while gluteal hip girth was 98.36 cm (SD= ± 10.71), calculating to an average WHR of 0.81 (SD= ± 0.06). According to WHR norms, standards for health risk vary with age and sex. For instance, health risk is very high for young men when WHR is more than 0.95. For young women, a WHR of more than 0.86 is considered a high health risk. For 60-69 year olds, WHR values greater than 1.03 for men and WHR values greater than 0.90 for women fall within the same high risk classification (ACSM, 2006:59). In this study both male and female subjects mean WHR fell within normal limits.

6.3.3 Descriptive statistics for standing lumbar posture

With respect to standing posture angles of the lumbar spine, the mean standing lumbosacral angle for men was calculated at 11° (SD= ± 4.14), while the mean thoracolumbar angle was determined at 20° (SD= ± 6.91). The standing lumbosacral angle for men fell outside the normal angles of 15°-30° (Saunders 1998:5), while the standing thoracolumbar angle calculated more than ten degrees below the minimum required lordosis of 30°, indicating loss of lumbar lordosis for the male subjects. The mean standing lumbosacral angle for the female subjects was 16° (SD= ± 5.01), while their average standing thoracolumbar angle was calculated at 29° (SD= ± 8.07). Female subjects' lumbosacral angle calculated within the norms of 15°-30°, while their standing thoracolumbar angle calculated just outside the minimum norm for a normal lordosis of 30°-40° (Saunders 1998:5). These parameters indicated that the standing lumbar posture of the females fell within the minimum required angles for a normal lordosis.

6.3.4 Descriptive statistics for the Oswestry Disability Index

The Oswestry Disability Index (ODI) for each subject was determined and then calculated for healthy and low back pain subjects. Asymptomatic subjects (n=25) ODI score was calculated at an average of 3.13% (SD= ± 4.17), which was interpreted as minimal disability (0%-20%), while the low back pain subjects (n=18) scored an average of 19.33% (SD= ± 11.06), which bordered on the norms for moderate disability. According to Sung (2003:1314), interpretation of disability scores is as follows: 0%-20%, minimal disability; 20%-40%, moderate disability; 40%-60%, severe disability; 60%-80%, crippled; 80%-100%, bed bound or the patient is greatly exaggerating his/her symptoms.

6.4 PRESSURE AIR BIOFEEDBACK DEVICE CALIBRATION

The results of the PAB calibration measurements were reported and discussed in Chapter Four (Section 4.3). The calibration results formed an integral part of the technology development and validation of the PAB device, this being the reason why they were reported and discussed in Chapter Four. In summary, it was found that the PAB device is a valid and reliable device when tested with calibrated weights. Therefore, the results of PAB force (mb) and applied external force comparisons (calibrated weights in kg) demonstrated high agreement or validity ($r=0.997$, $p<0.01$) between measures and the associated criterion. Furthermore, the ICC calculation of 0.997 (SEM=1.55) between the two sets of PAB force calibration measurements between day one and two indicated a significant correlation and therefore very good reliability of the PAB device.

6.5 REAL TIME BIOFEEDBACK GRAPHS FOR PRESSURE AIR BIOFEEDBACK, ELECTROMYOGRAPHY AND ULTRASOUND

For the purpose of the development of new technology muscle testing instruments, in this case the PAB device, real time biofeedback graphs of four different subjects in this research study were reported. The four different subjects, each with their own specific morphological and diagnostic profile, were

specifically selected from the group of 43 subjects. This was done to illustrate the ability of the PAB device to assess asymptomatic-, low back pain-, very old- as well as different gender subjects. According to Pienaar and Webster (2009), the PAB device was designed and developed for “strong man and old lady testing”, to illustrate the versatility of the PAB device and to reliably measure different groups of people, as reported in this study.

6.5.1 First case – an asymptomatic 42 year old male

Although they were very different in morphological and diagnostic appearance, it appeared that the PAB could measure MVIC (100% effort) and submaximal MVICs (80% and 50% effort) of all four cases, accurately and without any risk to the subjects. The EMG graphs were reported in RMS values. In the first mentioned case (an asymptomatic 42 year old male), the real-time biofeedback graphs for PAB (mb) and EMG (μV) in the first trial on day one are shown in Figures 6.1a and 6.1b.

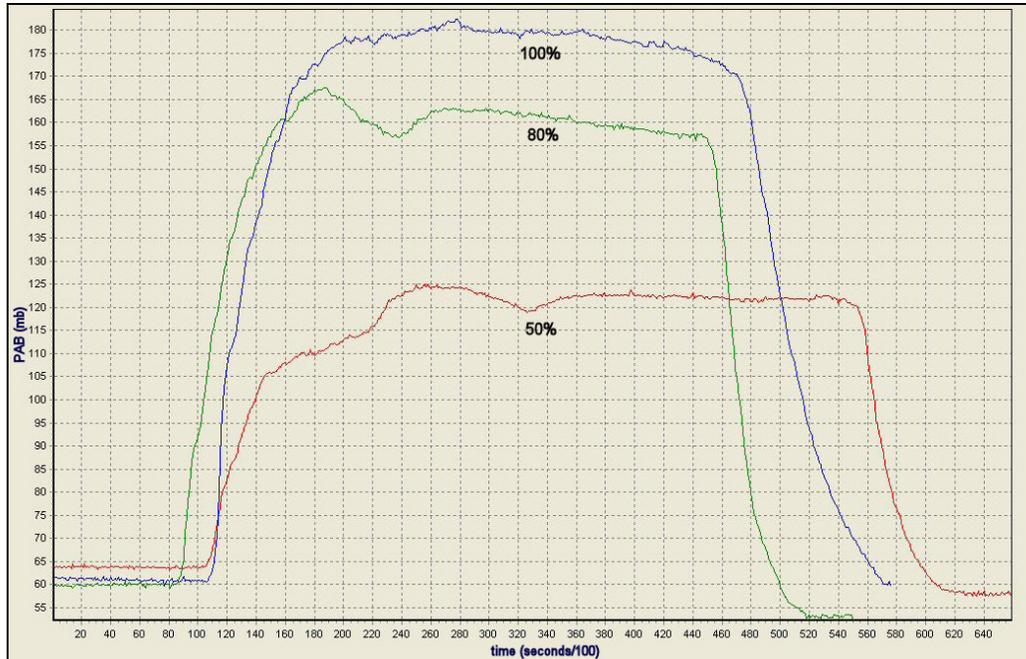


Figure 6.1a PAB force real-time graph (mb) in an asymptomatic 42 year old male, illustrating high air pressure output values for MVIC, 80% and 50% MVIC.

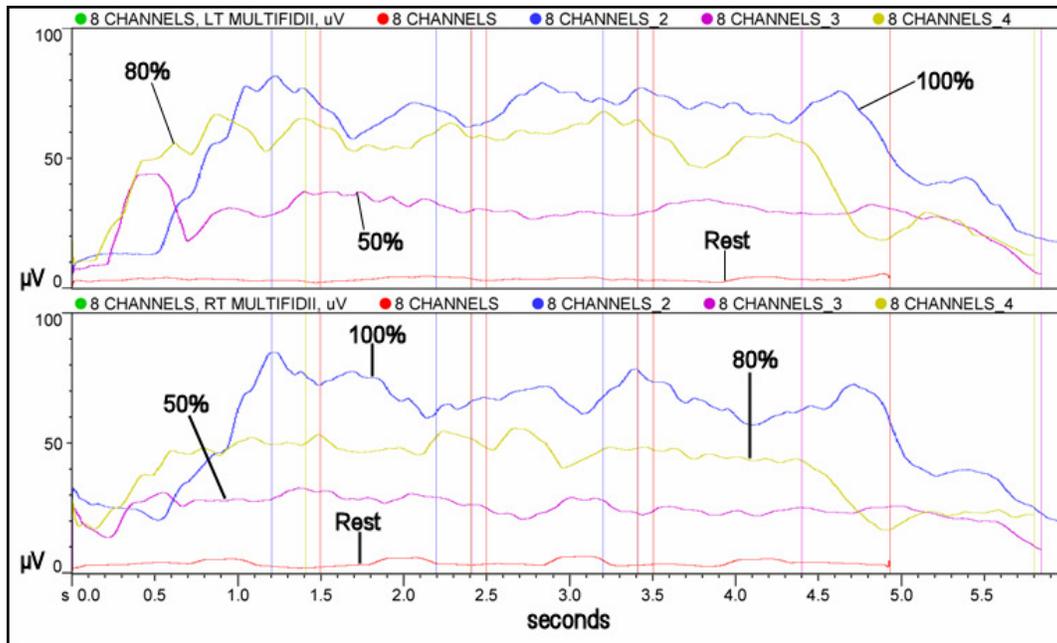


Figure 6.1b Electromyographic RMS values (μV) of the same 42 year old male at different effort levels. Left (LT) *m. lumbar multifidus* on the top is shown in relation to right (RT) *m. lumbar multifidus* at the bottom.

6.5.2 Second case – a 49 year old low back pain woman

The second case presented was a 49 year old female patient who suffered from low back pain, post-surgery. A transpedicular fixation was done at L5/S1 with an L5 laminectomy. At the time of testing, she was 19 months post-surgery. Real-time biofeedback graphs for PAB (mb) and EMG (μV) at MVIC, 80% and 50% MVIC in the second trial on day two are shown in Figures 6.2a and 6.2b.

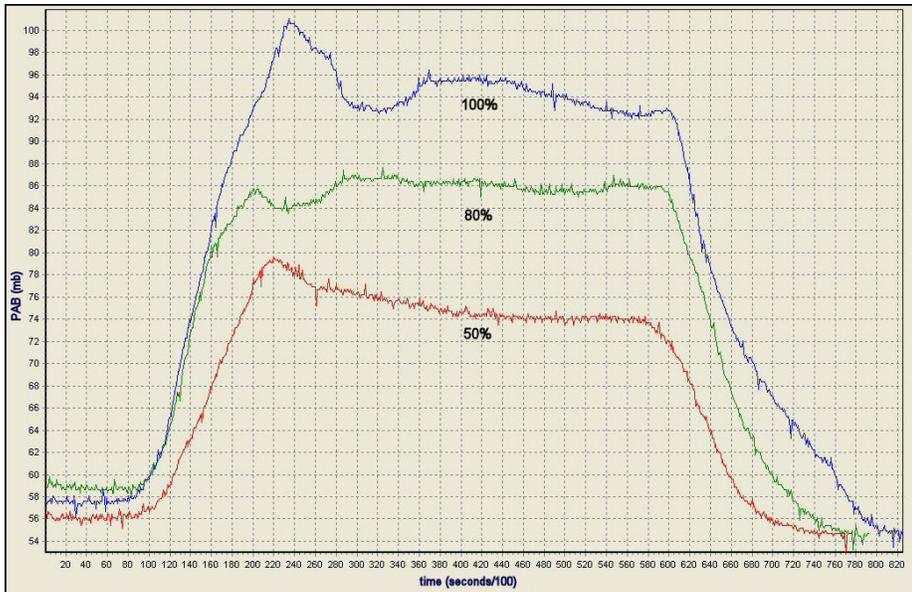


Figure 6.2a PAB force real-time graph (mb) in a 49 year old low back pain woman, illustrating weak low back extension strength according to the air pressure values shown, at three different effort levels.

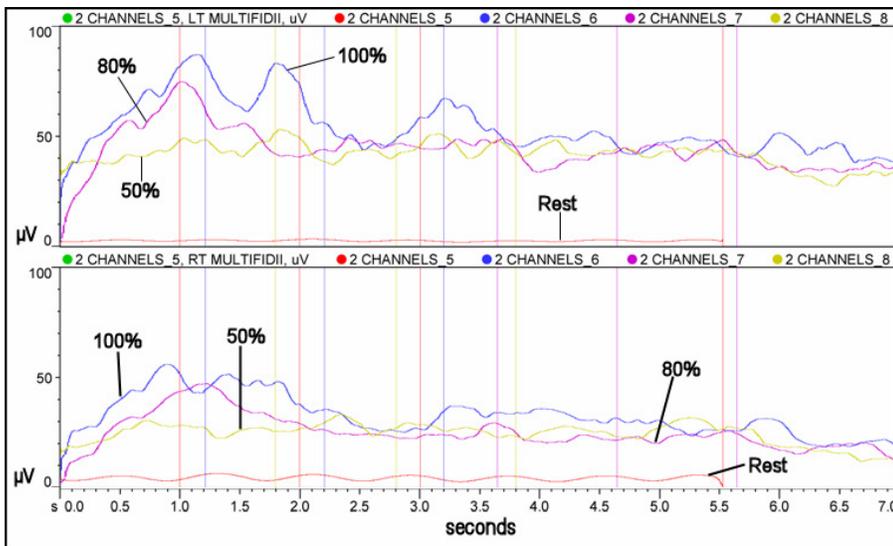


Figure 6.2b RMS values of EMG (μV) of different strength contractions for left (LT) and right (RT) *m. lumbar multifidus* of the same 49 year old, low back pain woman. Note the EMG imbalance between left (LT) and right (RT) sides, demonstrating muscle denervation of right *m. lumbar multifidus*.

6.5.3 Third case – an 80 year old asymptomatic male

In the third case, PAB (mb) and EMG (μV) real-time graphs (first trial, day one) of an 80 year old asymptomatic male are shown in Figures 6.3a and 6.3b.



Figure 6.3a Example of weak low back strength as shown in the low air pressure output levels of the PAB force graph (mb) of an 80 year old asymptomatic male.

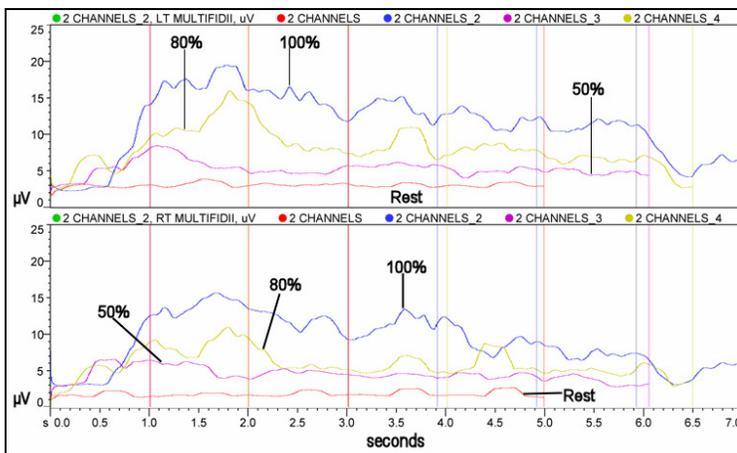


Figure 6.3b RMS values of EMG (μV) of left (LT) and right (RT) *m. lumbar multifidus* of the same 80 year old asymptomatic male. EMG difference between LT and RT demonstrate muscle denervation of the right side.

6.5.4 Fourth case – a 66 year old asymptomatic male

Furthermore, to illustrate how *m. lumbar multifidus* CSA on the US imaging scan has been measured, the results of a 66 year old asymptomatic male were randomly selected and reported. The real-time biofeedback results of PAB (mb) and real-time US (CSA in cm²) in the first trial on day one are shown in Figures 6.4a and 6.4b.

