USING OPTOELECTRONIC PLETHYSMOGRAPHY TO OBJECTIVELY CHARACTERISE AND MANAGE BREATHING PATTERNS IN ATHLETES WITH AND WITHOUT RESPIRATORY DISORDERS

This thesis is presented for the degree of Doctor of Philosophy
at the University of Kent

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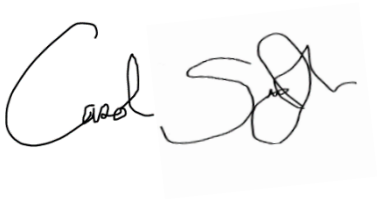
School of Sport and Exercise Sciences

University of Kent

2020

DECLARATION

No part of this thesis has been submitted in support of an application for any degree or other qualification of the University of Kent, or any other University of Institution of learning.

Signature:  Date: 14/12/2020

ABSTRACT

Optoelectronic plethysmography (OEP) is a non-invasive 3D motion capture-based system that tracks chest wall movement. The purpose of this thesis was to characterise the breathing patterns of athletes with and without respiratory disorders, namely exercise-induced bronchoconstriction and dysfunctional breathing, at rest and during exercise. OEP breath volumes agreed well with breath-by-breath gas analysis in support of previous findings. Normative ranges for OEP-derived breathing pattern parameters for a non-pathological population at rest and during exercise were presented, with candidate timing and phase angle parameters differing between rest and exercise identified (Chapter 4). It was subsequently found that phase angle breathing parameters significantly differed between a healthy population group at rest and athletes with exercise-induced bronchoconstriction (EIB) during bronchoconstriction (Chapter 5). Following this, significant differences in regional contribution and phase angle breathing parameters were found between athletes with and without dysfunctional breathing at rest and during exercise (Chapter 6).

A novel real-time OEP feedback system was developed and used as part of an acute breathing retraining intervention for athletes with dysfunctional breathing (Chapter 7). Significant reductions in phase angles values of RcAbPhase (ribcage versus abdomen), RCpAbPhase (upper ribcage versus lower ribcage and abdomen), and AbSPhase (abdomen versus shoulders) were found post-intervention indicating real-time OEP feedback may be a useful breathing retraining tool during high intensity exercise.

Although OEP has been shown to be a powerful breathing pattern measurement tool, it is not without its limitations. Therefore, potential portable alternatives to OEP were explored including an inspiratory muscle training device (Chapter 8) and an impedance pneumography (IP) system (Chapter 9). Both systems displayed some promise by correlating with the respiratory rate obtained from the OEP system at rest ($r > 0.8$; $ICC > 0.9$). For the IP, the agreement was reduced during exercise ($r > 0.4$; $ICC > 0.5$). Overall, OEP was shown to potentially be a powerful tool for diagnosing and management of respiratory disorders in athletes.

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ABBREIVATIONS

AB	Abdomen
AbCT	Abdomen volume contribution to total volume
AbHTA	Left and right abdomen hemi-thoracic asynchrony
ABP	Asynchronous breathing pattern
AbSPhase	Shoulder Phase Angle: Phase Angle between the Volume of the top row of the Pulmonary Ribcage and the Abdomen Volume (degrees)
BbB	Breath by breath
brpm	Breaths per minute
BPAT	Brompton breathing pattern assessment tool
CO_2	Carbon dioxide
COPD	Chronic obstructive pulmonary disease
DBP	Dysfunctional breathing pattern
deg	degrees
ECG	Electrocardiogram
EIB	Exercise-induced bronchoconstriction
EVH	Eucapnic voluntary hyperpnoea
FeNO	Fractional concentration exhaled nitric oxide
FEV_1	Forced expiratory volume in one second
FVC	Forced vital capacity
HBP	Healthy breathing pattern
HTA	Left and right hemi-thoracic asynchrony of the torso
HVS	Hyperventilation syndrome
ICC	Intra-class correlation coefficient
IE50	Ratio of inspiratory to expiratory flow at 50% of tidal volume
IP	Impedance pneumography
MARM	Manual assessment of respiratory motion
MRI	Magnetic resonance imaging
NBP	Normal breathing pattern
NQ	Nijmegen questionnaire
O_2	Oxygen
OEP	Optoelectronic plethysmography

OHS	Hypoventilation syndrome
PEF	Peak expiratory flow
ppb	Parts per billion
r	Pearson's correlation coefficient
RC	Ribcage
RCa	Abdominal ribcage
RCaAbCT volume	Combined abdominal ribcage and abdomen volume contribution to total
RcAbIndex	Ribcage volume / abdomen volume index
RcAbPhase	Phase angle between the ribcage volume and abdomen volume
RCaCT	Abdominal ribcage volume contribution to total volume
RCaHTA	Left and right abdominal ribcage hemi-thoracic asynchrony
RCaSPhase	Shoulder Phase Angle: Phase Angle between the Volume of the top row of the Pulmonary Ribcage and the Abdominal Ribcage Volume (degrees)
RcCT	Ribcage volume contribution to total volume
RcHTA	Left and right ribcage hemi-thoracic asynchrony
RCp	Pulmonary ribcage
RCpAbPhase	Phase angle between the pulmonary ribcage volume and the combined abdominal ribcage and abdomen volume
RCpCT	Pulmonary ribcage volume contribution to total volume
RCpHTA	Left and Right pulmonary ribcage hemi-thoracic asynchrony
RCpRCaPhase	Phase angle between the pulmonary ribcage volume and abdominal ribcage volume
RCpSPhase	Shoulder Phase Angle: Phase Angle between the Volume of the top row of the Pulmonary Ribcage and the bottom row of the Pulmonary Ribcage (degrees)
RER	Respiratory exchange ratio
RIP	Respiratory inductive plethysmography
RR	Respiratory Rate
s	seconds
SD	Standard deviation
SEBQ	Self-evaluation of breathing questionnaire
sEMG	surface electromyography
SLP	Structured light plethysmography
TBP	Thoracic dominant breathing pattern

tE	Expiratory time
tI	Inspiratory time
tI/tE	Inspiratory time / expiratory time ratio
tI/tTot	Inspiratory time / total breath time ratio
tPTEF	Time taken to reach peak expiratory flow
tPTIF	Time taken to reach peak inspiratory flow
tTot	Total breath time
V_E	Minute ventilation
Vt	Tidal volume
W	Watts

LIST OF PUBLICATIONS

Conference Proceedings:

Smyth, C., Winter, S.L., & Dickinson, J.W. 2018. Validation of breathing frequency algorithms in impedance pneumography. European Respiratory Journal 2018, 52: Suppl. 62, PA2417.

Smyth, C., Dickinson, J.W., & Winter, S.L. 2019. OEP derived thoracoabdominal movement parameters significantly distinguish between rest, various exercise intensities, and recovery. International Society of Biomechanics and American Society of Biomechanics, Calgary, Canada.

Smyth, C.M.E., Winter, S.L., & Dickinson, J.W. 2020. Optoelectronic plethysmography derived breathing parameters can differ between athletes with and without a dysfunctional breathing pattern during exercise. 2020 IEEE International Workshop on Metrology for Industry 4.0 & IoT, Roma, Italy, 2020, pp. 54-58.

ACKNOWLEDGMENTS

I would firstly like to express my gratitude and appreciation for my supervisors, Prof. John Dickinson and Dr Sam Winter. The last three years have certainly been challenging, with each year seemingly more difficult than the previous. I would not have survived without your unwavering support, guidance, and kindness, so thank you. It has been a pleasure to work with you both, I couldn't have asked for better supervisors. I would also like to congratulate John on your recent promotion to professor and Sam on your new position at Loughborough, both very well deserved.

A huge thank you to all of the SSES PhD students for your incredible kindness and friendship. Special shoutout to Will for always being my OEP guinea pig.

I would also like to thank all of the individuals who participated in my studies, this thesis would not have been possible without you.

Finally, I would like to thank all my family and friends for their continued support. I'd like to specially thank my best friend, Jennifer. Although we're in different countries, I couldn't have got through the last few years without your support. From our first day of playschool together to the last day of our PhDs, it's only apt that we finished our theses together!

Chapter 1. Introduction

1.1 Introduction

Respiratory symptoms are the most frequent complaint from athletes. However, the symptoms are not specific to one condition, which makes the diagnosis and management of respiratory conditions challenging. Exercise-induced bronchoconstriction (EIB) is a term which describes the temporary and reversible narrowing of the airways during exercise (Hull *et al.*, 2016). It is most commonly seen in athletes, with the prevalence of EIB in elite athletes reported to be up to 69% depending on the sport (Dickinson *et al.*, 2005; Levai et al. 2016). Correct diagnosis is vital due to the potential impact this condition can have on both the athlete's health and athletic performance (Price *et al.*, 2014). Despite its high prevalence, EIB is complex to diagnose as there is no gold standard diagnostic measure.

Reports suggests that approximately 20% of athletes who report significant exercise respiratory symptoms receive an inappropriate diagnosis of asthma or EIB (Dickinson *et al.*, 2005). Ansley and colleagues (2012) reported that mis-diagnosis of EIB in elite football can be up to 50%. A dysfunctional breathing pattern is a major differential diagnosis versus EIB for respiratory symptoms. Dysfunctional breathing can be used to describe a collection of breathing disorders in which individuals display chronic breathing pattern changes. The prevalence of dysfunctional breathing is approximately 8% for the general population in the UK. Approximately 29 – 80% of individuals with asthma also present with a dysfunctional breathing pattern. Currently, dysfunctional breathing is diagnosed by the exclusion of other pathologies using various breathing function measures because of this, individuals with a dysfunctional breathing pattern often have a delayed diagnosis of approximately 2 – 7 years (Depiazzi & Everard, 2016). There have been previous attempts to categorise or classify dysfunctional breathing patterns (Barker & Everard, 2015; Boulding *et al.*, 2016). However, there has not been an attempt to objectively quantify dysfunctional breathing patterns. When treated correctly, individuals with a dysfunctional breathing pattern display improvement in quality of life scores and reduced breathing difficulties during daily life activities and exercise (Hagman *et al.*, 2011; Thomas *et al.*, 2003).

Breathing pattern retraining is considered to be the first line of treatment for dysfunctional breathing (Bott, 2009). Breathing retraining aims to progressively alter and retrain an individual's breathing pattern through the use of breathing exercises (Jones *et al.,* 2013). Breathing retraining can involve techniques such as the Buteyko breathing (Bowler *et al.,* 1998), diaphragmatic breathing (Hagman *et al.,* 2011), Papworth method (Holloway & West, 2007), and yoga breathing (Singh *et al.,* 1990). However, there has been no consensus on the use of breathing exercises within clinical practice (Barker *et al.,* 2013; Jones *et al.,* 2013). Additionally, flow-incentive and volume-incentive spirometry feedback have been investigated as potential breathing training aids, with volume-incentive spirometry promoting deeper and slower breathing (de Moraes Paisani *et al.,* 2013; Tomich *et al.,* 2010). This indicates that different forms of feedback used in breathing training can have different impacts on breathing pattern.

Breathing pattern measurement systems are capable of measuring the movement of the chest wall which can include measuring circumference changes of the chest wall, measuring alterations in the cross-sectional area of the chest, measuring the linear displacements of the front of the chest, or measuring the chest wall volume change. The three most widely used breathing pattern measurement systems in clinical and research environments are respiratory inductive plethysmography (RIP), structured light plethysmography (SLP), and optoelectronic plethysmography (OEP). OEP has been established as a valid breathing pattern measurement system when compared to traditional breath-by-breath analysis (Aliverti *et al.,* 2000; Aliverti *et al.,* 2004; Cala *et al.,* 1996; Dellaca *et al.,* 2010; Vogiatzis *et al.,* 2005a). OEP also has the advantage over other systems as it has also been validated during exercise (Kenyon *et al.,* 1997; Layton *et al.,* 2013; Vogiatzis *et al.,* 2005b), and is often used as the gold standard to compare other breathing pattern measurement systems against. OEP is a non-invasive method for measuring thoracic movement. It measures the movement of the anterior and posterior chest wall using three-dimensional motion capture, allowing for the compartmentalisation of the torso into the abdomen, upper and lower ribcage.

Previously, OEP has been used to assess the breathing patterns of a wide range of populations including healthy individuals (Cala *et al.,* 1996), chronic obstructive pulmonary disease (COPD) patients (Aliverti *et al.,* 2004), asthmatics (Fregonez *et al.,* 2018), the elderly (Lunardi *et al.,* 2014), and infants (Dellaca *et al.,* 2010), to name a few. To the best of our knowledge, there have only been a few studies using OEP on individuals with asthma. OEP

was used to assess breathing pattern of asthmatics pre and post bronchoconstriction, demonstrating hyperinflation and increases in end-expiratory volume during bronchoconstriction predominantly due to ribcage volume increases (Filippelli *et al.*,2003; Gorini *et al.,* 1999). Feitosa and colleagues (2019) constructed a receiver operating characteristic curve for OEP-derived end-expiratory volume of the chest wall for children with exercise induced asthma. They achieved 93.75% sensitivity with 83.33% specificity for a cut-off point of 0.185% for end-expiratory volume. It was established that OEP can accurately be used to identify exercise induced asthma in children when compared to traditional spirometry. OEP has shown promise in identifying asthma from these limited studies, however, it has yet to be used to assess the breathing patterns of athletes with EIB or dysfunctional breathing.

Despite its wide use, OEP has a number of limitations as it is expensive, time-consuming, requires specialist expertise, and is confined to the laboratory setting. The ideal breathing pattern measurement system would be portable, allow the monitoring of breathing pattern continuously overtime in any/natural conditions and provide personalised real-time feedback to the user. Wearable technology may provide a solution to this issue. Currently, some wearable technology systems have had some success in measuring respiratory rate and tidal volume include, integrated RIP (Wilhelm *et al.,* 2003) and impedance pneumography (Mlyncazk & Cybulski 2012). However, there has been minimal investigation into measuring breathing pattern using wearable technology. The discreet continuous monitoring of breathing patterns may reveal significant diagnostic information for athletes with respiratory disorders including dysfunctional breathing and may be implemented as a breathing pattern retraining aid with real-time biofeedback.

This thesis will firstly attempt to objectively quantify, using OEP, the breathing patterns of athletes without any respiratory disease/disorder, athletes with exercise-induced bronchoconstriction, and athletes with a dysfunctional breathing pattern during rest and evoked conditions, in order to establish normative ranges for a collection of breathing parameters for each population group. It will be investigated if OEP can identify differences in both established and novel breathing pattern parameters between rest and evoked conditions within each population group, and also between non-pathological and pathological population groups specifically, individuals with EIB and dysfunctional breathing. Novel OEP parameters that are specifically related to the symptoms of those conditions will be assessed in these groups. Secondly, a novel real-time OEP feedback

system will be developed and assessed at rest and during exercise as part of an acute breathing retraining intervention for athletes with a dysfunctional breathing pattern, to determine if using real-time OEP feedback can elicit improvements in breathing pattern parameters. Finally, potentially portable and wearable alternatives to OEP, specifically an inspiratory muscle training device and a garment based impedance pneumography system, will be investigated to determine if it is feasible to further develop these systems for measuring breathing pattern parameters similar to those obtained by OEP outside of the laboratory setting.

Outline of this Thesis

The overall purpose of this thesis is to objectively characterise breathing patterns of pathological and non-pathological active populations, potentially expanding the use of optoelectronic plethysmography, and to explore potential methods of managing dysfunctional breathing in athletes.

Chapter 2

Chapter 2 provides a comprehensive literature review of the current methods of measuring respiratory function, specifically breathing patterns. It covers the current methods for measuring breathing patterns including structured light plethysmography, respiratory inductive plethysmography, optoelectronic plethysmography, and a range of alternative wearable technology.

Chapter 3

Chapter 3 provides in detail the optoelectronic plethysmography procedure used within the experimental chapters of this thesis including the equipment set up and breathing parameter calculations.

Chapter 4

Chapter 4 introduces 1) normative ranges for OEP-derived breathing pattern parameters for a non-pathological population, and 2) investigates how these breathing parameters alter during exercise.

Chapter 5

Chapter 5 firstly, investigates how OEP-derived breathing pattern parameters change during bronchoconstriction induced by eucapnic voluntary hyperpnoea in athletes with exercise-

induced bronchoconstriction, and secondly, determines if these breathing parameters differ between athletes with and without exercise-induced bronchoconstriction.

Chapter 6

Chapter 6 determines a breathing pattern exercise response in athletes with a dysfunctional breathing pattern, and investigates if OEP-derived breathing parameters differ between athletes with and without dysfunctional breathing at rest and during exercise.

Chapter 7

Chapter 7 assesses if a novel real-time OEP phase angle feedback can reduce the phase angle between the ribcage and the abdomen when compared to real-time OEP volume-orientated feedback in athletes with dysfunctional breathing at rest and during exercise.

Chapter 8

Chapter 8 investigates if OEP-derived phase angles increase with simulated dysfunctional breathing patterns, and secondly, determines if POWERbreathe outputs alter with simulated dysfunctional breathing patterns.

Chapter 9

Chapter 9 compares the respiratory rates and impedance change obtained from an impedance pneumography system to the respiratory rates and tidal volumes from both the OEP system and traditional breath-by-breath analysis.

Chapter 2. Literature Review

2.1 Introduction

The purpose of this chapter is to provide a review of the current methods of measuring respiratory function, specifically breathing patterns. A normal breathing pattern has previously been described as the pulmonary ribcage, abdominal ribcage, and the abdomen moving in synchronisation (Bradley & Esformes, 2014). Pressure gradients created by the respiratory muscles changing the volume of the lungs drives the air flow in and out of the lungs. Typically, lung function is assessed by measuring lung volumes, lung capacities, and flow rates. There are a number of parameters that can be used to assess lung function. Respiratory rate is the number of breaths taken per minute with the average resting respiratory rate for a healthy individual being 12 breaths per minute. Tidal volume is the volume of air that moves in and out of the lungs during a single breath. The average tidal volume for a healthy person at rest is ~500 ml. Minute ventilation is the total amount of air that flows in and out of the lungs in a minute. At rest, the average minute ventilation for a healthy individual is ~6000 ml/min (Stanfield, 2011). These parameters can be obtained using breath-by-breath analysis within a clinical setting. Additionally, spirometry is a simple method to test for respiratory disorders and can help to distinguish between obstructive and restrictive pulmonary diseases. Obstructive pulmonary diseases increase airway resistance, for example, asthma, while restrictive pulmonary diseases reduce lung compliance, for example, pulmonary fibrosis. Forced vital capacity and forced expiratory volume can be obtained from spirometry. A low forced vital capacity is indicative of a restrictive pulmonary disease, while a low force expiratory volume is indicative of an obstructive pulmonary disease. Normal forced expiratory volume in one second, which is a measure of the percentage of forced vital capacity that can be exhaled within one second, is ~80% (Stanfield, 2011).

Measuring lung function is important in the assessment and management of many respiratory problems in patients and athletes alike. Parameters including respiratory rate and tidal volume are fundamental in the assessment of lung function in a patient (Aliverti, 2017), for example, it has been shown that respiratory rate can predict certain adverse health incidences including cardiac arrest (Fieselmann *et al.*, 1993; Goldhill *et al.*, 2005). If respiratory rate is abnormal, it may also be used to monitor the progression of various illnesses and act as a

fundamental prognostic marker. Stable and patients at risk can be differentiated more easily using respiration rate over other measures including heart rate and blood pressure (Cretikos *et al.,* 2008). It has been shown that high risk patients can be identified up to 24 hours prior an adverse clinical event with 95% specificity using changes in respiration rate (Cretikos *et al.,* 2007). Respiration rate can be affected by numerous heart and lung diseases and thus should be monitored continuously over time in certain patient populations, for example, post-operative patients or patients with cardiopulmonary diseases, to monitor and predict trends that may potentially lead to complications or exacerbations (Churpek *et al.,* 2012). Similarly, continuous monitoring of respiration rate during sleep is essential to diagnosing and treating sleep disorders such as apnoea and hypopneas (Aliverti, 2017).

Minute ventilation and its components, respiration rate and tidal volume, are used as ventilatory output measures for athletes during exercise. During exercise, respiration rate is a strong indicator of physical effort. It has been shown to reflect physical effort better than other more traditional measures including blood lactate (Davies *et al.,* 2011) or glycogen depletion (Busse *et al.,* 1991). During various exercise tests including intermittent, continuous, maximal, and self-paced tests, respiration rate increases linearly over time with maximum respiration rate values occurring at test completion (Nicolo *et al.,* 2017a). Respiration rate has also been shown to closely correlate to rating of perceived exertion (Marcora *et al.,* 2008; Nicolo *et al.,* 2014a; 2016; 2017b). Therefore, it may have the potential to be used as a predictor of time to exhaustion and aid in the understanding of the distribution of effort during exercise tests (Pires *et al.,* 2011; Nicolo *et al.,* 2014). In comparison to other physiological measures such as heart rate, respiration rate has a rapid response to the onset and offset of exercise, as well as to rapid changes in physical effort which can commonly be seen in high-intensity interval training (Nicolo *et al.,* 2014b; 2017b).

Spirometry

One of the most commonly used lung function measurement systems is spirometry which measures the speed, volume, and frequency of the ventilatory cycle on a breath-by-breath basis. Spirometers can use different methods of measurement including pneumotachograph, turbine, ultrasound, and hot wire anemometer with turbine spirometers, with pneumotachograph being the most commonly used type (Schermer *et al.,* 2012). Although spirometry is widely used and thought to be reliable, there are various issues associated with it. The mouthpiece required for this measurement technique has the potential to alter an

individual's breathing pattern as it creates an awareness of respiration. This has been shown to have a potential effect on natural breathing frequency, dead space ventilation, and the tidal volume of an individual (Bloch *et al.,* 1995; Layton *et al.,* 2013). Thus, the altered breathing pattern may be non-representative for the individual or for a patient's condition (Layton *et al.,* 2011).

Within a clinical setting, spirometry is often used to diagnose and monitor various diseases (Levy *et al.,* 2006). In general, clinical practices utilise different types and brands of spirometers (Johns *et al.,* 2006). Typically, these are less expensive and less advanced than equipment used in a laboratory setting (Schermer *et al.,* 2012). However, within a laboratory, calibration and maintenance of spirometers may be below the standards required (Johns *et al.,* 2006). This deficiency and wear of the spirometer over time may affect the performance of the spirometer (Townsend *et al.,* 2004) which, therefore, may impact the interpretation of the results obtained (Schermer *et al.,* 2012).

These simple lung function measures can provide valuable insight into an individual's lung function and are widely used; however, they are only a small amount of the potential lung function parameters that can be obtained and do not provide any insight into an individual's breathing pattern i.e., the movement of the chest wall.

2.2 Breathing Diseases/Disorders in Athletes

One of the main difficulties respiratory measurement systems face is that many respiratory symptoms are non-specific and variable, making it difficult to distinguish between certain respiratory disorders/diseases. Respiratory disorders such as exercise-induced bronchoconstriction (EIB) and dysfunctional breathing have similar respiratory symptoms but different treatment methods (Dickinson *et al.,* 2007). Symptoms common to both dysfunctional breathing and EIB include chest tightness, wheezing, breathlessness, coughing, and chest pain (Keeley & Osman, 2001). However, individuals with dysfunctional breathing often experience these symptoms during exercise and symptoms resolve post-exercise, while for individuals with EIB the symptoms present and worsen post-exercise. Misdiagnosis of one of these disorders and receiving the subsequent incorrect treatment can have a negative impact on factors such as an athlete's performance (Price *et al.,* 2014).

Therefore, having specific and sensitive respiratory diagnostic systems are necessary to avoid misdiagnosis.

Exercise-Induced Bronchoconstriction

EIB is a term which describes the temporary and reversible narrowing of the airways occurring during exercise and often limiting expiration (Hull *et al.,* 2016). EIB has a prevalence of approximately 20% in elite athletes, however, this value is greater in sports such as swimming and cycling with a prevalence up to 69% (Dickinson *et al.,* 2005; Levai *et al.,* 2016a). Currently, there is no gold standard diagnostic measure for EIB and is diagnosed using one or more direct or indirect bronchial provocation test. Examples of provocation tests include the eucapnic voluntary hyperpnoea (EVH) challenge, the exercise challenge, the mannitol dry powder test, and the methacholine challenge, with lung function measurements such as forced expiratory volume in one second (FEV1) and forced vital capacity (FVC) being recorded pre and post each challenge.

The EVH challenge involves individuals inhaling a dry gas mixture of 5% CO_2, 21% O_2 and 74% N_2 for six minutes with the aim of achieving a target ventilation rate of approximately 85% maximal voluntary ventilation (FEV_1 x 30). FEV_1 is measured pre EVH challenge and post-EVH challenge at various time increments up to 20 mins using a spirometer. A positive EVH challenge is defined as a fall of FEV_1 from baseline by ≥10% at two consecutive time points (Anderson & Brannan, 2003). The American Thoracic Society (2000) has a recommended protocol for the exercise challenge. It involves individuals performing a stepped protocol on a cycle ergometer for eight minutes with the intensity for the final four minutes set to reach a heart rate > 85% of maximum. Similar to the EVH challenge, FEV_1 is recorded pre and post challenge at various time intervals up to 20 minutes with a positive test defined as a fall in FEV_1 of ≥10% at two consecutive time points. The mannitol dry powder test involves consecutive doses of mannitol dry powder up to a maximum cumulative dose of 635 mg via inhalator. A fall in FEV_1 of ≥15% from baseline is a positive test (Vakali *et al.,* 2016). Finally, the methacholine challenge involves the inhalation of a methacholine chloride and saline solution with a positive test defined as a reduction in FEV_1 of 20% (American Thoracic Society, 2000).

Vakali and colleagues (2016) assessed the response rate of each of the provocation challenges in elite athletes with and without a previous asthma diagnosis. The methacholine challenge displayed a high positive response rate in elite athletes. Previous literature suggests

that the effectiveness of the methacholine challenge can be dependent on the type of athlete i.e., summer versus winter sport athletes, with lower response rates in summer sport athletes (Sue-Chu *et al.,* 2010; Pedersen *et al.,* 2008; Vakali *et al.,* 2016). Using the methacholine challenge alone may result in under-diagnosis of EIB (Anderson *et al.,* 2009; Holzer *et al.,* 2002). Therefore, more than one provocation challenge may be necessary to correctly diagnose an athlete as they may have a positive response to only one of the tests (Vakali *et al.,* 2016). The EVH is recommended by the International Olympic Committee for diagnosing EIB (Anderson & Brannan, 2003). Vakali and colleagues (2016) found that the EVH had a high positive response rate in elite athletes with a previous diagnosis of asthma. The EVH challenges response also correlated with both indirect and direct bronchial provocation tests in athletes with a previous diagnosis of asthma (Vakali *et al.,* 2016). The EVH challenge does provoke bronchoconstriction and elicits similar symptoms, induces equivalent ventilation rates and maximum airway response induced by exercise (Anderson *et al.,* 2001a) therefore, is considered a surrogate for exercise (Parsons *et al.,* 2013).

Laboratory-based tests such as the EVH test can never completely replicate the broncho-provocative stimulus athletes experience during exercise (Anderson *et al.,* 2001b). Field-based tests, such as the exercise challenge test, are difficult to control due to factors including ambient conditions and exercise intensity (Dryden *et al.,* 2010). Slight alterations in exercise intensity can have a significant impact on the prevalence of EIB in athletes (Anderson *et al.,* 2001b). Therefore, although field-based tests may be more specific, they may lack the sensitivity required for diagnosing EIB in athletes (Dickinson *et al.,* 2006; Molphy *et al.,*2014).

EIB may be over-diagnosed when an objective test is not used and the diagnosis is based on an athlete's history and symptoms alone (Weiler *et al.,* 2014). Specific respiratory symptoms questionnaires have been developed as an alternative diagnostic method; however, they have poor accuracy and reproducibility (Hallstrand *et al.,* 2002; Weiler *et al.,* 2014). Ansley and colleagues (2012) found that approximately 50% of elite soccer players examined did not display any objective evidence of airway hyperresponsiveness or reversible airway obstruction to indirect stimuli but had a previous diagnosis of EIB or asthma from a physician. There is a high occurrence of both false-negative and false-positive EIB diagnosis in athletes when self-reported symptoms are used (Bougault *et al.,* 2010). Airway narrowing may also occur without symptoms, therefore, symptom-based diagnosis is not a reliable method of diagnosing EIB (Rundell *et al.,* 2001). Athletes misdiagnosed with asthma or EIB

may report symptoms during high intensity exercise more related to other breathing disorders such as dysfunctional breathing, and may be using β_2 agonists unnecessarily and to no benefit (Dickinson *et al.,* 2007). As well as the misdiagnosis of EIB, Dickinson and colleagues (2010) demonstrated that a large proportion of elite athletes with EIB are undiagnosed, therefore elite athletes should be routinely screened for EIB, however, this is difficult due to the lack of assessable diagnostic measures.

Dysfunctional Breathing

Dysfunctional breathing can be used to describe a collection of breathing disorders in which individuals display chronic breathing pattern changes without, or in excess of, another respiratory disease or disorder. Unlike other respiratory diseases, dysfunctional breathing is diagnosed through the elimination of other pathologies using multiple breathing function measures rather than the confirmation of a dysfunctional breathing pattern. The lack of gold standard diagnostic criteria and method makes it difficult to establish an accurate prevalence.

There have been previous attempts to classify or categorise different forms of dysfunctional breathing patterns. Boulding and colleagues (2016) suggested five classifications including hyperventilation syndrome, periodic deep sighing, thoracic dominant breathing, forced abdominal expiration, and thoracoabdominal asynchrony. Barker & Everard (2015) suggested that dysfunctional breathing be divided into four classifications including thoracic dysfunctional breathing, extra-thoracic dysfunctional breathing, functional dysfunctional breathing, and structural dysfunctional breathing. The lack of consensus adds to the difficulty in diagnosing dysfunctional breathing. The main component of dysfunctional breathing across these suggested classifications is a disordered breathing pattern (Barker & Everard, 2015; Vidotto *et al.*, 2019).

Hyperventilation syndrome (HVS) is the most well-known form of dysfunctional breathing and may be defined as individuals who display acute or chronic hyperventilation at rest or during stress or exercise which may be caused by psychological or behavioural factors such as anxiety or depression (Vidotto *et al.,* 2019). There is no one conclusive diagnostic method for HVS. The hyperventilation provocation test was once considered to be the gold standard diagnostic test for HVS. This test involves voluntary hyperventilation for multiple minutes with a positive test being confirmed if the symptoms of HVS were reproducible (Boulding et al., 2016). However, Hornsveld and Garssen (1996) demonstrated that this test is not

sensitive enough to distinguish between individuals with HVS and individuals without HVS. However, it may be used alongside other tests including the breath hold assessment, capnography, and arterial blood gas estimation to aid diagnosis (Warburton & Jack, 2006).

Alongside respiratory symptoms and function, a number of other factors need to be considered when screening for dysfunctional breathing including biomechanical, biochemical, and psychological factors (Vidotto *et al.,* 2019). Clinicians currently utilise a number of resources to diagnose dysfunctional breathing including questionnaires (e.g., Nijmegen Questionnaire and the self-evaluation of breathing questionnaire (SEBQ)), and evaluation of breathing pattern techniques (e.g., manual assessment of respiratory motion (MARM)). The Nijmegen Questionnaire (NQ) is a commonly used screening tool for dysfunctional breathing (van Dixhoorn & Folgering, 2015). It was originally developed for identifying hyperventilation syndrome (Boulding *et al.,* 2016). It has been used to detect dysfunctional breathing in the general population (Courtney *et al.,* 2011a), and in those with concurrent respiratory diseases/disorders such as asthma (Grammatopoulou *et al.,* 2014). Similarly, the self-evaluation of breathing questionnaire has been used to assess dysfunctional breathing symptoms. Although this questionnaire has high test re-test reliability (Mitchell *et al.,* 2016), normative ranges and cut-offs scores for dysfunctional breathing have not been established (Courtney *et al.,* 2019). Hagman and colleagues (2008) developed a 10-criterion list to identify dysfunctional breathing symptoms and successfully used it to compare individuals with dysfunctional breathing to individuals with well-controlled asthma. Kiesel and colleagues (2017) developed a screening protocol for dysfunctional breathing diagnosis which involved breath hold time test, a four-question questionnaire consisting of two questions from the SEBQ and two questions from the Nijmegen questionnaire. They determined that if this protocol yielded a negative result, there was an 89% chance that dysfunctional breathing was not present and a positive result would prompt further assessments.

The MARM is a manual technique used to assess and quantify the contribution of the upper and lower ribcage and the abdomen to the movement of breathing (Boulding *et al.,* 2016). Inter-examiner reliability has been established for the MARM (Courtney *et al.,* 2009) and it has been successfully evaluated against respiratory inductive plethysmography (RIP) (Courtney *et al.,* 2008). Despite their wide use, minimal clinically important differences have not been established for these assessments (Courtney *et al.,* 2019).

The Brompton breathing pattern assessment tool (BPAT) is a practical assessment of dysfunctional breathing intended to be used within daily clinical practice (Todd *et al.,* 2018). It involves visually assessing an individual's breathing using seven components scored from 0-2. It must be performed in a static seated position. Some of the features of the Hagman *et al.,* (2008) criterion lists are incorporated into this tool. As this is a relatively new assessment tool, further investigation is required to establish it as a robust clinical assessment tool. However, Todd *et al.* (2018) report sensitivity and specificity values of 0.92 and 0.75, respectively, with a BPAT score of four or more.

The different categories of assessments, i.e., questionnaires and breathing pattern assessments, do not have established correlations (Boulding *et al.,* 2016). Courtney and colleagues (2011b) demonstrated that the Nijmegen questionnaire and the SEBQ correlated well with each other when assessing individuals with mild dysfunctional breathing. Each of these assessments require some form of subjectivity and interpretation, therefore, there is a need for an objective breathing pattern measurement system for the diagnosis of dysfunctional breathing.

2.3 Methods of Measuring Breathing Pattern

It is becoming increasingly popular to use objective measurement systems to quantify the breathing pattern of individuals with different respiratory disorders. Measuring the movement of the chest wall is an objective method of assessing an individual's breathing function. Chest wall movement can be measured using physical devices placed on the chest or external imaging techniques (Seddon, 2014). Measuring the movement of the chest wall can include measuring circumference changes of the chest wall, measuring alterations in the cross-sectional area of the chest, measuring linear displacement of the front of the chest, and finally measuring chest volume changes.

The three most widely used breathing pattern measurement systems in clinical and research environments are structured light plethysmography (SLP), respiratory inductive plethysmography (RIP), and optoelectronic plethysmography (OEP).

Structured Light Plethysmography

Structured light plethysmography (SLP) is non-invasive, non-contact measurement system which tracks the displacement of the chest wall during quiet breathing. Its popularity in the clinical environment is growing and has been used to assess a number of conditions and clinical interventions (Motamedi-Fakhr *et al.*, 2017a).

SLP involves projecting a structured grid of light onto the front of the torso to measure the movement of the chest wall which is recorded by two digital cameras. The centre of the grid must be lined up to the xiphisternum which allows the grid to be subdivided into the ribcage and the abdomen. The movement-over-time trace recorded by the cameras can be used to derive timing and flow-related parameters. This measurement system, as it is non-contact, allows for the assessment of various populations where other measurement systems would be difficult to perform such as elderly individuals, children, new-borns, immobile patients, and patients post-surgery. However, the alignment of the grid offers a number of limitations. The grid is divided equally to cover the ribcage and the abdomen; however, the lengths of the ribcage and abdomen will vary between individuals. As no other anatomical landmarks are used, the grid is likely to not cover the entire chest wall resulting in inaccuracies. Participants are required to wear a tight white T-shirt to ensure the grid is projected clearly, however, any creases on the T-shirt results in poor displacement traces. As well as this, some individuals may find it difficult to put on the tight T-shirt such as elderly individuals with limited range of motion or individuals with shoulder injuries. SLP is very sensitive to

movement including head and limb movement, therefore, can only be used during static positions. Poor displacement traces caused by movement or creases in the T-shirt are visible in the SLP software, allowing the participant to be repositioned to collect better quality data. However, SLP also has slight tracking errors that may not be visible in the SLP trace (Hmeidi *et al.,* 2017) making it difficult to ensure optimal data is collected, potentially resulting in some data having to be excluded.

SLP allows for timing parameters such as RR (respiratory rate), tI (inspiratory time), tE (expiratory time), tTot (total breath time), tI/tE, tI/tTot, IE50 (ratio of inspiratory to expiratory flow at 50% of tidal volume), contribution parameters including ribcage, abdomen, as well as left and right side of the torso, and the phase angle between the ribcage and the abdomen. It has been used to investigated the breathing patterns of children (Ghessi *et al*., 2017; Hmeidi *et al.,* 2017), healthy individuals, COPD patients (Motamedi-Fakhr *et al.* 2017b; Nierat *et al.,* 2017), and to assess the impact of bronchodilators (Hmeidi *et al.,* 2017). The contribution and phase angle parameters cannot be obtained using traditional spirometry.

Motamedi and colleagues (2017a) compared breathing parameters including RR, tI, tE, tTot, tI/tE ratio, tI/tTot ratio, and IE50 obtained from SLP to those obtained from pneumotachography in healthy individuals and in a variety of individuals with breathing disorders and thus a range of breathing patterns. It was found that SLP correlated well with pneumography as breath-by-breath agreement for RR was ± 1.4 breaths per minute, tI was ± 0.2 seconds, tE was ± 0.3 seconds, and tTot was ± 0.3 seconds. When averaged over a period of many breaths these agreements improved. As RR increased, agreement for tI, tE, and tTot significantly improved, but did not significantly change for tI/tE, tI/tTot, and for IE50. However, agreement significantly reduced for RR on a breath-by-breath bases but did not significantly change when averaged of a period of time. It was also found that all parameters at rest did not significantly differ in repeated trials for both breath-by-breath and averaged analysis, demonstrating the repeatability of SLP at rest.

Nierat and colleagues (2017) evaluated the disruption of breathing pattern in healthy and COPD patients cause by the mouthpiece required in pneumotachograph by comparing it to SLP during tidal breathing. Firstly, it was found that RR, tI, tE, and tTot were not significantly different between the two measurement systems. Secondly, it was found that breathing variability was significantly different between the two measurement systems with the variability significantly lower by ~20-30% in healthy individuals and ~25-38% in COPD

patients using pneumotachograph. Breathing variability decreases with increased mechanical loads (Brack *et al.,* 2002) and is influenced by the intensity of the neutral drive to breathe (Nierat *et al.,* 2017). Therefore, the authors of this study assumed that reduction in breathing variability seen with pneumotachograph compared to SLP, indicates that SLP avoided disruption of the breathing pattern during tidal breathing.

Motamedi-Fakhr and colleagues (2017b) investigated the differences in breathing pattern between healthy individuals and individuals with COPD during tidal breathing. It was found that tI, tI/tE, tI/tTot, and tPTEF displayed significantly lower values in COPD patients. While IE50 and phase angle between the ribcage and the abdomen were found to be significantly greater in COPD patients compared to healthy individuals. This demonstrates that SLP has the ability to distinguish between populations during tidal breathing.

Similarly, Hmeidi and colleagues (2017) investigated the differences between healthy children and children with asthma pre and post bronchodilator using SLP. tI/tE, tI/tTot, tPTEF, IE50 and the variability of IE50 were found to be significantly different between the two population groups during tidal breathing with children with asthma displaying lower values for tI/tE and tI/tTot, and higher values for tPTEF (time taken to reach peak expiratory flow) and IE50. The variability of IE50 was significantly greater in children with asthma. This again demonstrates that SLP has the ability to distinguish between population groups during tidal breathing. In addition to this, SLP also was able to quantify the response to bronchodilator in the children with asthma. IE50 and the variability of IE50 both significantly decreased post-bronchodilator with all other breathing parameters displaying non-significant differences. A complimentary study by the same research group, investigated the impact of bronchodilators on children as young as 2 years of age with acute asthma/wheeze using SLP. RR, tI, tE, tTot, IE50, phase angle between the left and right side of the torso (HTA), and the phase angle between the ribcage and the abdomen (RcAbPhase) were found to be significantly different between children with acute asthma/wheeze and healthy controls. The variability of tI, tE, tTot, tI/tE, tI/tTot, tPTIF, HTA, and RcAbPhase were also found to be significantly different between the groups. The within-subject variability of tPTEF was significantly reduced post bronchodilator. This study demonstrates that SLP can be used on a wider population including children as young as 2 years of age.

SLP has also been used to successfully identify changes in breathing pattern in patients post-lung resection (Elshafie *et al.,* 2016). The operated side of the torso displayed significantly less motion post-surgery compared to pre-surgery with a significantly increase in the non-

operated side. The phase angle between the ribcage and the abdomen also displayed significant change with an increase post-surgery while there were no significant differences between the phase angle between the right and left sides of the torso. This indicates that although contribution of the right and left sides of the torso significantly changed, the timing of the movement was still synchronous.

Ghezzi and colleagues (2017) performed a case study to assess the use of SLP in the management and follow up of children with pneumonia. SLP was performed on the patients at admission, 48 hours post treatment, and 1-month post treatment. IE50 displayed a consistent decrease post treatment, with the contribution of both the right and left side of the torso becoming closer to 50%. The phase angle between the left and right side of the torso became closer to perfect synchrony (0 degrees) after treatment and displayed lower activity of the left side on admission. This study provided a basis for using SLP as a method to assess the effectiveness of clinical treatments. However, due to the limitations of SLP, its use is restricted to clinical environments where a patient can remain in a static position and is not suitable for dynamic environments such as exercise.

Respiratory Inductive Plethysmography

Respiratory inductive plethysmography (RIP) is a non-invasive wearable respiratory monitoring device. RIP is one of the more widely used wearable respiratory monitors in clinical and research settings and is considered to have higher accuracy and sensitivity when compared to other measures such as electric impedance plethysmography and strain gauges (Bradley *et al.,* 2011; Fiamma *et al.,* 2007; Carry *et al.,* 1997).

It was first introduced by Cohn (1982) which consisted of two elastic bands consisting of coils of wire placed around the ribcage, in line with the axilla, and the abdomen, in line with umbilicus. Respiratory timing and volumetric parameters are measured through the inductance of the bands from linear changes in the chest wall circumference caused by respiration (Cohn *et al.,* 1982; Zhang *et al.,* 2012). Prior to use, it is necessary to calibrate the RIP system typically using spirometry. A method of calibration typically used is called qualitative diagnostic calibration during tidal breathing (Matos *et al.,* 2012; Sackner et al., 1989; Tomich *et al.,* 2010). Qualitative diagnostic calibration is a two-step calibration procedure consisting of first correctly partitioning the ribcage and abdomen electrical gains of the RIP amplifiers during natural breathing adjusting the spirometer output to correspond to the values of the plethysmograph. Secondly, with the participant breathing into the spirometer for approximately 30 to 60 seconds, the spirometer output is then used to calibrate the sum signal for absolute volume obtained from the RIP system in ml (Matos *et al.,* 2012). This calibration requirement is a limitation of RIP as in order to use RIP, access to a spirometer is necessary. Another limitation of this calibration procedure is that the body position must remain constant post calibration in order to ensure satisfactory tidal volume values (Chadha et al., 1982; Brullmann et al., 2010). These restrictions limit the potential uses of RIP. Brullmann and colleagues (2010) demonstrated an RIP calibration technique using position sensors for position-specific calibration. This method allowed for the recalibration of RIP with each change in body position monitored using accelerometers and compared to breath-by-breath pneumotachograph values, the RIP derived tidal volumes displayed limits of agreement of ± 20% across all positions.

RIP currently allows for the following variables to be obtained: respiratory rate, tidal volume, minute ventilation (V_E), ratio of inspiratory time to total breath time (tI/tTot), percentage of ribcage and abdomen motion, and the phase angle between the ribcage and the abdomen. RIP has been used to investigate the breathing patterns of various populations including healthy individuals (Panka *et al.,* 2010; Parreira *et al.,* 2010), obese patients post

bariatric surgery (Matos *et al.,* 2012; Tomich *et al.,* 2010), COPD (Alves *et al.,* 2008), patients with restrictive lung disease (Brullmann *et al.,* 2010), infants (Stick *et al.,* 1992; Robles-Rubio *et al.,* 2020), individuals with sleep disorders (Yang *et al.,* 2011).

Early investigation validated RIP at rest and during exercise in adults and children (Chadha *et al.,* 1982). More recently, it was established that across various exercise intensities for healthy individuals, COPD patients, and patients with congestive heart failure, RIP, when compared to breath-by-breath analysis through pneumotachograph, displayed small inaccuracies (within 7%) for tidal volume, minute ventilation, and respiratory rate. However, if averaged over many breaths no significant differences were seen between measurement systems (Clarenbach *et al.,* 2005). Similarly, Heyde and colleagues (2014) investigated the validation of prior calibrated RIP during exercise on healthy participants. During an incremental running test, RIP estimated tidal volume within ± 20% of traditional breath-by-breath analysis. Similarly, tidal volume, inspiratory time, and expiratory time obtained via RIP have been shown to correlate with breath-by-breath analysis at rest and during walking in healthy adults (Retory *et al.,* 2016).

Parreira and colleagues (2010) investigated the influence of age and sex on breathing pattern and thoracoabdominal movement in a healthy population at rest using RIP. It was found that parameters including the ratio of inspiratory time to total breath time are influenced by sex, with values presenting significantly lower for 20 to 39-year old females. While thoracoabdominal movement parameters were found to be influenced by age, with the asynchrony between the ribcage and the abdomen displaying significantly greater values for individuals over the age of 60.

Panka and colleagues (2010) used RIP along with surface electromyography (sEMG) to investigate changes in breathing pattern during unsupported arm elevation during activities of daily living. sEMG demonstrated an increase in sternocleidomastoid muscle activity with RIP demonstrating an increase in tidal volume, minute ventilation, respiratory rate, mean inspiratory flow, and thoracoabdominal asynchrony. These results may provide a basis for the mechanism behind why patients with COPD experience dyspnea during daily living activities.

Breathing pattern changes in obese patients post bariatric surgery has been investigated using RIP (Matos *et al.,* 2012). It was found that prior to the surgery, the obese patients displayed significantly greater minute ventilation and phase angle values compared to the control

group. Post-surgery these values became closer to those seen in the control group, as well as this, the obese individuals displayed significant reduction in tidal volume without alterations in respiratory rate thus leading to a reduction in minute ventilation. There were no significant changes in the percentage of ribcage and abdominal motion pre and post-surgery and also compared to the control group. Therefore, RIP was able to identify changes in breathing pattern in obese individuals pre- and post- bariatric surgery. Tomich and colleagues (2010) used RIP to investigate the effect of breathing exercises on the breathing patterns of individuals post-bariatric surgery. RIP was sensitive enough to measure the differences in the effects of three different breathing exercises in this population, demonstrating that incentive spirometry with volume-orientated device promoted slower and deeper inhalations compared to the other breathing exercise techniques.

RIP has also been used to investigate the breathing patterns of COPD patients during exercise. It was found that during exercise COPD patients displayed greater abdominal motion compared to ribcage motion with increases in tidal volume, respiratory rate, and phase angle. The authors suggested that cycling may be an effective method of enhancing COPD patient's ventilation (Alves *et al.,* 2008).

Due to its widespread use, it has become one of the leading methods for measuring breathing patterns non-invasively (Wilhelm *et al.,* 2003). However, the accuracy of RIP is affected by an individual's body position and movement (type of exercise), motion artefact, and movement of the RIP bands themselves which limit the potential uses of RIP.

Optoelectronic Plethysmography

Optoelectronic plethysmography (OEP) is a non-invasive method of measuring the movement of the chest wall using a 3-dimensional motion capture system. It allows the compartmentalisation of the torso, providing significantly more information regarding an individual's breathing mechanics compared to other measurement systems.

The original OEP motion capture system set-up consisted of 86 markers (34 posterior, 10 lateral and 42 anterior) and four motion capture cameras, two posterior and two anterior (Cala *et al.*, 1996; Camevali *et al.*, 1996). When compared to spirometry, the 86-marker set had an error of < 3.5% (Camevali *et al.*,1996). To improve on this, Aliverti and colleagues (2001c) utilised six cameras and 89 markers (37 posterior, 10 lateral and 42 anterior) which has now become the standard marker set and placement for OEP. The markers are attached to the skin in a grid-like pattern using bi-adhesive hypoallergenic tape. The anterior grid is comprised of seven horizontal rows from the clavicle to the anterior superior iliac crest. Additionally, there is a lateral column of five markers on each midaxillary line. The posterior grid consists of seven horizontal rows between C7 and the posterior superior iliac spine (Layton *et al.*, 2011). However, the set-up of OEP although non-invasive (Meric *et al.*, 2016), requires substantial time due to the number of markers required. Marker placement can also be a complicated process where individuals would require training to correctly identify the relevant anatomical locations and an in-depth understanding of the testing protocol and data processing (Layton *et al.*, 2011). Due to the precise locations of the markers, small errors in the placement may result in the marker being unrecognisable by the motion capture system software. If this occurs, due to the larger number of markers being recorded at a high sampling frequency, the data processing time is further increased as manual adjustments must be made to obtain accurate data (Layton *et al.*, 2011). However, the use of multiple precise locations, although time consuming, allows for optimal coverage of the torso based on an individual's anatomical landmarks, thus improving accuracy. In contrast, other measurement systems such as SLP only use one anatomical landmark and the nature of the system means that parts of the torso may not be measured depending on an individual's ribcage and abdomen sizes.

Along with this, OEP requires a series of cameras, thus, can only be performed in a laboratory setting due to camera cost and camera set up requirement. For optimal marker visibility, each camera needs to be positioned around a specific capture volume and the

setting of each need to be manually adjusted. Although, OEP is currently the most accurate method to measure breathing mechanics, it is not without its limitations.

OEP data is based on an estimation of the torso volume. For the calculation of this volume, the torso is divided into three separate compartments consisting of the pulmonary (upper) rib cage (RCp), the abdominal (lower) rib cage (RCa), and the abdomen (AB). RCp consists of the area from the clavicles and jugular notch to the xiphoid, RCa consists of the area between the xiphoid process to the lower costal margin, and finally, AB consists of the area from the lower costal margin to the anterior superior iliac crest. The torso can further be divided into the right and left sides of the chest (Layton *et al.,* 2011). Torso volume can be calculated using a number of different methods. On initial development of OEP, an algorithm originally created to measure limb and joint movements was utilised to calculate chest wall movements (Ferrigno & Pedotti, 1985). However, when this method was compared to spirometry it yielded an error ± 21.3% (Cala *et al.,* 1996). A method based on Gauss theorem was then developed and allowed the compartmentalisation of the torso into the six compartments. This method estimates the surface of the torso by connecting the markers to create a mesh of 56 triangles. The volume enclosed within this mesh is then calculated using a procedure that is the equivalent to the Gauss theorem (Cala *et al.,* 1996). This method improved the accuracy of the measurement of the chest wall with an error of < 3.5% (Aliverti *et al.,* 2004; Cala *et al.,* 1996) and was labelled as the 'conventional method' for calculating OEP-derived torso volume. However, it does have a number of limitations including virtual markers have to be created to enable the triangulation of the markers which may lead to an over- or under-estimation and it is limited to the calculation of the six compartments only.

More recently, Massaroni and colleagues (2017a) developed a novel method to calculate the torso volume which divides the torso into prisms. This method involves creating seven centroids which are midpoints of seven elliptical shapes and are created by the seven horizontal rows of markers. 82 prisms are then created using four surface markers and two centroid markers to form each prism. This volume of each prism is then calculated using the Lagrange formula (Massaroni *et al.,* 2017a). When comparing both methods against spirometry, it was found that both displayed similar patterns, with the prism-based method displaying slightly better agreement to the spirometer compared to the conventional method for tidal volume and temporal values. Both methods over-estimated tidal volume compared to spirometer with the conventional method having the larger over-estimation (3.56%) compared to the prism-based method (2.23%).

Two convex hull based methods, convex hull with boundary conditions and convex hull with Delaunay triangulation, have also been assessed against the conventional and the prism based methods (Massaroni *et al.*, 2017b). The convex hull with boundary conditions method is based on the method proposed by Ferrigno and colleagues (1994) by extending it to 89 markers creating 46 volume sections by grouping 8 markers which allows for the compartmentalisation of the torso. The convex hull with Delaunay triangulation method is based on the computation of the volume enclosed by a mesh surface created by the triangulation of the markers, however, this method does not allow for the compartmentalisation of the torso. When comparing all four computational methods with spirometry, it was found that volume trace obtained from all methods were comparable to that of the spirometry. The prism-based and convex hull with boundary conditions methods both displayed reduced bias in tidal volume estimations when compared to the conventional and convex hull with Delaunay triangulation methods. All methods except the convex hull with Delaunay triangulation overestimated the volume compared to spirometry. When the prism-based method and the convex hull with boundary conditions were compared to the conventional method measuring thoraco-abdominal movement pattern, the prism-based method demonstrated values closer to the conventional method with maximum differences of 30 ml in the abdomen compartment. Therefore, this study suggests that the prism-based method demonstrated the most accuracy with spirometry during tidal breathing.

OEP Validation

Several studies have compared OEP with spirometry as the gold standard. A good linear relationship between OEP and spirometry has been found with r^2 values greater than 0.9 across various populations including healthy individuals (Cala *et al.*, 1996), intensive care patients (Aliverti *et al.*, 2000), chronic obstructive pulmonary disease patients (Aliverti *et al.*, 2004; Vogiatzis *et al.*, 2005a) and new-borns (Dellaca *et al.*, 2010). During rest and submaximal exercise, tidal volume obtained from OEP has been validated against spirometry tidal volume with r^2 values greater than 0.95 (Kenyon *et al.*, 1997; Vogiatzis *et al.*, 2005b). Layton and colleagues (2013) also found high correlation values ($r = 0.963$, $p < 0.001$) during a maximal cycle test in healthy individuals. Another study compared chest wall volume measured by spirometry, pneumotachograph, and OEP during seated quiet and deep breathing. High correlations were found between the three techniques with discrepancies of 4.9 ± 6.4%, -1.6 ± 5.4% and 1.7 ± 5.9% between spirometry and pneumotachograph, pneumotachograph and OEP, and spirometry and OEP, respectively (Aliverti *et al.*, 2000).

Therefore, OEP has been validated across a variety of populations under a variety of conditions. Previously there has only been one known study which has assessed the inter- and intra-rater reliability of OEP which found intra-class correlation coefficient above 0.70 and a coefficient of variation below 11% in a healthy population at rest and during exercise (Vieira *et al.,* 2013).

OEP offers significantly more information regarding an individual's breathing mechanics/pattern through the measurement of various breathing parameters compared to simple breath-by-breath analysis (Pereira *et al.,* 2017). These breathing parameters include timing and ratio parameters, compartmental volume contribution parameters, and phase angle parameters (Massaroni *et al.,* 2017c). Many of these breathing parameters can be obtained from the three main measurement systems: RIP (Alves *et al.,* 2008; Matos *et al.,* 2012; Panka *et al.,* 2010; Tomich *et al.,* 2010), SLP (Hmeidi *et al.,* 2017), and OEP (Aliverti *et al.,* 2009; Bruni *et al.,* 2012; Gagliardi *et al.,* 2013; Lunardi *et al.,* 2014; Priori *et al.,* 2013; Vieira *et al.,* 2014; Zoumot *et al.,* 2015). However, it is difficult to make comparisons with and between studies investigating these breathing parameters due to differences in protocols with factors including the measurement system, posture, and participant type (Parreira *et al.,* 2010). These parameters are calculated slightly differently depending on the measurement system as the breathing parameters estimated from SLP are derived from the displacement of the front of the chest and from RIP are derived from electrical signals while the breathing parameters estimated from OEP are derived from the torso volume changes. Pereira and colleagues (2017) used thoracoabdominal asynchrony to compare RIP and OEP in healthy individuals, patients with interstitial lung disease, and patients with COPD. It was concluded that the OEP system yielded similar thoracoabdominal asynchrony values to RIP and displayed satisfactory agreement during rest and exercise across the three groups.

Pereira and colleagues (2017) compared RIP to optoelectronic plethysmography (OEP) in three different populations including healthy individuals, COPD patients, and interstitial lung disease patients, at rest and during exercise. Overall, comparing the phase angle between the pulmonary ribcage and the abdomen, there were no significant differences between both measurement systems at rest and during exercise (50% and 75% of maximal workload). Healthy individuals displayed good agreement across all conditions, while for the COPD patients and patients with interstitial lung disease displayed good agreement except when the phase angle demonstrated high variability. Although this study displayed

good agreement between OEP and RIP, the results should be interpreted with caution as the measurement systems were not used simultaneously.

Previous studies have measured these OEP-derived breathing parameters in a supine position (Matos *et al.,* 2012; Parreria *et al.,* 2010; Vieira *et al.,* 2014), a seated position (Lunardi *et al.,* 2014; Panka *et al.,* 2010; Tomich *et al.,* 2010), and during cycling (Aliverti *et al.,* 2009; Alves *et al.,* 2008; Bruni *et al.,* 2012; Gagliardi *et al.,* 2013; Zoumot *et al.,* 2015). Priori and colleagues (2013) investigated the effect of posture on thoracoabdominal asynchrony in individuals with COPD using OEP. It was found that posture significantly influenced thoracoabdominal asynchrony in COPD patients and in healthy controls. The asynchrony between RCp and RCa reduced when supine compared to a seated position while, the asynchrony between RCp and AB was found to increase when in a supine position. OEP has been used to examine various different body positions in healthy participants, both male and female. It was found that the standing and sitting positions, quiet breathing is mostly contributed to by the abdominals while in the supine position is it predominately thoracic (Lee *et al.,* 2010). Similarly, Romei and colleagues (2010) investigated five different seated postures at various inclines, with and without back support, of supine position. It was found that with increased inclination of the torso, there was a decrease in ribcage displacement as well as a reduction in tidal volume and minute ventilation.

OEP has also been used to assess the breathing patterns of healthy individuals in relation to age and sex (Mendes *et al.,* 2020). Early studies found no significant differences in thoraco-abdominal displacement between males and females during quiet breathing (Sharp *et al.,* 1975; Verschakelen & Demedts, 1995). In contrast, similar studies found larger abdominal displacement in males compared to females (Fugel-Meyer, 1974; Gilbert *et al.,* 1981). This is supported by more recent studies where it was found, using chest radiographs (Bellemare *et al.,* 2003) and OEP (Romei *et al.,* 2010), that females have less changes in abdominal displacement and abdominal contribution to tidal volume in comparison to males. Bellemare and colleagues (2003) also used esophageal and gastric pressure measurements to show that males have less inspiratory ribcage muscle contribution compared to females at rest. However, when corrected for body size differences, there were no differences in tidal volume and minute ventilation between males and females (Kilbride *et al.,* 2003; Romei *et al.,* 2010). Aging has been shown to significantly reduce ribcage contribution to tidal volume and increase end-inspiratory and end-expiratory volumes in healthy subjects at rest (Mendes *et*

al., 2020). De Souza and colleagues (2016) demonstrated differences in abdominal contribution between young and elderly females during different inspiratory resistance.

OEP allows breathing function assessments to be performed on patients who would otherwise not be able to have such an assessment, including individuals unable to achieve a tight seal on a mouthpiece for spirometry, or those who are unconscious (Layton *et al.,* 2011). Therefore, OEP has been used to assess a variety of pathological populations including COPD patients (Aliverti *et al.,* 2004; 2009; Priori *et al.,* 2013), obese individuals (De Melo Barcelar *et al.,* 2014), intensive care patients (Aliverti *et al.,* 2000), asthmatics (Fregonez *et al.,* 2018), patients with hemiplegia (Lanini *et al.,* 2003), individuals with Duchenne muscular dystrophy (LoMauro *et al.,* 2010), individuals with interstitial lung disease (Pereira *et al.,* 2017), patients with pectus excavatum (Redlinger *et al.,* 2011; 2012), patients with late-onset Pompe disease (Meric *et al.,* 2016), and patients with emphysema (Zoumot *et al.,* 2015), as well as elderly individuals (Lunardi *et al.,* 2014) and new-borns (Dellaca *et al.,* 2010). OEP has demonstrated that thoracic surgery can alter the biomechanics of the chest wall, particularly during the early post-surgery period, including thoracotomy (Bastianini *et al.,* 2009; 2010; Lo Mauro *et al.,* 2017).

Vogiatzis and colleagues (2005a) found patients with severe COPD display two different breathing patterns during exercise consisting of early and late hyperinflation. Patients with early hyperinflation require longer recovery periods post exercise before the hyperinflation is completely reversed. Similarly, Aliverti and colleagues (2004) have demonstrated with OEP, that patients with severe COPD present with dynamic hyperinflation during an incremental exercise test with varying alterations in chest wall volume and movement. COPD patients with hyperinflation are likely to present with abnormal ribcage movement in particular paradoxical lower ribcage, however, this movement cannot entirely be explained by hyperinflation and is likely to be an important factor for the onset of dyspnoea and exercise intolerance in individuals with COPD (Massaroni *et al.,* 2017c). There have been a number of studies which have investigated using OEP the impact of different treatment methods for patients with COPD including exercise programmes (Gagliardi *et al.,* 2014), oxygen therapy (Bruni *et al.,* 2012), inspiratory loaded breathing (Myrrha *et al.,* 2013), and pulmonary rehabilitative exercises (Georgiadou *et al.,* 2007). Dal Negro and colleagues (2012) also found that OEP can also detect and distinguish between the effects of different bronchodilators in moderate to severe COPD patients.