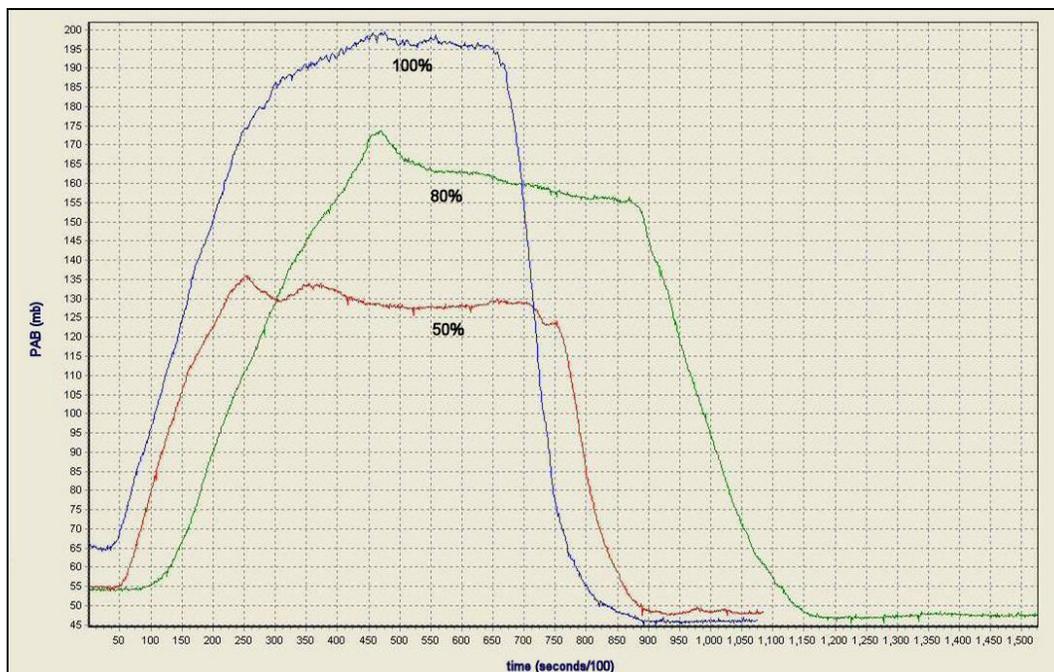


Figure 6.4a PAB force real-time graph (mb) in a 66 year old asymptomatic male, illustrating high air pressure output values for MVIC, 80% and 50% MVIC.

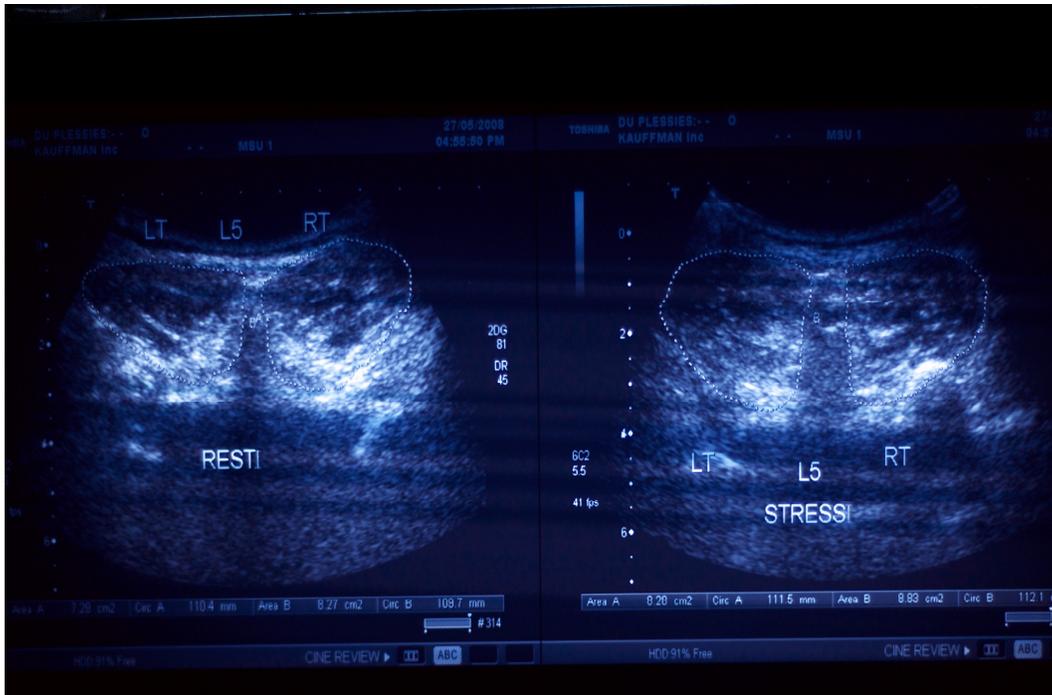


Figure 6.4b Axial US image of *m. lumbar multifidus* (CSA in cm²) of resting and MVIC (stress) of the same 66 year old asymptomatic male. Left (LT) and right (RT) *m. lumbar multifidus* are shown at the fifth lumbar vertebrae (L5). At rest, the LT side measured 7.29 cm² and the RT side, 8.27 cm². At MVIC (stress) the LT side measured 8.28 cm² and the RT side, 8.83 cm².

6.6 INTRACLASS CORRELATION COEFFICIENT AGREEMENT RESULTS OF ELECTROMYOGRAPHY, PRESSURE AIR BIOFEEDBACK AND ULTRASOUND VARIABLES

Table 6.1 show intraclass correlation coefficient (ICC) agreement results of root mean square (RMS) values of EMG between left and right *m. lumbar multifidus* at different effort levels of the 43 subjects. Table 6.1 further reports the RMS values of EMG measured over two days taken per second (s) over a three second period. Four effort levels, namely; resting, MVIC, 80% MVIC and 50% MVIC were done in an upright seated back extension test on day one as well as day two. EMG day one (rest) showed an ICC agreement of 0.83 (SEM=1.55), while day two (rest) calculated 0.82 (SEM=2.37). EMG day one (MVIC) showed

an ICC agreement result of 0.88 (SEM=12.67), while day two reported a result of 0.91 (SEM=8.46).

Table 6.1 ICC results of RMS values of EMG during four effort levels measured on day one and two between left (L) and right (R) *m. lumbar multifidus* over a three second period (n=43).

	Variables	ICC Agreement	95% CF	Standard error of measurement (SEM)
1	EMG day1 rest L1,2,3 s	0.83	0.75 - 0.89	1.55
2	EMG day1 rest R1,2,3 s			
1	EMG day1 MVIC L1,2,3 s	0.88	0.82 - 0.93	12.67
2	EMG day1 MVIC R1,2,3 s			
1	EMG day1 80% MVIC L1,2,3 s	0.92	0.88 - 0.95	5.63
2	EMG day1 80% MVIC R1,2,3 s			
1	EMG day1 50% MVIC L1,2,3 s	0.85	0.79 - 0.91	5.50
2	EMG day1 50% MVIC R1,2,3 s			
1	EMG day2 rest L 1,2,3 s	0.82	0.74 - 0.88	2.37
2	EMG day2 rest R 1,2,3 s			
1	EMG day2 MVIC L 1,2,3 s	0.91	0.86 – 0.94	8.46
2	EMG day2 MVIC R 1,2,3 s			
1	EMG day2 80% MVIC L1,2,3 s	0.89	0.84 – 0.93	8.32
2	EMG day2 80% MVIC R1,2,3 s			
1	EMG day2 50% MVIC L1,2,3 s	0.87	0.82 – 0.92	6.43
2	EMG day2 50% MVIC R1,2,3 s			

Intraclass correlation coefficient (ICC) agreement results for EMG day one and two at 80% MVIC were 0.92 (SEM=5.63) and 0.89 (SEM=8.32) respectively. EMG day one (50% MVIC) calculated an ICC of 0.85 (SEM=5.50), while day two reported 0.87 (SEM=6.43). As can be seen, the ICC for all EMG variables indicated that the EMG test was statistically reliable as it was performed in a systematic manner in a test-retest protocol. This may indicate the reliability of the Noraxon EMG device when used in the closed chain, seated back extension test.

With respect to Table 6.2, ICC agreement results of various PAB effort levels (during EMG testing) are reported. PAB (EMG) day one (MVIC) reported an ICC of 0.99 (SEM=4.44), while day two (MVIC) reported the same result of 0.99 (SEM=3.22).

Table 6.2 ICC of PAB values at three effort levels (during EMG testing) measured over two days, taken per second (s) over a three second period (n=43).

	Variables	ICC agreement	95% CF	Standard error of measurement (SEM)
1	PAB(EMG) day1 MVIC 1,2,3 s	0.99	0.98 – 0.99	4.44
2	PAB(EMG) day1 80% MVIC 1,2,3 s	0.99	0.99 - 1.00	2.31
3	PAB(EMG) day1 50% MVIC 1,2,3 s	0.99	0.99 – 1.00	1.69
1	PAB(EMG) day2 MVIC 1,2,3 s	0.99	0.98 – 1.00	3.22
2	PAB(EMG) day2 80% MVIC 1,2,3 s	0.99	0.98 – 0.99	3.00
3	PAB(EMG) day2 50% MVIC 1,2,3 s	0.99	0.98 – 0.99	2.04

PAB (EMG) day one (80% MVIC) scored 0.99 (SEM=2.31), while day two (80% MVIC) also calculated 0.99 (SEM=3.00). The ICC agreement calculation for PAB (EMG) day one (50% MVIC) was 0.99 (SEM=1.69), while day two (50% MVIC) again calculated 0.99 (SEM=2.04). Test-retest correlations for all PAB variables showed good reliability at 0.99 indicating the PAB testing protocol to be statistically reliable over the two days. This indicates that the PAB device is a reliable instrument when used in an upright seated, closed chain back extension test.

With respect to US data, ICC agreement results of different US variables between left and right *m. lumbar multifidus* are shown in Table 6.3. ICC for US day one (rest) was 0.91 (SEM=0.40), while day two (rest) calculated 0.90 (SEM=0.33). US day one (MVIC) reported an ICC of 0.89 (SEM=0.59), while day two (MVIC) also reported 0.89 (SEM=0.52). US day one (80% MVIC)

scored 0.94 (SEM=0.40), while day two (80% MVIC) scored 0.88 (SEM=0.48). ICC agreement calculations for US day one and two at 50% MVIC were 0.92 (SEM=0.44) and 0.94 (SEM=0.38) respectively.

Table 6.3 ICC of US values during four effort levels measured on day one and two between left (L) and right (R) *m. lumbar multifidus* (n=28).

	Variables	ICC agreement	95% CF	Standard error of measurement (SEM)
1	US day1 rest L & R	0.91	0.81 – 0.96	0.40
2	US day1 MVIC L & R	0.89	0.77 – 0.95	0.59
3	US day1 80% MVIC L & R	0.94	0.83 – 0.98	0.40
4	US day1 50% MVIC L & R	0.92	0.83 – 0.96	0.44
1	US day2 rest L & R	0.90	0.78 – 0.95	0.33
2	US day2 MVIC L & R	0.89	0.78 – 0.95	0.52
3	US day2 80% MVIC L & R	0.88	0.64 – 0.95	0.48
4	US day2 50% MVIC L & R	0.94	0.84 – 0.97	0.38

Intraclass correlation coefficient (ICC) calculations for all US variables varied between 0.88-0.94, which indicated that the US scanner showed good reliability when measuring CSA of the *m. lumbar multifidus* during an upright seated, closed chain back extension test.

6.7 ADJUSTED TESTING PROTOCOL

Before discussing any further results, it is important to briefly mention the difference in number of subjects assessed by US (n=28) and EMG (n=43). After assessing 28 subjects with PAB and US, it was suspected that the majority of US assessments showed no increase in cm² of *m. lumbar multifidus* CSA. Therefore, it was decided to statistically analyze the results on the basis of the non-increase in CSA in *m. lumbar multifidus*. Because Pearson correlation calculations were non-significant (at the 0.05 level of significance) between US

and PAB for all three isometric effort levels for both days (see Paragraph 6.8.6), the decision was taken to terminate US assessments. However, it was decided to continue with EMG assessments, because the Pearson correlation between RMS values of EMG (μV) and PAB force (mb) for the 28 subjects at this stage varied between $r=0.43$ ($p<0.01$) and $r=0.54$ ($p<0.01$) for the various effort tests (day one). Day two correlations varied between 0.49 and 0.59 ($p<0.01$). An additional 15 subjects were randomly selected and tested to bring the total of PAB and EMG tests to 43. This explains the 28 reported results for US and the 43 reported results for EMG.

6.8 RESULTS OF DAY-TO-DAY REPEATABILITY ANALYSIS

6.8.1 Pressure air biofeedback day one versus day two during electromyography testing

Intraclass correlation coefficient (ICC) results of PAB (EMG) day one versus PAB (EMG) day two measurements from 43 subjects who participated in the day-to-day intrarater reliability component of the study are shown in Figures 6.5 – 6.7.

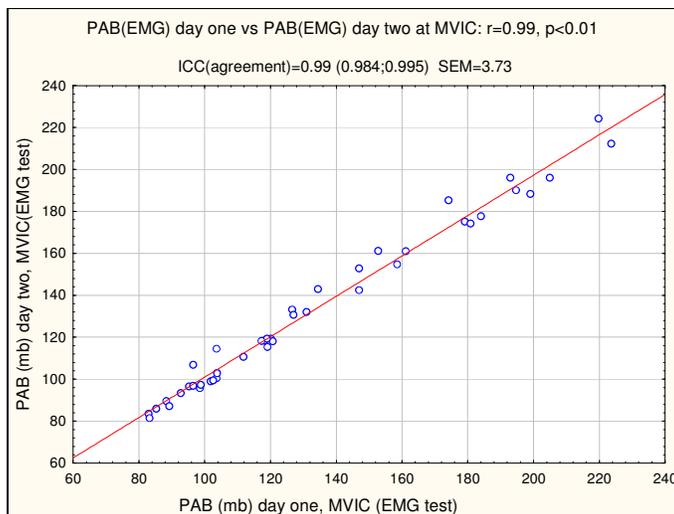


Figure 6.5 PAB (mb) day one versus PAB (mb) day two, reflecting a MVIC during EMG testing ($n=43$).

An ICC agreement of 0.99 (SEM=3.73) was calculated for PAB (MVIC) values between day one and two. ICC calculations for PAB (80% MVIC) values between day one and two were 0.99 (SEM=2.55), while the ICC for PAB (50% MVIC) calculated to 0.98 (SEM=3.32), showing good reliability for all results.

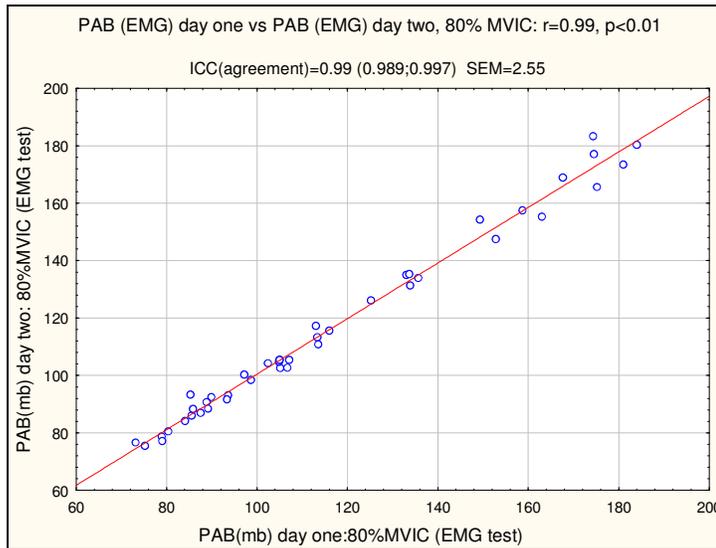


Figure 6.6 PAB (mb) day one versus PAB (mb) day two, reflecting an 80% MVIC during EMG testing (n=43).