Using OEP, it was found that patients with stroke demonstrate changes in the control of breathing and hemiparetic stroke leads to asymmetric ventilation (Lanini *et al.*, 2003). Aliverti and colleagues (2000) investigated whether OEP could be utilised as a measurement technique of chest wall volume alterations in intensive care patients by comparing the OEP volumes to volumes obtained using spirometry and pneumotachograph. With high correlations between the measurement techniques, OEP was deemed a feasible method of measuring chest wall volume in intensive care patients, specifically paralyzed patients with acute respiratory distress syndrome and acute lung injury. In patients with Duchenne muscular dystrophy, OEP-derived abdominal volume contribution has been used to assess respiratory muscle impairments with abdominal contribution being a marker of disease progression and an indicator for hypoxaemia during sleep (LoMauro *et al.*, 2010).

From an extensive literature search, there have only been four studies using OEP on individuals with asthma. Gorini and colleagues (1999) used OEP to assess breathing pattern of asthmatics pre and post histamine induced bronchoconstriction. During bronchoconstriction, hyperinflation occurred with the ribcage accounting for the majority of the volume with an increase in ribcage end-expiratory volume. Similarly, during methacholine induced bronchoconstriction, hyperinflation also occurred with significant increases in end-expiratory volume predominantly due to an increase in ribcage volume (Filippelli *et al.*, 2003). Feitosa and colleagues (2019) constructed a receiver operating characteristic curve for OEP-derived end-expiratory volume of the chest wall for children with exercise induced asthma. It was established that OEP can accurately be used to identify exercise induced asthma in children when compared to traditional spirometry. Fregonezi and colleagues (2018) utilised OEP to successfully investigate the breathing patterns of individuals with mild asthmas but without EIB during mild exercise.

Overall, the current literature demonstrates OEP as a non-invasive technique for measuring breathing patterns across a wide range of applications in the clinical and research setting. However, there have been minimal OEP related studies investigating the breathing pattern of individuals with asthma and no studies to date assessing athletes with EIB or dysfunction breathing.

Other Methods

Devices such as accelerometers, magnetometers, ultrasound signal transmission delay times, and linear displacement lasers have been used to measure the linear displacement of the rib cage and abdomen. These devices have a number of advantages as they are simple, small, portable, and relatively inexpensive (Seddon, 2015). However, the major disadvantage of all of these devices is the assumption that the alterations in one dimension are representative of the complex 3-dimensional movement of the chest. Early research found that the relationship between the measured linear displacement of the chest and the volume change of the chest varied across healthy participants (Reich & McHenry, 1990), with the larger the displacement the less reliable this measurement technique became (Russell & Helms 1994). Another disadvantage of using accelerometers to measure chest wall displacement is it can only be used in static conditions due to motion artefacts (Aliverti, 2017). However, accelerometers have been coupled with gyroscopes (Yoon *et al.,* 2014) or magnetometers (Gollee *et al.,* 2007) in an attempt to overcome these limitations. More recently, it was found that using a configuration of three micro electro-mechanical system accelerometers, the peak tidal expiratory flow over total expiratory time was accurately estimated in healthy adults when compared to oronasal flow signal and strain gauge belts (Dehkordi *et al.,* 2014).

Along with RIP, alterations in the circumference of the chest wall maybe measured using a number of direct and indirect techniques including strain gauges, fibre optic respiratory plethysmography, impedance pneumography, and electrical impedance tomography. These measurement techniques may be more accurate at estimating chest volume compared to the other methods measuring linear displacement only. However, these methods still have a number of assumptions including that the cross-sectional shape of the chest remains constant during breathing and a measurement at one point of the rib cage and one point of the abdomen is representative of the entire thorax. Both assumptions may not be true particularly for individuals with severe respiratory or neuromuscular disease, or for infants (Seddon, 2015).

Impedance pneumography (IP) is a non-invasive technique that has emerging applications in respiratory measurement. IP involves measuring electrical impedance changes of the thorax during breathing using skin electrodes (Gupta, 2011). This electrical impedance consists of a relatively constant value and a fluctuating value. The constant value is thoracic impedance or baseline impedance and is normally 500 Ω (Grenvik *et al.,* 1972; Gupta, 2011). The fluctuating value may be referred to as respiratory impedance. The fluctuations of

respiratory impedance are caused by two main effects: changes in the gas volume of the chest relative to the fluid volume, an increase in this will cause a decrease in conductivity, and changes in the conductance path length due to chest expansion. An increase in both effects result in an increase in the electrical impedance (Kelkar *et al.,* 2008). One of the issues with IP is that it is difficult to identify the contribution of these effects to a signal as they may change on an individual basis. However, a linear relationship between changes in impedance and changes in lung volume and thus pulmonary flow rate signal derivation has previously been demonstrated (Seppa *et al.,* 2010). Seppa and colleagues (2013a) demonstrated that IP was capable of accurately measuring tidal breathing flow-volume curve in breathing pattern alteration induced by expiratory loading, when compared to traditional breath-by-breath analysis. IP has also been used to track methacholine-induced bronchoconstriction in children with breathing difficulties, demonstrating agreement with direct breath-by-breath analysis (Seppa *et al.,* 2013c). IP has also been used to assess tidal breathing flow profiles of infants (Gracia-Tabuenca *et al.,* 2019; Malmberg *et al.,* 2017). Malmberg and colleagues (2017) demonstrated agreement between IP and pneumotachograph, although this agreement was affected by infants with reduced lung function (Malmberg *et al.*, 2017). However, parameters derived from the tidal breathing flow volume were not shown to be interchangeable with those obtained from pneumotachograph (Malmberg *et al.,* 2017). Using IP, tidal breathing flow volume profiles vary in wheezing infants during the different sleep phases (Gracia-Tabuenca *et al.,* 2019). IP has not generally integrated into wearable devices (Charlton *et al.,* 2016), however, it seems to be a prospective method for non-invasive, long term measurement of respiration (Mlynczak & Cybulski, 2012).

Fiber Bragg gratings are growing in popularity for measuring vital signs in recent years (LoPresti *et al,* 2020). Fiber Bragg gratings have a number of advantages as they are small, lightweight, non-invasive, and can obtain information such as respiratory rate, heart, rate, blood pressure, and body temperature (LoPresti *et al.*,2020). For measuring respiratory rate, fiber Bragg gratings have been used alone or integrated into clothing such as chest bands or t-shirts. A smart t-shirt with up to twelve gratings has been used to measure breathing parameters including respiratory rate, tidal volume, compartmental volume, inspiratory and expiratory time (Ciocchetti *et al.,* 2015; Massaroni *et al.,* 2016; LoPresti *et al.,* 2017) with percentage errors in respiration rate estimation reducing ($\leq 0.4\%$) with increased number of grating sensors (LoPresti *et al.,* 2017). Fiber Bragg gratings have been validated in healthy males and females against OEP (Ciocchetti *et al.,* 2015; Massaroni *et al.,* 2016; LoPresti *et*

al., 2018), against a flow sensor (LoPresti *et al.*, 2019), and against spirometry (Iacoponi *et al.*, 2018).

Strain gauges measure changes in the circumferences of the thorax and abdomen through changes in electrical resistance of the sensors due to stretching (Al-Khalidi *et al.*, 2011). Chu and colleagues (2019) have demonstrated that using small disposable wearable strain sensors on the abdomen and ribcage, respiratory rate and volume can be obtained which highly correlate with breath-by-breath spirometry at rest in a reclined position. Although strain gauges have previously been used for respiratory monitoring (Kwak *et al.*, 2018), they have predominately been replaced by RIP (Aliverti, 2017).

Respiratory rate may be obtained from a variety of non-contract methods. ECG-derived respiration rate is based on the fact that respiration rate has a modulating effect on QRS slopes and R-wave angles (Aliverti, 2017; Larzaro *et al.*, 2014). Alternatively, respiration rate may be determined from photoplethysmogram signal from pulse oximeters. Both of these signal types are affected by three types of modulation namely baseline drift, amplitude modulation, and frequency modulation. Although the mechanisms behind these modulations are poorly understood (Karlen *et al.*, 2013; Tarassenko *et al.*, 2014), respiration rate has been predicted through using the three different modulations (Orphanidou *et al.*, 2013; Zhu *et al.*, 2015). This technique is often used within a range of wearable devices, for example, smartwatches, as it allows respiratory rate to be estimated without additional sensors and thus, adding value to the existing data obtained (Charlton *et al.*, 2016). The issue with these methods is hundreds of different algorithms have been proposed to estimate respiration rate, thus, making it difficult to determine which, if any, are valid and reliable (Charlton *et al.*, 2016).

There are a number of imaging techniques that can be used to measure chest wall displacement, lung volumes, and hemidiaphragm excursion including computed tomography (CT), sonography (Matamis *et al.*, 2013), diaphragmatic video fluoroscopy (McCool *et al.*, 2012), and magnetic resonance imaging (MRI). CTs can estimate lung volumes, tissue and air, and the volume of the lung occupied by increased or decreased density. Similarly, MRI may be used to measure chest wall movement and measures such as lung volume. MRI allows lung volumes to be obtained within a single breath due to the high sampling frequency, and within different regions. However, the major disadvantage to these techniques is the considerable cost, time, and specialist equipment which can only be performed in a clinical/laboratory setting with substantial technical skills by the operator

(Meric *et al.*, 2016). Noncontact laser displacement sensors have also been used to measure rib cage and abdominal displacement, yielding similar tidal volume results as RIP (Hargrove *et al.*, 2009). The cameras on smartphones have also been used to estimate respiratory rate, Reyes and colleagues (2016) found a mean error of 0.02 breaths per minute when comparing the respiratory rate obtained from a smartphone camera and a spirometer. Similarly, webcams have also been used to estimate vital signs including respiratory rate, demonstrating good agreement with finger blood volume pulse (Poh *et al.*, 2011). Similarly, good agreement has been demonstrated when comparing a single webcam against a differential pressure sensor, with an average mean absolute error of 0.55 breaths per min at 720 resolution using a time domain algorithm, and perfect agreement using a frequency domain algorithm (Massaroni *et al.*, 2018).

Other non-contact method for measuring respiration is acoustic measurements. Microphones have been used to monitor breathing for long period of time during sleep with the capability of estimating respiratory rate, identifying apnea events (Massaroni *et al.*, 2020a), and agonal breathing, which can be a sign of cardiac arrest (Chan *et al.*, 2019). The microphones used for monitoring breathing can be built-in smartphone microphones (Azam *et al.*, 2018; Larson *et al.*, 2012), Bluetooth headset microphone (Hao *et al.*, 2015), or devices such as a smart speaker (Chan *et al.*, 2019). There is a lack of validation against gold standard methods for this measurement technique, and it can have interference from the surrounding environment making it difficult to obtain robust data (Massaroni *et al.*, 2020b).

Wearable Technology

Wearable devices are becoming increasingly popular within the public. The use of wearable devices is of particular interest for monitoring individuals within sport and fitness, wellbeing, and healthcare settings. The ideal wearable device should report all information wirelessly, be installed once or continuously, have a low consumption, long lasting battery life, all without hindering or constraining the wearer in any way (Aliverti, 2017; Chan *et al.*, 2012). Currently, most measurement tools within healthcare only offer a snapshot in time of an individual's health. However, monitoring an individual continuously over time under natural conditions would provide a much more beneficial insight into their health status with real-time feedback (Aliverti, 2017). There are a number of benefits to using wearable devices over traditional monitoring techniques. Within healthcare, wearable devices will allow monitoring of patients at home thus improving efficiency, patient comfort, and independent living for elderly, and reducing medical staff workload and cost of healthcare services. Along with this, post-operative rehabilitation and recovery may be improved. Within the general public, these devices will allow monitoring of physical activity, tracking trends and performance improvements. For professional athletes, performance may be tracked aiding the athletes and coaches to make informed decisions regarding training strategies (Aliverti, 2017). To date, there have been a number of wearable commercial solutions to measuring respiratory rate with the integration of current technology such as RIP (Wilhelm *et al.,* 2003), pressure sensors, strain gauges (Chu *et al.,* 2019), and motion sensors.

RIP, discussed previously in this chapter, has been embedded into clothing with one example of this being the LifeShirt. The LifeShirt (Vivometrics, Inc., Ventura, CA, USA) is a smart shirt which utilises RIP. Additionally, it integrates a tri-axial accelerometer and a three-lead ECG into the garment, allowing breathing rate, heart rate, posture, and activity to be measured simultaneously (Wilhelm *et al.,* 2003). Grossman and colleagues (2010) investigated the use of the LifeShirt during a protocol simulating daily activities in healthy adults and found that the LifeShirt displayed high correlation values ($r > 0.9$) for RR, Vt, and V_E when compared to mobile egrospirometry, except in a supine position. Hollier and colleagues (2014) compared the LifeShirt against spirometry in healthy individuals and obese individuals with untreated hypoventilation syndrome (OHS). It was found that there was unacceptable error between the two measurement systems for quantifying respiratory volumes, with the percentage error being ± 20% and ±36% for the healthy and OHS group, respectively.

Similarly, the Hexoskin (Hexoskin, Carre Technolohies Inc, Canada) is a smart t-shirt capable of measuring a number of parameters including breathing rate, tidal volume and minute ventilation using RIP to measure changes in the cross-sectional area of the chest. It has the ability to detect breathing rates from 3 to 80 breaths per minute. The t-shirt incorporates textile sensors which create electrically conductive loops around the torso. Changes in the cross-sectional area of the torso result alterations in the impedance of the loops (Aliverti, 2017). The validity of respiratory rate obtained by the Hexoskin has been assessed during different postures (Villar *et al.*,2015), walking (Montes *et al.*, 2018), and cycling (Elliot *et al.*, 2019), with respiratory rate displaying low coefficient of variation and high intraclass correlation values (Villar *et al.*, 2015), correlation values above 0.7 (Montes *et al*, 2018), and low typical errors of ~1-6% (Elliot *et al.*, 2019) when compared to a breath-by-breath analyser.

Zephyr (Zephyr BioHarness, Zephyr Technology, New Zealand) is also a wearable solution to monitoring breathing rate through a capacitive pressure sensor. Additionally, this garment has the ability to monitor other measures include heart rate through electrodes place on the chest wall, skin temperature through infra-red technology, tri-axial accelerometery and posture through a piezoelectric element (Johnstone *et al.*,2012a). Hailstone and Kilding (2011) investigated the reliability and validity of the Zephyr BioHarness during an incremental treadmill test to volitional exhaustion, reporting percentage errors of 4.4-8.7% when compared to traditional breath-by-breath analysis. Out of the possible measures, respiratory rate has been shown to have lower reliability agreement intra and inter device ($r < 0.7$), high coefficient of variation (<17.4) (Johnstone *et al.*, 2012a), and reduced accuracy at higher treadmill velocities (Johnstone *et al.*, 2012b).

AirGo (www.myairgo.com) is a wearable chest strap used to monitor respiratory function using integrated strain gauges. Using a sliver-coated strap worn over the lower ribcage and changes in chest circumference can be measured over time (Aliverti, 2017). It claims to be ‘medical grade’, however, to our knowledge there is only one recent peer-reviewed study which assessed the validation of the device against breath-by-breath analysis. Antonelli and colleagues (2020) presented a relative error median percentage of 0% for respiratory rate at rest and during exercise. Minute ventilation was also obtained with relative error median percentages ranging between ~1-40% with error increasing with exercise intensity.

The Spire Health Tag and Spiro Stone (www.spire.io) are sensors that can be placed on or clipped onto clothing such as the waist band. These systems use motion sensors to assess breathing patterns with regards to breathing rate and waveform, although the actual measurement system is not fully disclosed. Similarly, the Vitali Smart Bra & Gem (www.vitaliwear.com) also measures breathing patterns using a motion sensor system providing biofeedback to the wearer (Aliverti, 2017).

Overall, there are a number of wearable devices within the market which claim to measure lung function including breathing rate, minute ventilation, and ventilatory threshold. There are a number of difficulties in assessing wearable devices within the market. The companies and manufacturers often do not fully disclose the type of measurement principle used within their devices. The quality of the data obtained should be viewed with caution as the data and research methods are often not disseminated within the public or research community thus, many devices lack peer reviewed studies demonstrating the validity and reliability of the device and the data collected. Many wearable devices claim to be medical grade however, without any evidence to support this. For the few validation studies available, they often include small sample sizes or exclude data sets either due to device malfunctions or irregularities in the signal, possibly caused by abnormal breathing patterns. This suggests that some wearable devices are only appropriate for use in specific conditions and unable to cope with any type of irregular breathing patterns thus, may not be useful for individuals with a respiratory disease/disorder such as dysfunctional breathing. There is an undeniable need for most wearable devices to be clinically evaluated and the results published in peer-reviewed scientific journals as wearable devices have the potential to provide an individual with a personalised monitoring system over long period of time, in a variety of conditions.

Summary

It is clear that there is a need for specific, sensitive, and objective breathing pattern measurement systems to avoid misdiagnosis of respiratory disorders and subsequent inappropriate treatment. Currently, there are a large range of measurement systems available, with RIP, SLP, and OEP being the most widely used systems. OEP has been established as a valid breathing pattern measurement system, particularly during exercise, and is often used as the gold standard to compare other breathing pattern measurement systems against. OEP has been used to assess a variety of different population groups, however, OEP has not been used to assess athletes with respiratory disorders which are commonly misdiagnosed including exercise-induced bronchoconstriction and dysfunctional breathing. Monitoring an individual's breathing pattern continuously over time under natural conditions would provide beneficial insight into their health status and could also provide real-time feedback, which would require the development of a portable, wearable breathing pattern measurement device. Based on the previous literature, development into the measurement of breathing patterns in athletes is required to avoid misdiagnosis of respiratory disorders within this population group.

Thesis Aims

The overall aim of this thesis is to objectively characterise breathing patterns of pathological and non-pathological athletic populations, and to explore potential methods of managing dysfunctional breathing in athletes using OEP and potential alternatives.

More specifically, the main aims of this thesis were to:

1. Establish normative ranges for OEP-derived breathing pattern parameters for a healthy, physically active population at rest and investigate how these parameters change with exercise.
2. Investigate if OEP-derived breathing pattern parameters differ between active individuals with and without exercise-induced bronchoconstriction during bronchoconstriction induced by eucapnic voluntary hyperpnoea.
3. Determine if OEP-derived breathing pattern parameters differ between active individuals with and without dysfunctional breathing at rest and during an exercise challenge.
4. Investigate if a novel real-time OEP phase angle feedback system could be used as part of an acute breathing retraining intervention to reduce the phase angle between

the ribcage and abdomen in individuals with dysfunctional breathing at rest and during an exercise challenge.

5. Determine if the outputs from an inspiratory muscle training device (POWERbreathe) differ with simulated dysfunctional breathing patterns.
6. Compare the outputs from an impedance pneumography system to OEP and traditional breath-by-breath analysis including respiratory rate and tidal volume.

Chapter 3. General Methods

This chapter describes in detail the optoelectronic plethysmography (OEP) procedure used within the experimental chapters of this thesis.

3.1 Camera Set-Up

For each OEP data collection, eleven Qualisys cameras (Oqus, Qualisys AB, Goteborg, Sweden) were positioned around a suitable capture volume centred around a stationary bicycle (Lode – Corival, Netherlands) consisting of the performance volume and an additional margin around this volume (Figure 3.1). Each camera was configured for optimal marker visibility by adjusting the camera focus, marker threshold, gain, exposure, and flash time with the Qualisys Track Manager software. Camera calibration was performed prior to each participant's testing session. An L-frame (wand kit 750 mm, Qualisys Medical AB, Sweden) was placed near the bottom of the capture volume, at the level of the bicycle seat, to set the volume origin and global axes orientation (Figure 3.2a) with the x-axis and y-axis in line with the frontal and sagittal plane of the torso, respectively. A wand calibration was performed (length 501.5 mm, Qualisys Medical AB, Sweden) and accepted if the average residuals for each camera were below 1.0 mm. The sampling frequency was set to 100 Hz.

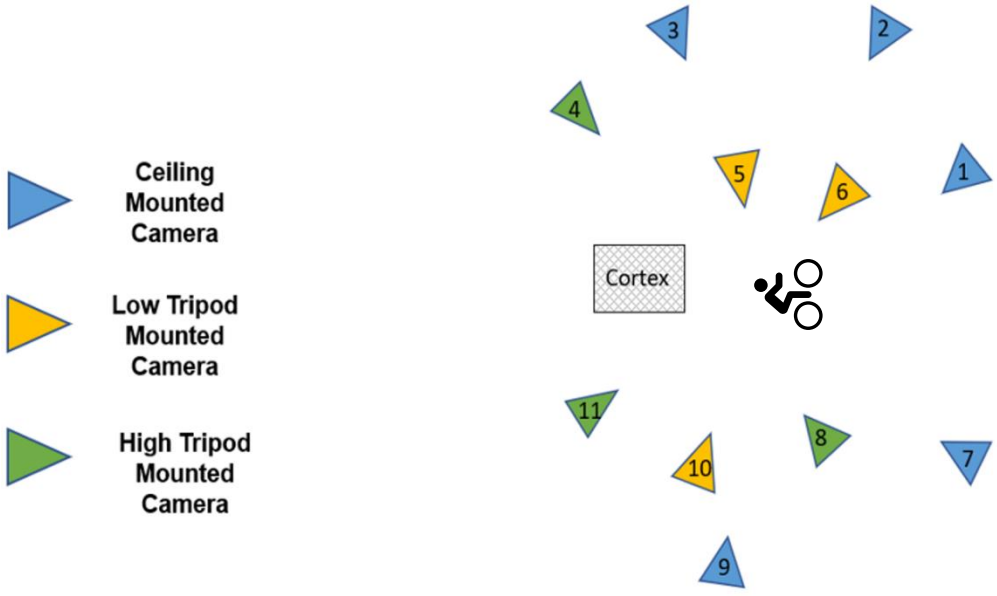


Figure 3.1 Camera set up with 11 Qualisys cameras positioned around a suitable capture volume.

a) b)

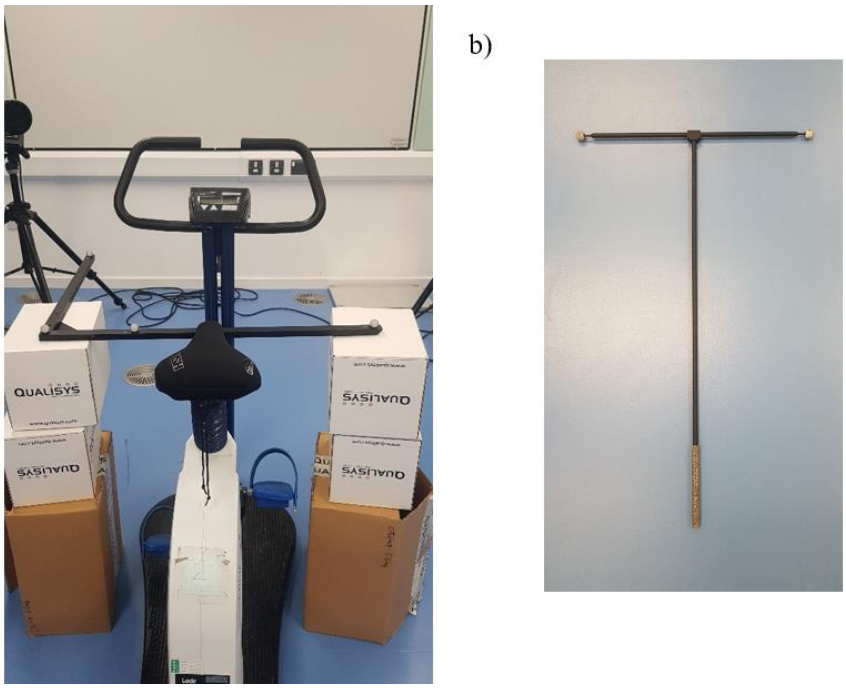


Figure 3.2 Calibration set up with L-frame (a) and T-wand (b)

3.2 Marker Set

A marker set (Figure 3.3a) consisting of 90 (38 posterior, 10 lateral, 42 anterior) spherical reflective markers (6- and 9-mm diameters) were attached to each participant using bi-adhesive hypoallergenic tape. This 90 marker set consists of a standard 89 marker set (Aliverti *et al.*, 2000; Massaroni *et al.*, 2017b) with an additional marker on C7 which adds an anatomical reference point. The larger diameter markers were placed on areas of the torso where the marker visibility was poorest, specifically the lateral sides of the torso under the participant's arms. The marker set (Figure 3.3b) was positioned slightly differently on females compared to male to minimise the effect of soft tissue movement in females (Layton *et al.*, 2011).

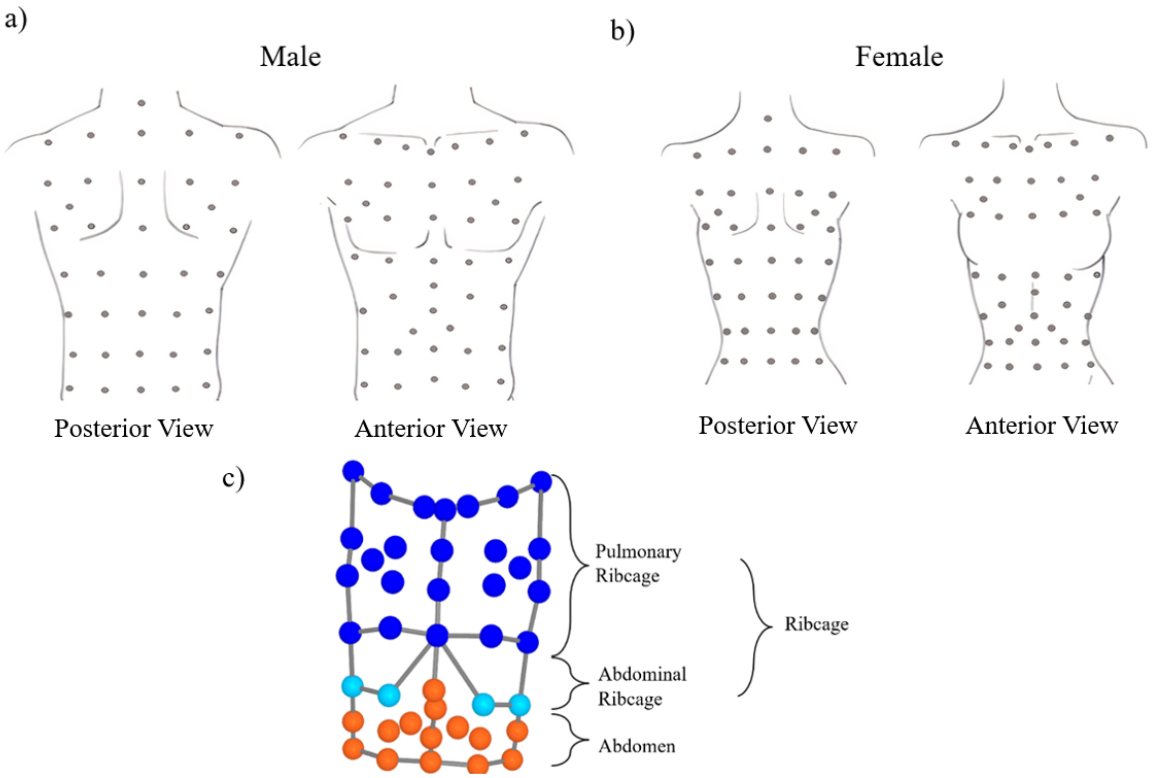


Figure 3.3 (a) 90 marker set positions on males, b) 90 marker set positions on females, (c) division of the torso into three compartments

Markers were placed in a grid-like pattern. This allowed for the division of the torso into three compartments; the pulmonary ribcage (RCp), the abdominal ribcage (RCa), and the abdomen (AB), as well as left and right sides of each compartment (Figure 3.3c). The RCp was defined as the area between the first row in line with the clavicles to the row in line with the xiphoid process. The RCa was defined as the area between the row in line with the xiphoid process to the lower costal margin. The AB was defined as the area between the row on the lower costal margin to the bottom row at the anterior superior iliac crest.

3.3 Participant Set-Up

Participants were positioned on the stationary bicycle in an upright position to maintain the torso at 90 degrees of hip flexion. This ensured optimal marker visibility. Initially, participants would place their hands on the bicycle handles (Figure 3.4), however, if this position obscured any of the markers, particularly the markers on the later sides of the torso, arms were extended to the sides and rested on stands. This would improve marker visibility and also attempted to avoid muscular contraction required to hold arms out in an extended position which may alter an individual's breathing pattern.

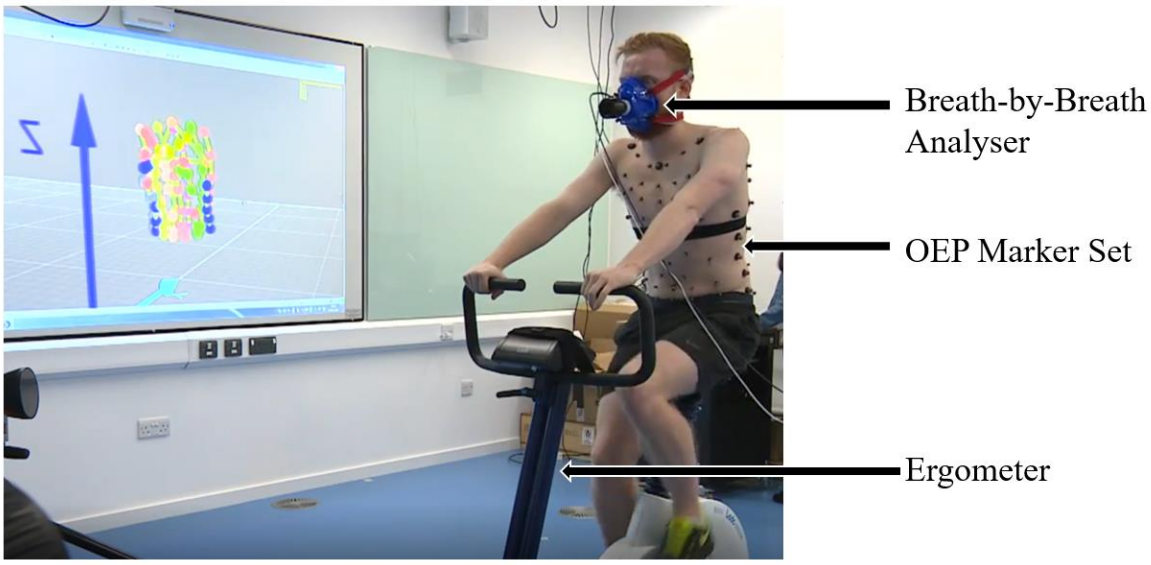


Figure 3.4 Participant positioning on the cycle ergometer

3.4 Volume Calculation

The marker configuration used allowed for the total volume of the torso to be calculated using the prism-based method (Massaroni *et al.*, 2017a). Seven centroids were created by calculating the mid-point between the front centre column markers and the spine markers corresponding to the seven rows of the marker set (Figure 3.5). Using three or four surface markers and one or two centroids, 82 prism shapes were created. The location of each prism can be seen in Figure 3.5.

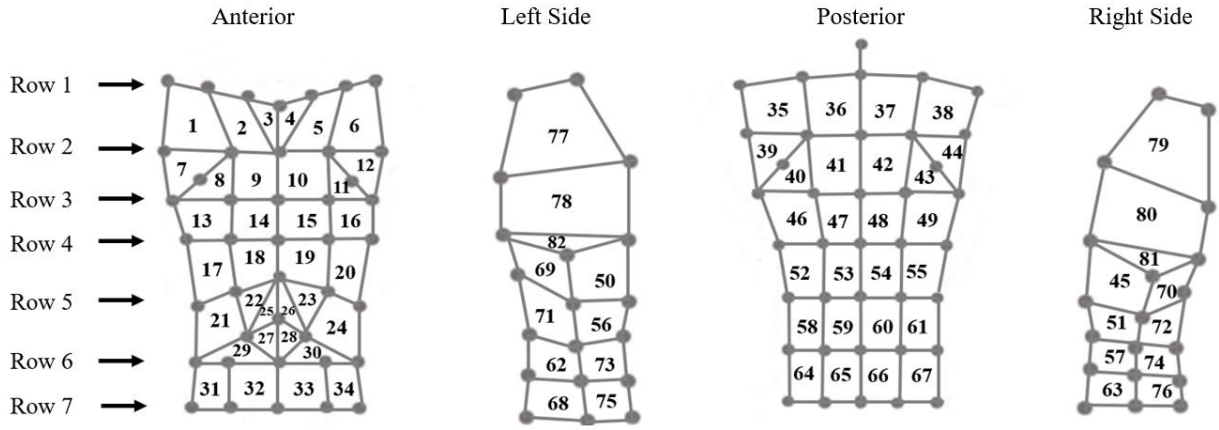


Figure 3.5 The seven rows used to create the seven centroids and the division of the surface markers to create 82 prism shapes.

Each prism was then divided into three tetrahedrons (Figure 3.6). Each marker consisted of x, y, and z coordinates. Surface marker A consisted of coordinates X_A, Y_A, and Z_A, surface marker B consisted of X_B, Y_B, and Z_B, surface marker C consisted of X_C, Y_C, and Z_C, surface marker D consisted of X_D, Y_D, and Z_D, centroid marker O1 consisted of X_{O1}, Y_{O1}, and Z_{O1}, and centroid marker O2 consisted of X_{O2}, Y_{O2}, and Z_{O2} (Massaroni *et al.*, 2017a). The volume of each tetrahedron was then calculated and summed to obtain the volume of a single prism using the Lagrange formula, using the following equation;

$$V_{Prism_{ABCDO1O2}} = V_{tetrahedron_{ABCO1}} + V_{tetrahedron_{ACDO2}} + V_{tetrahedron_{CDO1O2}}$$

$$= \frac{1}{6}.abs\left(\begin{vmatrix} 1 & X_A & Y_A & Z_A \\ 1 & X_B & Y_B & Z_B \\ 1 & X_C & Y_C & Z_C \\ 1 & X_{O1} & Y_{O1} & Z_{O1} \end{vmatrix}\right) + \frac{1}{6}.abs\left(\begin{vmatrix} 1 & X_A & Y_A & Z_A \\ 1 & X_C & Y_C & Z_C \\ 1 & X_D & Y_D & Z_D \\ 1 & X_{O2} & Y_{O2} & Z_{O2} \end{vmatrix}\right) + \frac{1}{6}.abs\left(\begin{vmatrix} 1 & X_C & Y_C & Z_C \\ 1 & X_D & Y_D & Z_D \\ 1 & X_{O1} & Y_{O1} & Z_{O1} \\ 1 & X_{O2} & Y_{O2} & Z_{O2} \end{vmatrix}\right) \quad (1)$$

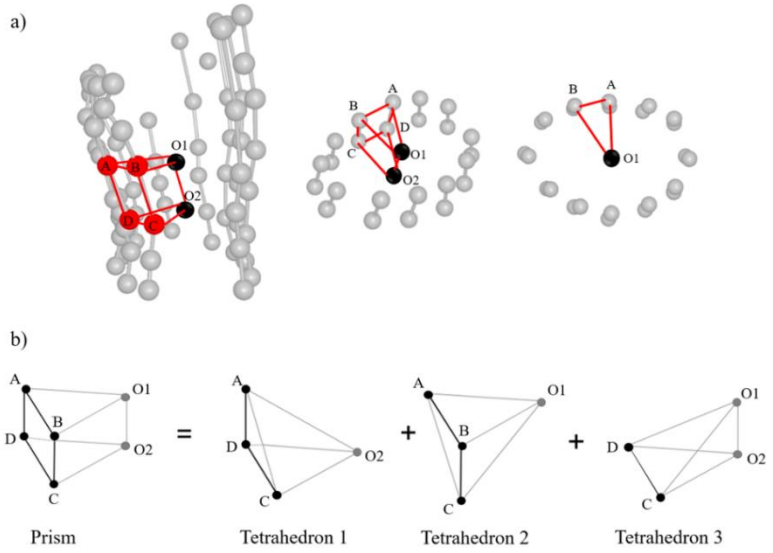


Figure 3.6 a) Different views of the creation of the prism with four surface markers (red) A, B, C, D, and two centroids (black) O1 and O2; b) Division of a single prism into three tetrahedrons.

Compartmental volumes were then calculated by summing the prisms within each compartment or relative section. The volume trace was initially filtered using a low-pass second order zero-phase Butterworth filter with a cut-off frequency of 2 Hz. This allowed rough estimates for the end of exhale and end of inhale indexes to be obtained. The original volume trace was then filtered using the same filter but with a cut-off frequency of 5 Hz (Massaroni *et al.*, 2017b). The indexes obtained from the 2 Hz filtered trace were then used to obtain the actual end of exhale and end of inhale values from the 5 Hz filtered trace by

obtaining min and max values, respectively, within a range (± 20 frames) around the matching indexes. Tidal volume was obtained by calculating the difference between end-expiratory and end-inspiratory volumes.

3.5 Breathing Parameters

Breathing parameters were then calculated for each breath from the tidal volume. Respiration rate was obtained from the volume data by counting the number of peaks within the trial. Timing indices and ratio parameters including inspiratory time (tI), expiratory time (tE), total breath time (tTot), tI/tE, tI/tTot. time to reach peak tidal expiratory flow (tPTEF) normalised to tE, time to reach peak tidal inspiratory flow (tPTIF) normalised to tI, and ratio of inspiratory to expiratory flow at 50% of Vt (IE50). Regional contribution parameters were calculated as a percentage of the total volume which included abdomen contribution of the total volume (AbCT), ribcage contribution of the total volume (RcCT), pulmonary ribcage contribution of the total volume (RCpCT), abdominal ribcage contribution of the total volume (RCaCT), combined abdominal ribcage and abdomen contribution of the total volume (RCaAbCT). A ribcage and abdomen volume index (RcAbIndex) was calculated by dividing the ribcage volume by the abdomen volume.

Phase angle represents the temporal movement of one thoracic compartment in relation to another compartment during each breathing cycle which allows the synchrony between the two compartments to be quantified. Perfect synchrony is represented by a phase angle of 0 degrees, while perfect asynchrony is represented by a phase angle of ± 90 degrees. Therefore, phase angle values have a range of 180 degrees between – 90 and + 90 degrees (Hmeidi *et al.*, 2017). The value of the phase angle indicates the amount of asynchrony while the sign of the value (positive or negative) indicates which compartment moves first. The order in which the compartments are inputted into the phase angle calculation will impact the sign of the phase angle output. Table 3.2 lists the phase angle value signs associated with the initiating compartment which will be used to define the phase angles throughout this thesis. Phase angles can be visually represented as Konno-Mead loops (Figure 3.7) and are calculated using the following equation;

$$Phase\ angle = \arcsin\left(\frac{m}{s}\right) \tag{2}$$

where s is the range of the first compartment displacement and m is the width of the loop at 50% of the second compartment displacement.

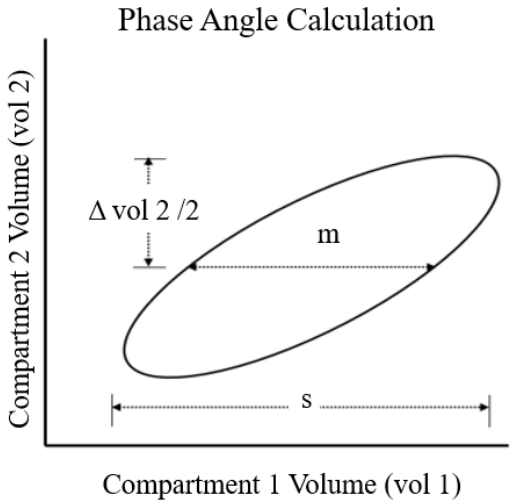


Figure 3.7 Illustration of the phase angle calculation with s representing the range of compartment one displacement and m representing the width of a single breath loop at 50% of the compartment 2 volume displacement. Phase angle is calculated as the arcsine(m/s).

Phase angles calculated in this thesis included the hemi-thoracic asynchronies between left and right sides of compartments, phase angles between compartments, and phase angles between shoulder movement and other compartments. The hemi-thoracic asynchronies included the phase angle between the left and right sides of the torso (HTA), the hemi-thoracic asynchrony between the left and right abdomen compartments (AbHTA), the hemi-thoracic asynchrony between the left and right ribcage compartments (RcHTA), the hemi-thoracic asynchrony between the left and right pulmonary ribcage compartments (RCpHTA), and the hemi-thoracic asynchrony between the left and right abdominal ribcage compartments (RCaHTA). The phase angles included the phase angle between the ribcage and the abdomen (RcAbPhase), between the pulmonary ribcage and the combined abdominal ribcage and abdomen (RCpAbPhase), and between the pulmonary ribcage and the abdominal ribcage (RCpRCaPhase).

The shoulder phase angles included the phase angles between the volume of the top two rows of markers of the marker set and the abdominal ribcage (RCaSPhase), the volume of the bottom two rows of the pulmonary ribcage (RCpSPhase), and the abdomen (AbSPhase). Figure 3.8 displays illustrations of the compartments relevant for each phase angle. Table 3.1 summaries the definitions for each breathing parameter calculated using OEP in this thesis.

Table 3.1 Breathing Parameter Definitions

Breathing Parameter	Description
Timing and Ratio Parameters	
RR	Respiratory Rate (breaths/minute)
Vt	Tidal Volume (litres)
V_E	Minute Ventilation (litres/minute)
tI	Inspiratory Time (seconds)
tE	Expiratory Time (seconds)
tTot	Total Breath Time (seconds)
tI/tE	Inspiratory Time / Expiratory Time Ratio
tI/tTot	Inspiratory Time / Total Breath Time Ratio
tPTEF	Normalised Time Taken to Reach Peak Expiratory Flow (seconds)
tPTIF	Normalised Time Taken to Reach Peak Inspiratory Flow (seconds)
IE50	Ratio of Inspiratory to Expiratory Flow at 50% of Tidal Volume
Regional Contribution Parameters	
AbCT	Abdomen Volume Contribution to Total Volume (%)
RcCT	Ribcage Volume Contribution to Total Volume (%)
RCpCT	Pulmonary Ribcage Volume Contribution to Total Volume (%)
RCaCT	Abdominal Ribcage Volume Contribution to Total Volume (%)
RCaAbCT	Combined Abdominal Ribcage and Abdomen Volume Contribution to Total Volume (%)
RcAbIndex	Ribcage Volume / Abdomen Volume Index
Phase Angle Parameters	
HTA	Left and Right Hemi-Thoracic Asynchrony (degrees)
AbHTA	Left and Right Abdomen Hemi-Thoracic Asynchrony (degrees)
RcHTA	Left and Right Ribcage Hemi-Thoracic Asynchrony (degrees)
RCpHTA	Left and Right Pulmonary Ribcage Hemi-Thoracic Asynchrony (degrees)
RCaHTA	Left and Right Abdominal Ribcage Hemi-Thoracic Asynchrony (degrees)
RcAbPhase	Phase Angle between Ribcage Volume and Abdomen Volume (degrees)
RCpAbPhase	Phase Angle between Pulmonary Ribcage Volume and the Combined Abdominal Ribcage and Abdomen Volume (degrees)
RCpRCaPhase	Phase Angle between Pulmonary Ribcage Volume and Abdominal Ribcage Volume (degrees)
AbSPhase	Shoulder Phase Angle: Phase Angle between the Volume of the top row of the Pulmonary Ribcage and the Abdomen Volume (degrees)
RCaSPhase	Shoulder Phase Angle: Phase Angle between the Volume of the top row of the Pulmonary Ribcage and the Abdominal Ribcage Volume (degrees)
RCpSPhase	Shoulder Phase Angle: Phase Angle between the Volume of the top row of the Pulmonary Ribcage and the bottom row of the Pulmonary Ribcage (degrees)

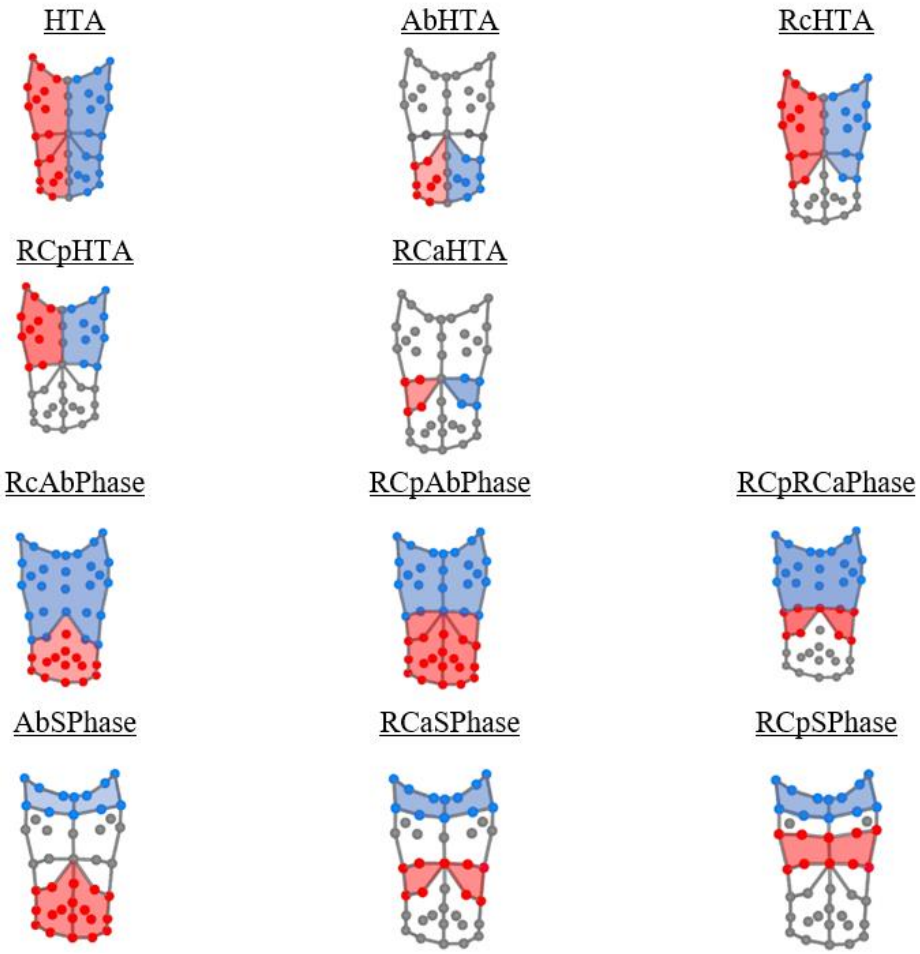


Figure 3.8 Illustration of the compartments related to each phase angle parameter. The blue and red areas show which sections of the thorax are involved in calculating each phase angle. The phase angle represents the synchrony between the two areas shown as described above.

Table 3.2 Phase angle value sign associated with the initiating compartment for each phase angle.

Phase Angle	Positive Sign (+)	Negative Sign (-)
HTA	Left side of torso	Right side of torso
AbHTA	Left side of Ab	Right side of Ab
RcHTA	Left side of Rc	Right side of Rc
RCpHTA	Left side of RCp	Right side of RCp
RCaHTA	Left side of RCa	Right side of RCa
RcAbPhase	Rc	Ab
RCpAbPhase	RCp	RCa + Ab
RCpRCaPhase	RCp	RCa
AbSPhase	Shoulders	Ab
RCaSPhase	Shoulders	RCa
RCpSPhase	Shoulder	RCp

Chapter 4. Thoraco-abdominal movement parameters derived from optoelectronic plethysmography successfully distinguish between exercise intensities in healthy individuals

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4.1 Introduction

The measurement of respiratory function is essential for the diagnosis and management of various respiratory disorders and diseases including asthma, chronic obstructive pulmonary disease (COPD) and dysfunctional breathing patterns. Spirometry via breath by breath (BbB) analysis is a valuable assessment method to assist in the diagnosis and management of pulmonary dysfunction, however, the method may alter natural breathing pattern due to the necessary mouthpiece (Layton *et al.,* 2013). Additionally, some individuals may not be able to perform the required breathing manoeuvres due to factors such as age and severity of disorder/disease (Hmeidi *et al.,* 2017). To overcome these issues various other non-invasive measurement techniques have been employed such as respiratory inductive plethysmography (RIP), structured light plethysmography (SLP), and optoelectronic plethysmography (OEP).

OEP measures the movement of the anterior and posterior chest wall using a three-dimensional motion capture system, allowing the compartmentalisation of the torso. It has been widely used to assess breathing patterns of several populations (Aliverti *et al.,* 2000; Dellaca *et al.,* 2010; Vogiatzis *et al.,* 2005a). OEP has previously been validated against BbB volume records at rest and during exercise (Layton *et al.,* 2013). This non-invasive measurement technique offers significantly richer information regarding an individual's breathing mechanics through the measurement of various breathing parameters when compared to simple BbB analysis (Pereira *et al.,* 2017). These breathing parameters consist of compartmental timing indices and ratios, displacement with time-derived parameters, and phase and regional contribution parameters, providing greater insight into the movement of the chest wall. There have been attempts to simplify and reduce the OEP marker set (Massaroni *et al.,* 2018), however, this leads to a reduction in the number of parameters that can be obtained by OEP. With the full 89 marker set, all of the parameters typically obtained

in a clinical environment using other methodologies such as SLP (Hmeidi *et al.*, 2017) can be obtained using OEP. These parameters include measures of right versus left asynchrony and rib cage versus abdominal asynchrony (e.g., Zoumot *et al.*, 2015). OEP has the advantage over SLP and RIP as it can be used during movement and exercise making it more suited to a wider population (e.g., children and athletes) and wider range of pathologies (e.g., exercise-induced bronchoconstriction).

A normal breathing pattern has previously been described as the pulmonary ribcage, abdominal ribcage, and the abdomen moving in synchronisation (Bradley & Esformes, 2014), and may be influenced by factors including posture, age, sex, disease, and exercise (Levai *et al.*, 2016b; Parreira et al., 2010). Differences in synchronisation measures have been shown between pathological populations and control groups at rest with SLP (Hmeidi *et al.*, 2017; Priori *et al.*, 2013) and with RIP (Panka *et al.*, 2010; Parreira *et al.*, 2010; Tomich *et al.*, 2010), demonstrating the potential of such measures to differentiate between populations. OEP has shown to have good agreement with RIP (Pereira *et al.*, 2017), and has been used to examine pathological populations during exercise using various different measures of asynchrony (Aliverti *et al.*, 2009; Lunardi *et al.*, 2014; Pereira *et al.*, 2017) but typically sample sizes of ten or fewer are used. However, there is a vast range of parameters that can be calculated using OEP data. Different studies have utilised different parameters, for example, some studies use rib cage versus abdominal phase angle, and others use pulmonary rib cage versus abdominal phase angle. In order to establish this motion-capture based technology for use in distinguishing between healthy and pathological populations and improved symptom management, normative values need to be established and the sensitivity and utility of different parameters needs to be considered.

Therefore, the purpose of this study was 1) to determine normative ranges for OEP breathing parameters from a non-pathological physically active population using a reasonably large sample of 47 participants and 2) to establish which of these OEP parameters alter with changes in chest wall mechanics during different exercise intensities. Establishing which of these parameters are capable of distinguishing between exercise intensities in a non-pathological population, will likely inform the selection of candidate parameters for objectively quantifying, distinguishing and interpreting pathological breathing patterns.

4.2 Methods

Participants

Forty-seven physically active healthy males (n = 28; age 34.3 ± 7.5 years; height 1.7 ± 0.1 m; body mass 77.8 ± 11.5 kg) and females (n = 19; age 28.4 ± 9.3 years; height 1.7 ± 0.1 cm; body mass 61.3 ± 5.5 kg) gave informed written consent to participate in this study. Participants were required to have a consistent training regime with a minimum of 5 hours of physical activity per week. Each participant performed a maximal lung function test and an incremental exercise challenge. This study was approved by the School of Sport & Exercise Sciences Research Ethics Advisory Group at the University of Kent (Prop 44_2017_18).

Equipment

Respiratory data was collected using OEP and a BbB analyser (CORTEX Biophysik MetaLyzer 3B-R2 stationary CPX system) simultaneously. The OEP data was collected following the methods described in Chapter 3. Briefly, the OEP data was collected using 11 Qualisys cameras (Oqus, Qualisys AB, Goteborg, Sweden) sampling at 100 Hz.

A marker set consisting of 90 (38 posterior, 10 lateral, 42 anterior) spherical reflective markers (6, 9 mm diameter) were attached to each participant. Markers were placed in a grid-like pattern (Cala *et al.,* 1996). This allowed the division of the torso into three compartments, namely the abdomen (AB), the abdominal rib cage (RCa), and the pulmonary rib cage (RCp). AB was defined as the area between the row of markers placed on the lower costal margin to the final row at the anterior superior iliac crest. RCa was defined by the row from the xiphoid process to the lower costal margin. Finally, RCp was defined from the first row at the clavicles to the row at the xiphoid process. The marker set was positioned slightly different on females compared to males in order to reduce the effect of soft tissue movement (Layton *et al.,* 2011). For the BbB analyser calibration of the pressure analyser, volume transducer, and O_2 and CO_2 gas analyser was performed prior to each participant.

Protocol

With the BbB analyser flow sensor and face mask in place, participants performed tidal breathing, and then performed an incremental cycle test, followed by a period of recovery. Participants were instructed to sit as upright as possible to maintain torso at 90^{o} of hip flexion. Initially, hands were placed on the handles of the cycle ergometer (Lode – Corival, Netherlands), however, if the arms obscured any markers from the cameras, arms were placed on stands to keep them extended to the sides while avoiding muscular contraction which may alter breathing mechanics. This posture was maintained during rest, recovery, and throughout the exercise challenge. The BbB analyser recorded respiratory data throughout the test, namely respiratory rate (RR), tidal volume (Vt), and respiratory exchange ratio (RER). The OEP system only recorded data for approximately 30 – 60 s at rest, low intensity, moderate intensity, high intensity exercise and recovery post exercise. Participants began cycling at 50 W, this was then increased by 30 W every minute. The exercise intensities were defined using the RER obtained from the BbB analyser, with low, moderate, and high intensity being defined as a RER value of 0.8, 1.0, >1.1, respectively (Deuster & Heled, 2008). Once a participant reached each intensity, OEP data was recorded for approximately 30 – 60 seconds. Once the high intensity data was recorded, the cycle test was terminated. The period of recovery began immediately post-exercise and lasted until the participant's RR returned to their resting values. OEP data was also recorded pre and post exercise challenge to obtain resting and recovery data for approximately 3 minutes.

Data Analysis

The OEP trials were gap filled for up to 10 frames using Qualisys Track Manager v2.17 and exported to MATLAB. The BbB analyser data was recorded and exported using CORTEX Biophysik MetaSoft CPX testing software. The BbB analyser trials were then cut to match the timings of the OEP data trials.

Volume and breathing parameters were calculated following the method described in Chapter 3. In summary, from the data obtained by the OEP system, volume and changes in volume were calculated using the prism-based method (Massaroni *et al.*, 2017a) using custom-written MATLAB (2017) scripts. Timing, compartment contribution and phase angle parameters were calculated from the OEP volume trace (see Table 3.1).

Additionally, novel shoulder phase angle parameters were created to assess the temporal movement of the shoulders in relation to the various thoracoabdominal regions. This study

included the phase angle between the shoulders and the following compartments: the abdominal ribcage volume (RCaSPhase), the bottom two rows of the RCp (RCpSPhase), and the abdomen (AbSPhase). The synchrony/asynchrony of each phase angle were visually represented as Konno-Mead loops (Figure 4.1a, 4.1b).

Statistical Analysis

The RRs obtained simultaneously from both measures were averaged over 30 seconds for each trial. Using SPSS (IBM SPSS Statistics 25), Pearson's correlation coefficient and intraclass correlation coefficients were calculated between the RRs and Vts obtained from the BbB analyser and the OEP system. A repeated measures ANOVA was used to analyse the influence of exercise intensity on the breathing parameters and 95% confidence intervals were computed for each OEP parameter. This study design and statistical analysis allows for control of between subject variability and does not impose the assumption that there will be linear changes in the parameters by treating the conditions as categorical.

4.3 Results

Table 4.1 displays the correlation coefficient values comparing the RR and Vt obtained by the OEP system to those obtained from the BbB analyser. At rest and across all exercise intensities, both OEP RR ($r > 0.90$, $p < 0.05$) and OEP Vt ($r > 0.80$, $p < 0.05$) significantly correlated with the BbB analyser. A high degree of reliability was found between both systems, with intraclass correlation coefficient values above 0.9 across all conditions. Bland-Altman analysis also confirmed that there was good agreement between the BbB analyser and the OEP derived parameters.

Table 4.1 Mean and standard deviation (SD) respiratory rate and tidal volumes obtained from the OEP system and BbB analyser with Pearson's correlation coefficient (r) and intraclass correlation coefficients (ICC) comparing respiratory rate and tidal volumes obtained from both systems (* represents $p \leq 0.05$).

	OEP		**BbB**		**Correlation**	
Respiratory Rate (breaths per min)	**Mean**	**SD**	**Mean**	**SD**	**r**	**ICC**
Rest	15.36	3.51	15.55	3.80	0.928*	0.959*
Low Intensity	21.83	4.80	22.59	5.08	0.928*	0.962*
Moderate Intensity	24.84	5.01	25.64	4.97	0.942*	0.962*
High Intensity	33.15	6.59	33.56	6.27	0.959*	0.977*
Recovery	23.37	4.69	24.08	4.15	0.915*	0.955*
Tidal Volume (litres)						
Rest	0.75	0.33	0.60	0.29	0.921*	0.957*
Low Intensity	1.36	0.29	1.16	0.23	0.835*	0.906*
Moderate Intensity	2.09	0.67	1.72	0.49	0.943*	0.953*
High Intensity	2.64	0.86	2.13	0.69	0.946*	0.963*
Recovery	1.83	0.53	1.56	0.44	0.875*	0.918*

Tables 4.2, 4.3 and 4.4 display the mean, standard deviation and upper and lower 95% confidence intervals for each breathing parameter calculated at rest, low, moderate, and high intensity exercise, and recovery post-exercise. Generally, as exercise intensity increased, there was an increase in the following parameters: RR, Vt, VE, tI/tE, tI/tTot, RcCT, RcAbIndex, and RCaHTA. Conversely, as the exercise intensity increased, the following parameters decreased: tI, tE, tTot, IE50, AbCT, RcAbPhase, RCpAbPhase, RCpRCaPhase, RCaSPhase, and AbSPhase. All other parameters did not display a consistent pattern with changes in exercise intensity.

Table 4.2 Healthy population group mean, standard deviation (SD) and 95% confidence intervals (CI) for the timing and ratio breathing parameters across each condition. Significant differences between conditions are represented by R*, L*, M*, H*, and Rc* as a mean significant ($p \leq 0.05$) difference from rest, low, moderate, high intensity exercise, and recovery, respectively.

	Rest				Low Intensity				Moderate Intensity				High Intensity				Recovery			
			95% CI				95% CI				95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
RR (brpm)	15.36	3.51	14.36	16.36	21.83	4.80	20.46	23.20	24.84	5.01	23.40	26.27	33.15	6.59	31.26	35.03	23.37	4.69	22.03	24.71
	L*, M*, H*, Rc*				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
Vt (litres)	0.75	0.33	0.65	0.84	1.36	0.29	1.28	1.45	2.09	0.67	1.90	2.29	2.64	0.86	2.40	2.89	1.83	0.53	1.68	1.98
	L*, M*, H*, Rc*				R*, M*, H, Rc*				R*, L*, H*				R*, L*, M*, Rc *				R*, L*, H*			
V_E (lit/min)	11.19	4.42	9.93	12.46	28.75	6.66	26.85	30.66	49.96	14.98	45.68	54.25	85.35	32.27	76.13	94.58	41.38	14.69	37.18	45.58
	L*, M*, H*, Rc*				R*, M*, H*, Rc*				R*, L*, H*				R*, L*, M*, Rc *				R*, L*, H*			
tI (s)	1.85	0.47	1.72	1.99	1.37	0.33	1.28	1.46	1.22	0.24	1.15	1.29	0.94	0.24	0.87	1.01	1.31	0.37	1.20	1.41
	L*, M*, H*, Rc*				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
tE (s)	2.37	0.66	2.18	2.56	1.54	0.39	1.43	1.65	1.32	0.28	1.24	1.40	0.99	0.27	0.91	1.07	1.44	0.42	1.32	1.56
	L*, M*, H*, Rc*				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
tTot (s)	4.23	1.08	3.92	4.54	2.90	0.69	2.70	3.10	2.54	0.53	2.39	2.69	1.94	0.52	1.79	2.09	2.75	0.77	2.52	2.97
	L*, M*, H*, Rc*				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H *			
tI/tE	0.80	0.14	0.75	0.84	0.90	0.08	0.88	0.92	0.93	0.06	0.91	0.95	0.95	0.04	0.94	0.96	0.91	0.09	0.89	0.94
	L*, M*, H*, Rc*				R*, H*				R*				R*, L*				R*			
tI/tTot	0.44	0.05	0.43	0.45	0.47	0.02	0.47	0.48	0.48	0.02	0.48	0.49	0.49	0.01	0.49	0.49	0.48	0.02	0.47	0.48
	L*, M*, H*, Rc*				R*, H*				R*				R*, L*				R*			
tPTEF	0.37	0.09	0.34	0.40	0.50	0.06	0.48	0.52	0.50	0.04	0.49	0.51	0.50	0.02	0.49	0.51	0.47	0.06	0.46	0.49
	L*, M*, H*, Rc*				R*				R*				R*				R*			
tPTIF	0.50	0.06	0.49	0.52	0.49	0.04	0.47	0.50	0.48	0.04	0.47	0.49	0.50	0.02	0.49	0.50	0.49	0.04	0.48	0.50
	M*								R*											
IE50	1.33	0.27	1.25	1.41	1.16	0.14	1.12	1.20	1.09	0.10	1.06	1.11	1.05	0.07	1.03	1.07	1.11	0.14	1.08	1.15
	L*, M*, H*, Rc *				R*, M*, H*				R*, L*				R*, L*, Rc*				R*, H*			

Table 4.3. Healthy population group mean, standard deviation (SD) and 95% confidence intervals (CI) for the regional contribution breathing parameters across each condition. Significant differences between conditions are represented by R*, L*, M*, H*, and Rc* as a mean significant ($p \leq 0.05$) difference from rest, low, moderate, high intensity exercise, and recovery, respectively.