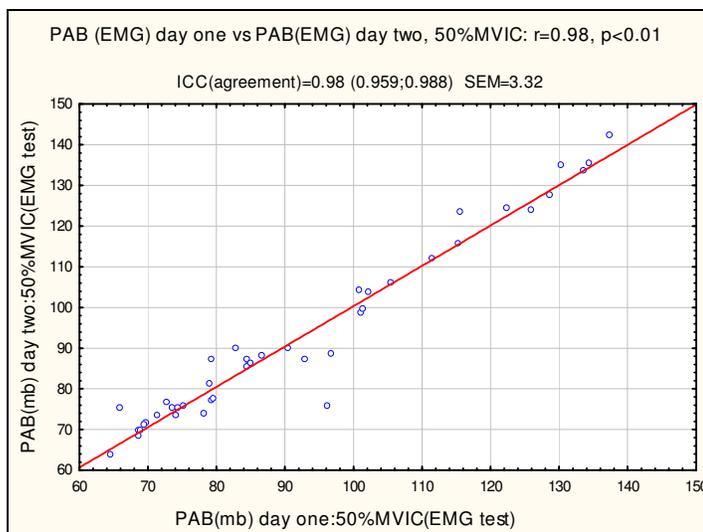


Figure 6.7 PAB (mb) day one versus PAB (mb) day two, reflecting a 50% MVIC during EMG testing (n=43).

6.8.2 Pressure air biofeedback day one versus day two during ultrasound testing

Figures 6.8 – 6.10 reflect the ICC results of PAB (US) day one versus PAB (US) day two from 28 of the 43 subjects who participated in the between-day intrarater reliability component of the study.

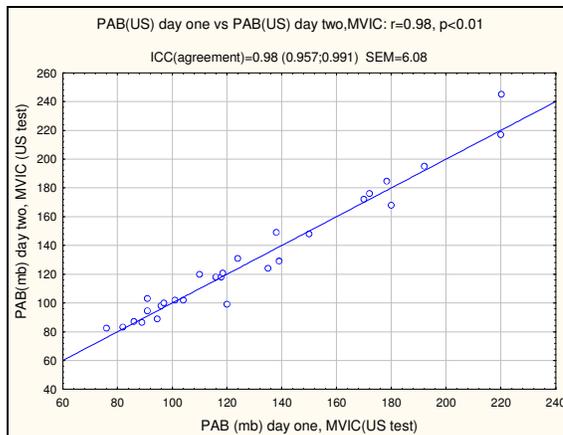


Figure 6.8 PAB (mb) day one versus PAB (mb) day two, reflecting a MVIC during US testing (n=28).

An ICC agreement value of 0.98 (SEM=6.08) was calculated for PAB (MVIC), for PAB (80% MVIC) the ICC was 0.91 (SEM=10.71) and for PAB (50% MVIC) it scored 0.98 (SEM=3.21) indicating good reliability for the PAB values during US testing.

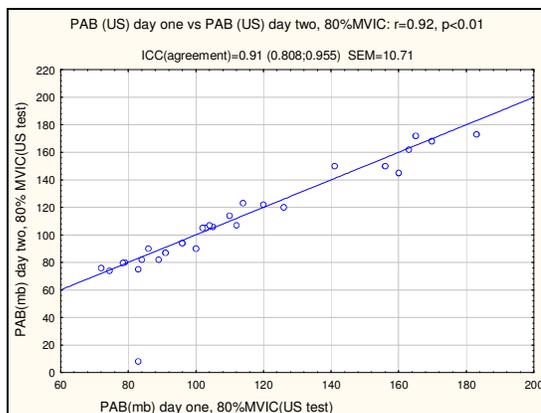


Figure 6.9 PAB (mb) day one versus PAB (mb) day two at 80% MVIC (n=28).

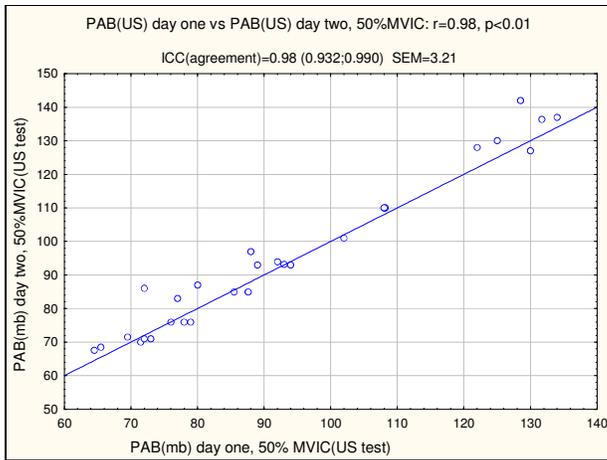


Figure 6.10 PAB (mb) day one versus PAB (mb) day two at 50% MVIC ($n=28$).

6.8.3 Pressure air biofeedback during the trial-to-trial tests of electromyography and ultrasound on day one and two

Results of PAB (EMG) day one versus PAB (US) day one of 28 of the 43 subjects who participated in the trial-to-trial intrarater reliability component of the study are shown in Figures 6.11 – 6.13. Pearson correlation coefficients (r) were calculated to examine the correlation between PAB pressure levels during the trial-to-trial tests of EMG and US on day one and two. An alpha level of 0.05 was selected for statistical significance.

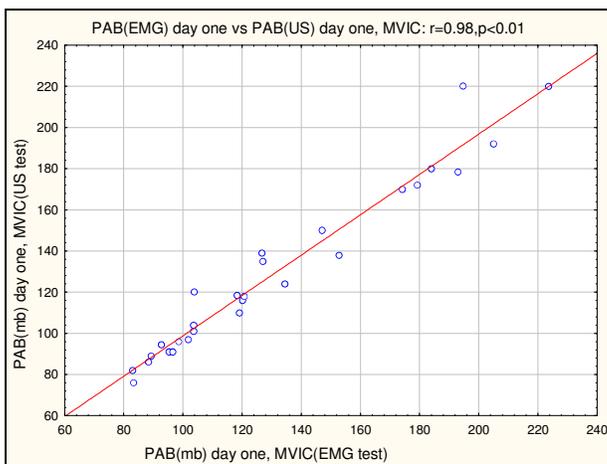


Figure 6.11 PAB (mb) versus PAB (mb) day one, reflecting a MVIC during EMG and US tests on the same day ($n=28$).

Correlation results of day one trial-to-trial PAB pressure data (during EMG and US tests) were statistically significant for MVIC=0.98 ($p<0.01$), 80% MVIC=0.99 ($p<0.01$) and 50% MVIC=0.97 ($p<0.01$). The calculated correlation for PAB trial-to-trial measurements on day two (during EMG and US tests) again were statistically significant at $r=0.96$ ($p<0.01$) for MVIC, $r=0.91$ ($p<0.01$) for 80% MVIC and $r=0.97$ ($p<0.01$) for 50% MVIC. Therefore, the correlation for the PAB assessments in the seated back extension test was significant, indicating good reliability for the trial-to-trial PAB test.

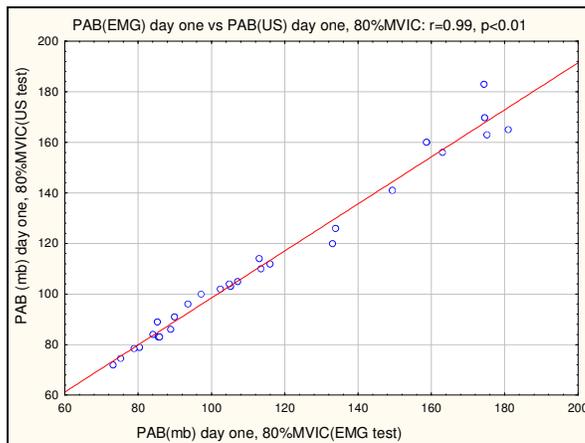


Figure 6.12 PAB (mb) versus PAB (mb) day one, reflecting an 80% MVIC during EMG and US tests on the same day ($n=28$).

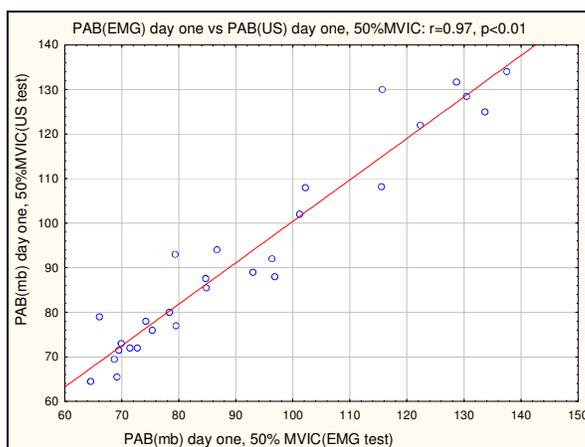


Figure 6.13 PAB (mb) versus PAB (mb) day one, reflecting a 50% MVIC during EMG and US tests on the same day ($n=28$).

6.8.4 Electromyography root mean square values day one versus day two

Average RMS values of EMG day one versus day two measurements from the 43 subjects who participated in the between-day intrarater reliability component of the study were investigated. The ICC for RMS values of EMG (μV) was calculated in order to examine the reliability between the RMS values of EMG measurements of day one versus day two during different isometric effort levels. ICC agreement values calculated to 0.79 (0.588; 0.891) with SEM=1.85 for resting EMG; 0.85 (0.744; 0.917) with SEM=12.54 for MVIC; 0.89 (0.798; 0.939) with SEM=7.15 for 80% MVIC and 0.80 (0.657; 0.886) with SEM=6.88 for 50% MVIC, which indicated good reliability for all results.

6.8.5 Ultrasound day one versus ultrasound day two

The ICC results of average US values day one versus US day two from 28 of the 43 subjects who participated in the between-day intrarater reliability component of the study were reported. An ICC agreement of 0.73 (SEM=0.62) was calculated for US (rest) values between day one and two. ICC calculations for US (MVIC) values between day one and two were 0.82 (SEM=0.72), for US (80% MVIC) it scored 0.86 (SEM=0.65) while the ICC for US (50% MVIC) calculated to 0.83 (SEM=0.66), which indicated good reliability for all results.

6.8.6 Ultrasound day one and two versus pressure air biofeedback (ultrasound testing) day one and two

Results of day one (US versus PAB) during US testing, as well as day two measurements from 28 of the 43 subjects who participated in the between-day reliability component of the study was reported. At the 0.05 level of significance, Pearson correlation calculations showed no correlation between US and PAB on day one, as well as on day two (Table 6.4).

Table 6.4 Pearson correlation calculations between ultrasound and pressure air biofeedback for different effort levels on day one and day two.

	DAY ONE - US versus PAB		DAY TWO – US versus PAB	
	Pearson (r)	p-value	Pearson (r)	p-value
MVIC	r=0.36	p<0.06	r=0.31	p<0.11
80% MVIC	r=0.38	p<0.05	r=0.27	p<0.17
50% MVIC	r=0.47	p<0.01	r=0.27	P<0.17

The non-significant correlation found between US and PAB tested over the different isometric effort levels across two days indicated that US was non-reliable in measuring the *m. lumbar multifidus* strength contraction in an upright sitting neutral spine posture. Figure 6.14 illustrates the non-significant linear relationship between US and PAB for MVIC on day two. As mentioned in Paragraph 6.7, on the basis of the non-significant correlation between US and PAB for all isometric levels tested on both days, it was decided to terminate US assessments and continue with EMG and PAB assessments only. This explains the reported results of 28 out of the 43 subjects that were assessed by US.

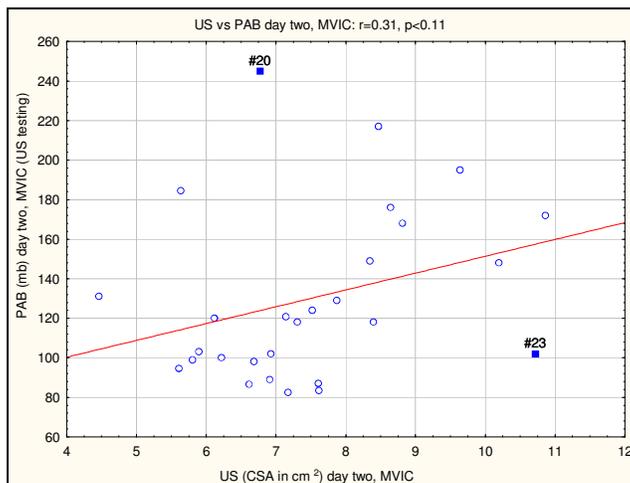


Figure 6.14 PAB (mb) versus US (CSA in cm²) day two, reflecting a non-linear relationship at MVIC during US testing (n=28). See Paragraph 7.5 for the explanation of the discrepancy between subjects 20 and 23.

6.8.7 Electromyography versus pressure air biofeedback (electromyography testing) day one and two for the whole group

Figures 6.15 – 6.20 show the MVIC, 80% and 50% MVIC results of day one (EMG versus PAB), as well as day two (EMG versus PAB) during EMG testing from 42 subjects who participated in the between-day intrarater reliability component of the study. One asymptomatic subject was not able to produce controlled contractions during the PAB (EMG) trials. This coincided with one of the exclusion criteria for participation in this study. Furthermore, the subject's EMG data showed a discrepancy of 35% between day one and day two which could not be explained in terms of the specific experimental preparation and testing procedure that had been followed. The subject's data was therefore excluded in the EMG versus PAB (during EMG testing) calculations, leaving 42 subjects to be analyzed (whole group) and 24 subjects in the asymptomatic group.

Table 6.5 Pearson correlation calculations between PAB force (mb) and RMS values of EMG (μ V) for different effort levels on day one and day two for the whole group (n=42).

	DAY ONE – EMG versus PAB		DAY TWO – EMG versus PAB	
	Pearson (r)	p-value	Pearson (r)	p-value
MVIC	r=0.75	p<0.01	r=0.63	p<0.01
80% MVIC	r=0.75	p<0.01	r=0.64	p<0.01
50% MVIC	r=0.63	p<0.01	r=0.54	p<0.01

According to the table for critical values of correlation coefficients (Thomas & Nelson (1985:344) and the degrees of freedom (df=n-2), a significant correlation (at the 0.05 level of significance) was calculated between EMG and PAB for the whole group (n=42) for all effort levels over the two days.

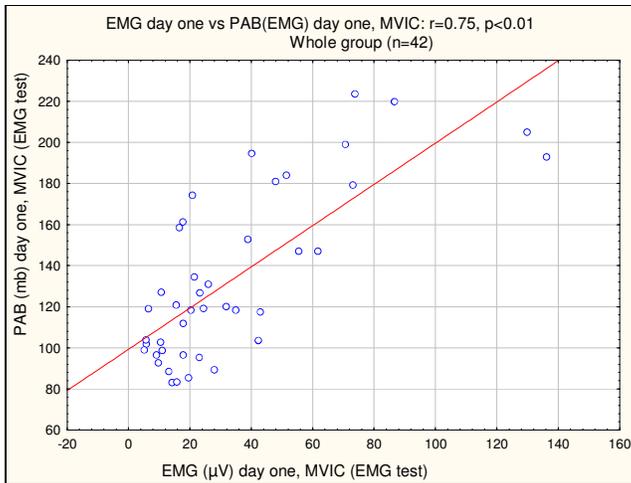


Figure 6.15 Day one, EMG (μV) versus PAB (mb) at MVIC (n=42).