	Rest				**Low Intensity**				**Moderate Intensity**				**High Intensity**				**Recovery**			
			95% CI				**95% CI**				**95% CI**				**95% CI**				**95% CI**	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
RCpCT (%)	57.87 L*, M*, H*, Rc*	4.56	56.56	59.17	58.55 R*	4.56	57.25	59.85	58.47 R*	4.93	57.06	59.88	58.50 R*	4.83	57.12	59.88	58.46 R*	4.89	57.06	59.86
RCaCT (%)	13.30	2.42	12.60	13.99	13.21 H*	2.28	12.56	13.86	13.40	2.51	12.68	14.12	13.49 **L***	2.38	12.81	14.17	13.21	2.39	12.52	13.89
RcCT (%)	71.17 L*, M*, H*, Rc*	3.54	70.16	72.18	71.77 R*, M*, H*	3.64	70.73	72.81	71.88 R*	3.81	70.80	72.97	72.01 R*, Rc*	3.91	70.89	73.12	71.67 R*, H*	3.97	70.53	72.80
AbCT (%)	28.83 L*, M*, H* Rc*	3.54	27.82	29.84	28.23 R*	3.64	27.19	29.27	28.12 R*	3.81	27.03	29.20	27.99 R*, Rc*	3.91	26.88	29.11	28.33 R*, H*	3.97	27.20	29.47
RCaAbCT (%)	42.13 L*, M*, H*, Rc*	4.56	40.83	43.44	41.45 R*	4.56	40.15	42.75	41.53 R*	4.93	40.12	42.94	41.50 R*	4.83	40.12	42.88	41.54 R*	4.89	40.14	42.94
RcAbIndex	2.52 L*, M*, H*, Rc*	0.44	2.39	2.65	2.60 R*	0.49	2.46	2.74	2.62 R*	0.51	2.48	2.77	2.64 R*, Rc*	0.54	2.49	2.80	2.60 R*, H*	0.53	2.45	2.75

Table 4.4. Healthy population group mean, standard deviation (SD) and 95% confidence intervals (CI) for the phase angle breathing parameters across each condition. Significant differences between conditions are represented by R*, L*, M*, H*, and Rc* as a mean significant ($p \leq 0.05$) difference from rest, low, moderate, high intensity exercise, and recovery, respectively.

	Rest				Low Intensity				Moderate Intensity				High Intensity				Recovery			
			95% CI				95% CI				95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
HTA (deg)	0.02	0.09	0.00	0.05	0.01	0.27	-0.07	0.08	0.02	0.35	-0.08	0.12	0.13	0.30	0.04	0.22	0.05	0.12	0.01	0.08
RcHTA (deg)	0.03	0.10	0.00	0.06	-0.02 H*	0.25	-0.09	0.05	0.08	0.36	-0.02	0.18	0.14 L*	0.28	0.06	0.22	0.05	0.15	0.01	0.09
AbHTA (deg)	0.01	0.14	-0.03	0.05	-0.15	0.75	-0.36	0.07	-0.20	0.87	-0.45	0.05	-0.25	0.70	-0.45	-0.05	-0.03	0.38	-0.14	0.08
RCpHTA (deg)	0.02	0.09	0.00	0.05	-0.05	0.31	-0.14	0.04	-0.02	0.48	-0.16	0.12	0.00	0.29	-0.08	0.09	0.00	0.16	-0.04	0.05
RCaHTA (deg)	0.00 H*	0.13	-0.04	0.04	0.06 H*	0.37	-0.05	0.17	0.15 H*	0.50	0.01	0.30	0.54 R*, L*, M*, Rc*	0.66	0.35	0.73	0.19 H*	0.35	0.09	0.29
RcAbPhase (deg)	-0.20 L*, M*, H*, Rc*	0.39	-0.31	-0.09	-0.84 R*, L*, M*	0.76	-1.06	-0.62	-1.66 R*, L*, Rc*	0.99	-1.95	-1.38	-1.90 R*, L*, Rc *	1.06	-2.20	-1.60	-1.07 R*, M*, H *	0.90	-1.33	-0.81
RCpAbPhase (deg)	-0.20 L*, M*, H*, Rc*	0.42	-0.32	-0.08	-0.64 R*, M*, H*	0.70	-0.84	-0.44	-1.37 R*, L*, H*, Rc *	0.87	-1.61	-1.12	-1.79 R*, L*, M*, Rc *	1.28	-2.16	-1.43	-0.81 R*, M*, H*	0.94	-1.08	-0.54
RCpRCaPhase (deg)	-0.07 H*	0.28	-0.15	0.01	-0.11	0.57	-0.27	0.06	-0.35	0.82	-0.58	-0.11	-0.45 R*, Rc*	1.08	-0.75	-0.14	-0.06 H*	0.80	-0.29	0.17
RCaSPhase (deg)	0.02	0.27	-0.05	0.10	0.00	0.57	-0.16	0.16	-0.27	0.91	-0.53	-0.01	-0.33 Rc*	1.26	-0.69	0.03	0.06 H*	0.93	-0.21	0.33
RCpSPhase(deg)	0.12	0.19	0.07	0.18	0.16	0.36	0.06	0.26	0.22	0.49	0.09	0.36	0.17	0.67	-0.02	0.36	0.31	0.45	0.18	0.44
AbSPhase (deg)	-0.06 L*, M*, H*, Rc*	0.43	-0.18	0.07	-0.68 R*, M*, H*	0.89	-0.94	-0.43	-1.61 R*, L*, H*, Rc*	1.25	-1.96	-1.25	-2.26 R*, L*, M*, Rc*	1.70	-2.74	-1.77	-0.97 R*, M*, H*	1.20	-1.31	-0.63

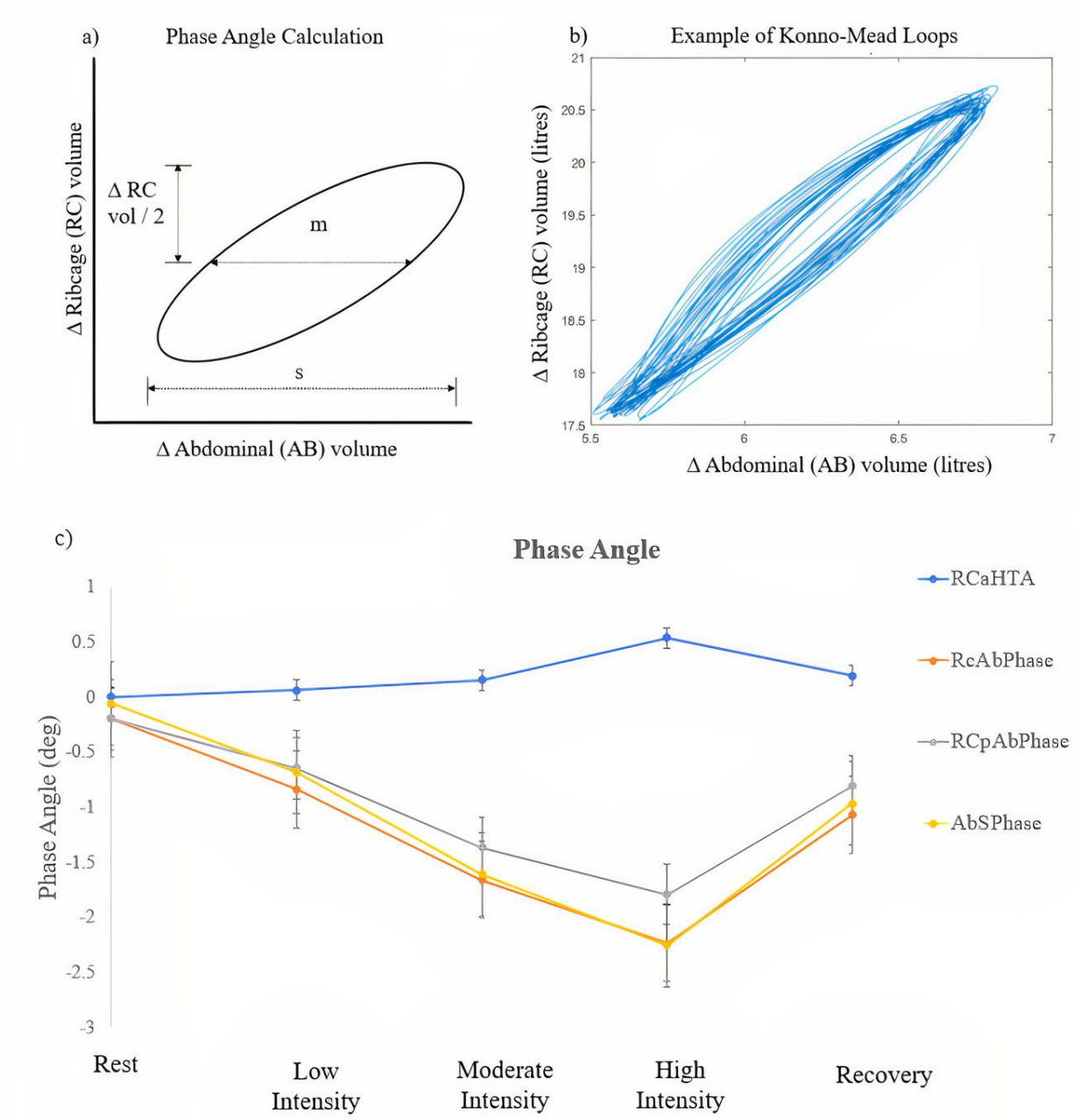


Figure 4.1 a) Illustration of the phase angle calculation with m representing the width of a single breath loop at 50% of the ribcage change in volume, s represents the range of abdominal volume change, and phase angle is calculated as the arcsine(m/s); b) illustrates an example of an individual's Konno-Mead loops for the phase angle between the ribcage and abdomen (RcAbPhase) over a period of high intensity exercise; c) displays how each phase angle changes with increased exercise intensity and recovery post-exercise.

Data for comparisons of the breathing parameters across exercise intensities are also shown in Table 4.2, 4.3 and 4.4. The following timing and ratio parameters were found to be significantly different at any condition (exercise intensities and post-exercise recovery) compared to rest: RR, Vt, VE, tI, tE, tTot, tI/tE, tI/tTot, tPTEF, and IE50 (Table 4.2). The following relative volume contribution parameters were found to be significantly different at any condition compared to rest: RcCT, RCpCT, AbCT, RcAbCT, and RcAbIndex (Table 4.3). The following phase angle parameters displayed a significant difference between any condition and rest: RcAbPhase, RCpAbPhase and AbSPhase. The phase angles RCpAbPhase, and AbSPhase displayed significant differences between high intensity exercise and all other conditions. More specifically, RCpAbPhase and AbSPhase demonstrated significant differences between each exercise condition (Table 4.4). RCaHTA displayed significant differences between high intensity and every other condition. No significant differences were found for the remaining parameters.

4.4 Discussion

For the first time this study has established 95% confidence intervals for OEP derived breathing parameters for a healthy population and determined how these values change during different exercise intensities. A significant novel finding was that two new OEP derived parameters, namely RcAbIndex and AbSPhase, were able to distinguish between rest and exercise. These metrics relate to compartment contribution and timings that are useful in diagnosis of disorders such as dysfunctional breathing pattern (Todd *et al.,* 2016). The present study further shows that such metrics can be determined during exercise. The breathing patterns described by these metrics alter with exercise intensity, which is potentially important for identifying exercise-induced dysfunctional breathing. The shoulder-based metrics can only be calculated using this motion-capture based technology.

The results of this study support the previously established agreement between OEP and BbB analyser. Both RR and Vt obtained from the OEP system highly correlated with those obtained by the BbB analyser with correlations of $r > 0.90$, $p < 0.05$ and $r > 0.80$, $p < 0.05$, respectively, across all conditions. Additionally, intraclass correlation values were all above 0.9 for RR and Vt indicating excellent reliability (Table 4.1). Similar to this study, a good linear relationship between both methods has been found with r^2 values greater than 0.9 across various populations including healthy individuals (Cala *et al.,* 1996), intensive care patients (Aliverti *et al.,* 2000), COPD patients (Aliverti *et al.,* 2004; Vogiatzis *et al.,* 2005a) and new-borns (Dellaca *et al.,* 2010). During rest and submaximal exercise conditions, Vt obtained from OEP has been validated against spirometry with r^2 values greater than 0.95 (Kenyon *et al.,* 1997; Vogiatzis *et al.,* 2005b). Layton and colleagues (2013) found high correlation values ($r = 0.963$, $p < 0.001$) during a maximal cycle test in healthy individuals which is comparable to this study.

The OEP derived values for the timing and ratio parameters obtained in this study are comparable to previous studies using OEP, SLP and RIP at rest including RR (Aliverti *et al.,* 2009; Matos *et al.,* 2012; Parreira *et al.,* 2010; Priori *et al.,* 2013), tI (Panka *et al.,* 2010), tTot (Lunardi *et al.,* 2014), and tI/tTot (Hmeidi *et al.,* 2017; Lunardi *et al.,* 2014; Matos *et al.,* 2012; Parreira *et al.,* 2010). In contrast to the other measurement systems (SLP and RIP), OEP allows these parameters to be measured during exercise. Therefore, few studies have investigated these parameters in non-pathological participants during exercise (Aliverti *et al.,* 2009; Pereira *et al.,* 2017). RR, Vt, VE, tI, tE, tTot, tI/tE, tI/tTot, tPTEF, and IE50 were found to be significantly different at any exercise intensity when compared to rest (Table

4.2). As expected, RR and tPTEF increased and tI, tE and tTot decreased with exercise intensity. The changes in tI/tE and tI/tTot indicate that tI became closer to 50% of the tTot due to a greater reduction in tE as exercise intensity increased. IE50 indicated that the expiratory flow is always greater than the inspiratory flow at 50% of tidal volume and as exercise increased the difference between the flow values decreased. RR, Vt, VE, tI, tE, tTot, tI/tE, tI/tTot, and IE50 each display a positive or negative correlation with exercise intensity with significant differences between rest and exercise. Therefore, these parameters may be used to distinguish whether an individual is exercising or at rest.

The following relative contribution parameters were significantly different at any exercise intensity when compared to rest: RCpCT, RcCT, AbCT, and RcAbCT (Table 4.3). Priori and colleagues' (2013) reported similar values for AbCT, RCpCT, and RCpCT to those obtained in this study using OEP in a non-pathological population at rest. The reported normative values for non-pathological individuals in this study may be useful to distinguish between non-pathological and pathological populations for example, individuals with asthma or COPD in future studies.

This study established a novel relative contribution parameter, RcAbIndex. a measure of the ribcage volume relative to the abdomen volume. As expected, the ribcage volume was greater than the abdomen volume throughout all conditions (Table 4.3). RcAbIndex was significantly different at rest when compared to all other conditions including recovery. RcAbIndex increased during high intensity indicating that the ribcage volume contribution increased with a decrease in the abdomen volume contribution. This demonstrates the potential use of this novel parameter to not only distinguish between rest and exercise but also between high intensity exercise and recovery post exercise.

This study established a second novel parameter, shoulder phase angle (AbSPhase). This parameter is a measure of the phase angle between the volume of top of the pulmonary ribcage, which represents the movement of the shoulders and the abdomen volume. AbSPhase indicates that the abdomen moves first across all of the conditions in this non-pathological population. AbSPhase demonstrated significant differences between rest and all other conditions, and also between exercise conditions. This indicates that this novel parameter was sensitive enough to distinguish not only between rest and exercise but also between exercise intensities. These differences indicate that the 'shoulders' and the abdomen were closer to perfect synchrony during rest, low exercise intensity, and recovery, and became slightly more out of sync during moderate and high exercise intensity. Other phase

angle parameters including RcAbPhase, and RCpAbPhase displayed significant differences between rest and all other conditions (Table 4.4). RCpRCaPhase presented no significant differences between rest and low and moderate exercise intensity which in line with the results found by Aliverti and colleagues (2009) where RCpRCaPhase did not exceed 14 degrees and did not significantly change with exercise.

The pattern seen with AbSPhase of the abdomen moving first is also demonstrated by RcAbPhase and RCpAbPhase with increased angles in the negative direction as exercise intensity increased, and a subsequent reduction in asynchrony during the post-exercise recovery phase (figure 4.1c). Similarly, Pereira and colleagues (2017) found an increase in RcAbPhase with negative angles in healthy individuals as exercise intensity increased. They attributed the negative angles to the early recruitment of the abdominal muscles, possibly related to more expansion of the chest wall during exercise. The use of these phase angle parameters as measures to distinguish between rest and exercise intensities may be useful for a variety of groups including healthy individuals, individuals with respiratory disorders such as asthma, COPD and athletes with dysfunctional breathing. Alves and colleagues (2008) demonstrated that cycling is an effective exercise to improve ventilation in COPD patients and similar to this study, RcAbPhase increased with exercise. This increase in RcAbPhase has also been found in the elderly during an increase in daily living activities (Panka *et al.*, 2010). Other systems, including SLP and RIP, can only be used accurately in a stationary position. This limits their use in terms of environment and population. OEP not only offers the benefit of being used during exercise, it also is the only system with the ability to measure the novel parameter AbSPhase. Therefore, these OEP-derived parameters may be used to aid with breathing function training and monitor progression in a variety of populations.

Previous studies have used some of these breathing parameters to investigate the effects of surgery (Matos *et al.*, 2012; Zoumot *et al.*, 2015), the influence of breathing exercises in a healthy population (Vieira *et al.*, 2014), as part of a post-operative treatment (Tomich *et al.*, 2010), and the effects of exercise on the elderly (Panka *et al.*, 2010) and COPD patients (Alves *et al.*, 2008). Vieira and colleagues (2014) showed that breathing pattern parameters, namely RR, Vt, VE, RcCT, and RcAbPhase can be altered by breathing exercises including diaphragmatic breathing, inspiratory sighs, sustained maximal inspiration, and intercostal exercises in healthy individuals at rest when compared to quiet breathing. It has also been shown that breathing exercises can influence breathing pattern and thoracoabdominal

movement post-surgery in obese individuals (Tomich *et al.*, 2010). The present study further demonstrates the value of OEP-derived parameters in understanding breathing mechanics and the potential change with different pathological conditions.

The participants of this study were not exhaustively screened for respiratory disorders. All participants had no known respiratory conditions Some individuals may not know if they have a dysfunctional breathing pattern, however, each participant performed spirometry based maximal lung function test and no participant revealed any symptoms of dysfunctional breathing during rest or exercise. Assessing different pathological populations with these parameters established here will allow for direct comparisons to be made between the groups and determine how these breathing parameters change not only with exercise intensity but also with respiratory disorder. The novel parameters and ranges provided in this study can be used to begin to establish and understand pathological breathing patterns and to profile different populations. This is not only important for diagnosis but also to monitor the impact of treatment for various conditions.

In conclusion, having established agreement between OEP and the BbB analyser, normative ranges for OEP derived breathing pattern and thoracoabdominal movement parameters have been presented for the first time for a non-pathological population including novel parameters RcAbIndex, and AbSPhase. The OEP derived parameters RR, Vt, VE, tI, tE, tTot, tI/tE, tI/tTot, IE50, RcAbIndex, RcAbPhase, RCpAbPhase, and AbSPhase successfully distinguished between rest and exercise, and exercise intensities including recovery. Our findings provide the basis for future studies to investigate these parameters to differentiate between healthy individuals and those with respiratory disease or breathing pattern disorder. This adds to the potential use of OEP as a breathing function measurement system as other measurement systems typically used such as SLP and RIP are much more difficult to administer, particularly during exercise.

Chapter 5. OEP breathing pattern response to the eucapnic voluntary hyperpnoea (EVH) test in athletes with exercise induced bronchoconstriction

5.1 Introduction

Exercise-induced bronchoconstriction (EIB) is a term which describes the temporary and reversible narrowing of the airways often occurring during exercise, limiting expiration (Hull *et al.,* 2016). EIB is reversible spontaneously post-exercise or with the inhalation of β_2 agonists. It is most commonly seen in athletes and diagnosis is vital due to the potential impact this condition can have on both the athletes' health and athletic performance (Price *et al.,* 2014). The prevalence of EIB in elite athletes has been reported to be approximately 20%, with this value increasing for sports such as swimming and cycling, with prevalence reported to be up to 69% (Dickinson *et al.,* 2005; Levai *et al.,* 2016a).

Despite its prevalence, EIB is complex to diagnose as there is no gold standard diagnostic measure. EIB diagnosis often involves the use of direct or indirect bronchial provocation tests, where baseline and post-provocation lung function values are compared, as discussed in the literature review. However, athletes may only demonstrate a positive response to one of the provocation tests (Vakali *et al.,* 2017). Therefore, more than one objective measure along with the athlete's history of symptoms may be required to confirm EIB diagnosis in athletes. The eucapnic voluntary hyperpnoea (EVH) test is a sensitive method for diagnosing EIB in athletes, and is a surrogate for exercise (Parsons *et al.,* 2013). Similar to other provocation tests, spirometry is used to measure maximal lung function pre and post EVH, with a reduction in forced expiratory volume in one second (FEV_1) of > 10% at two consecutive time points as the diagnostic criteria. However, spirometry required to perform the EVH test involves maximal effort breathing manoeuvres which may not be possible for certain individuals based on type or severity of disease (Hmeidi *et al.,* 2017).

Optoelectronic plethysmography (OEP) has previously been used to investigate the effects of histamine-induced bronchoconstriction (Gorini *et al.,* 1999) and methacholine induced bronchoconstriction (Filippelli *et al.,* 2003) on breathing pattern in asthmatics. Similarly, the breathing pattern of individuals with mild asthma but without EIB has been successfully investigated during mild exercise using OEP (Fregonezi *et al.,* 2018). Each of these studies demonstrated altered breathing pattern during bronchoprovocation and exercise, more

specifically altered compartment volume contributions and thoracoabdominal asynchrony. Feitosa and colleagues (2019) constructed a receiver operating characteristic curve for OEP-derived end-expiratory volume of the chest wall for children with exercise induced asthma. It was established that OEP can accurately be used to identify exercise induced asthma in children. In the previous chapter, we demonstrated that OEP is capable of identifying alterations in breathing pattern between rest and exercise in a healthy active population. Unlike other pattern measurement systems such as structured light plethysmography (Hmeidi *et al.,* 2017; Hmeidi *et al.,* 2018), OEP can be used during exercise. OEP provides detailed information of breathing pattern mechanics which cannot be obtained from spirometry and can be useful for individuals unable to perform the maximal effort breathing manoeuvres required for spirometry in the diagnosis of EIB (Feitosa *et al.,* 2019).

Therefore, the purpose of this chapter was 1) to investigate how OEP-derived breathing pattern parameters alter during bronchoconstriction induced by EVH in athletes with EIB and 2) to establish if these breathing parameters differ between athletes with and without EIB.

5.2 Methods

Participants

Eighty-five active participants gave informed written consent to participate in this study. This study was approved by the School of Sport & Exercise Sciences Research Ethics Advisory Group at the University of Kent (Prop 21_2018_19). Participants were required to have a consistent training regime with a minimum of 5 hours of physical activity per week. Each participant completed a questionnaire (Appendix 2) regarding respiratory symptoms during or post-exercise and in different environmental conditions. Participants were asked to report on the following symptoms: coughing, wheezing breathing in and/or out, chest tightness breathing in and/or out, dyspnoea, and excess mucus production during or post-exercise. Individuals who reported no respiratory symptoms and had no history of any respiratory disease/disorder were assumed to have a healthy breathing pattern. Individuals who reported respiratory symptoms and/or had a previous/current diagnosis of asthma/EIB performed an EVH challenge. Participants who demonstrated a positive EVH result were considered to have EIB. A summary of participant information can be seen in Table 5.1. This study's protocol is summarised in Figure 5.1.

Table 5.1 Participant Information (mean ± standard deviation)

	N	Age (yrs)	Height (m)	Weight (kg)	Activity Level (hrs/week)
Healthy Group n = 47					
Male	28	34.3 ± 12.1	1.8 ± 0.1	77.8 ± 11.5	5.6 ± 2.7
Female	19	28.4 ± 9.3	1.7 ± 0.1	61.3 ± 5.5	5.3 ± 2.6
EIB Group n = 10					
Male	8	35.5 ± 13.7	1.8 ± 0.1	81.0 ± 10.6	6.2 ± 2.7
Female	2	30.5 ± 3.5	1.7 ± 0.1	71.0 ± 18.4	5.5 ± 0.71

Protocol

Participants currently using asthma medication were required to withhold use prior to the EVH challenge; montelukasts for 7 days prior, inhaled corticosteroid and combination inhalers for at least 3 days prior, long acting β_2-agonists for 2 days prior, and short acting β_2-agonists for 1-day prior testing (Parsons *et al.,* 2013). Forced exhaled nitric oxide (FeNO) was measured using nitric oxide monitor (NIOX VERO, Circassia) following guidelines from the American Thoracic Society (2006).

OEP data was collected pre- and post-EVH challenge, and post-inhaler in participants who experienced a bronchoconstriction post EVH in a standing position. OEP data was recorded for the healthy group during tidal breathing.

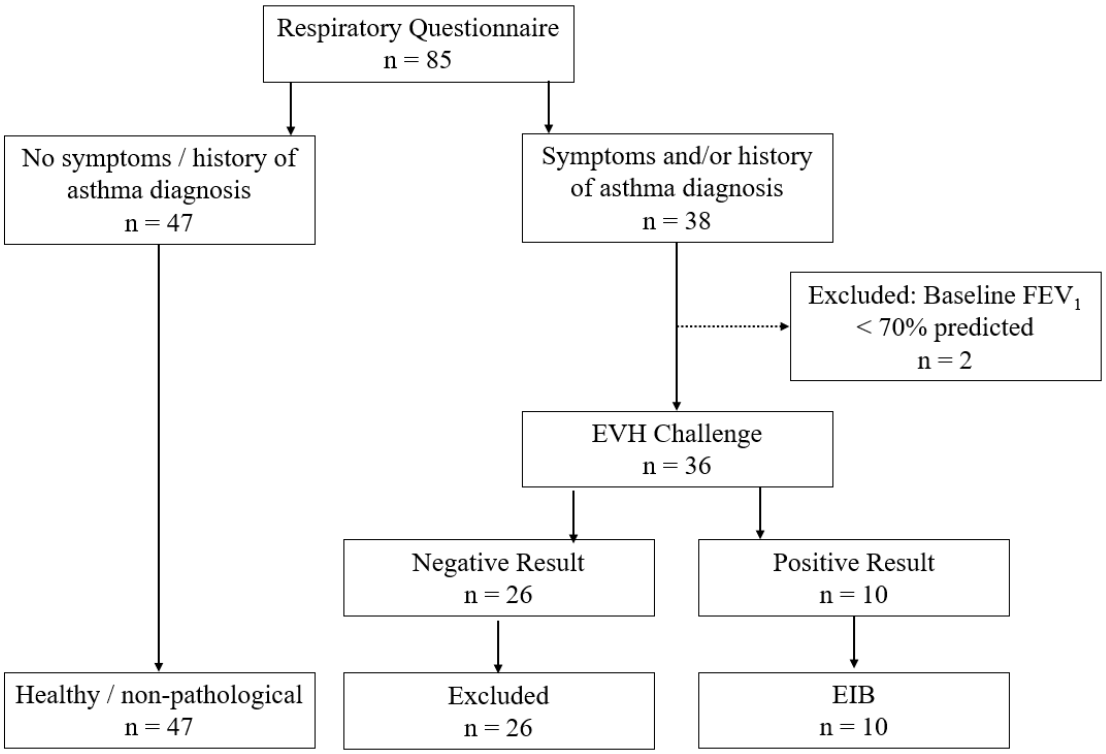


Figure 5.1 Summary of participant grouping and study protocol

Optoelectronic Plethysmography

OEP data was collected using 11 Qualisys cameras (Oqus 3, Qualisys AB, Goteborg, Sweden) sampling at 100 Hz. A marker set of 90 (38 posterior, 10 lateral, 42 anterior) reflective markers were placed on the participants torso in a grid-like pattern (Aliverti *et al.*, 2000; Massaroni *et al.*, 2017a). This allowed for the torso to be split into three compartments namely, the abdomen (AB), abdominal ribcage (RCa), and the pulmonary ribcage (RCp). A detailed description of the OEP set-up and calculations can be found in Chapter 3.

Eucapnic Voluntary Hyperpnoea (EVH) test

A spirometer (ML3500 Micro Medical Spirometer, Cardinal Health, UK) was used to measure baseline and post-EVH lung function. Forced expiratory volume in one second (FEV_1), forced vital capacity (FVC), FEV_1:FVC ratio (FEV_1/FVC), peak expiratory flow (PEF), and forced expiratory flow between 25 and 75% of FVC (FEF_{25-75}) were recorded in accordance with the European Respiratory Society criteria (Miller *et al.,* 2005). Baseline FEV_1 was established from the best of three manoeuvres. To be eligible to perform the EVH challenge participants must have had a baseline FEV_1 value greater than 70% of their predicted value, if below this they were excluded from this study.

The EVH challenge involved participants, who had a previous asthma diagnosis and/or reported respiratory symptoms during exercise, inhaling a dry gas mixture consisting of 74% nitrogen, 21% oxygen, and 5% carbon dioxide for six minutes. Participants aimed to achieve a target minute ventilation (V_E) of 85% of their predicted maximal voluntary ventilation rate (baseline FEV_1 multiplied by 30) (Spiering *et al.,* 2004). V_E was recorded using a dry gas meter. Participants were given visual feedback of their V_E every minute and verbal encouragement throughout the challenge. Two maximal flow-volume loops were recorded 3 and 5 minutes post-EVH challenge with the flow-volume loop with the highest FEV_1 being recorded. A positive EVH result was defined as a fall by at least 10% in FEV_1 from baseline at two consecutive time points (Parsons *et al.,* 2013). If participants displayed this fall in FEV_1, they were given 200 µg inhaled β_2-agonist and performed flow-volume loop manoeuvres 10 minutes post bronchodilator. Participants were instructed to stay in the laboratory until their FEV_1 returned to within 10% of their baseline.

Data Analysis

OEP data was processed using Qualisys Track Manager (v2019.2 Build 4610). Custom-built MATLAB (version R2019a) scripts were used to calculate OEP derived-breathing parameters for both population groups. Timing, compartment contribution and phase angle parameters were calculated from the OEP volume trace (see Table 3.1 for full list and description) using a custom-built MATLAB script.

Statistical Analysis

95% upper and lower confidence intervals were calculated for each OEP parameters across each condition i.e., rest, evoked, and recovery. A repeated measures ANOVA was used to analyse the differences between EVH conditions within EIB group. The healthy group at rest was compared to the EIB group for each of the EVH conditions using t-tests.

5.3 Results

From the 38 individuals recruited who reported respiratory symptoms and/or had a previous diagnosis of asthma (n = 16), 10 participants had a positive EVH challenge. Baseline FEV_1 was established from the best of three manoeuvres prior to performing the EVH which ranged between 80 – 124% of the predicted value across the EIB group. The participants included in the EIB group all experienced a reduction of ≥10% in FEV_1 from baseline at two consecutive time points i.e., positive EVH result, confirming the presence of EIB. The mean and standard deviation FeNO for the EIB group was 61 ± 37 ppb.

From the spirometry data (Table 5.2), the EIB group displayed a significant reduction in FEV_1, FVC, and FEV_1/FVC post-bronchoconstriction from the resting baseline ($p < 0.05$). Post-inhaler, FEV_1 ($p = 0.02$), FVC, and FEV_1/FVC ($p < 0.01$) displayed an increase from the post-bronchoconstriction condition. FEV_1/FVC for the healthy group's baseline rest was significantly different to the EIB group across all of the EVH challenge time points i.e., rest, post-bronchoconstriction and recovery ($p < 0.01$). FEV_1 and FVC were significantly different between the healthy group at rest and the EIB group post-bronchoconstriction ($p < 0.01$).

Table 5.2 Mean and standard deviation (SD) spirometry data for the healthy group at rest and the EIB group across the EVH conditions. * represent significant difference between the EIB condition and the healthy group at rest ($p \leq 0.05$).

	FEV_1 (litres)		**Predicted FEV_1 (%)**		**FVC (litres)**		**Predicted FVC (%)**		**FEV_1/FVC (%)**	
	Mean	**SD**	**Mean**	**SD**	**Mean**	**SD**	**Mean**	**SD**	**Mean**	**SD**
Healthy Group										
Rest	3.88	0.82	101	14	4.63	1.10	101	14	85	6
EIB Group										
Rest	4.01	0.71	101	11	5.36	1.34	111	16	76	11*
Post-EVH	2.90	0.58*	74	16*	4.76	1.10	99	10	62	12*
Recovery (post bronchodilator)	3.57	0.85	92	12	4.93	1.63	106	13	73	11*

Tables 5.3, 5.4, and 5.5 display the mean, standard deviation, and upper and lower 95% confidence intervals for each breathing parameter during each condition of the EVH challenge for individuals with EIB. These tables also display the significant changes across the EVH challenge time points within the EIB group. V_E, RCpRCaPhase, RCaSPhase, and RCpSPhase displayed significant differences ($p < 0.01$) between rest and post-bronchoconstriction. RCpSPhase also displayed a significant difference ($p = 0.03$) between post-bronchoconstriction and recovery. All other parameters were non-significant across rest, post-bronchoconstriction, and recovery.

Table 5.3 Exercise-induced bronchoconstriction group mean, standard deviation (SD) and 95% confidence intervals (CI) for the timing and ratio breathing parameters across each EVH challenge time points. Mean significant differences ($p \leq 0.05$) between time points are also displayed with a significant difference from rest, post-bronchoconstriction, and recovery represented by R*, PB*, and Rc*, respectively.

	Rest				**Post-Bronchoconstriction**				**Recovery**			
			95% CI				**95% CI**				**95% CI**	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
RR (brpm)	11.40	2.77	9.69	13.12	13.47	3.05	11.58	15.36	13.45	3.41	11.33	15.56
Vt (litres)	0.91	0.44	0.64	1.19	1.05	0.34	0.84	1.27	0.96	0.29	0.78	1.14
V_E (lit/min)	8.82	2.14 PB*	7.49	10.15	13.89 R*	3.82	11.52	16.26	12.23	3.47	10.08	14.38
tI (s)	2.23	0.64	1.83	2.63	1.94	0.34	1.72	2.15	1.96	0.35	1.74	2.18
tE (s)	2.92	0.79	2.43	3.40	2.44	0.55	2.09	2.78	2.37	0.49	2.06	2.68
tTot (s)	5.14	1.33	4.32	5.97	4.45	0.83	3.94	4.97	4.40	0.78	3.91	4.88
tI/tE	0.75	0.17	0.64	0.85	0.82	0.13	0.74	0.90	0.84	0.08	0.79	0.89
tI/tTot	0.42	0.06	0.39	0.46	0.45	0.04	0.42	0.47	0.45	0.02	0.44	0.47
tPTEF	0.37	0.11	0.30	0.44	0.36	0.08	0.31	0.41	0.39	0.08	0.34	0.44
tPTIF	0.52	0.05	0.49	0.55	0.51	0.04	0.49	0.54	0.52	0.03	0.50	0.54
IE50	1.16	0.15	1.07	1.25	1.27	0.22	1.13	1.41	1.19	0.19	1.07	1.31

Table 5.4 Exercise-induced bronchoconstriction group mean, standard deviation (SD) and 95% confidence intervals (CI) for the regional contribution breathing parameters across each condition.

	Rest				Post-Bronchoconstriction				Recovery			
			95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
RCpCT (%)	56.96	3.76	54.63	59.29	56.71	3.70	54.42	59.00	56.46	3.77	54.13	58.80
RCaCT (%)	13.04	1.90	11.86	14.22	13.34	1.71	12.27	14.40	13.60	1.51	12.66	14.54
RcCT (%)	70.01	3.59	67.78	72.24	70.03	3.62	67.79	72.28	70.06	4.10	67.51	72.60
AbCT (%)	29.99	3.59	27.76	32.22	29.97	3.62	27.72	32.21	29.94	4.10	27.40	32.49
RCaAbCT (%)	43.04	3.76	40.71	45.37	43.29	3.70	41.00	45.58	43.54	3.77	41.20	45.87
RcAbIndex	2.38	0.41	2.12	2.63	2.38	0.41	2.13	2.63	2.40	0.47	2.10	2.69

Table 5.5 Exercise-induced bronchoconstriction group mean, standard deviation (SD) and 95% confidence intervals (CI) for the phase angle breathing parameters across each condition. Mean significant differences ($p \leq 0.05$) between time points are also displayed with a significant difference from rest, post-bronchoconstriction, and recovery represented by R*, PB*, and Rc*, respectively.

	Rest				Post-Bronchoconstriction				Recovery			
			95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
HTA (deg)	-0.02	0.06	-0.05	0.02	0.03	0.09	-0.02	0.09	-0.01	0.08	-0.06	0.04
RcHTA (deg)	-0.01	0.08	-0.06	0.04	0.04	0.10	-0.02	0.10	-0.01	0.10	-0.07	0.06
AbHTA (deg)	0.00	0.14	-0.09	0.09	0.03	0.27	-0.13	0.20	0.04	0.32	-0.09	0.17
RCpHTA (deg)	-0.03	0.10	-0.10	0.03	0.01	0.12	-0.07	0.08	0.00	0.12	-0.07	0.07
RCaHTA (deg)	0.00	0.12	-0.08	0.07	0.14	0.23	0.00	0.29	0.04	0.09	-0.02	0.10
RcAbPhase (deg)	-0.03	0.38	-0.27	0.20	-0.46	0.67	-0.87	-0.04	-0.16	0.63	-0.55	0.23
RCpAbPhase (deg)	-0.07	0.31	-0.27	0.12	-0.14	0.43	-0.41	0.12	-0.12	0.43	-0.38	0.14
RCpRCaPhase (deg)	-0.11 PB*	0.23	-0.25	0.04 R*	0.42	0.54	0.08	0.76	0.09	0.27	-0.08	0.26
RCaSPhase (deg)	-0.04 PB*	0.24	-0.19	0.11 R*	0.70	0.69	0.28	1.13	0.19	0.38	-0.05	0.42
RCpSPhase(deg)	0.10 PB*	0.16	0.00	0.20	0.66 R*, Rc*	0.54	0.32	0.99	0.29 PB*	0.31	0.09	0.48
AbSPhase (deg)	0.06	0.49	-0.24	0.36	-0.10	0.50	-0.41	0.21	-0.06	0.61	-0.44	0.32

Table 5.6 displays the comparison between healthy individuals at rest and individuals with EIB at rest, post-bronchoconstriction, and recovery. RR, tE and tTot displayed significant differences between the two groups at rest. All other breathing parameters were non-significant between the two groups at rest. Vt, RCaHTA, RCpRCaPhase, RCaSPhase, and AbSPhase were significantly different between healthy individuals at rest and individuals with EIB post-bronchoconstriction. When comparing the healthy group at rest to the EIB group during recovery, RCpSPhase was significantly larger for the EIB group. All other breathing parameters were non-significant.

Table 5.6 Comparison of the EIB group at rest, evoked, and during recovery to the healthy group at rest. * represents significant with $p \leq 0.05$.

	Healthy Group		EIB Group								
	Rest		Rest			Post-Bronchoconstriction			Recovery		
	Mean	SD	Mean	SD	Sig	Mean	SD	Sig	Mean	SD	Sig
RR (brpm)	15.36	3.51	11.40	2.77	< 0.01 *	13.47	3.05	0.12	13.45	3.41	0.14
Vt (litres)	0.75	0.33	0.92	0.45	0.35	1.05	0.34	< 0.01*	0.96	0.29	0.08
V_E (lit/min)	11.19	4.42	8.82	2.14	0.27	13.89	3.82	0.08	12.23	3.47	0.51
tI (s)	1.86	0.47	2.23	0.64	0.23	1.94	0.34	0.60	1.96	0.36	0.52
tE (s)	2.37	0.66	2.92	0.79	0.05 *	2.44	0.56	0.76	2.37	0.49	0.99
tTot (s)	4.23	1.08	5.14	1.33	0.04 *	4.46	0.83	0.54	4.40	0.78	0.66
tI/tE	0.80	0.14	0.75	0.17	0.59	0.82	0.13	0.66	0.84	0.08	0.39
tI/tTot	0.44	0.05	0.42	0.06	0.54	0.45	0.04	0.63	0.46	0.03	0.33
tPTEF	0.37	0.09	0.37	0.11	0.10	0.36	0.08	0.77	0.39	0.08	0.50
tPTIF	0.51	0.06	0.52	0.05	0.68	0.51	0.04	0.69	0.52	0.03	0.54
IE50	1.33	0.27	1.16	0.15	0.18	1.27	0.22	0.52	1.19	0.19	0.15
RCpCT (%)	57.88	4.56	56.96	3.77	0.83	56.71	3.70	0.46	56.46	3.77	0.39
RCaCT (%)	13.30	2.42	13.04	1.91	0.94	13.34	1.71	0.96	13.60	1.51	0.72
RcCT (%)	71.17	3.54	70.01	3.59	0.61	70.04	3.63	0.36	70.06	4.10	0.40
AbCT (%)	28.83	3.54	29.99	3.59	0.61	29.97	3.62	0.36	29.95	4.10	0.40
RCaAbCT (%)	42.13	4.56	43.04	3.77	0.83	43.29	3.70	0.46	43.54	3.77	0.39
RcAbIndex	2.52	0.44	2.38	0.41	0.60	2.38	0.41	0.36	2.40	0.47	0.45
HTA (deg)	0.03	0.09	-0.02	0.06	0.32	0.03	0.09	0.77	-0.01	0.08	0.31
RcHTA (deg)	0.03	0.10	-0.01	0.08	0.43	0.04	0.10	0.66	-0.01	0.10	0.32
AbHTA (deg)	0.01	0.14	-0.00	0.14	0.96	0.03	0.27	0.74	0.04	0.21	0.64
RCpHTA (deg)	0.02	0.09	-0.04	0.10	0.20	0.01	0.12	0.61	0.00	0.12	0.53
RCaHTA (deg)	0.00	0.13	-0.00	0.12	0.99	0.14	0.23	< 0.01*	0.043	0.09	0.36
RcAbPhase (deg)	-0.20	0.39	-0.03	0.38	0.53	-0.46	0.67	0.10	-0.16	0.63	0.81
RCpAbPhase (deg)	-0.20	0.42	-0.07	0.31	0.67	-0.14	0.43	0.71	-0.12	0.43	0.60
RCpRCaPhase (deg)	-0.07	0.28	-0.11	0.23	0.92	0.42	0.54	< 0.01*	0.09	0.27	0.12
RCaSPhase (deg)	0.02	0.27	-0.04	0.24	0.79	0.70	0.69	< 0.01*	0.19	0.38	0.12
RCpSPhase (deg)	0.12	0.19	0.10	0.16	0.96	0.66	0.54	< 0.01*	0.29	0.31	0.04*
AbSPhase (deg)	-0.06	0.43	0.06	0.49	0.78	-0.10	0.50	0.79	-0.06	0.61	0.98

5.4 Discussion

This study, for the first time, demonstrates alterations in key OEP-derived breathing pattern parameters during bronchoconstriction induced by the EVH challenge in athletes with EIB, including V_E, RCpRCaPhase, RCaSPhase, and RCpSPhase. The alterations in OEP parameters were in line with those obtained from the traditional spirometry. Our findings also identify significant differences in the OEP-derived Vt, RCaHTA, RCpRCaPhase, RCaSPhase, and RCpSPhase between athletes without EIB at rest and with EIB during bronchoconstriction.

IE50 displayed an increase post-bronchoconstriction followed by a subsequent decrease post-inhaler, although not significant (Table 5.3). Both OEP derived tPTEF and tPTIF demonstrated a decrease post-bronchoconstriction with a subsequent increase during recovery (Table 5.3). This is an expected response as an increase in airway resistance would be anticipated post-bronchoconstriction, and therefore, a reduction in tidal expiratory flow at 50% of tidal volume (TEF50). As TEF50 is a component of IE50, it would be expected that a reduction in TEF50 would elicit an increase in IE50 (Hmeidi *et al.,* 2017). This is further supported by the mean reduction in FEV_1, FEV_1/FVC, and PEF obtained from the spirometer (Table 5.3) at rest and bronchoconstriction (-1.1 L, -14%, and -190 L/min, respectively). IE50 has previously been used to successfully distinguish between healthy controls and individuals with asthma (Hmeidi *et al.,* 2017) and individuals with chronic obstructive pulmonary disease (COPD) (Motamedi-Fakhr *et al.*, 2016) using SLP. In this study, the EIB group did not display any significant differences in IE50 throughout the EVH challenge to the healthy group at rest (Table 5.6). It has previously been shown with SLP that individuals with asthma have significantly higher IE50 values compared to healthy controls and can be reduced with a bronchodilator (Hmeidi *et al.,* 2017). Similarly, individuals with COPD have also displayed higher IE50 values compared with healthy controls (Motamedi-Fakhr *et al.,* 2016). With further investigation, OEP-derived IE50 may be capable of identifying an increase in airway resistance during the bronchoconstriction in athletes with EIB which is in line with the spirometer.

All of the regional contribution parameters in this study showed no significant changes across the conditions of the EVH challenge within the EIB group (Table 5.4). Similarly, when comparing individuals with and without EIB, there were no significant differences in the regional contribution parameters (Table 5.6). This is comparable to previous research where Hmeidi and colleagues (2017) found no significant differences between regional

parameters when comparing asthmatic and healthy children at rest using SLP. Similarly, regional parameters, such as RcCT, have been shown to display no significant changes with the use of a bronchodilator in asthmatics when compared to a resting state (Hmeidi *et al.*, 2017). The findings of this study indicate that OEP-derived regional contribution parameters do not change post-bronchoprovocation during the EVH challenge in athletes with EIB, therefore, these parameters may not be essential for monitoring athletes with EIB.

Hemi-thoracic asynchrony (HTA) is a measure of the asynchrony between the movement of the left and right sides of the torso compartments. Throughout the EVH challenge, each HTA parameter did not significantly change (Table 5.5). Previously it has been reported that HTA is not significantly different between healthy and asthmatic individuals (Hmeidi *et al.*, 2017). Our findings support this, as HTA between the torso and for each compartment, except RCa, for the EIB group were not significantly different from the non-EIB group (Table 5.6). In contrast, RCaHTA was significantly different between EIB bronchoconstriction and healthy group at rest. The EIB group displayed more asynchrony during bronchoconstriction compared to the healthy group, although this was not significantly different from rest or post-inhaler within the EIB group. Within individuals with different respiratory diseases, in order to cope with different ventilatory demands, such as exercise, different breathing pattern adjustments occur (Wilkens *et al.*, 2010). This alteration in RCaHTA may be a result of the breathing pattern adjustments individuals EIB use to cope with the demands of bronchoconstriction or may be due to chance because of the small sample size.

The phase angles between the upper and lower ribcage (RCpRCaPhase), between the shoulders and the lower ribcage (RCaSPhase), and between the shoulders and the upper ribcage (RCpSPhase) displayed a significant increase in value from rest to bronchoconstriction in the EIB group. RCpSPhase also demonstrated a significant reduction in phase angle during recovery compared to the bronchoconstriction condition (Table 5.5). This indicates that during bronchoconstriction athletes with EIB display more asynchrony in relation to these phase angles compared to rest, with the upper ribcage (RCp) moving prior to the lower ribcage (RCa), and the shoulders moving before both the upper and lower ribcage. When comparing the athletes with and without EIB at rest, there were no significant differences in any of the phase angle parameters (Table 5.6). Similarly, Hmeidi and colleagues (2017) found no significant differences between asthmatics and healthy individuals when measuring RcAbPhase with SLP at rest. When comparing the EIB group during bronchoconstriction to the healthy group at rest, RCpRCaPhase, RCaSPhase, and

RCpSPhase were significantly greater in the EIB group during bronchoconstriction indicating greater asynchrony between these compartments. This increase in asynchrony during bronchoconstriction demonstrates that individuals with EIB alter their breathing pattern during bronchoconstriction which may be a compensatory mechanism in order to cope with the airway obstruction. OEP has previously been used to assess breathing pattern changes in asthmatics during bronchoconstriction induced by both histamine (Gorini *et al.*, 1999) and methacholine (Filippelli *et al.,* 2003). However, phase angle parameters were not included in these studies. Post-inhaler, RCpSPhase was significantly greater in the EIB group compared to the healthy group at rest (Table 5.6). As RCpSPhase was also significantly lower post-inhaler in comparison to during bronchoconstriction (Table 5.5), this indicates that this phase angle began to revert to its resting value post-inhaler but had not recovered completely. The other phase angles seemed to have reverted back to their resting state post-inhaler, with RCpSPhase as the only phase angle potentially demonstrating a delayed response to the inhaler. The differences found in this study indicate that these OEP-derived phase angle parameters may have the potential use of identifying EIB in athletes and monitoring response to inhaler therapy.

This study demonstrates that although the compartment contributions were not significantly different between the groups (Table 5.6), the temporal movement of these compartments differ during bronchoconstriction. The order in which the compartments move during a breath may provide the basis for distinguishing between the groups. At rest, for the healthy breathing pattern group the abdomen is the first compartment to move with the upper ribcage being the final compartment. For the EIB group at rest, the lower ribcage moves first and the upper ribcage is the last to move. Therefore, for both groups at rest, the upper ribcage is the final compartment to move. In contrast, during bronchoconstriction, the compartment order for the EIB group changes, with the abdomen now initiating the breath and the lower ribcage is the final compartment to move. This demonstrates that when bronchoconstriction occurs the individuals with EIB display an altered breathing pattern in response to bronchoconstriction, with delayed movement of the lower ribcage.

The recruitment of athletes with EIB is difficult due to its under, over, and mis-diagnosis (Kavanagh *et al.,* 2019). From the ten participants in this study, five individuals had a previous diagnosis of EIB/asthma and were currently taking asthmatic medication, while five individuals did not have a previous diagnosis. Additionally, six individuals with a previous diagnosis of EIB/asthma and currently taking medication were recruited but

demonstrated a negative EVH test and therefore, were excluded from this study. This is in line with previous research which has demonstrated that physician-diagnosed asthma has a high rate of misdiagnosis, approximately 33% (Aaron *et al.,* 2017). Two of the recruited individuals had a previous diagnosis but were unable to perform the EVH challenge as their baseline FEV_1 was below 70%. The difficulty in recruiting individuals with EIB is not an uncommon issue due to the difficulty in diagnosing EIB (Rundell *et al.,* 2001). Previous studies have found that ~15 – 44% of athletes recruited who reported exercise-induced respiratory symptoms demonstrated a negative provocation challenge response, while ~8-24% of elite athletes demonstrated a positive provocation challenge response without reporting any exercise-induced respiratory symptoms (Bougault *et al.,* 2010; Vakali *et al.,* 2017). Although this study provides some interesting insight into breathing pattern changes in athletes with EIB, a larger sample size would be required to establish distinct cut-off points for these parameters in order to develop OEP as an alternative to spirometry and as a potential diagnostic aid for EIB in athletes. Additionally, this study only included those with mild to moderate EIB. Individuals with severe, chronic asthma may present with a similar or a unique breathing pattern profile at rest, however, in order to determine this further exploration is required using OEP. Finally, individuals within different age-ranges should be investigated as this study only includes individuals between 18 – 45 years of age.

In conclusion, OEP breathing pattern response to the bronchoconstriction via EVH in athletes with EIB has been presented for the first time in this study, with alterations in breathing pattern coordination in line with traditional spirometry. OEP successfully identified significant differences between rest and bronchoconstriction with the following parameters: V_E, RCpRCaPhase, RCaSPhase, and RCpSPhase indicating greater compartment asynchronisation during bronchoconstriction leading to altered breathing mechanics. Additionally, when compared to athletes without EIB at rest, athletes with EIB demonstrated significantly larger Vt, RCaHTA, RCpRCaPhase, RCaSPhase, and RCpSPhase during bronchoconstriction. Although this preliminary data provides insight into the breathing patterns associated with EIB, a larger, more varied sample size would be required to establish these OEP-derived parameters, allowing OEP to distinguish between healthy athletes and athletes with EIB. If established, OEP would potentially provide an alternative method of EIB diagnosis which would be useful for those unable to perform the required breathing manoeuvres for traditional spirometry.

Chapter 6. Using OEP to successfully distinguish between athletes with and without a dysfunctional breathing pattern at rest and during exercise

6.1 Introduction

Many respiratory symptoms are not specific to a single respiratory disease or disorder. Symptoms including dyspnoea, chest tightness, & wheezing are all symptoms often related to respiratory disorders such as dysfunctional breathing and exercise-induced bronchoconstriction (Dickinson *et al.,* 2010; Molphy *et al.,* 2014). Reports suggest that approximately 20% of athletes who report significant exercise respiratory symptoms receive an inappropriate diagnosis of asthma or exercise induced bronchoconstriction (EIB) (Dickinson *et al.,* 2005). This highlights the need for diagnostic measures to be sensitive and specific enough to distinguish between respiratory diseases and disorders to avoid misdiagnosis and the incorrect use of medication or unnecessary treatments. A differential diagnosis for these athletes may be a dysfunctional breathing pattern.

Dysfunctional breathing can be used to describe a collection of breathing disorders in which individuals display chronic breathing pattern changes. This can be without, or in addition to, another respiratory disease or disorder. Characteristics of these breathing pattern changes have been identified including apical/thoracic dominant breathing (Barker & Everard, 2015; Courtney *et al.,* 2011), thoracoabdominal asynchrony (Boulding *et al.,* 2016), or irregular breathing rhythm, and have been used to diagnose dysfunctional breathing patterns (Todd *et al.,* 2017). However, there is no formal definition or gold standard diagnostic method or diagnostic criteria for dysfunctional breathing (Boulding *et al.,* 2016). It is therefore difficult to establish an accurate prevalence of dysfunctional breathing. However, Thomas and colleagues (2001) reported that dysfunctional breathing has an occurrence of ~ 2% and ~14% in males and females, respectively, in the general population using a questionnaire-based method. While in elite athletes, airway dysfunction can have a prevalence of ~25 to 75% depending on the sport and the diagnostic measure (Price *et al.,* 2013). Dysfunctional breathing can occur alongside other respiratory disorders including asthma, with approximately 30-65% of individuals with asthma also displaying symptoms associated with dysfunctional breathing (Stanton *et al.,* 2008; Thomas *et al.,* 2001). For children with

asthma, the incidence rates follow those of the general population with ~2% and ~13% in males and females, respectively (de Groot *et al.*, 2013).

Currently, dysfunctional breathing diagnosis relies on the exclusion of other pathologies using various breathing function measures and may be treated using non-pharmacological strategies (Dickinson *et al.,* 2007). Clinicians utilise a number of resources to diagnose dysfunctional breathing including questionnaires (e.g., Nijmegen Questionnaire and breathing pattern assessment tool (BPAT)), and evaluation of breathing pattern techniques (e.g., manual assessment of respiratory motion (MARM)). These assessments aim to identify one or more dysfunctional breathing pattern characteristic or symptom. Despite their wide use, minimal clinically meaningful differences have not been established for these assessments (Courtney *et al.,* 2019). Due to the lack of standard diagnostic measures, individuals with dysfunctional breathing often have a delayed diagnosis or a misdiagnosis, with the average length of time until a correct diagnosis being between 2 – 7 years (Depiazzi & Everard, 2016). Diagnosis is complicated by similarities between other diseases/disorders, the co-existing of dysfunctional breathing with other disease/disorders, and symptoms of dysfunctional breathing may not be present at all times, for example, the symptoms may be exercise or stress-induced (Boulding *et al.*,2016). Assessments such as the MARM can only be utilised statically, making it unsuitable for assessing dysfunctional breathing in athletes during exercise. There are also many terms which are used synonymously with dysfunctional breathing, making it more difficult for clinicians to access the appropriate information and resources. It is important to diagnose dysfunctional breathing as soon as possible to avoid the incorrect use of medication or unnecessary treatments. When dysfunctional breathing is treated correctly, individuals show improvement in quality of life and reduced breathing difficulties during daily life and exercise (Hagman *et al.,* 2011; Thomas *et al.,* 2003).

The eucapnic voluntary hyperpnoea (EVH) challenge is a common method of diagnosing exercise-induced bronchoconstriction (EIB). In the previous chapter, it was identified individuals with a negative EVH test while still reporting respiratory symptoms during exercise. It is expected that these individuals may have a dysfunctional breathing pattern which is causing them to present with respiratory symptoms during exercise. Individuals may display a number of these symptoms concurrently, although one may predominate (Depiazzi & Everard, 2016). While these symptoms are also related to EIB, they present differently between the two disorders, as discussed in Chapter 2.

Optoelectronic plethysmography (OEP) is a non-invasive method of objectively assessing breathing pattern. Unlike other similar techniques such as respiratory inductive plethysmography (RIP) and structured light plethysmography (SLP), OEP can be utilised during movement and exercise, making it suitable for measuring the breathing function of athletes during exercise. Previous chapters of this thesis have demonstrated that OEP is capable of identifying differences between rest and exercise intensities in an active healthy population, and can identify differences between athletes with and without EIB. Due to the symptom similarities between DBP and other diseases, a focus on the breathing pattern characteristics may be more useful for diagnosis. As dysfunctional breathing can be exercise-induced, it would be expected that dysfunctional pattern is likely to onset at higher exercise intensities. Athletes with a dysfunctional breathing pattern will likely present with dysfunctional characteristics currently used for diagnosis (e.g., BPAT and RIP) such as apical or asynchronous movement of the chest during exercise (Todd *et al.*, 2017) but potentially not at rest when compared to individuals with a healthy breathing pattern. OEP allows for the objective measure of breathing pattern characteristics through the measurement of regional contribution and phase angle parameters which allow for apical and asynchronous breathing patterns to be quantified.

Therefore, the purpose of this chapter was 1) to establish a breathing pattern exercise response in athletes with a dysfunctional breathing pattern using OEP, and 2) to investigate if OEP-derived breathing parameters, specifically regional contribution and phase angle parameters, differ between athletes with a suspected dysfunctional breathing pattern from athletes without any known breathing disorders and athletes with exercise-induced bronchoconstriction during an evoked state.

6.2 Methods

Participants

Eighty-five active participants gave informed written consent to participate in this study. This study was approved by the School of Sport & Exercise Sciences Research Ethics Advisory Group at the University of Kent (Prop 21_2018_19). Participants were required to have a consistent training regime with a minimum of 5 hours of physical activity per week Each participant completed a questionnaire (Appendix 2) regarding respiratory symptoms during or post-exercise and in various environmental conditions. Individuals who reported no respiratory symptoms and had no history of any respiratory disease were assumed to have a healthy breathing pattern. Individuals who reported respiratory symptoms and/or had a previous/current diagnosis of asthma/EIB performed an EVH challenge. Individuals who demonstrated a negative EVH result but reported exercise respiratory symptoms and otherwise healthy, were suspected to have a dysfunctional breathing pattern during exercise. The healthy (HBP) and dysfunctional breathing pattern (DBP) group performed an exercise challenge. A summary of participant information can be seen in Table 6.1 and a summary of the protocol is displayed in Figure 6.1.