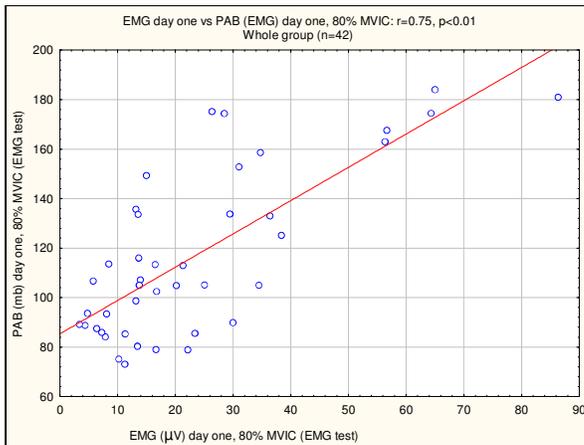


Figure 6.16 Day one, EMG (μV) versus PAB (mb) at 80% MVIC (n=42).

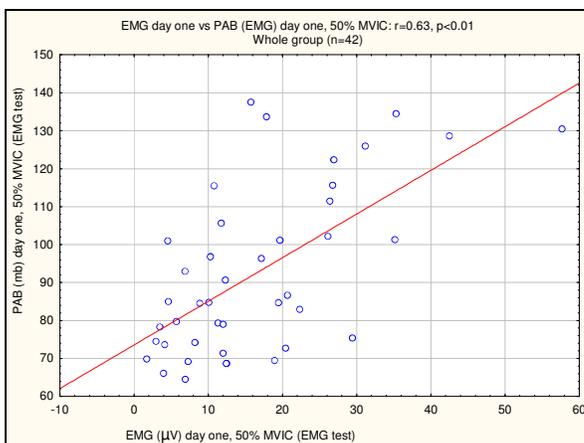


Figure 6.17 Day one, EMG (μV) versus PAB (mb) at 50% MVIC (n=42).

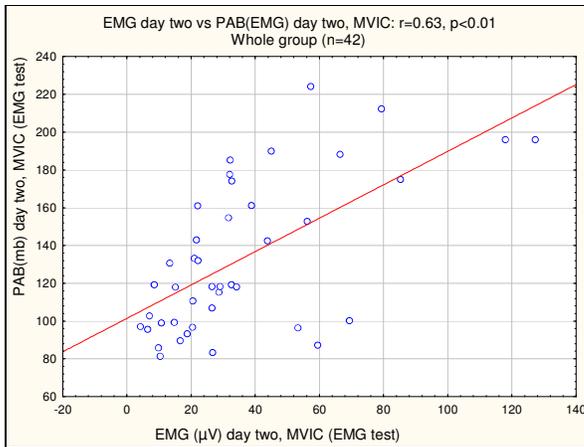


Figure 6.18 Day two, EMG (μV) versus PAB (mb) at MVIC (n=42).

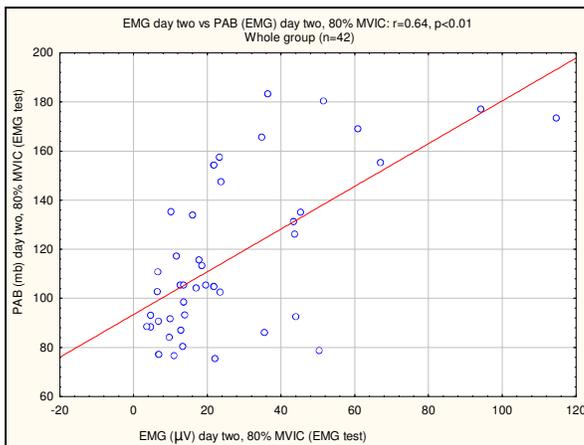


Figure 6.19 Day two, EMG (μV) versus PAB (mb) at 80% MVIC (n=42).

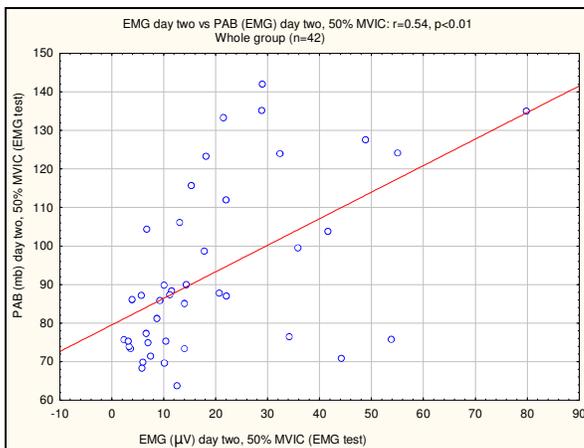


Figure 6.20 Day two, EMG (μV) versus PAB (mb) at 50% MVIC (n=42).

6.8.8 Electromyography versus pressure air biofeedback day one and two for two subgroups

Pearson correlation coefficients (r) were calculated to examine the correlation between EMG and PAB (during EMG testing) for the low back pain group over the two days (Table 6.6). According to the table for critical values of correlation coefficients (Thomas & Nelson 1985:344) and the degrees of freedom ($df=n-2$), a non-significant correlation (at the 0.05 level of significance) was found between the RMS values of EMG and PAB for all the effort levels in the low back pain group ($n=18$).

Table 6.6 Pearson correlation calculations between PAB force (mb) and RMS values of EMG (μV) for different effort levels on day one and day two for the low back pain group ($n=18$).

	DAY ONE – EMG versus PAB		DAY TWO – EMG versus PAB	
	Pearson (r)	p-value	Pearson (r)	p-value
MVIC	$r=0.26$	$p<0.29$	$r= -0.11$	$p<0.68$
80% MVIC	$r=0.16$	$p<0.52$	$r= -0.16$	$p<0.54$
50% MVIC	$r=0.18$	$p<0.46$	$r= -0.02$	$p<0.94$

Day one EMG versus PAB (during EMG testing), as well as day two measurements from 18 low back pain subjects and 24 asymptomatic subjects who participated in the between-day reliability component of the study were reported. Figures 6.21 - 6.22 are showing MVIC results of EMG versus PAB (during EMG testing) over the two days from 18 low back pain subjects.

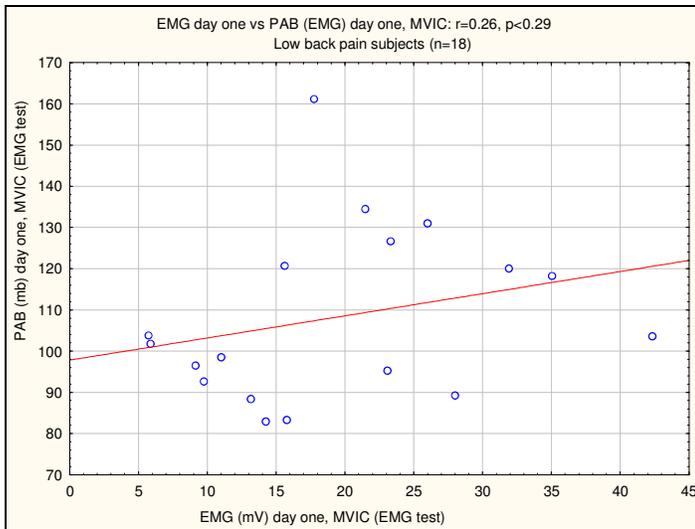


Figure 6.21 A non-significant relationship ($r=0.26$, $p<0.29$) between EMG (μV) and PAB (mb) at MVIC for low back pain subjects ($n=18$) on day one, is illustrated.

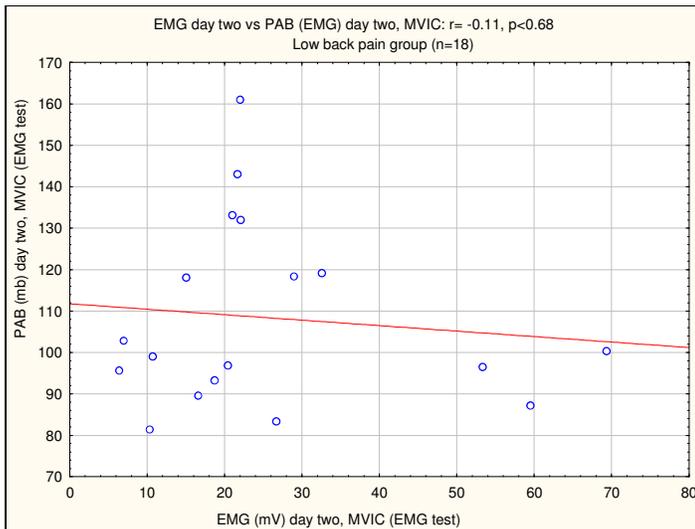


Figure 6.22 A highly, non-significant relationship ($r=-0.11$, $p<0.68$) between EMG (μV) and PAB (mb) at MVIC for 18 low back subjects on day two is illustrated, again.

With reference to the objectives set out in Chapter One, the second objective aimed to determine if a correlation exists between PAB force (mb) and EMG activity (μV) of *m. lumbar multifidus* contraction in a closed chain loaded, upright

sitting, back extension test in asymptomatic subjects and low back patients. As seen with the low back pain group, correlation calculations between EMG (RMS) and PAB for all effort levels for the 18 subjects over the two days were non-significant. With respect to the asymptomatic group (n=24), day one and two showed significant correlation results between EMG (RMS) and PAB for all effort levels (Table 6.7).

Table 6.7 Pearson correlation calculations between pressure air biofeedback and EMG (RMS) values for different effort levels on day one and day two for the asymptomatic group (n=24).

	DAY ONE – EMG versus PAB		DAY TWO – EMG versus PAB	
	Pearson (r)	p-value	Pearson (r)	p-value
MVIC	r=0.75	p<0.01	r=0.73	p<0.01
80% MVIC	r=0.76	p<0.01	r=0.72	p<0.01
50% MVIC	r=0.63	p<0.01	r=0.67	p<0.01

Figures 6.23 and 6.24 show the significant correlation results between EMG (RMS) and PAB for MVIC for both days.

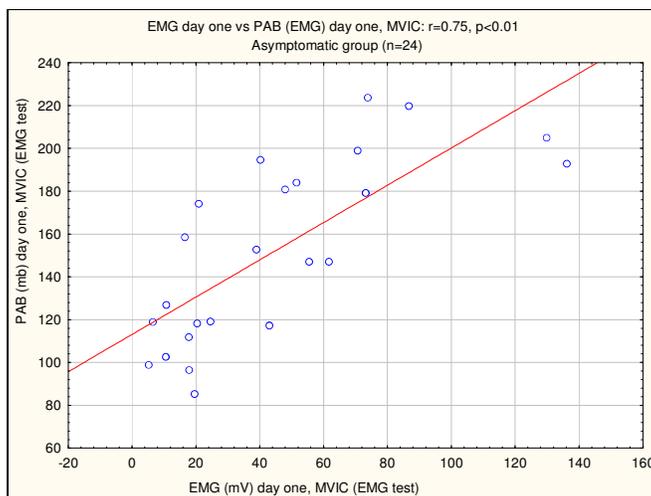


Figure 6.23 A very good linear relationship is demonstrated (r=0.75, p<0.01) between EMG (μ V) and PAB (mb) on day one, for MVIC (n=24).

In terms of the critical values of correlation coefficients (at the 0.05 level of significance) for the asymptomatic group of 24 subjects and the degrees of freedom ($df=n-2$) (Thomas & Nelson 1985:106, 344), correlation calculations between RMS values of EMG and PAB over the two days for all the isometric tests were significant ($p<0.01$) and showed good to excellent reliability.

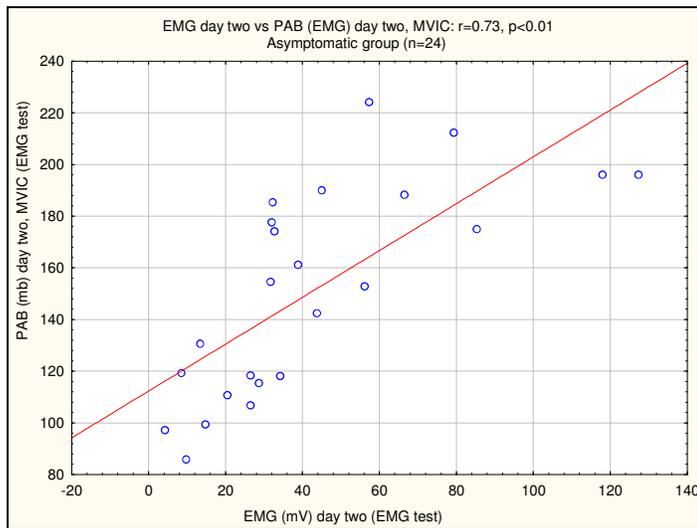


Figure 6.24 A very good linear relationship is also demonstrated ($r=0.73$, $p<0.01$) on day two, between EMG (μV) and PAB (mb) for MVIC ($n=24$).

6.9 LOW BACK STRENGTH IN ASYMPTOMATIC AND LOW BACK PAIN SUBJECTS

With respect to the difference in lumbar extension strength between the two groups, repeated measures ANOVA were performed to determine the difference in lumbar extension strength, expressed in PAB force values of mb, kilogram force (kgf) or Newton (N) between the low back pain ($n=18$) and asymptomatic group ($n=24$) during MVIC, 80% MVIC and 50% MVIC (during EMG testing). Effects were considered significant at $p<0.05$.

The reason for expressing PAB force values as mb or kgf or N has been explained in detail in Chapter Four. In short, the highly significant linear force

pressure characteristic that has been found between calibrated weights increase and corresponding PAB force increase in mb, can also be explained in that any measured increase in maximum or submaximal PAB force values signifies a proportional increase in peak or submaximal external force (kgf) or corresponding muscle strength. In terms of the SI metric conversion, the air pressure output of the PAB may be expressed as PAB force in kgf or N (Appendix E).

The lumbar extension strength of asymptomatic subjects, assessed in an upright seated, closed chain back extension test, reported a mean PAB force pressure of 150.33 mb (SD=42.44) or 110 kgf (1079 N) at MVIC. In the same seated, closed chain test, low back pain subjects could only manage a mean PAB force pressure of 108.24 mb (SD=20.99) or 57.5 kg force (564.08 N) at MVIC. In terms of displaying the PAB force in units of kilograms or Newton, the difference between the two groups calculated to 52.5 kgf or 515.02 N ($p<0.01$) in actual terms (see Appendix E). These low back strength levels represented a 100% isometric effort or MVIC for both groups (Figure 6.25).

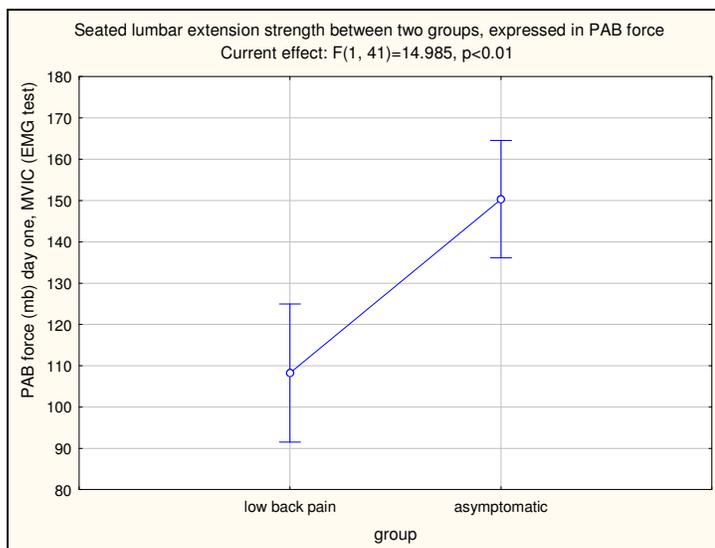


Figure 6.25 Illustrated, is a PAB force difference of 42.09 mb ($p<0.01$) between asymptomatic ($n=24$) and low back pain subjects ($n=18$) at MVIC.

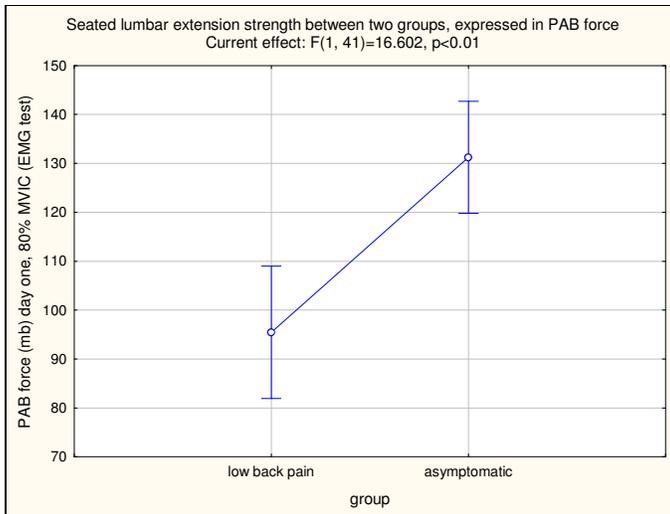


Figure 6.26 A PAB force difference of 35.8 mb ($p<0.01$) is illustrated between asymptomatic ($n=24$) and low back pain subjects ($n=18$) at 80% MVIC.

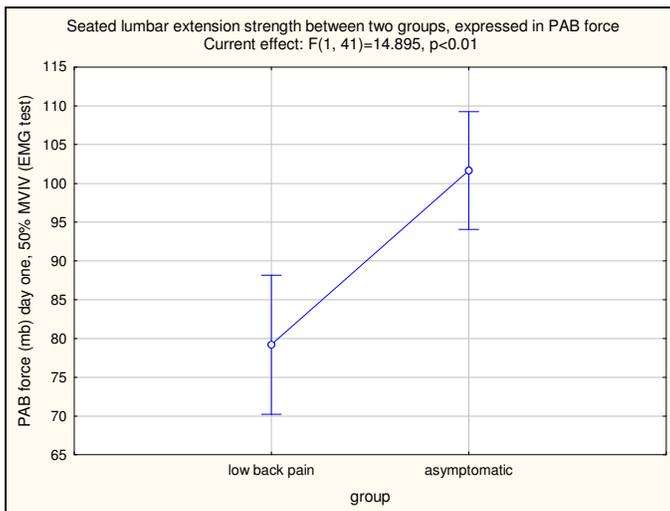


Figure 6.27 Illustrated, is a PAB force difference of 22.66 mb ($p<0.01$) between asymptomatic ($n=24$) and low back pain subjects ($n=18$) at 50% MVIC.

Eighty percent (80%) MVIC for the asymptomatic group calculated to a mean PAB force of 131.25 mb (SD=34.32) or 85 kgf, while the low back group's mean result was 95.45 mb (SD=16.91) or 42.5 kgf. This calculated to a strength

difference of 42.5 kgf or 416.92 N ($p<0.01$) between the two groups (Figure 6.26). With respect to lumbar extension strength at 50% MVIC, the asymptomatic group achieved an average PAB force of 101.65 mb (SD=22.46) or 50 kgf, while the low back group's mean result was 79.19 mb (SD=11.93) or 25 kgf, calculating to a PAB force difference of 25 kgf or 245.25 N ($p<0.01$) between the two groups (figure 6.27). These PAB force differences were calculated from the PAB/EMG tests done on day one. Similar results were achieved on the second day.

In terms of the difference in lumbar extension strength between the two groups, the low back pain group lacked significant lumbar extension strength ($p<0.01$) as measured at the L5 level, compared to the asymptomatic group. Significant lumbar extension strength differences were also reported for the 80% and 50% MVIC levels ($p<0.01$). Therefore, the PAB test appears to be reliable and valid by differentiating between the lumbar extension strength levels of low back pain and asymptomatic subjects.

CHAPTER SEVEN

DISCUSSION

- 7.1 RESULTS OF MORPHOLOGY AND THE OSWESTRY DISABILITY INDEX
- 7.2 ADJUSTED TESTING PROTOCOL
- 7.3 PRESSURE AIR BIOFEEDBACK RELIABILITY AND VALIDITY TESTS
- 7.4 ULTRASOUND AND ELECTROMYOGRAPHY RELIABILITY
- 7.5 ULTRASOUND VERSUS THE PRESSURE AIR BIOFEEDBACK TEST
- 7.6 ELECTROMYOGRAPHY VERSUS THE PRESSURE AIR BIOFEEDBACK TEST
- 7.7 LOW BACK STRENGTH - ASYMPTOMATIC VERSUS LOW BACK PAIN SUBJECTS
- 7.8 THE PRESSURE AIR BIOFEEDBACK TEST VERSUS THE BIERING-SORENSEN TEST
- 7.9 CONCLUSION

7.1 RESULTS OF MORPHOLOGY AND THE OSWESTRY DISABILITY INDEX

With reference to the subjects tested, biological and morphological results of the 43 subjects indicated similar characteristics for the different sub-groups for age and body mass index (BMI in kg/m²). They were divided into sub-groups of 21 male (mean age of 47.48 years, SD= ±20.47) and 22 female (mean age of 47.68 years, SD= ±17.08). Furthermore, they were divided in 25 asymptomatic (mean age of 46.64 years, SD= ±21.43) and 18 low back pain subjects (mean age of 48.89 years, SD= ±14.22). Mean BMI for males was 25.99 kg/m² (SD= ±2.77) and was classified as overweight, while females mean BMI of 23.96 kg/m² (SD= ±4.21) fell within the normal BMI range as indicated by the American College of Sports Medicine (ACSM, 2006:58). Furthermore, the waist-to-hip ratio (WHR) for males calculated to 0.93 (SD= ±0.07), while females scored 0.81 (SD= ±0.06). The WHR for both genders fell within normal limits (ACSM, 2006:59). Also, no significant difference ($p>0.05$) was found in terms of comparing BMI and WHR between the low back pain and asymptomatic groups, which indicated good homogeneity for comparison purposes.

With regard to the standing lumbopelvic posture, it appeared that the average male experienced a loss (decrease) of lumbar lordosis, while the average female fell within the minimum required angles for a normal lordosis. However, no significant difference ($p>0.05$) was found when comparing lumbopelvic posture angles (lumbosacral angle and standing lumbar posture) between the low back pain and asymptomatic groups. It appears that in the random selection of subjects for either the low back pain or asymptomatic group, homogenous characteristics featured strongly in both groups in terms of morphological assessments done.

Furthermore, the Oswestry Disability Index (ODI) score for the average asymptomatic person was 3.13% (SD= ±4.17) (minimal disability), while the average low back pain person scored 19.33% (SD= ±11.06), which bordered on moderate disability. The relative low percentage score for the low back pain group may be explained by the fact that all the low back pain subjects were

active members of a gymnasium or followed different exercise regimes like walking or prescribed home exercises.

7.2 ADJUSTED TESTING PROTOCOL

Although briefly mentioned in Chapter Six, it is important to explain the reason for doing an adjusted testing protocol as it pertains to the difference in numbers of subjects tested for ultrasound (US, $n=28$) and electromyography (EMG, $n=43$). On the basis of the original testing protocol, subjects were assessed by pressure air biofeedback (PAB force in mb) and electromyography (EMG in μV) as part of trial one on day one, while the US test (CSA in cm^2) and PAB test were part of trial two on day one. The same protocol was followed for day two.

After assessing 28 subjects according to the original protocol, no real increase in cross-sectional area (CSA in cm^2) of *m. lumbar multifidus* was observed on the real time US scan. Therefore, it was decided to statistically analyze the results on the basis of the non-increase in cm^2 in *m. lumbar multifidus* CSA as measured by US. The recorded results were sent for statistical analysis to the Stellenbosch University. After analyzing the results between US and PAB, it was confirmed that the correlation between these two parameters was very low during day one as well as day two. Pearson correlation calculations were non-significant between US (CSA in cm^2) and PAB force (mb) for all three isometric effort levels of maximum voluntary isometric contraction (MVIC), 80% MVIC and 50% MVIC for both days (tested at the 0.05 level of significance). Table 6.4 and Figure 6.14 illustrate this non-significant correlation between the PAB force values and US data and it was therefore decided to terminate US assessments.

Because of the non significant correlation between PAB and US, an adjusted testing protocol was suggested whereby the researcher would continue testing, but only between EMG and PAB. This was because of the higher correlation that existed between the results of PAB and EMG. Root mean square (RMS) values of EMG and PAB force values showed a stronger correlation as indicated by Pearson correlation calculations. Calculated correlations between PAB force (mb) and RMS values of EMG (μV) for the 28 subjects on day one

varied between $r=0.43$ ($p<0.01$) and $r=0.54$ ($p<0.01$) for the various effort tests, while day two correlations varied between 0.49 and 0.59 ($p<0.01$). It was therefore decided to continue with PAB and EMG assessments. The same set-up and testing protocol for PAB and EMG was continued. This explains the reported results of 28 out of the 43 subjects that were assessed by US. An additional 15 subjects were randomly selected and assessed by EMG and PAB only. This brought the total of PAB and EMG tests to 43, hence the reported results of 43 subjects.

7.3 PRESSURE AIR BIOFEEDBACK RELIABILITY AND VALIDITY TESTS

With respect to reliability measurements of the PAB force values in this study, significant intraclass correlation coefficient (ICC) agreement values of PAB force (mb) at the three isometric contraction effort levels (during EMG testing) measured over two days, taken per second over a three second period were reported ($n=43$). According to Table 6.2, ICC agreement values for PAB force scored 0.99 (with associated SEMs) for all the variables tested over the two days.

With reference to average PAB force values tested between day one and day two during EMG testing ($n=43$), an ICC agreement of 0.99 (SEM=3.73) was calculated for PAB (MVIC) values between day one and two. ICC calculations for PAB (80% MVIC) values between day one and two were 0.99 (SEM=2.55) while the ICC for PAB (50% MVIC) calculated to 0.98 (SEM=3.32), showing good reliability for all results. Also, ICC results for averaged PAB force variables tested on day one versus day two during US testing ($n=28$), calculated to 0.98 (SEM=6.08) for PAB at MVIC. At 80% MVIC the ICC agreement was 0.91 (SEM=10.71) while 50% MVIC scored 0.98 with a SEM=3.21. These results indicated good reliability for the PAB force values.

Furthermore, Pearson correlation calculations for 28 of the 43 subjects who participated in the trial-to-trial tests on day one (PAB and EMG test, trial one versus PAB and US test, trial two) showed significant results that varied between 0.97-0.99 ($p<0.01$) for the three isometric effort tests done (see

Paragraph 6.8.3). Day two again reported significant results that varied between 0.91-0.97 ($p < 0.01$) for the three effort tests done, indicating good reliability for the trial-to-trial PAB test. Therefore, the very good reliability results indicated that the variation between PAB force variables was very small because of the high agreement found between measures.

Validity assessments and results of the PAB device have been discussed in detail in Chapter Four. However, only a brief discussing of the results will be given. Pearson correlation coefficients (r) indicated that a significant linear relationship emerged between air pressure output (PAB force in mb) and the whole range of applied external forces (calibrated kg weights) in the two calibration tests done on day one ($r = 0.995$, $p < 0.01$) and day two ($r = 0.998$, $p < 0.01$). Secondly, the ICC agreement result of the PAB force versus PAB force values of the two calibration tests was also calculated (Figure 4.8). The ICC calculation of 0.997, with a small SEM of 1.55 mb between the average PAB force values calculated over the two days in relation to calibrated weights, indicated very good reliability of the PAB device.

Finally, the results of PAB force (mb) and applied external force comparisons (calibrated weights in kg) demonstrated high agreement or validity between measures (PAB force in mb) and a criterion (calibrated weights in kg) and is in agreement with the results of a similar study where air pressure and calibrated weights were used as measurement units for neck muscle strength (Axen *et al.*, 1992:7). The very good reliability and validity results of the PAB device may also be explained by the mechanical design of the PAB device. For example, the relationship that exists between the external force applied by the subject through the fibreglass shells to the elastic ball and the internal pressure developed within the system is determined by the area of the apposition between the two rigid fibreglass shells and the PAB ball. This means that the external force is applied proportionally through the two fibreglass shells to the opposite outer surface segments of the PAB ball preventing distortion of the PAB ball and possible unreliable PAB force output results.

The implication of this linear force pressure characteristic is that a measured increase in mb PAB force signifies a proportional increase in peak external force (kg) or corresponding muscle strength. These significant results indicated that air pressure may be used as a valid and reliable medium for assessing the muscle extension strength of the lower lumbar spine.

7.4 ULTRASOUND AND ELECTROMYOGRAPHY RELIABILITY

With respect to US reliability measurements, ICC calculations of US values (CSA in cm²) during four effort levels measured on day one and on day two between left and right *m. lumbar multifidus* (n=28) were significant and varied between 0.88-0.94 with associated SEMs that varied only between 0.33-0.59 cm² (Table 6.3). This indicated good reliability. Also, average US variables (CSA in cm²) tested between day one and day two reported ICC calculations that varied between 0.73-0.86 (associated SEMs varied between 0.62-0.72 cm²), indicating that the US imaging scanner showed good reliability in measuring *m. lumbar multifidus* CSA in cm².

Firstly, the use of the Toshiba 6000C Ultrasound imaging scanner that holds international accreditation and requirements for medical systems (Toshiba, 2009), may have contributed to the reliability of the US results and secondly, a highly skilled diagnostic radiologist scanned and measured the CSA of *m. lumbar multifidus* at lumbar level five (L5).

Reliability measurements of root mean square (RMS) values of EMG (μ V) in 43 subjects showed significant ICC agreement results that varied between 0.82-0.92, with associated SEMs that showed slight variation of 1.55-12.67 μ V (see Table 6.1). These results reflect the RMS values of EMG during four effort levels measured on day one and two between left and right *m. lumbar multifidus* over a three second period. Also, significant ICC agreement results of average RMS values of EMG day one versus EMG day two were reported. The ICC calculations varied between 0.79-0.89 with associated SEMs that varied only between 1.85-12.54 μ V, indicating good reliability for all results (Paragraph 6.8.4). The Noraxon Myotrace 400 electromyography device might have

contributed to the good reliability of the EMG results because of the credibility of the Noraxon EMG and Sensor Systems that are used in more than 350 leading international universities, clinics, research laboratories and training centres in over 30 countries (Noraxon, 2008).

7.5 ULTRASOUND VERSUS THE PRESSURE AIR BIOFEEDBACK TEST

When CSA (cm^2) of US *m. lumbar multifidus* was tested against PAB force values (mb), a non-significant correlation (tested at the 0.05 level of significance) was calculated between the CSA (cm^2) of the *m. lumbar multifidus* and respective PAB force (mb) values (Chapter Six, Table 6.4). This did not necessarily mean that the US imaging scanner is not reliable in measuring CSA of *m. lumbar multifidus*. It arguably demonstrated the reliability of the US scanner to measure the “non-contraction” of *m. lumbar multifidus* at the different isometric effort levels accurately over the two days. Before discussing the non-significant increase in CSA of *m. lumbar multifidus*, it is necessary to address the discrepancy in CSA (cm^2) and PAB force (mb) results that has been shown between two subjects (no's 20 and 23) in Figure 6.14.