Table 6.1 Participant Information (mean ± standard deviation)

	N	**Age (yrs)**	**Height (m)**	**Weight (kg)**	**Activity Level (hrs/week)**
Healthy Breathing Pattern Group n = 47					
Male	28	34.4 ± 12.1	1.8 ± 0.1	77.8 ± 11.5	5.6 ± 2.7
Female	19	28.4 ± 9.3	1.7 ± 0.1	61.3 ± 5.5	5.3 ± 2.6
Dysfunctional Breathing Pattern Group n = 26					
Male	12	34.8 ± 9.6	1.8 ± 0.1	75.7 ± 7.4	5.2 ± 2.5
Female	14	30.9 ± 11.5	1.6 ± 0.1	58.8 ± 6.4	7.1 ± 4.2

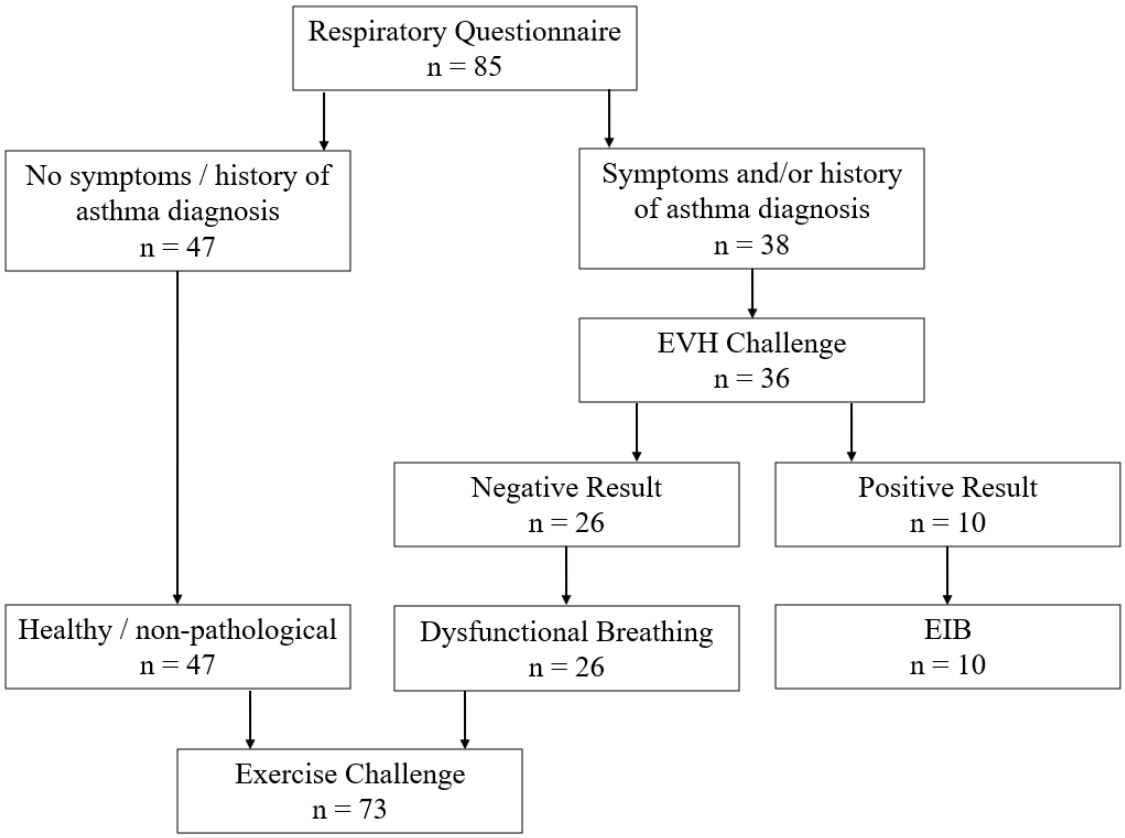


Figure 6.1 Summary of the protocol

Equipment

OEP data was collected as described in Chapter 3, but briefly the data collection involved participants in an upright seated position on a cycle ergometer (Lode – Corival, Netherlands) surrounded by 11 Qualisys cameras (Oqus 3, Qualisys AB, Goteborg, Sweden) sampling at 100 Hz. A marker set of 90 (38 posterior, 10 lateral, 42 anterior) reflective markers were placed on the participants torso in a grid-like pattern (Aliverti *et al.*, 2000; Massaroni *et al.*, 2017a). The marker set was positioned slightly differently on females to reduce the influence of soft tissue movement (Layton *et al.*, 2011).

Protocol

Participants began with a period of tidal breathing, followed by a sub-maximal incremental cycle challenge. The cycle challenge began at 50 W which was increased by 30 W every minute. The exercise intensities used in this study were defined using a rate of perceived exertion (RPE) scale with low, moderate, and high intensity being defined as an RPE values of 11, 13/14, and 17/18, respectively (Muyor, 2013). Once a participant reached each intensity, OEP data was recorded for approximately 30 – 60 seconds. Once the high intensity data was recorded, the cycle test was terminated. The period of recovery began immediately post-exercise and lasted until the participant's RR returned to their resting values. OEP data was also recorded pre and post exercise challenge to obtain resting and recovery data for approximately 3 minutes.

Data Analysis

OEP data was processed using Qualisys Track Manager (v2019.2 Build 4610). Custom-built MATLAB (version R2019a) scripts were used to calculate all of the OEP breathing parameters. Timing, compartment contribution and phase angle parameters were calculated from the OEP volume trace (see Table 3.1 for full list and description) using a custom-built MATLAB script.

A novel parameter was developed in order to measure the deviation of the phase angle trace from the 'perfect' straight 45 degree diagonal line during the inhale (inhale % deviation) and exhale (exhale % deviation) using the following equation:

$$Percentage\ Deviation = (AbVol50/AbVol100) - 0.5) * 100 \quad (1)$$

Where AbVol50 is the abdomen volume at 50% of the ribcage volume, and AbVol100 is the abdomen volume at 100% of the ribcage volume during an inhale and then repeated for exhale.

Statistical Analysis

95% upper and lower confidence intervals were calculated for each OEP parameters across each condition i.e., rest, low, moderate, high intensity exercise, and recovery post exercise. A repeated measures ANOVA was used to analyse the differences between exercise conditions within dysfunctional breathing pattern group. The healthy group was compared to the dysfunctional breathing group during each exercise conditions using t-tests.

6.3 Results

Participants reported respiratory symptoms including coughing, wheezing breathing in and/or out, chest tightness breathing in and/or out, and dyspnoea during or post-exercise.

Tables 6.2, 6.3 and 6.4 display the mean, standard deviation, and upper and lower 95% confidence intervals for each breathing parameter during every condition of the exercise challenge i.e., rest, low, moderate and high exercise intensity, and recovery post-exercise, for individuals with suspected dysfunctional breathing pattern.

Table 6.2 Dysfunctional breathing pattern group mean, standard deviation (SD) and 95% confidence intervals (CI) for the timing and ratio breathing parameters across each condition. Significant differences between conditions are represented by R*, L*, M*, H*, and Rc* as a mean significant ($p \leq 0.05$) difference from rest, low, moderate, high intensity exercise, and recovery, respectively.

	Rest				Low Intensity				Moderate Intensity				High Intensity				Recovery			
			95% CI				95% CI				95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
RR (brpm)	15.62	3.94	14.11	17.14	22.63	4.36	20.96	24.31	26.92	4.96	25.01	28.82	36.93	8.37	33.71	40.15	23.29	5.70	21.10	25.48
	L*, M*, H*, Rc *				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
Vt (litres)	0.70	0.33	0.57	0.83	1.72	0.63	1.47	1.96	2.18	0.76	1.89	2.47	2.60	0.95	2.24	2.97	1.95	0.71	1.68	2.23
	L*, M*, H*, Rc *				R*, M*, H				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
V_E (lit/min)	10.20	4.78	8.37	12.04	37.94	14.07	32.54	43.35	55.76	17.11	49.18	62.34	93.70	38.40	78.94	108.46	46.23	22.55	37.56	54.90
	L*, M*, H*, Rc *				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
tI (s)	1.86	0.33	1.73	1.98	1.24	0.19	1.17	1.31	1.08	0.16	1.02	1.14	0.80	0.15	0.75	0.86	1.24	0.28	1.13	1.35
	L*, M*, H*, Rc *				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H*			
tE (s)	2.38	0.60	2.15	2.61	1.35	0.25	1.25	1.45	1.12	0.19	1.05	1.20	0.84	0.17	0.77	0.90	1.33	0.30	1.21	1.45
	L*, M*, H*, Rc *				R*, H*				R*, H*				R*, L*, M*, Rc *				R*, H*			
tTot (s)	4.31	0.93	3.95	4.67	2.57	0.43	2.40	2.73	2.20	0.35	2.06	2.33	1.65	0.31	1.53	1.77	2.57	0.58	2.35	2.79
	L*, M*, H*, Rc *				R*, M*, H*				R*, L*, H*				R*, L*, M*, Rc *				R*, H *			
tI/tE	0.81	0.11	0.77	0.85	0.93	0.06	0.91	0.95	0.96	0.04	0.94	0.97	0.96	0.04	0.94	0.97	0.94	0.06	0.91	0.96
	L*, M*, H*, Rc *				R*				R*				R*				R*			
tI/tTot	0.45	0.03	0.43	0.46	0.48	0.02	0.47	0.49	0.49	0.01	0.48	0.49	0.49	0.01	0.48	0.49	0.48	0.02	0.48	0.49
	L*, M*, H*, Rc *				R*, M*, H*, Rc *				R*, L*, H*, Rc *				R*, L*, M*, Rc *				R*, L*, M*, H*			
tPTEF	0.38	0.10	0.35	0.42	0.50	0.04	0.48	0.52	0.50	0.01	0.50	0.50	0.50	0.02	0.49	0.51	0.48	0.05	0.46	0.49
	L*, M*, H*, Rc *				R*				R*				R*				R*			
tPTIF	0.53	0.06	0.51	0.55	0.50	0.02	0.49	0.51	0.50	0.02	0.49	0.50	0.50	0.01	0.50	0.51	0.50	0.03	0.49	0.51
	L*, M*, H*, Rc *				R*, M*, H*, Rc *				R*, L*, H*, Rc *				R*, L*, M*, Rc *				R*, L*, M*, H*			
IE50	1.30	0.22	1.21	1.38	1.09	0.11	1.04	1.14	1.06	0.12	1.02	1.11	1.02	0.07	1.00	1.05	1.09	0.10	1.05	1.13
	L*, M*, H*, Rc *				R*				R*				R*				R*			

Table 6.3 Dysfunctional breathing pattern group mean, standard deviation (SD) and 95% confidence intervals (CI) for the regional contribution breathing parameters across each condition. Significant differences between conditions are represented by R*, L*, M*, H*, and Rc* as a mean significant ($p \leq 0.05$) difference from rest, low, moderate, high intensity exercise, and recovery, respectively.

	Rest				Low Intensity				Moderate Intensity				High Intensity				Recovery			
			95% CI				95% CI				95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
RCpCT (%)	57.45	4.70	55.64	59.26	57.79	5.12	55.83	59.76	57.83	5.18	55.84	59.82	57.96	5.22	55.95	59.97	58.06	5.29	56.03	60.09
RCaCT (%)	11.38	1.89	10.66	12.11	11.52	2.08	10.72	12.32	11.60	2.16	10.78	12.43	11.78 Rc*	2.15	10.95	12.61	11.31 H*	2.17	10.48	12.15
RcCT (%)	68.84 L*, M*, H*	3.44	67.52	70.16	69.34 R*	3.80	67.88	70.80	69.45 R*	3.89	67.96	70.95	69.77 R*	3.89	68.27	71.26	69.40	3.86	67.91	70.88
AbCT (%)	31.16 L*, M*, H*	3.44	29.84	32.48	30.66 R*	3.80	29.20	32.12	30.55 R*	3.89	29.05	32.04	30.23 R*	3.89	28.74	31.72	30.60	3.86	29.12	32.09
RCaAbCT (%)	42.55 L*, M*, H*	4.70	40.74	44.36	42.21 R*	5.12	40.24	44.17	42.17 R*	5.18	40.18	44.17	42.04 R*	5.22	40.03	44.05	41.94	5.29	39.91	43.97
RcAbIndex	2.25 L*, M*, H*	0.37	2.11	2.39	2.31 R*	0.43	2.15	2.48	2.33 R*	0.44	2.16	2.50	2.36 R*	0.45	2.19	2.54	2.32	0.43	2.15	2.48

Table 6.4 Dysfunctional breathing pattern group mean, standard deviation (SD) and 95% confidence intervals (CI) for the phase angle breathing parameters across each condition. Significant differences between conditions are represented by R*, L*, M*, H*, and Rc* as a mean significant ($p \leq 0.05$) difference from rest, low, moderate, high intensity exercise, and recovery, respectively.

	Rest				Low Intensity				Moderate Intensity				High Intensity				Recovery			
			95% CI				95% CI				95% CI				95% CI				95% CI	
Parameter	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**	**Mean**	**SD**	**Lower**	**Upper**
HTA (deg)	-0.04	0.10	-0.07	0.00	0.05	0.35	-0.08	0.19	0.12	0.37	-0.02	0.26	0.12	0.33	0.00	0.26	0.05	0.11	0.00	0.09
RcHTA (deg)	-0.02	0.10	-0.07	0.01	0.04	0.30	-0.07	0.16	0.11	0.29	-0.00	0.22	0.16	0.35	0.02	0.29	0.06	0.16	0.00	0.12
AbHTA (deg)	-0.02	0.20	-0.10	0.05	-0.39 M*	0.93	-0.75	-0.03	0.29 L*	0.85	-0.04	0.61	-0.08	0.63	-0.32	0.17	0.03	0.31	-0.09	0.15
RCpHTA (deg)	-0.02 H*	0.10	-0.06	0.02	0.06	0.36	-0.08	0.20	0.16	0.44	-0.01	0.33	0.22 R*	0.37	0.09	0.37	0.05	0.16	-0.01	0.11
RCaHTA (deg)	0.01	0.21	-0.07	0.09	-0.10 H*	0.50	-0.30	0.09	0.08	0.83	-0.24	0.40	0.42 L*	0.99	0.04	0.80	0.20	0.35	0.07	0.34
RcAbPhase (deg)	-0.12 L*, M*, H*	0.53	-0.32	0.08	-1.19 R*, H*	0.92	-1.54	-0.83	-1.88 R*, H*, Rc *	1.01	-2.27	-1.49	-2.70 R*, L*, M*, Rc *	1.71	-3.36	-2.04	-0.91 M*, H *	1.12	-1.34	-0.48
RCpAbPhase (deg)	-0.14 L*, M*, H*	0.44	-0.31	0.03	-0.81 R*, H*	0.76	-1.10	-0.51	-1.27 R*, Rc *	0.99	-1.65	-0.89	-1.64 R*, L*, Rc *	1.43	-2.19	-1.20	-0.50 M*, H*	0.95	-0.87	-0.14
RCpRCaPhase (deg)	0.02 H*	0.24	-0.08	0.11	0.09	0.61	-0.15	0.32	0.28	0.80	-0.03	0.58	0.46 R*	0.82	0.14	0.77	0.42	0.72	0.14	0.70
RCaSPhase (deg)	0.06 M*	0.29	-0.05	0.17	0.29	0.81	-0.02	0.60	0.62 R*	0.80	0.31	0.93	0.55	0.93	0.19	0.91	0.58	0.91	0.23	0.93
RCpSPhase(deg)	0.14 L*, M*, H*, Rc *	0.22	0.06	0.23	0.40 R*	0.42	0.23	0.56	0.56 R*	0.48	0.38	0.74	0.58 R*	0.54	0.37	0.79	0.52 R*	0.64	0.27	0.77
AbSPhase (deg)	-0.14 M*, H*	0.61	-0.37	0.10	-0.83 H*	1.14	-1.27	-0.39	-1.42 R*	1.05	-1.82	-1.01	-2.18 R*, L*, Rc *	1.79	-2.87	-1.49	-0.72 H*	1.32	-1.23	-0.21
Inhale % Deviation	1.33 L*, M*, H*, Rc*	14.82	1.94	4.10	10.39 R*	11.17	6.51	14.39	13.66 R*	11.49	8.25	18.50	19.72 R*	8.60	11.87	21.94	12.06 R*	12.50	6.85	18.38
Exhale % Deviation	-3.45 H*	14.30	-1.76	1.12	-8.38	11.34	-4.77	-5.03	-10.18 Rc*	11.34	-5.99	-6.52	-17.60 R*, Rc*	7.50	-10.95	-13.47	-2.00 M*, H*	16.20	0.90	-1.26

Table 6.2, 6.3, 6.4 also display the changes across exercise conditions within the group of individuals with suspected dysfunctional breathing pattern. The following breathing parameters displayed significant differences between rest and exercise: RR, Vt, V_E, tI, tE, tTot, tI/tE, tI/tTot, tPTEF, tPTIF, IE50, RcCT, AbCT, RcAbIndex, RcAbPhase, RCpAbPhase, RCpSPhase and AbSPhase. RCpRCaPhase and RCpHTA were found to be significantly different between rest and high exercise intensity.

RR, Vt, V_E, tI, tE, tTot, tI/tE, tPTIF, RcAbPhase, and RCpAbPhase were significantly different between high exercise intensity and other exercise intensities.

Finally, RR, Vt, V_E, tI, tE, tTot, tI/tTot, tPTIF, IE50, and RCpSPhase were found to be significantly different between rest and recovery, with RR, Vt, VE, tI, tE, tTot, tI/tTot, tPTIF, RCaCT, RcAbPhase, RCpAbPhase, and AbSPhase significantly different between recovery and some exercise intensities.

Tables 6.5 displays the comparison between healthy individuals and individuals with suspected DBP at rest and recovery post-exercise, with Table 6.6 displaying the comparison between the two groups during exercise. RCaCT. RcCT, AbCT, RcAbIndex, and HTA were found to be significantly different between the two groups at rest. All other breathing parameters were found to be non-significant at rest. The parameters tI, tE, tTot, RCaCT, RCpHTA, RcAbPhase, RCpRCaPhase, RCaSPhase, and RCpSPhase displayed significant differences between healthy individuals and individuals with dysfunctional breathing during high intensity exercise (Table 6.6). All other breathing parameters were found to be non-significant. RCaCT, RCpRCaPhase, and RCaSPhase were found to be significantly different between the two groups during recovery post-exercise with all other parameters demonstrating no significant difference (Table 6.5). Figure 6.2 displays how these parameters changed across conditions for both groups.

In the previous chapter, a comparison was made between the healthy breathing pattern group and the EIB group. Table 6.7 displays the comparison of the EIB data from the previous chapter and the DBP group. At rest, only RR was significantly different between the EIB and the DBP group, with all other parameters displaying non-significant differences. During the 'evoked' state, RR, Vt, V_E, tI, tE, tTot, tI/tE, tI/tTot, tPTEF, IE50, RCpHTA, RcAbPhase, RCpAbPhase, and AbSPhase were found to be significant different between the two groups. Finally, RR, Vt, V_E, tI, tE, tTot, tI/tE, tI/tTot, tPTEF, and RCaCT displayed significant differences between the two group during the recovery condition.

Table 6.5 Comparison between healthy (HBP) and DBP at rest and recovery post-exercise. * significant with $p \leq 0.05$.

	Rest					Recovery				
	HBP Group		DBP Group			HBP Group		DBP Group		
Parameter	**Mean**	**SD**	**Mean**	**SD**	**Sig**	**Mean**	**SD**	**Mean**	**SD**	**Sig**
RR (brpm)	15.36	3.51	15.62	3.94	0.95	23.37	4.69	23.29	5.70	0.99
Vt (litres)	0.75	0.33	0.70	0.33	0.83	1.83	0.53	1.95	0.71	0.72
V_E (lit/min)	11.19	4.42	10.20	4.78	0.62	41.38	14.69	46.23	22.55	0.59
tI (s)	1.86	0.47	1.86	0.33	1.00	1.32	0.37	1.24	0.29	0.72
tE (s)	2.37	0.66	2.38	0.60	0.99	1.44	0.42	1.33	0.30	0.48
tTot (s)	4.23	1.08	4.31	0.93	0.95	2.75	0.77	2.57	0.58	0.59
tI/tE	0.80	0.14	0.81	0.11	0.90	0.91	0.09	0.94	0.06	0.49
tI/tTot	0.44	0.05	0.45	0.03	0.84	0.48	0.03	0.48	0.02	0.46
tPTEF	0.37	0.09	0.38	0.10	0.85	0.47	0.06	0.47	0.05	0.98
tPTIF	0.51	0.06	0.53	0.06	0.25	0.49	0.04	0.50	0.03	0.55
IE50	1.33	0.27	1.30	0.22	0.85	1.12	0.14	1.09	0.10	0.74
RCpCT (%)	57.88	4.56	57.45	4.70	0.93	58.46	4.89	58.06	5.29	0.94
RCaCT (%)	13.30	2.42	11.38	1.89	< 0.01 *	13.21	2.39	11.31	2.17	< 0.01 *
RcCT (%)	71.17	3.54	68.84	3.44	0.02 *	71.67	3.97	69.40	3.86	0.05
AbCT (%)	28.83	3.54	31.16	3.44	0.02 *	28.33	3.97	30.60	3.86	0.05
RCaAbCT (%)	42.13	4.56	42.55	4.70	0.93	41.54	4.89	41.94	5.29	0.94
RcAbIndex	2.52	0.44	2.25	0.37	0.03 *	2.60	0.53	2.32	0.43	0.06
HTA (deg)	0.03	0.09	-0.04	0.10	0.01 *	0.05	0.12	0.05	0.11	0.99
RcHTA (deg)	0.03	0.10	-0.03	0.10	0.06	0.05	0.15	0.06	0.16	0.95
AbHTA (deg)	0.01	0.14	-0.02	0.20	0.61	-0.03	0.38	0.03	0.31	0.80
RCpHTA (deg)	0.02	0.09	-0.02	0.10	0.13	0.00	0.16	0.05	0.16	0.39
RCaHTA (deg)	0.00	0.13	0.01	0.21	0.96	0.19	0.35	0.20	0.35	0.99
RcAbPhase (deg)	-0.20	0.39	-0.12	0.53	0.75	-1.07	0.90	-0.91	1.12	0.77
RCpAbPhase (deg)	-0.20	0.42	-0.14	0.44	0.83	-0.81	0.95	-0.50	0.95	0.36
RCpRCaPhase (deg)	-0.07	0.28	0.02	0.24	0.33	-0.06	0.80	0.42	0.72	0.03 *
RCaSPhase (deg)	0.02	0.27	0.06	0.29	0.81	0.06	0.93	0.58	0.91	0.04 *
RCpSPhase(deg)	0.12	0.19	0.14	0.22	0.89	0.31	0.45	0.52	0.64	0.20
AbSPhase (deg)	-0.06	0.43	-0.14	0.61	0.78	-0.97	1.20	-0.72	1.32	0.67
Inhale % Deviation	-0.11	11.07	1.33	14.82	0.02 *	13.06	8.64	12.06	12.50	0.09
Exhale % Deviation	-5.48	9.50	-3.45	14.30	0.99	-4.66	11.54	-2.00	16.20	0.94

Table 6.6 Comparison between healthy (HBP) and DBP during low, moderate, and high intensity exercise. * represents significant differences with $p \leq 0.05$ between groups at each intensity.

	Low Intensity Exercise					Moderate Intensity Exercise					High Intensity Exercise				
	HBP Group		DBP Group			HBP Group		DBP Group			HBP Group		DBP Group		
Parameter	**Mean**	**SD**	**Mean**	**SD**	**Sig**	**Mean**	**SD**	**Mean**	**SD**	**Sig**	**Mean**	**SD**	**Mean**	**SD**	**Sig**
RR (brpm)	21.82	4.80	22.63	4.36	0.48	24.84	5.01	26.92	4.96	0.09	33.15	6.59	36.93	8.37	0.13
Vt (litres)	1.36	0.30	1.72	0.63	< 0.01 *	2.09	0.67	2.18	0.76	0.64	2.64	0.86	2.60	0.95	0.99
V_E (lit/min)	28.75	6.66	37.94	14.07	< 0.01 *	49.96	14.98	55.76	17.11	0.14	85.35	32.26	93.70	38.40	0.62
tI (s)	1.37	0.33	1.24	0.19	0.03 *	1.22	0.24	1.08	0.16	< 0.01 *	0.94	0.24	0.80	0.15	< 0.01 *
tE (s)	1.54	0.39	1.35	0.25	< 0.01 *	1.32	0.28	1.12	0.19	< 0.01 *	0.99	0.27	0.84	0.17	< 0.01 *
tTot (s)	2.90	0.70	2.57	0.43	< 0.01 *	2.54	0.53	2.20	0.35	< 0.01 *	1.94	0.52	1.65	0.31	< 0.01 *
tI/tE	0.90	0.08	0.93	0.06	0.10	0.93	0.06	1.00	0.04	0.05	0.95	0.04	0.96	0.04	0.99
tI/tTot	0.47	0.02	0.48	0.01	0.10	0.48	0.02	0.49	0.01	0.04 *	0.49	0.01	0.49	0.01	1.00
tPTEF	0.50	0.06	0.50	0.04	0.92	0.50	0.04	0.50	0.01	0.77	0.45	0.02	0.50	0.02	1.00
tPTIF	0.49	0.04	0.50	0.02	0.05	0.48	0.04	0.50	0.02	0.04 *	0.50	0.02	0.50	0.01	0.22
IE50	1.16	0.14	1.09	0.11	0.032*	1.09	0.10	1.06	0.12	0.40	1.05	0.07	1.02	0.07	0.26
RCpCT (%)	58.55	4.55	57.79	5.12	0.52	58.47	4.93	57.83	5.18	0.60	58.50	4.83	57.96	5.22	0.89
RCaCT (%)	13.21	2.56	11.52	2.08	< 0.01 *	13.40	2.51	11.60	2.16	< 0.01 *	13.49	2.38	11.78	2.15	< 0.01 *
RcCT (%)	71.77	3.64	69.34	3.80	< 0.01 *	71.88	3.81	69.45	3.89	< 0.01 *	72.01	3.91	69.77	3.89	0.05
AbCT (%)	28.23	3.64	30.66	3.80	< 0.01 *	28.12	3.81	30.55	3.89	< 0.01 *	27.99	3.91	30.23	3.89	0.05
RCaAbCT (%)	41.45	4.56	42.21	5.12	0.52	41.53	4.93	42.17	5.18	0.60	41.50	4.83	42.04	5.22	0.89
RcAbIndex	2.60	0.49	2.31	0.43	< 0.01 *	2.62	0.51	2.33	0.44	0.02 *	2.64	0.54	2.36	0.45	0.06
HTA (deg)	0.01	0.27	0.05	0.35	0.52	0.02	0.35	0.12	0.37	0.24	0.13	0.30	0.13	0.33	1.00
RcHTA (deg)	-0.02	0.25	0.04	0.30	0.35	0.08	0.36	0.11	0.29	0.75	0.14	0.28	0.16	0.35	0.30
AbHTA (deg)	-0.15	0.75	-0.39	0.93	0.23	-0.20	0.87	0.29	0.85	0.03 *	-0.25	0.70	-0.08	0.63	0.51
RCpHTA (deg)	-0.05	0.31	0.06	0.36	0.19	-0.02	0.48	0.16	0.44	0.13	0.00	0.30	0.23	0.37	0.03 *
RCaHTA (deg)	0.06	0.37	-0.10	0.50	0.12	0.15	0.50	0.08	0.83	0.64	0.54	0.66	0.42	0.99	0.86
RcAbPhase (deg)	-0.84	0.76	-1.19	0.93	0.09	-1.66	0.99	-1.88	1.01	0.38	-1.90	1.06	-2.70	1.71	0.04 *
RCpAbPhase (deg)	-0.65	0.70	-0.81	0.76	0.36	-1.37	0.87	-1.27	0.99	0.67	-1.79	1.28	-1.64	1.43	0.90
RCpRCaPhase (deg)	-0.11	0.59	0.09	0.61	0.18	-0.35	0.82	0.28	0.80	< 0.01 *	-0.45	1.08	0.46	0.82	< 0.01 *
RCaSPhase (deg)	-0.00	0.57	0.29	0.81	0.08	-0.27	0.91	0.62	0.80	< 0.01 *	-0.33	1.26	0.55	0.93	< 0.01 *
RCpSPhase(deg)	0.16	0.36	0.40	0.43	< 0.01 *	0.23	0.49	0.56	0.48	< 0.01 *	0.17	0.67	0.58	0.54	0.02 *
AbSPhase (deg)	-0.68	0.89	-0.83	1.14	0.55	-1.61	1.25	-1.42	1.05	0.51	-2.26	1.70	-2.18	1.79	0.98
Inhale % deviation	4.64	10.27	10.39	11.17	< 0.01 *	9.60	10.36	13.66	11.49	< 0.01 *	8.80	5.20	19.72	8.60	< 0.01 *
Exhale % deviation	-10.64	10.09	-8.38	11.34	0.84	-13.47	10.66	-10.18	11.34	0.67	10.90	6.60	-17.60	7.50	< 0.01 *

Table 6.7 Comparison between EIB (data from chapter 5) and DBP at rest, evoked state and recovery. * represents significant differences with $p \leq 0.05$ between groups at each intensity.