Figure 6.14 illustrates a significant discrepancy in CSA (cm^2) and PAB force (mb) results in relation to the linear regression line. Subject 20 represents a high PAB force value with a relative small CSA of *m. lumbar multifidus*, while subject 23 represents a low PAB force result with a relative big CSA in *m. lumbar multifidus*. These results were registered during the US versus PAB test on day two and may be explained as follows. The first subject (no. 20), a 27 year old asymptomatic woman (BMI of 22.4 kg/m^2), recorded a PAB force of 245 mb at maximum voluntary isometric contraction (MVIC) with a simultaneous US recording of 6.77 cm^2 in CSA of the L5 *m. lumbar multifidus*. Her resting *m. lumbar multifidus* CSA measured 7.04 cm^2 on the US image. Analysis of her day one values showed a PAB force of 220.17 mb at MVIC, with an *m. lumbar multifidus* CSA recording of 6.20 cm^2 . Her resting *m. lumbar multifidus* CSA measured 6.12 cm^2 on day one.

The second subject (no. 23), a 36 year old asymptomatic woman (BMI of 19.7 kg/m²), recorded a PAB force of 100 mb at MVIC with a CSA contraction of 10.72 cm² of *m. lumbar multifidus*. Her US *m. lumbar multifidus* CSA at rest measured 8.35 cm². Day one US analysis showed a CSA contraction of 8.63 cm² at a PAB force of 104 mb (at MVIC), while her resting value measured 7.78 cm². Similar *m. lumbar multifidus* CSA (cm²) measurements have been reported by Hides (2005:156), Stokes *et al.* (2005:120-122) and Lee *et al.* (2006:2259-2260). The discrepancy in these two results presented itself as a paradox, e.g. subject one presented with a relative small *m. lumbar multifidus* CSA contraction of 6.77 cm², but with a strong lower back extension result of 245 mb. Subject two presented with a relatively large *m. lumbar multifidus* CSA contraction of 10.72 cm², with a weak lower back extension result of 100 mb. Also, it should be mentioned that the 27 year old woman has been a serious gym trainer, arguably the reason for her strong lower back, although with a smaller *m. lumbar multifidus*. The 36 year old woman has been a long distance runner for years. Her weak low back extension strength, but larger *m. lumbar multifidus* CSA at L5, may be due to a lack of strength training in a predominantly cardiovascular training programme.

The smaller and bigger CSA in the L5 *m. lumbar multifidus* between these two women may also be explained as a purely biological size difference in their L5 *m. lumbar multifidus*. Finally, the serious gym trainer arguably knew how to use her lower back extension muscles in the upright seated, closed chain PAB test (similar to the dead lifting technique), hence the high PAB force result for this subject.

A few US reliability studies have been done on *m. lumbar multifidus* thickness and CSA. It was found that researchers have done assessments in different testing positions and that they measured either *m. lumbar multifidus* muscle thickness (in cm) or CSA (cm²) but they did not quantify low back muscle strength (Hides *et al.*, 1992:19; Hides *et al.*, 1995:54; Hides, 2005:154-155; Stokes *et al.*, 2005:125; Lee *et al.*, 2006:2261; Vasseljen *et al.*, 2006:911; Kiesel *et al.*, 2007:164). However, in this study the US assessment was done in an upright seated, closed chain PAB test, the CSA (cm²) of *m. lumbar multifidus*

was measured at lumbar level five (L5) and low back strength was quantified in mb PAB force. However, the non-significant correlation ($p > 0.05$) between the CSA increase in L5 *m. lumbar multifidus* and PAB force data in this study (Table 6.4), indicated that the US measurement of CSA changes in *m. lumbar multifidus* may not be reliable to indicate the strength contraction of *m. lumbar multifidus* in an upright seated, closed chain PAB test.

It should be explained that subjects were sitting in an upright, neutral spine position, which allowed *m. lumbar multifidus* to contract for segmental and postural support as indicated in the study of Lee *et al.* (2006:2261). The study of O'Sullivan *et al.* (2006:E707) also reported significantly greater EMG muscle activity of *m. lumbar multifidus* compared to slump sitting. This may suggest that *m. lumbar multifidus* was already in contraction (increased CSA) when the subjects started the upright seated, closed chain PAB test, the reason for insignificant CSA increase in the L5 *m. lumbar multifidus* in this study. Hodges *et al.* (2003:268) in their study also postulated that an upright position may cause an initial increase in *m. lumbar multifidus* activity to create a protective trunk splinting response. Because the *m. lumbar multifidus* is active in the upright posture, reflecting its tonic postural role (Hides, 2005:63), it contributes significantly to controlling lumbar lordosis (O'Sullivan *et al.*, 1997:2964; Hides, 2005:68-70) as well as controlling stability of the lumbar segment (Panjabi *et al.*, 1989:194; Kay, 2001:33).

A further explanation may concern the functional subdivision between back muscles. For instance, to train stabilisers like *m. lumbar multifidus* in their holding and controlling capacity, the magnitude of resistance must be at least 30% of maximum contraction, while higher levels of activity are required to optimize the torque-producing muscles (Danneels *et al.*, 2002:18). In the upright seated PAB test, *m. lumbar multifidus* may have already reached 30% of maximum contraction for its lumbar holding and controlling effect before the PAB test started. Furthermore, Kiesel *et al.* (2008:136) mentioned that there is a structural and functional differentiation between the deep and superficial fibres of *m. lumbar multifidus*. This may allow the deep *m. lumbar multifidus* to act as a lumbar stabiliser, while the superficial *m. lumbar multifidus* may act as a

lumbar extensor. This dual or structural-functional contraction of *m. lumbar multifidus* may have contributed to the non-increase in CSA of this muscle.

These studies support the view of Du Toit (2008) that the dual anatomical contraction of *m. lumbar multifidus* may create a “torsion contraction” which may not increase *m. lumbar multifidus* CSA as seen on the US scan results in this study. This may also explain the relative unchanging geometry of *m. lumbar multifidus* through a range of postures in a three-dimensional study done by McGill (1991:813). According to Hides (2005:68), this unchanging geometry may indicate that the purpose of this muscle is to adjust vertebrae with small movements rather than to function as a prime mover.

Lee *et al.* (2006:2258) hypothesized that changes in the CSA of a muscle, with reference to *m. lumbar multifidus*, can be a good indicator of showing the contractile function of the back muscle. In assessing the CSA of *m. lumbar multifidus* in different postures, they found that the sharp increase in CSA at standing suggested that an increase of force has been exerted by *m. lumbar multifidus* to stabilize the lumbar segments at upright postures. This result was consistent with the study of Bogduk *et al.* (1992:897). However, this increase in CSA was reflected only in the postural and stabilizing contraction of *m. lumbar multifidus* from prone lying to upright standing. When a lumbar extension strength test (PAB test) was applied to the upright posture, as in this study (n=28), no further CSA increase of *m. lumbar multifidus* (at L5) took place when compared to the resting CSA as measured by US in upright sitting (see Table 6.4).

Therefore, the result of this study has indicated that the contractile function of the *m. lumbar multifidus* in an upright seated back extension test did not increase its CSA. This non-increase in CSA may not be a good indicator of the statement by Bogduk *et al.* (1992:897) and Lee *et al.* (2006:2258) that: “the maximum force exerted by a muscle is proportional to its size, including CSA.”

Studies of *m. lumbar multifidus* CSA measured in upright sitting during low back extension tests are scarce (Kiesel *et al.*, 2008:136). This needs to be

investigated considering that *m. lumbar multifidus* plays such an important role in lumbar segmental stability, postural control, and that the biggest muscle bulk of *m. lumbar multifidus* overlays the L5 segment which also has the highest incidence of pathology in low back disorders (Panjabi *et al.*, 1989:194; Kay, 2000:104; Kay, 2001:33; Hides, 2005:66, 68; O'Sullivan *et al.*, 2006:E711). Although no CSA increase occurred in the *m. lumbar multifidus*, isometric extension strength of the lower lumbar muscle at L5 was significant (as indicated by increased EMG activity and PAB force values) in the asymptomatic and low back pain groups. This increase of the lower lumbar (L5) isometric extension strength may be reflected in the structural and functional differentiation between the deep and superficial fibres of *m. lumbar multifidus* (Kiesel *et al.*, (2008:136). This may allow the deep *m. lumbar multifidus* to act as a lumbar stabiliser while the superficial *m. lumbar multifidus* may act as a lumbar extensor.

7.6 ELECTROMYOGRAPHY VERSUS THE PRESSURE AIR BIOFEEDBACK TEST

When comparing the RMS values of EMG (μV) and PAB force results (mb) for the whole group ($n=42$), a moderately significant result ($p<0.01$) for all effort levels was achieved (Table 6.5 and Figures 6.15-6.20). This moderate correlation between EMG and PAB force for the whole group ($n=42$) was arguably due to the low back pain group's ($n=18$) highly non-significant EMG/PAB force correlation ($p>0.05$) when compared in sub-groups.

More specifically, the *m. lumbar multifidus* EMG activity during maximum and sub-maximum isometric contractions in the low back pain subjects indicated a significant non-linear relationship between the bioelectrical output (μV) and the PAB force output (mb) of the *m. lumbar multifidus* in the upright seated, closed chain PAB test (see Table 6.6 and Figures 6.21-6.22). Clinically, it may be explained that 66.7% (12 of 18 subjects) of the low back pain group experienced unilateral denervation of *m. lumbar multifidus* as shown in their real-time EMG graphs. One such example of a low back patient's EMG real-time graph is illustrated in Figure 6.2b, showing the EMG imbalance between

the right and left *m. lumbar multifidus*. In this case, the left *m. lumbar multifidus* recorded more EMG activity than the right *m. lumbar multifidus*.

Various other studies have also reported findings of decreased *m. lumbar multifidus* activation (Haig *et al.*, 1993:482; Hides, 1994:165; Larivière *et al.*, 2003:314, Hides, 2005:154). Furthermore, Hyun *et al.* (2007:E598) reported asymmetric atrophy of *m. lumbar multifidus* in patients with unilateral lumbosacral radiculopathy, and contributed it to the denervation of the lumbar multifidi. The study of Wallwark (2006) also indicated that chronic low back patients have poorer muscle activation and greater muscle wasting of the multifidi at the lower lumbar levels compared to normal subjects and highlighted the importance of restoring *m. lumbar multifidus* activation and size.

With relevance to the current study, Richardson (2005:93) highlighted the importance of the anti-gravity muscle support system of the trunk and limbs for joint protection of the lumbopelvic region. Furthermore, Hides (2005:124-125) reported that significant muscle atrophy is commonly observed in quadriceps depth, mainly *m. vastus medialis* that functions as an anti-gravity muscle. This anti-gravity muscle may have a greater amount of type 1 fibres that are most vulnerable to atrophy and dysfunction during periods of immobilization, and has also been reflected in the *m. lumbar multifidus*. It may explain the finding of the non-significant bioelectrical output of the *m. lumbar multifidus* in the 18 low back patients in this study, caused by reflex inhibition and which has similar muscular characteristics as *m. vastus medialis*. It may appear that the *m. lumbar multifidus* has an anti-gravity function (Hides, 2005:70-72) to control upright posture, to maintain neutral lumbar lordosis and to provide lumbar segmental stability during dynamic movements.

Therefore, to achieve optimal bioelectrical activation of *m. lumbar multifidus*, it is important to be aware of the dynamics of the anti-gravity muscle system with reference to the local and one-joint (anti-gravity) or weightbearing muscles. Hodges and Richardson (1993:57) in their EMG study reported that co-activation of *m. adductor magnus* (one-joint hip muscle) and *m. vastus medialis oblique* (local muscle of the knee) was significantly higher in weightbearing

(closed chain) than non-weightbearing (open chain) loading. Little research has focused on muscle force production of the anti-gravity extensors in an erect loaded posture or closed chain loading, although according to Richardson (2005:96-97), Tesch has shown the high use of anti-gravity muscles in closed chain loading positions. Therefore, it is important to develop a more functional closed chain, back extension strength test to optimise the recruitment of the anti-gravity lumbar extensor muscles. This optimal recruitment of anti-gravity lumbar extensor muscles (*m. lumbar multifidus*) might have been created in the upright seated, closed chain PAB test with respect to the significant correlation that was found between the RMS values of EMG (μV) and PAB force (mb) results in the asymptomatic subjects.

When the asymptomatic group's ($n=24$) data was analyzed, the correlation between EMG activity and PAB force indicated a significant linear relationship ($p<0.01$) as reported in Table 6.7 and Figures 6.23-6.24. Similar linear relationships between increased EMG and increased muscle strength were reported in the studies of Leisman *et al.* (1995:973), Ng *et al.* (1997:959), Arnall *et al.* (2002:761) and Humphrey *et al.* (2005:181). Therefore, the difference in *m. lumbar multifidus* EMG activity and PAB force between the low back pain and asymptomatic groups may indicate a normal stabilisation or anti-gravity contraction of *m. lumbar multifidus* in the asymptomatic group (Hides, 2004: 72; Richardson, 2005:94) and dysfunction due to bioelectrical denervation of *m. lumbar multifidus* in the low back pain group (Biedermann *et al.*, 1991:1179; Kay, 2001:17; Fryer *et al.*, 2004:354; Hides, 2005: 125; Richardson, 2005:107).

This is the reason why this research study has focussed on the *m. lumbar multifidus*'s anti-gravity extension contraction specifically in an upright seated position. The two lumbar torque producers, the global *m. longissimus thoracis pars lumborum* (L5) and *m. iliocostalis lumborum pars lumborum* (L4), on the lateral border of the erector spinae in line with the L4 and L5 spinous processes just above the postero-superior iliac spine (Coorevits *et al.*, 2005:446; Hides, 2005:62), were excluded.

Also, there appeared to be differences (tremor appearance) in the PAB force traces as reflected on the PAB force graphs between weaker (low back pain and old persons) and stronger (asymptomatic) subjects tested. An example is the difference in the traces of PAB force graphs between a 42 year old asymptomatic subject and a 49 year old low back pain subject (Figure 6.1a and Figure 6.2a). This may indicate possible motor control dysfunction of *m. lumbar multifidus* in the low back pain patient (Hodges, 2005:28; Richardson & Hides, 2005:88), as well as in older persons (also see tremor traces in Figure 6.3a), but this clinical result has not been investigated in this study. Furthermore, *m. lumbar multifidus* imbalance (left > right) in respect of muscle activation at the L5 level was also indicated in both the EMG graphs of the low back pain and older subject (Figures 6.2b and 6.3b). This may indicate unilateral motor control dysfunction that may lead to possible asymmetrical atrophy of the L5 *m. lumbar multifidus* (Hides, 2005:159; Kiesel *et al.*, 2007:161). Again, this was not investigated in this study.

In summary, the international standard measurement tool for assessing muscle activation is kinesiological EMG (Soderberg and Knutson, 2000:486; Konrad, 2005; Kiesel *et al.*, 2007:162 and Konrad, 2008). Therefore, the significant correlation between EMG (μV) and PAB force (mb) in *m. lumbar multifidus* contraction in this study indicated that the PAB device may be recommended for measuring the back extension strength contraction of the lower lumbar spine in the upright seated, closed chain PAB test.

7.7 LOW BACK STRENGTH - ASYMPTOMATIC VERSUS LOW BACK PAIN SUBJECTS

The result in low back strength levels between asymptomatic and low back pain subjects in this study indicated that the low back pain group lacked significant lower back strength ($p < 0.01$), measured at lumbar level five (L5) compared to the asymptomatic group (see Figures 6.25-6.27). Low back extension strength results have been expressed in PAB force (mb), kilogram force (kgf) and/or Newton (N) as explained in Chapter Six.

Significant differences ($p < 0.01$) in lumbar extension strength between low back pain and asymptomatic subjects were reported for PAB force (mb) at all effort levels (MVIC, 80% MVIC and 50% MVIC). The biggest difference in lumbar extension strength between the two groups was 52.5 kgf or 515.02 N ($p < 0.01$). This was recorded at MVIC. Significant strength differences ($p < 0.01$) between the two groups were also recorded at 80% and 50% MVIC. These results could be explained in parallel with the EMG versus PAB force results in that increased muscle contraction (increased EMG activation) signifies a proportional increase in muscle strength expressed in the proportional increase of PAB force.

The difference in lumbar extension strength between asymptomatic and low back pain subjects indicated a normal stabilisation contraction of *m. lumbar multifidus* in the asymptomatic group (Hides, 2005:72; Richardson, 2005:94). The significant weaker lumbar extension strength in the low back pain patients may be explained as dysfunction due to neuromuscular denervation of *m. lumbar multifidus* in the low back pain group (Biedermann *et al.*, 1991:1179; Kay, 2001:17; Fryer *et al.*, 2004:354; Hides, 2005:125; Richardson, 2005:107). Therefore, the PAB test appears to be reliable and valid by differentiating between the low back strength levels of low back pain versus asymptomatic subjects ($p < 0.01$).

7.8 THE PRESSURE AIR BIOFEEDBACK TEST VERSUS THE BIERING-SORENSEN TEST

It is only now possible to compare the PAB test with the Biering-Sorensen test with respect to the results that have been reported at the completion of this study. In comparing these two tests, it is important to assess them according to the five components (as mentioned in Chapter Five, Paragraph 5.1) that are important to adhere to when testing subjects (Helewa *et al.*, 1981:353; Helewa *et al.*, 1986:1044; Matheson *et al.*, 1993:66). Testing methods or devices that comply with these five components are considered reliable and valid.

- Safety: the evaluation should be completed without any risk of injury to the subject.

The Biering-Sorensen test:

Firstly, with reference to the Biering-Sorensen test, the injury mechanism of an extension injury to the lumbar spine can be better understood if we are more aware of the muscle activation levels and the resultant spinal load created by this movement. Callaghan *et al.* (1998:16) and McGill (2007:91) reported the compression load of performing an upper body extension exercise with legs fixed and the cantilevered upper body extending over a bench or roman chair (open chain loading). This extension movement activates the thoracic and lumbar portions of *m. longissimus* and *m. iliocostalis* (four extensors) which impose over 4 000 N (397 kg) on the lumbar spine. These calculations were based on the virtual spine model of the McGill group (McGill & Norman, 1985:883; McGill & Norman, 1986:666; Cholewicki & McGill, 1996:13; McGill, 2007:16-21). Therefore precaution should be taken when prescribing this type of test to patients.

Secondly, the Biering-Sorensen test can be classified as an open kinetic chain test where the body part or limb is free to move. Also, an open kinetic chain test is typically non-weightbearing with the movement occurring around a joint. If there is weight applied, it is applied to the distal portion of the limb or body part (Richardson, 2005:95-96, Wikipedia, 2009). According to the recruitment patterns of muscle synergists in open chain loading (Richardson, 2005:95-96) and the load of 4 000 N (397 kg) that is imposed on the lumbar spine (McGill, 2007:16-21), it may not be advisable or safe to test the low back patient in the prone, back extension holding position.

Thirdly, with reference to the stages of exercise management as described by Richardson *et al.* (2005:179), the Biering-Sorensen test may only be applied to the last stage of the exercise rehabilitation management protocol and is contraindicated for the earlier and safe assessment of the low back pain patient.

Fourthly, because the Biering-Sorensen test is testing the muscle endurance component of the lower back, it can place the patient at risk of injury since studies of Roy *et al.* (1990:463), Biedermann *et al.* (1991:1179) and Danneels

et al. (2002:17) have demonstrated that the *m. lumbar multifidus* showed greater fatigue rates in subjects with low back pain compared to normal subjects.

Fifthly, the parallelogram of vector forces for the Biering-Sorensen test has indicated that the resultant vector force intersects with the subject's trunk at the eleventh thoracic level (T11) and that the long moment arm, measured from *m. teres major* to the hip, may increase loading on the lumbar spine (Figure 5.5). Therefore, the Biering-Sorensen vector parallelogram indicates a high-risk test for the low back pain patient.

The PAB test:

Firstly, with reference to the PAB test, it can be classified as a closed kinetic chain test, where the feet remain in constant contact (fixed) with the ground, or the base of a testing machine. A closed kinetic chain test is typically a weightbearing test which involves joint compression where proximal and distal body segments move together to load longitudinally through the body and the feet (Richardson, 2005:95-96, Wikipedia, 2009). According to the recruitment patterns of the muscle synergists in closed chain loading (Richardson, 2005:96), it may be advisable to test the low back patient in the upright seated, closed chain PAB test without any risk to the low back patient.

Secondly, with reference to the stages of exercise management as described by Richardson *et al.* (2005:179), the PAB test may be applied at the end of the first stage of the exercise management protocol, plus the other two stages and is therefore indicated for the safe assessment of the low back pain patient in all three stages.

Thirdly, the parallelogram of vector forces for the PAB test indicated that the resultant vector force intersects with the subject's lumbar spine at the L4 level and that the short moment arm, measured from the subscapula to the hip joint, may minimize the loading on the lumbar spine (Figure 5.4) especially in the

closed chain loading PAB test. Therefore, the PAB vector parallelogram indicates a low-risk test for the low back pain patient.

- Reliability: the apparatus, as well as the examiner, must be able to repeat the test with consistent results.

The Biering-Sorensen test:

According to the review of Demoulin *et al.* (2006:47), the Biering-Sorensen test allows for the reproducible evaluation of the isometric endurance of the lower back.

The PAB test:

According to this research study, the trial-to-trial and day-to-day evaluations of the PAB test showed excellent reliability by repeating all the different tests with consistent results.

- Validity: each test needs to be specific in what it is testing and the results must be able to exactly measure that.

The Biering-Sorensen test:

The Biering-Sorensen test is testing the muscle endurance component of the lower back and therefore cannot quantify the low back muscle strength developed by the subject (Demoulin *et al.*, 2006:47). By not quantifying low back muscle strength, it is impossible to know at what effort level the low back patient is extending the lumbar spine. Also, patients are commonly informed that they have improved because they are able to “successfully” perform more advanced exercises. Precisely how clinicians are making such judgments without quantifying the effects (muscle strength improvements) of rehabilitation exercise programmes is not clear (Hagins *et al.*, 1999:547). The Biering-Sorensen test may not qualify as a valid test.

However, research has shown that the Biering-Sorensen test may discriminate between asymptomatic and low back pain subjects and may predict the occurrence of low back pain in the future (Demoulin *et al.*, 2006:47), which may increase the validity of the test.

The PAB test:

The PAB test is also a mechanical method of testing, which has a disadvantage in that it may not allow the specific investigation of particular muscles of the lower lumbar spine to be tested (Hides, 2005:150). However, the closed chain loaded PAB test has arguably applied an important concept of the anti-gravity exercise model of Richardson (2005:102) to the test in that closed chain loading is likely to facilitate weight bearing muscle function and “turn off” the more active non-weight bearing muscle system thereby isolating the particular muscle (*m. lumbar multifidus*) to be tested in the PAB test. Also, this research study has attempted to isolate the *m. lumbar multifidus* as far as possible according to the clinical and biomechanical approaches of Hermann & Barnes (2001:971), Hodges (2003:245), Richardson *et al.* (2005), O’Sullivan *et al.* (2006:E707) and McGill (2007), as well as many other researchers as set out in the application of scientific principles in Chapter Five. Furthermore, results of this research study showed a significant correlation ($p < 0.01$) between EMG (increased bioelectrical activity of *m. lumbar multifidus*) and PAB force (increased air pressure in mb, kgf or N), reflecting the corresponding increase in lumbar extension strength for all the tests done over the two days. The PAB test may therefore qualify for validity.

- Practical: is it cost-effective in terms of time efficiency?

The Biering-Sorensen test:

The Biering-Sorensen test is a rapid and simple test with respect to time taken to set up and do the test (Demoulin *et al.*, 2006:47). Testing equipment (bench or examination bed) is not expensive but may not be too practical to carry. However, the Biering-Sorensen test may be considered to be a practical test.

The PAB test:

The PAB test appears to be a rapid and simple test to do. Testing equipment is not expensive and is portable to take to any venue (clinical office setting, home, gymnasium, etc) to assess lumbar extension strength. The PAB test may be considered to be a practical test.

- Versatility: how applicable are the results of the evaluation?

The Biering-Sorensen test:

The Biering-Sorensen test results are applicable to the isometric endurance of the trunk extensor muscles (thoracic area included). Also, this test discriminates between low back pain and asymptomatic subjects and may predict the occurrence of low back pain in the near future (Demoulin *et al.*, 2006:47).

However, there are drawbacks that may influence the versatility of the Biering-Sorensen test and needs mentioning again e.g. the fact that females perform better than men, the influence of body weight on the trunk holding position, the unknown contribution of the hip extensor muscles, the lack of a standardized testing protocol and the decreased motivation factor caused by pain (Demoulin *et al.*, 2006:43).

The PAB test:

On the other hand, the PAB test results are applicable to the isometric extensor strength of the lumbar spinal muscles. It discriminates between the lumbar extensor strength of the low back pain and the asymptomatic subjects as reported in this study. Apart from discriminating between low back pain and asymptomatic lumbar extension strength, it also allows one to quantify maximum and sub-maximum strength levels for subjects to monitor themselves against, while training. It allows for the safe prescription of exercise rehabilitation programmes. Lumbar muscle strength results are immediately

available (biofeedback) while the subject is tested or trained and may help with the motivation factor. It allows for a standardized testing protocol to be set up.

However, the PAB test has only been studied in this research study and needs to be investigated further to gain scientific standing and reference for assessing isometric extension strength of the lumbar spine in low back pain and asymptomatic subjects.

7.9 CONCLUSION

The effects of the PAB test were tested against the 0.05 level of significance. This decision was taken because the PAB test was developed as a low risk, lumbar spine extension test that would not incur injury to low back patients or asymptomatic subjects. The results of the PAB test differentiated between low back pain and asymptomatic subjects lower lumbar extension strength (significant at $p < 0.01$), without any risk to the subject. Therefore, the PAB device appears to be reliable and valid and may be useful as a clinical testing instrument in the assessment of isometric extensor strength of the lower lumbar spine. Also, the findings of the PAB test indicated that the method used, in this case an upright seated, closed chain loaded test to assess lumbar extension strength at lumbar level five (L5), may be critical to determine the exact muscles measured in the lumbar spine.

Finally, with respect to the null and alternative hypotheses, the findings of this study have indicated that the null hypotheses is to be rejected in favour of the alternative hypotheses.

CHAPTER EIGHT

RECOMMENDATIONS AND LIMITATIONS

- 8.1 SUMMARY
- 8.2 RECOMMENDATIONS
- 8.3 LIMITATIONS

8.1 SUMMARY

Worldwide epidemiological findings strongly indicate low back pain as a growing epidemic despite the latest diagnostic and treatment methods used (Jellema *et al.*, 2001:377; Woolf & Pfleger, 2003:646; Kopec *et al.*, 2004:70; Frost & Sullivan, 2006; Dagenais *et al.*, 2008:9). From this clinical problem, a need arose to quantify lumbar muscle performance for the safe monitoring of assessments and rehabilitation programmes. However, muscle strength assessment devices in rehabilitation medicine have been too large and expensive to operate (Gubler-Hanna *et al.*, 2007:920), while poorly validated and unreliable testing devices have highlighted the need for new muscle testing technology (Alexander & Clarkson, 2000:53; Tousignant *et al.*, 2001:235).

Therefore, the need for portable, valid and reliable muscle testing devices for the spine have been growing over the last few decades. This is the reason why a few researchers have developed new or added improvements to such measuring instruments (Helewa *et al.*, 1981:353; Giles, 1984:36; Helewa *et al.*, 1986:1044; Helewa *et al.*, 1990:966; Axen *et al.*, 1992:2 and Richardson *et al.*, 1992:105). This growing need for portable, clinical muscle testing devices for the spine has been further stimulated by the recent rapid advancement of computer technology that has promoted the development of highly accurate, sophisticated digital testing devices (Nobori & Maruyama, 2007:10).

Apart from the need for a clinical testing device, certain methods of assessing low back muscle performance have been widely criticised and debated. Arguably the most widely used test in published studies is the Biering-Sorensen test, but it has several major drawbacks (Demoulin *et al.*, 2006:47). This has been addressed by the development of the pressure air biofeedback (PAB) device (Patent number, P42817ZP00 MR/mjm) as reported in this study.

Finally, this study demonstrated the reliability and validity of using a compressible ball linked to an air pressure-sensor to assess the isometric strength of the paraspinal muscles at lumbar level five (L5) in a clinical office setting. The findings of this study also provide a fundamental database for

prescribing the PAB test in the assessment of lumbar extension strength with specific reference to the *m. lumbar multifidus*. Furthermore, pressure air biofeedback has been used as an applicable, reliable and alternative testing method to the Biering-Sorensen and/or isokinetics tests, in quantifying muscle strength of the L5 *m. lumbar multifidus*.

8.2 RECOMMENDATIONS

- The quantification of lumbar muscle strength measured at the L5 level is possible only if the influence of control mechanisms is considered (Larivière *et al.*, 2003:306). Arguably, this has been achieved with adequate standardization of the closed chain, upright seated PAB test. However, for the PAB test to become a standard of reference for lumbar extension strength it is recommended that future assessments do not vary from the original PAB test as has been the case in the Biering-Sorensen test (Demoulin *et al.*, 2006:47).
- It is recommended that many more asymptomatic subjects and low back pain subjects be assessed by the PAB test to be able to create quantifiable norms for males and females, asymptomatic subjects and low back pain subjects, as well as for different age groups.
- The closed chain PAB test can be recommended for the safe and reliable estimation of maximal voluntary isometric contraction (MVIC) of the L5 lumbar extensor muscles of low back pain patients. The quantification of lumbar muscle MVIC can therefore assist rehabilitation specialists in the safe prescription and monitoring of low back rehabilitation programmes.
- Pressured air was used as the testing medium in the PAB test and was found to be very sensitive or responsive when projected on the PAB force graph. This is due to the pressure transducer's sampling rate of ± 20 micro second intervals and the PC's logged recording sample rate of 10 milliseconds. This led to a difference in PAB force graphs (tremor

appearance versus stable appearance) between weaker (low back pain subjects) and stronger (asymptomatic) subjects (see Figures 6.1a, 6.2a, 6.3a and 6.4a). It is recommended to investigate the difference in appearance in PAB force graphs between low back pain subjects and asymptomatic subjects in relation to EMG and what these differences may indicate.

- According to the force vector parallelograms of the PAB and Biering-Sorensen tests, it appears that the PAB test's resultant force vector projects through the L4 level, while the moment arm is much shorter than that of the Biering-Sorensen test. This may produce less compression load on the lumbar spine. With more than 4 000 N (397 kg) compressive loading on the lumbar spine during the Biering-Sorensen test (Callaghan *et al.*, 1998:16 and McGill, 2007:91), which surpasses the compression limit of 3 300 N (325 kg) (McGill, 2007:88), it is recommended to determine the compression load on the L4-L5 lumbar spine during the PAB test.
- Ultrasound imaging has been used as a clinical diagnostic and rehabilitation tool in the assessment of *m. lumbar multifidus*, with patients in the prone lying position (Richardson & Hides, 2005:90). However, this study has indicated that US should not be recommended to assess the muscle strength contraction, with respect to *m. lumbar multifidus* CSA increase during an upright seated (neutral spine), back extension test.
- It is recommended that an isometric muscle endurance test of 30-45 seconds be done to "isolate" the anti-gravity extension and posture holding *m. lumbar multifidus* from the lumbar torque producers, the *m. longissimus thoracis pars lumborum* and *m. iliocostalis lumborum pars lumborum* (Hides, 2005:62), specifically with reference to the upright seated, closed chain loaded PAB test.
- The PAB device used in this study was a prototype. It is therefore recommended to investigate any further technical and/or physical

changes to the PAB device that will make it more user-friendly for clinical office testing. This may contribute to the standardization of the physical and technical operation of the PAB device as a final product.

- The PAB device has been designed in such a way that it can also be applied to the assessment of several muscle groups in the legs, arms and shoulders. It is subject to further research.

8.3 LIMITATIONS

- It has been acknowledged that the PAB device is an inflatable, elastic ball which is not definitively calibrated because of its elasticity. This has been contained with two rigid hemispheric shells on opposite outer surface segments of the PAB ball. It is subject to further research.
- The histological make-up of predominantly type I fibres of the lumbar multifidus muscle indicates the tonic holding and thus supportive function of this muscle but type II fibres are also vulnerable to atrophy and dysfunction of *m. lumbar multifidus* (Yoshihara *et al.*, 2001:625; Yoshihara *et al.*, 2003:494 and Hides, 2005:63,125). Therefore, a muscle endurance component (e.g. a 30 second isometric back extension test at 60%-80% MVIC) should have been included in the new PAB testing protocol to specifically assess *m. lumbar multifidus* endurance contraction. The five (5) second isometric contraction period in this study was not sufficient enough, to specifically stimulate *m. lumbar multifidus* EMG activity, to assess possible motor control dysfunction between the low back pain- and asymptomatic group.
- It has been acknowledged that the torque producers of the lower lumbar spine, *m. longissimus thoracis pars lumborum* and *m. iliocostalis lumborum pars lumborum* (Hides, 2005:62), were excluded, so it was not possible to gauge their impact or contribution (through EMG analysis) on the PAB testing results. This is subject to further research.

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¹ The Harvard Reference System has been used.

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APPENDIX A**Informed Consent Document****STATEMENT BY RESEARCH SUBJECT OR HIS/HER LAWYER**

The project information sheet has been presented to me.....
, and explained by Mr André Pienaar in
 English / Afrikaans. I the research subject have full command of the specific
 language or it has been explained to me satisfactorily. I have been given the
 chance to ask questions and my questions have been answered satisfactorily.

I hereby give consent to participate out of my own freewill in the research study
 or / I hereby give my consent that the research subject may participate in the
 study. A copy of this form has been given to me.

.....
Name of Research Subject

.....
Name of Lawyer (if necessary)

.....
Signature of Research Subject / or Lawyer

.....
Date

STATEMENT BY RESEARCHER

I declare, that all the information included in this document has been explained
 to.....and/or
 his/her lawyer.....He/she
 has been encouraged and given enough time to ask questions. The
 conversation was conducted in [English/Afrikaans/Zulu/Other] and [no translator
 was used / the conversation was translated
 in.....by.....]

.....
Signature of Researcher

.....
Date

APPENDIX A – continue

The testing programme is voluntary. You are free to deny consent if you so desire. I.....(Name of participant of Phd- Research Project of the Department of Sport Science of the University of Stellenbosch) hereby agree that the researcher conducting the tests, as well as the Sport Science Department of the University of Stellenbosch shall not be liable for any loss, or damage, or injury of any nature whatsoever by myself, or any third party, of any nature whatsoever, whether direct or indirect, resulting from the participation, facilities, benefits or arrangements which are made available to the Phd- Research participant.

I hereby declare that, to the best of my knowledge, I am currently free from any medical condition or other complaint that would preclude me from undertaking any physiological and / or anthropometrical tests.

I understand that the researcher will explain the test protocols and procedures that I will have to follow.

I have read this form carefully and fully understand the test procedures. I consent to participate in these tests.

Signed at.....on.....day of.....month, 200...

Name Participant..... Signature Participant.....

Name Witness 1.....Signature Witness 1.....

Name Witness 2.....Signature Witness 2.....

APPENDIX B

Oswestry Disability Index

Could you please complete this questionnaire. It is designed to give us information as to how your back (or leg) trouble has affected your ability to manage in everyday life. Please answer every section. Mark one box only in each section that most closely describes you today.

Section 1: Pain intensity

- I have no pain at the moment.
- The pain is very mild at the moment.
- The pain is moderate at the moment.
- The pain is fairly severe at the moment.
- The pain is very severe at the moment.
- The pain is the worst imaginable at the moment.

Section 2: Personal care (washing, dressing, etc.)

- I can look after myself normally without causing extra pain.
- I can look after myself normally but it is very painful.
- It is painful to look after myself and I am slow and careful.
- I need some help but manage most of my personal care.
- I need help everyday in most aspects of self care.
- I do not get dressed, wash with difficulty and stay in bed.

Section 3: Lifting

- I can lift heavy weights without extra pain.
- I can lift heavy weights but it gives extra pain.
- Pain prevents me from lifting heavy weights off the floor but I can manage if they are conveniently positioned, e.g., on a table.
- Pain prevents me from lifting heavy weights but I can manage light to medium weights if they are conveniently positioned.
- I can lift only very light weights.
- I cannot lift or carry anything at all.

Section 4: Walking

- Pain does not prevent me walking any distance.
- Pain prevents me walking more than 1 mile.
- Pain prevents me walking more than a quarter of a mile.
- Pain prevents me walking more than 100 yards.
- I can only walk using a stick or crutches.
- I am in bed most of the time and have to crawl to the toilet.

Section 5: Sitting

- I can sit in any chair as long as I like.
- I can sit in my favourite chair as long as I like.
- Pain prevents me from sitting for more than 1 hour.
- Pain prevents me from sitting for more than half an hour.
- Pain prevents me from sitting for more than 10 minutes.
- Pain prevents me from sitting at all.

APPENDIX B - continue**Section 6: Standing**

- I can stand as long as I want without extra pain.
- I can stand as long as I want but it gives me extra pain.
- Pain prevents me from standing for more than 1 hour.
- Pain prevents me from standing for more than half an hour.
- Pain prevents me from standing for more than 10 minutes.
- Pain prevents me from standing at all.

Section 7: Sleeping

- My sleep is never disturbed by pain.
- My sleep is occasionally disturbed by pain.
- Because of pain I have less than 6 hours' sleep.
- Because of pain I have less than 4 hours' sleep.
- Because of pain I have less than 2 hours' sleep.
- Pain prevents me from sleeping at all.

Section 8: Sex life (if applicable)

- My sex life is normal and causes no extra pain.
- My sex life is normal but causes some extra pain
- My sex life is nearly normal but is very painful.
- My sex life is severely restricted by pain.
- My sex life is nearly absent because of pain.
- Pain prevents any sex life at all.

Section 9: Social life

- My social life is normal and causes me no extra pain.
- My social life is normal but increases the degree of pain.
- Pain has no significant effect on my social life apart from limiting my more energetic interests, eg., sport, etc.
- Pain has restricted my social life and I do not go out as often.
- Pain has restricted social life to my home.
- I have no social life because of pain.

Section 10: Travelling

- I can travel anywhere without pain
- I can travel anywhere but it gives extra pain.
- Pain is bad but I manage journeys over 2 hours.
- Pain restricts me to journeys of less than 1 hour.
- Pain restricts me to short necessary journeys under 30 minutes.
- Pain prevents me from travelling except to receive treatment.

APPENDIX C**Anthropometric testing protocol for PhD research**

NAME..... AGE.....DATE.....

SCALE

- 1 Body mass =kg
 2 Body height =m **BMI** =

MEASURE TAPE (WAIST-TO-HIP RATIO)

- 3 Waist girth =cm
 4 Gluteal (hip) girth =cm **Waist-to-hip ratio** =

LUMBOSACRAL ANGLE AND STANDING LUMBAR POSTURE

- 5 Saunders digital inclinometer: L-S joint =degrees
 T-L joint =degrees

OSWESTRY DISABILITY INDEX SCORE

- 6 ODI =.....%

*Applicable only to ladies: Will you be menstruating on testing day? YES/ NO

APPENDIX D**PAB Testing Protocol**

Subject:.....Date:.....Test:.....

TEST 1: Pressure Reading (PAB)**EMG (mV)****1 Rest test in upright sitting (5sec)** L+R multifidi=.....**2 Maximum voluntary isometric contraction (MVIC) test (5sec)**1st contraction =.....mmBar L+R multifidi=.....

1min rest

2nd contraction =.....mmBar L+R multifidi=.....

Two (2) second build-up to maximum then hold for three seconds. Highest pressure of the two MVICs is taken as 100% effort. Take three (3) minute break.

3 Sub-maximal tests (50% MVIC and 80% MVIC)

One (1) isometric contraction of 5 seconds each at 50% and 80% MVIC. Two (2) second build-up to sub-maximal pressure then hold for three (3) seconds. Its important to hold contraction within 10% of calculated target pressure value. Take two (2) minute break between tests.

Pressure Reading (PAB)**EMG (activity in mV)**

50% contraction =.....mmBar L+R multifidi=.....

2min rest

80% contraction =.....mmBar L+R multifidi=.....

APPENDIX D – continue**TESTS 2/3/4: Pressure Reading (PAB) US (CSA in cm²) or EMG (mV)**

1 Rest test in upright sitting (5 sec) = L+R multifidi=.....

2 One (1) MVIC test (5sec) =mmBar L+R multifidi=.....

Two (2) second build-up to maximum pressure then hold for 3 seconds. To hold contraction within 10% of calculated target pressure value. Take 2 min break.

3 Sub-maximal tests (50% and 80% MVIC)

One (1) isometric contraction of 5 seconds each at 50% and 80% MVIC. Two (2) second build up to target pressure then hold for three (3) seconds. Two (2) minutes rest between tests.

Pressure Reading (PAB)**US(CSA in cm²) or EMG (mV)**

50% contraction =mmBar L+R multifidi=.....

2min rest

80% contraction =mmBar L+R multifidi=.....

4 Recording of data

For EMG, three (3) data values are recorded over three seconds after target pressure value has been reached. For US, one (1) data value is recorded, two (2) seconds after target pressure has been reached.

APPENDIX E

Pull-compression calibration tests done on two separate days with force conversions

CALIBRATION	DAY ONE	DAY TWO	PAB-mb	#	FORCE
Weights-kg	PAB-mb	PAB-mb	Average	PAB-mb	Newton (N)
2.5	53.8	54.8	54.3	4.3	24.53
5	57.8	58.1	57.9	7.9	49.05
7.5	59	61.3	60.2	10.2	73.58
10	63.5	63.8	63.7	13.7	98.10
12.5	64.7	66.4	65.6	15.6	122.63
15	67.7	68.5	68.1	18.1	147.15
17.5	70.8	71.1	71	21	171.68
20	73.8	73.9	73.9	23.9	196.20
22.5	76.5	76.4	76.5	26.5	220.73
25	79.2	78.9	79	29	245.25
27.5	82	81.6	81.8	31.8	269.78
30	84.2	84	84.1	34.1	294.43
32.5	86.8	86.5	86.7	36.7	318.83
35	89	88.6	88.8	38.8	343.35
37.5	92.5	91	91.8	41.8	367.88
40	94.4	94	94.2	44.2	392.40
42.5	96.5	96.1	96.3	46.3	416.93
45	98.6	98	98.3	48.3	441.45
47.5	101	99.7	100.4	50.4	465.98
50	103	101	102	52	490.50
52.5	105.2	103	104.1	54.1	515.03
55	107.6	105.2	106.4	56.4	539.55
57.5	110	107	109	59	564.08
60	112.4	109.8	111.1	61.1	588.60
62.5	114.3	111	112.7	62.7	613.13
65	116.2	113.3	114.8	64.8	637.65
67.5	119.8	115.2	117.5	67.5	662.18
70	122.2	117.5	119.9	69.9	686.70
72.5	125	119	122	72	711.23
75	128	121.2	124.6	74.6	735.75
77.5	130	123.3	126.7	76.7	760.28
80	131.5	125.5	128.5	78.5	784.80
82.5	133.5	127.1	130.3	80.3	809.33
85	135.5	129.4	132.5	82.5	833.85
87.5	137.6	131.5	134.6	84.6	858.38
90	139.4	133.7	136.6	86.6	882.90

Actual PAB mb values represent the average PAB mb value minus the 50 mb pressure (constant) inside the ball.

APPENDIX E – continue

Calibration weights – kg	DAY ONE PAB - mb	DAY TWO PAB - mb	PAB - mb average	# ACTUAL PAB - mb	FORCE CONVERSION Newton (N)
92.5	140.6	135	137.8	87.8	907.43
95	141.7	137.3	139.5	89.5	931.95
97.5	143	139	141	91	956.48
100	144.3	142.8	143.3	93.3	981
102.5	146	144.2	145.1	95.1	1005.53
105	147.8	146	146.9	96.9	1030.05
107.5	149.5	148	148.8	98.8	1054.58
110	151.8	150.5	151.2	101.2	1079.10
112.5	152.8	151.6	152.2	102.2	1103.63
115	154	153.6	153.8	103.8	1128.15
117.5	156.5	155	155.8	105.8	1152.68
120	157.2	158.2	157.7	107.7	1177.20
122.5	159.8	159.2	159.5	109.5	1201.73
125	161.4	160.2	160.8	110.8	1226.25
127.5	163	162	162.5	112.5	1250.78
130	165.2	163.9	164.6	114.6	1275.30
132.5	167	164.8	165.9	115.9	1299.83
135	168.8	165.9	167.4	117.4	1324.35
137.5	170	167.8	168.9	118.9	1348.88
140	171.8	169.6	170.7	120.7	1373.40
142.5	172.8	171.5	172.2	122.2	1397.93
145	174.5	173.5	174	124	1422.45
147.5	176.5	174.5	175.5	125.5	1446.98
150	177.8	176	176.9	126.9	1471.5
152.5	179.9	177.5	178.7	128.7	1496.03
155	180.5	179.5	180	130	1520.55
157.5	182.8	180.6	181.7	131.7	1545.08
160	184	182	183	133	1569.60

APPENDIX F

Pressure air biofeedback real time biometric recording and reporting system specifications

General

The system comprises of a PC application, connected to a pressure transducer via an interface PCB cable. The PAB real time biometric recording and reporting system was selected and designed for “strong man and old lady testing”.

PC application specs

User interface for complete control of tests, (eg starting, stopping, system configuration), subject/patient detail storage and complete reporting.

The application has been custom designed and written in C++ Builder from Borland Corporation.

All data is stored in MSSQL database.

Sample rate of logged recording = 10 milliseconds.

Connection to transducer interface is done using ethernet connection at 100 Mega bits per second.

Druck PMP 1400 pressure transducer for prototyping

The selected transducer was selected for “strong man and old lady testing”.

Manufacturer: General Electric.

Combined non-linearity, hysteresis and repeatability: ± 0.15 percent (%) typical-maximum of $\pm 0.25\%$.

Long term stability: 0.2% full scale per annum typical.

Over pressure protection. Min 2 x rated pressure.

Temperature Range: -20 degrees Celsius ($^{\circ}\text{C}$) - 80°C .

Temperature compensated with total error due to ‘out of temp range’ at full scale is 1.5%.

Pressure transducer reader and interface PCB

Embedded 32 bit microprocessor technology.

Pressure sampling: ± 20 micro second intervals.

Up to 22 bit conversion resolution (pressure to electrical level). (Relative to person tested and transducer used).

Temperature compensated: -20°C - 85°C .

Repeatability of conversion = 99.8%.

With maximum of $\pm 0.2\%$ error between samples at full scale.