	Rest					Evoked State					Recovery				
	EIB Group		DBP Group			EIB Group		DBP Group			EIB Group		DBP Group		
Parameter	**Mean**	**SD**	**Mean**	**SD**	**Sig**	**Mean**	**SD**	**Mean**	**SD**	**Sig**	**Mean**	**SD**	**Mean**	**SD**	**Sig**
RR (brpm)	11.40	2.77	15.62	3.94	< 0.01 *	13.47	3.05	36.93	8.37	< 0.01 *	13.45	3.41	23.29	5.79	< 0.01 *
Vt (litres)	0.92	0.45	0.70	0.33	0.22	1.05	0.34	2.60	0.95	< 0.01 *	0.96	0.29	1.95	0.71	< 0.01 *
V_E (lit/min)	8.82	2.14	10.20	4.78	0.67	13.89	3.82	93.70	38.40	< 0.01 *	12.23	3.47	46.23	22.55	< 0.01 *
tI (s)	2.23	0.64	1.86	0.33	0.23	1.94	0.34	0.80	0.15	< 0.01 *	1.96	0.36	1.24	0.29	< 0.01 *
tE (s)	2.92	0.79	2.38	0.60	0.08	2.44	0.56	0.84	0.17	< 0.01 *	2.37	0.49	1.33	0.30	< 0.01 *
tTot (s)	5.14	1.33	4.31	0.93	0.10	4.46	0.83	1.65	0.31	< 0.01 *	4.40	0.78	2.57	0.58	< 0.01 *
tI/tE	0.75	0.17	0.81	0.11	0.46	0.82	0.13	0.96	0.04	0.02 *	0.84	0.08	0.94	0.06	< 0.01 *
tI/tTot	0.42	0.06	0.45	0.03	0.37	0.45	0.04	0.49	0.01	0.03 *	0.46	0.03	0.48	0.02	< 0.01 *
tPTEF	0.37	0.11	0.38	0.10	0.91	0.36	0.08	0.50	0.02	< 0.01 *	0.39	0.09	0.48	0.05	< 0.01 *
tPTIF	0.52	0.05	0.53	0.06	0.96	0.51	0.04	0.50	0.01	0.67	0.52	0.03	0.50	0.03	0.43
IE50	1.16	0.15	1.30	0.22	0.37	1.27	0.22	1.02	0.07	0.02 *	1.19	0.19	1.09	0.10	0.13
RCpCT (%)	56.96	3.77	57.45	4.70	0.95	56.71	3.70	57.96	5.22	0.77	56.46	3.77	58.06	5.29	0.68
RCaCT (%)	13.04	1.91	11.38	1.89	0.12	13.34	1.71	11.78	2.15	0.16	13.60	1.51	11.31	2.17	0.03 *
RcCT (%)	70.01	3.59	68.84	3.44	0.64	70.04	3.62	69.77	3.89	0.98	70.06	4.10	69.40	3.86	0.90
AbCT (%)	29.99	3.59	31.16	3.44	0.64	29.97	3.63	30.23	3.89	0.98	29.95	4.10	30.60	3.86	0.90
RCaAbCT (%)	43.04	3.77	42.55	4.70	0.95	43.29	3.70	42.04	5.22	0.77	43.54	3.77	41.94	5.29	0.68
RcAbIndex	2.38	0.41	2.25	0.37	0.68	2.38	0.41	2.36	0.45	1.00	2.40	0.47	2.32	0.43	0.91
HTA (deg)	-0.02	0.06	-0.04	0.10	0.85	0.03	0.09	0.13	0.33	0.38	-0.01	0.08	0.05	0.11	0.43
RcHTA (deg)	-0.01	0.08	-0.03	0.10	0.43	0.04	0.10	0.16	0.35	0.98	-0.01	0.10	0.06	0.16	0.46
AbHTA (deg)	-0.00	0.14	-0.02	0.20	0.92	0.03	0.27	-0.08	0.63	0.51	0.04	0.21	0.03	0.31	1.00
RCpHTA (deg)	-0.04	0.10	-0.02	0.10	0.94	0.01	0.12	0.23	0.37	0.03 *	0.00	0.12	0.05	0.16	0.68
RCaHTA (deg)	-0.00	0.12	0.01	0.21	0.97	0.14	0.23	0.42	0.99	0.38	0.04	0.09	0.20	0.35	0.10
RcAbPhase (deg)	-0.03	0.38	-0.12	0.53	0.86	-0.46	0.67	-2.70	1.71	< 0.01 *	-0.16	0.63	-0.91	1.12	0.11
RCpAbPhase (deg)	-0.07	0.31	-0.14	0.44	0.91	-0.14	0,43	-1.64	1.43	< 0.01 *	-0.12	0.43	-0.50	0.95	0.52
RCpRCaPhase (deg)	-0.11	0.23	0.02	0.24	0.40	0.42	0.54	0.46	0.82	0.99	0.09	0.27	0.42	0.72	0.48
RCaSPhase (deg)	-0.04	0.24	0.06	0.29	0.58	0.70	0.69	0.55	0.93	0.93	0.19	0.38	0.58	0.91	0.49
RCpSPhase(deg)	0.10	0.16	0.14	0.22	0.84	0.66	0.54	0.58	0.54	0.94	0.29	0.31	0.5	0.64	0.45
AbSPhase (deg)	0.06	0.49	-0.14	0.61	0.53	-0.10	0.50	-2.18	1.79	< 0.01 *	-0.06	0.61	-0.72	1.32	0.33
Inhale % Deviation	-1.20	2.11	1.33	14.82	0.08	13.95	10.74	19.72	8.58	0.09	-10.52	9.66	12.06	12.50	0.10
Exhale % Deviation	-3.13	14.49	-3.45	14.30	0.99	17.35	9.33	17.60	7.51	0.89	-3.54	9.66	-2.00	16.20	0.34

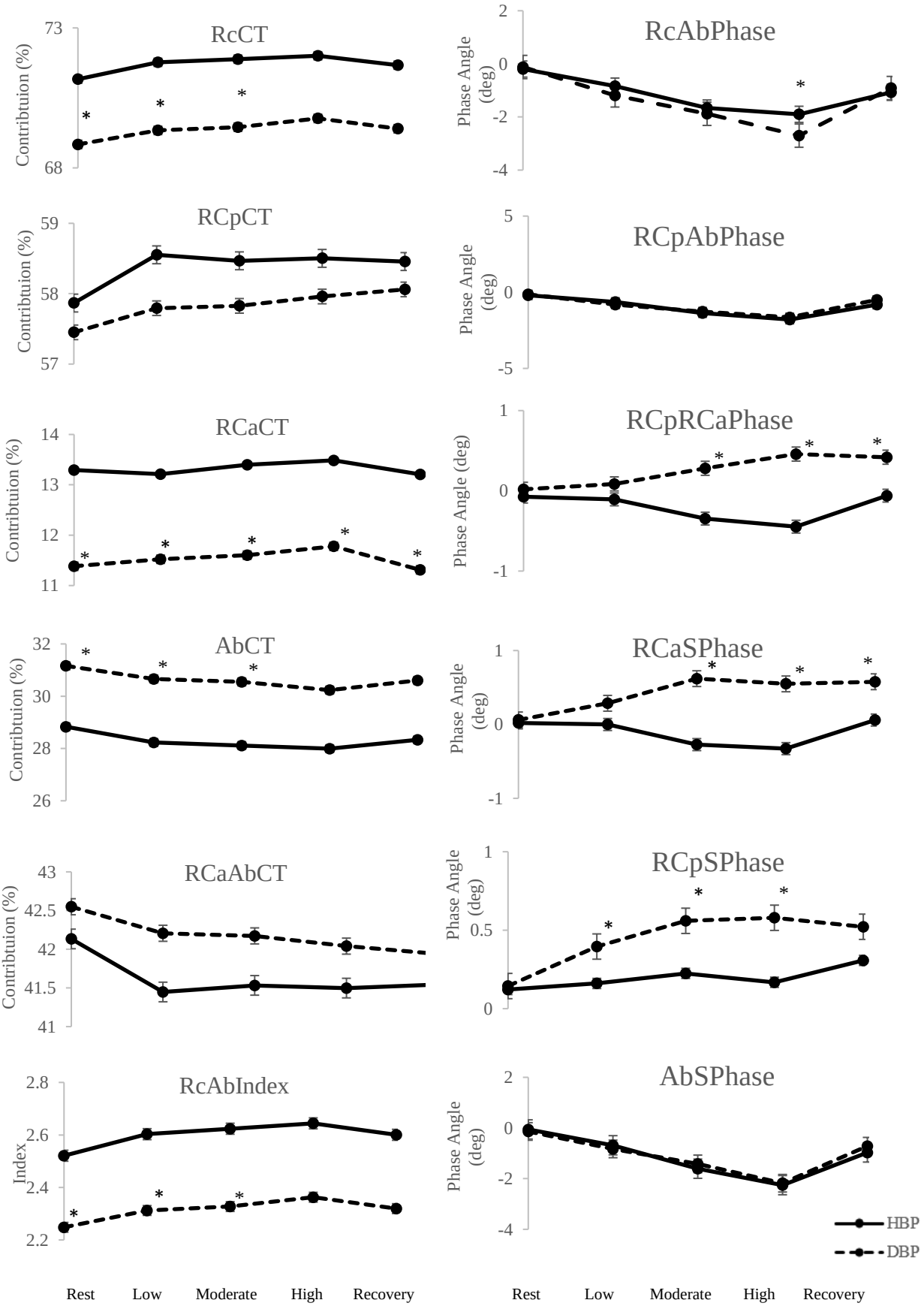
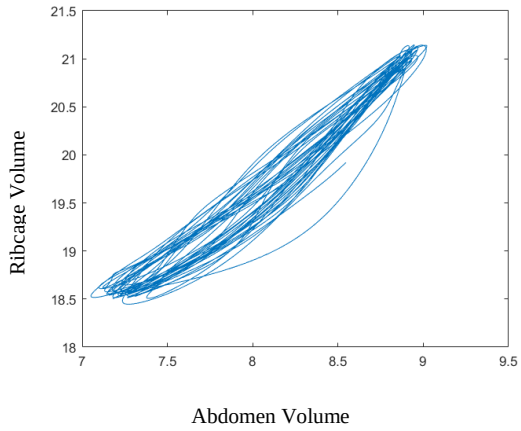


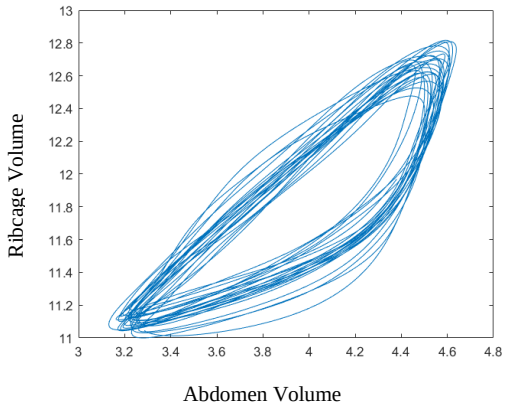
Figure 6.2 Illustration of contribution and phase angle parameters exercise response for the healthy breathing pattern group (solid line) and the dysfunctional breathing pattern group (dashed line). * indicates a significant difference between the groups with $p \leq 0.05$.

a) Healthy Breathing Pattern



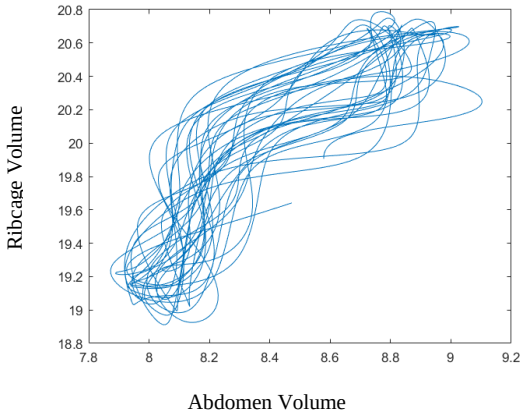
Phase Angle: - 1.19 degrees
Inhale % Deviation: - 12.95 %
Exhale % Deviation: - 2.11 %

b) Dysfunctional Breathing Pattern



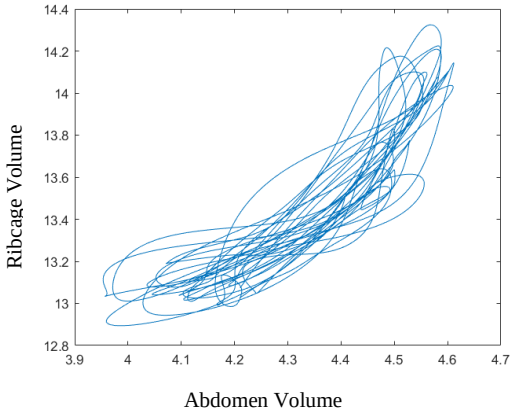
Phase Angle: -7.32 degrees
Inhale % Deviation: - 34.57 %
Exhale % Deviation: - 7.28 %

c) Dysfunctional Breathing Pattern



Phase Angle: - 0.19 degrees
Inhale % Deviation: -34.95 %
Exhale % Deviation: -37.68 %

d) Dysfunctional Breathing Pattern



Phase Angle: 0.32 degrees
Inhale % Deviation: 15.56 %
Exhale % Deviation: 22.16 %

Figure 6.3 Examples of RcAbPhase Konno-Mead loops for (a) a healthy breathing pattern with a low phase angle and low inhale and exhale percentage deviation, b) a dysfunctional breathing pattern with a high phase angle and large deviations and (c,d) dysfunctional breathing patterns with a low phase angle but large percentage deviations during high intensity exercise.

6.4 Discussion

This study, for the first time, demonstrates that OEP-derived breathing parameters can identify differences between athletes with and without a dysfunctional breathing pattern at rest and during exercise. The findings in this study also establish an exercise response for athletes with a dysfunctional breathing pattern.

The OEP-derived timing and ratio parameters RR, Vt, V_E, tI, tE, tTot, tI/tE, tI/tTot, tPTEF, tPTIF, and IE50 were significantly different at rest compared to exercise and recovery post-exercise within the DBP group (Table 6.2). RR, Vt, V_E, tI/tE, tI/tTot, and tPTEF demonstrated an expected exercise response with an increase in value during exercise followed by a subsequent decrease during recovery post-exercise (Table 6.2). Similarly, tI, tE, tTot, tPTIF, and IE50 decreased with exercise intensity and increased during recovery as expected (Table 6.2). This exercise response is similar to that seen in the healthy athletic population, however, the DBP group displayed significantly lower tI, tE, and tTot values during exercise (Table 6.6) with all other timing and ratio parameters displaying non-significant differences when compared to the HBP group. Similar to this study, Kiesel *et al.*, (2017) found no differences in RR between individuals with and without dysfunctional breathing at rest using capnography.

The regional parameters RcCT, AbCT, RCaAbCT, and RcAbIndex were significantly different at rest compared to each exercise intensity within the DBP group (Table 6.4). RcCT and RcAbIndex increased with exercise intensity, while AbCT and RCaAbCT decreased as exercise intensity increased (Table 6.3). RCaCT was significantly lower across all conditions i.e., rest, exercise, and recovery for the DBP group when compared to the HBP group (Table 6.5 and 6.6). Similarly, RcCT and RcAbIndex displayed significantly lower values, while AbCT displayed significantly higher values at rest, low and moderate intensity exercise only for the DBP group when compared to the HBP group (Table 6.5 and 6.6, Figure 6.2). Although this trend continued during high intensity exercise and recovery post-exercise, the differences between the groups for the parameters were non-significant during these conditions. These findings demonstrate that individuals with a DBP use more abdominal and less ribcage contribution to their total breath volume compared to the HBP group. Previous research has investigated these contributions parameters using SLP in asthmatic and chronic obstructive pulmonary disease (COPD) populations groups. Laveneziana and colleagues (2014) found a reduction in SLP-derived RcCT post-bronchodilation in individuals with

COPD. While, SLP-derived RcCT has also been shown to display non-significant differences between individuals with and without COPD (Motamedi-Fakhr *et al.,* 2016).

Phase angles RcAbPhase, RCpAbPhase, RCpSPhase, and AbSPhase demonstrated significant differences between rest and exercise within the DBP group (Table 6.4). Similarly, RCpRCaPhase and RCaSPhase only displayed significant differences between rest and both moderate and high exercise intensities, respectively (Table 6.4). For each of these phase angles, there was an increase in phase angle as exercise intensity increased (Table 6.4). This indicates that individuals with a DBP demonstrate a more asynchronous breathing pattern during exercise when compared to their resting state. Overall, the DBP group displayed larger phase angle values compared to HBP across each condition (Table 6.5 and 6.6). This indicates that the DBP group had a significantly more asynchronous breathing pattern during exercise and recovery post-exercise when compared to a HBP.

More specifically, RcAbPhase was significantly different between the two groups during high intensity exercise only, with the abdomen moving first followed by the ribcage in both groups i.e., negative values. This phase angle is a measure of thoracoabdominal asynchrony which is a characteristic previously used to identify DBP, for example, BPAT assessment. Previously, SLP-derived RcAbPhase has been used to distinguish between children with DBP compared to a healthy control group with the DBP group displaying lower phase angle values (Barker *et al.*, 2014), which is in contrast to the findings of this study. Although, Barker and colleagues (2014) reported the DBP group as having phase angles closer to zero compared to the control group, they did not comment on the shape of the corresponding Konno-Mead loops. RIP-derived RcAbPhase has been used to identify DBP within individuals with severe asthma and, similar to the present study, those with a DBP displayed larger phase angle values compared to healthy controls (Pyris-Picard *et al.,* 2004). Pereira and colleagues (2017) found that OEP and RIP derived RCpAbPhase displayed more asynchrony in COPD patients compared to healthy individuals during rest and exercise. Similarly, SLP-derived RcAbPhase has been shown to be significantly larger in patients with COPD (Motamedi-Fakhr *et al.*, 2016).

The phase angle RCpRCaPhase demonstrated significant differences between the two groups during moderate and high intensity exercise, and recovery post-exercise. OEP has previously been used to establish RCpRCaPhase exercise response in COPD patients (Aliverti *et al.*, 2009; Bruni *et al.*, 2012). It is thought that changes in phase angles within COPD patients is dependent on the position of the diaphragm and the degree of activation

of the inspiratory muscles (Aliverti *et al.*, 2009; Pereira *et al.*, 2017). For RCpRCaPhase, the DBP group displayed a breathing pattern with the RCp moving before the RCa, however, the HBP group displayed the opposite (Table 6.4). This indicates that the DBP displayed thoracic dominant breathing pattern which is a characteristic often used in assessments such as the BPAT to identify dysfunctional breathing. This OEP-derived phase angle offers an objective and quantitative measure of thoracic dominant breathing patterns.

During exercise, the phase angle RCpSPhase was significantly larger in the DBP group when compared to HBP group, although during rest and recovery post-exercise this difference was non-significant. RCaSPhase demonstrated significant differences between the two groups during moderate and high intensity exercise, and recovery post-exercise, only.

For RCpSPhase, the shoulders moved before the RCp in both groups across all conditions. For RCaSPhase, the shoulders moved first followed by the RCa across all conditions for the DBP group. However, for the HBP group, the RCa moved before the shoulders only during exercise. This indicates the movement of the breath for the DBP was not initiated by the RCa but seems to be a leading compartment in the HBP group during exercise. These shoulder phase angles are novel to this thesis and therefore, comparison to other studies is not possible. However, they may offer additional information for assessing apical/thoracic dominant breathing pattern which is a typical characteristic of dysfunctional breathing (Courtney *et al.*, 2011). Thoracic dominant breathing patterns involve an increase in the vertical motion of the ribcage with minimal abdomen movement (Boulding *et al.*, 2016). This motion can increase the activation of muscles of the upper ribcage such as the upper trapezius and may increase the elevation of the shoulders (Courtney, 2009). This study supports this mechanism with the DBP group displaying shoulder movement prior to lower ribcage movement during a breath across every condition. This characteristic of a dysfunctional breathing pattern may cause postural issues and/or shoulder pain (Courtney, 2009).

As discussed above, the DBP group displayed significantly higher RcAbPhase values during high intensity exercise compared to the HBP group (Table 6.6). Interestingly, some of the participants within the DBP group displayed unexpected low RcAbPhase values similar to that of a HBP, however, after visual inspection the corresponding Konno-Mead loops did not display a typical healthy breathing pattern i.e., close to a straight 45-degree diagonal line. The phase angle calculation is a measure of the synchrony between two compartments but more specifically it is a measure of the tightness of the Konno-Mead loop. For example,

Figure 6.3c, 6.3d displays a Konno-Mead loop from the dysfunctional breathing pattern group with a low phase angle however, it does not follow a straight diagonal line. Within this plot (Figure 6.3c) specifically, it indicates that the abdomen is moving first out of synchrony with the ribcage during the inhale and the opposite is happening during the exhale resulting in the tight loop and therefore, low phase angle value. This issue prompted the development of the novel parameters inhale and exhale percentage deviation, which are measures of how much the inhale and exhale deviate from the straight 45-degree diagonal line within a Konno-Mead loop. During high intensity exercise, the DBP group displayed significantly greater inhale ($p < 0.01$) and exhale ($p < 0.01$) percentage deviation values (Mean ± SD: inhale: 19.7 ± 8.6%, exhale: 17.6 ± 7.5 %) when compared to the HBP group (Mean ± SD: inhale: 8.8 ± 5.2 %, exhale: 10.9 ± 6.6 %). The differences in Konno-Mead loop shape within the dysfunctional breathing pattern group may be explained by different types of dysfunctional breathing patterns. Previous research has suggested that DBP may present differently between individuals such as thoraco-abdominal asynchrony or thoracic dominant breathing (Boulding *et al.,* 2016). The findings of this study provide a basis for further investigation into potential classifications within dysfunctional breathing.

The results thus far have shown that the OEP-derived contribution parameters seem to be useful in distinguishing between these two population groups at rest and during the lower exercise intensities but as exercise intensity increases become less important. In contrast, the phase angle parameters seem to be most useful during the higher exercise intensities and recovery post exercise, but less useful during rest and low exercise intensity. This indicates that there is not one singular OEP-derived breathing parameter that could be used to distinguish between healthy and dysfunctional breathing patterns at rest and during exercise. However, by using a combination of these parameters, a complete picture can be obtained of an individual's breathing pattern and used to distinguish between population groups. Generally, across rest and exercise, individuals with a DBP seem to display lower RC contribution and higher AB contribution compared to a HBP. They also demonstrate more asynchrony during exercise resulting in larger phase angle values with the shoulders, RCp and AB initiating the breath. However, depending on the type of dysfunctional breathing pattern, an individual may display an unexpected low phase angle, therefore the percentage deviation parameters should be used to distinguish between a dysfunctional low phase angle and a healthy low phase angle. These deviation parameters can also give greater insight into where in the breath cycle the asynchrony occurs. For example, Figure 6.3b displays an asynchronous DBP with greater deviation during the inhale (-34.57%) compared to the

exhale (-7.28%). Previous research has demonstrated that breathing pattern training within a DBP group improves symptoms and quality of life (Hagman *et al.*, 2011; Thomas *et al.*, 2003). These novel deviation parameters may allow for more targeted and individualised breathing pattern training.

As discussed in the previous chapter, individuals with EIB would be expected to demonstrate a high IE50 value post-EVH due to an increase in airway resistance. The DBP group consistently displayed lower IE50 values throughout all conditions when compared to the healthy breathing pattern group, although not significantly (Table 6.5 and 6.6). Similarly, the DBP group displayed lower IE50 values when compared to the EIB group from the previous chapter, with this difference being significant during the evoked state only (Table 6.7). This is further supported by greater reduction in FEV_1 obtained from the spirometer in the EIB group (-1.20 L) compared to the DBP group (-0.08 L) post-EVH from baseline. This demonstrates that the symptoms experienced by the individuals with a negative EVH test do not arise directly from airway restriction but may be caused by other factors such as an individual's breathing pattern.

A limitation of OEP is that in order to successfully track each marker, participants often have to extend their arms to the side. This is an unnatural position particularly during exercise and has the potential to alter the activation of inspiratory muscles and thus alter breathing pattern. To minimise this, participants rested their arms on stands rather than actively holding them out. Although this study provides insight into the differences between dysfunctional and healthy breathing patterns, in order to establish clear cut-off criteria to objectively support the diagnosis of dysfunctional breathing pattern, a more substantive data set would be required.

In conclusion, an exercise response for OEP derived breathing parameters have been presented for the first time in a DBP group. At rest, the regional contribution parameters (RcCT, RCaCT, AbCT, RcAbIndex) were significantly different between HBP and DBP with the DBP group displaying less ribcage and more abdominal contribution. During exercise, the DBP displayed significantly more asynchrony between various thoracic compartments (RcAbPhase, RCpRCaPhase, RCaSPhase, and RCpSPhase) such that thoracoabdominal asynchrony and thoracic dominant breathing patterns were identified in the DBP group. However, it is clear that there is not one singular OEP-derived breathing parameter that could be used to distinguish between healthy and dysfunctional breathing patterns at rest and during exercise, but by using a combination of these parameters greater

insight can be obtained and used to distinguish between these population groups. This includes the newly developed inhale and exhale percentage deviation parameters which successfully distinguished between the groups during high intensity exercise. Our findings suggest that OEP is capable of identifying differences between conditions within a population but also between populations, providing a basis for further investigation into OEP as a potential diagnostic and training tool for dysfunctional breathing.

Chapter 7. Real-time OEP phase angle feedback for dysfunctional breathing pattern re-training – an acute intervention study

7.1 Introduction

Dysfunctional breathing pattern has an impact on individual's quality of life with individuals experiencing breathing problems impairing daily life activities and/or exercise (Barker & Everard, 2015; Hagman *et al.,* 2011). Dysfunctional breathing is defined as chronic alterations in breathing pattern and may present as hyperventilation, an asynchronous breathing pattern, or a thoracic dominant breathing pattern (Boulding *et al.,* 2016). Dysfunctional breathing often occurs alongside other respiratory disorders/diseases further exacerbating the disorder/disease, for example, asthmatics with dysfunctional breathing experience increases in self-reported symptoms (Thomas *et al.,* 2001) and asthma medication use, and a reduction in asthma control (Courtney *et al.,* 2019; Veidal *et al.,* 2017). Breathing retraining is considered to be the first line of treatment for dysfunctional breathing (Bott, 2009). Breathing retraining aims to progressively alter and retrain an individual's breathing pattern, in order to restore and maintain a healthy, normal breathing pattern through the use of breathing exercises (Jones *et al.,* 2013). Breathing exercises can focus on a number of different factors including reducing respiratory rate and tidal volume (Bott, 2009), altering the pattern of breathing, improving respiratory muscle strength and/or endurance, improving posture, and/or increasing the range of motion of the thorax (Thomas & Bruton, 2014).

It has previously established within the literature that a normal breathing pattern can be described as the pulmonary ribcage, abdominal ribcage, and the abdomen moving in synchronisation (Bradley & Esformes, 2014). Achieving such a breathing pattern through breathing retraining has been investigated with techniques such as the Buteyko breathing (Bowler *et al.,* 1998), diaphragmatic breathing (Hagman *et al.,* 2011), Papworth method (Holloway & West, 2007), and yoga breathing (Singh *et al.,* 1990). However, there has been no consensus on the use of breathing exercises within clinical practice (Barker *et al.,* 2013; Jones *et al.,* 2013). Additionally, flow-incentive and volume-incentive spirometry feedback have been investigated as potential breathing training aids, with volume-incentive spirometry promoting deeper and slower breathing in patients post-gastroplasty (Tomich *et*

al., 2010) and healthy individuals (de Moraes Paisani *et al.,* 2013). This indicates that different forms of feedback used in breathing training can have different impacts on breathing pattern.

The primary outcomes for many breathing retraining studies are self-reported questionnaires such as the asthma quality of life questionnaire (Bruton *et al.,* 2018), the Nijmegen questionnaire (Hagman *et al.,* 2011), or the self-evaluation of breathing questionnaire (Courtney *et al.,* 2019). Bruton and colleagues (2018) demonstrated that quality of life scores improved in asthmatics receiving breathing training versus standard care and reduced adverse events. Similarly, a five year follow-up breathing retraining study for individuals with dysfunctional breathing demonstrated significantly improvements in quality of life scores, reduced symptoms, and reduced emergency room visits (Hagman *et al.,* 2011). Other studies have used, in addition to questionnaires, lung function measures including force expired volume in one second (FEV_1) via spirometry (Bruton *et al.,* 2018) and ultrasound to measure diaphragm excursion (Alaparthi *et al.,* 2016). Although breathing retraining altered quality of life scores in asthmatics, FEV_1 and exhaled nitric oxide were not significantly altered pre and post intervention as expected (Bruton *et al.,* 2018). When using ultrasound to measure diaphragmatic excursion, it was found that both diaphragmatic breathing exercises and volume incentive spirometry helped to preserve diaphragm excursion in patients post laparoscopic surgery when compared to no breathing training and flow incentive spirometry (Alaparthi *et al.,* 2016).

Courtney and colleagues (2019) utilised the manual assessment of respiratory motion (MARM) along with self-reported questionnaires and lung function measures to assess the influence of combined breathing retraining and manual therapy on dysfunctional breathing in asthmatics. Similar to other retraining studies, dysfunctional breathing symptoms were reduced post treatment. Additionally, the MARM indicated reduced asynchrony between the upper ribcage and the abdominal ribcage/abdomen (Courtney *et al.,* 2019). Unlike self-reported questionnaires, the MARM provides insight into the alterations of chest wall movement, however, it does require subjective interpretation by the examiner. From these results, breathing retraining can reduce the asynchrony associated with dysfunctional breathing which leads to a subsequent reduction in symptoms and improved quality of life scores.

OEP has previously been used, as an objective measure, to assess multiple different types of breathing interventions including breath/air stacking (De Melo Barcelar *et al.*, 2014; Sarmento *et al.*, 2017), incentive spirometry (de Moraes Paisani *et al.*, 2013), loaded inspiratory breathing (Brandao *et al.*, 2012), respiratory muscle stretching (de Sa *et al.*, 2017), and rehabilitation (Albuquerque *et al.*, 2016). Changes in breathing pattern during different types of breathing exercises including diaphragmatic breathing, inspiratory sighs, sustained maximal inspiration and intercostal breathing in healthy subjects has also been assessed using OEP (Vieira *et al.*, 2014). These breathing exercises were shown to alter OEP-derived respiratory rate, tidal volume, thoracic compartment contribution, and phase angles when compared to tidal breathing (Vieira *et al.*, 2014). More specifically, the phase angles between the ribcage and abdomen, and between the upper and lower ribcage have been shown to significantly alter with breathing exercises (Vieira *et al.*, 2014), while the phase angle between the upper ribcage and abdomen did not significantly change (de Moraes Paisani *et al.*, 2013) in healthy individuals. Similarly, changes in the phase angle between the ribcage and abdomen have been reported using respiratory inductive plethysmography (RIP) between diaphragmatic breathing, flow-oriented spirometry, and volume-oriented spirometry in patients with obesity post gastroplasty (Tomich *et al.*, 2010). Previously in this thesis, we have demonstrated that the phase angle between the ribcage and abdomen was the only phase angle to differ between individuals with dysfunctional breathing during high intensity exercise and both healthy individuals and individuals with exercise-induced bronchoconstriction (Chapter 6).

For individuals with asthma, breathing exercises improve quality of life scores but lung function and airway inflammation remain unchanged (Bruton *et al.*, 2018). However, evidence suggests that breathing retraining can be used to alter dysfunctional breathing patterns (Hagman *et al.*, 2011). From the previous literature, volume-based feedback systems are promoted over alternatives such as flow-based feedback for breathing training as it results in more controlled breaths and reduced breathing pattern asynchrony (de Moraes Paisani *et al.*, 2013; Tomich *et al.*, 2010). Objective breathing pattern measurement systems, including OEP, have not been used to assess the influence of breathing retraining on the breathing pattern of individuals with dysfunctional breathing nor has OEP been use to provide real-time feedback on breathing pattern to any population group. For the first time, a real-time OEP feedback system has been developed with the ability to display both volume and phase angle feedback.

To our knowledge, phase angle feedback has never been used in real-time as part of breathing retraining. It was hypothesised that real-time OEP phase angle feedback between the ribcage and the abdomen (RcAbPhase) along with real-time volume feedback, as part of a breathing retraining intervention, would result in a greater reduction the RcAbPhase phase angle value at rest and during exericse when compared to real-time OEP volume-based feedback alone, in individuals with dysfunctional breathing.

7.2 Methods

Participants

Eighteen participants (Table 7.1) with a suspected dysfunctional breathing pattern gave written informed consent to participate in this study. This study was approved by the School of Sport & Exercise Sciences Research Ethics Advisory Group at the University of Kent (Prop 92_2018_19). Participants were recruited from the previous study (Chapter 6) in this thesis. Inclusion criteria included reporting respiratory symptoms either at rest and/or during exercise, and a negative eucapnic voluntary hyperpnoea test. Participants were required to have a consistent training regime with a minimum of 5 hours of physical activity per week. Participants were randomly assigned to either the control group which included the volume-based OEP feedback, or to the experimental group which included both the volume-based and phase angle OEP feedback. Participants performed an exercise challenge, both groups received the same breathing training instructions, then repeated the same exercise challenge with real-time OEP feedback.

Table 7.1 Participant Information (mean ± standard deviation)

N	Age (yrs)	Height (m)	Body Mass (kg)	Activity Level (hrs/week)
Total				
18	36.8± 9.4	1.7 ± 0.1	66.9 ± 12.6	6.1 ± 3.6
Control Group				
9	38.6 ± 10.7	1.7 ± 0.1	63.6 ± 11.3	6.7 ± 4.6
Experimental Group				
9	35.0 ± 8.2	1.7 ± 0.1	70.2 ± 13.6	5.5 ± 2.2

Exercise Challenge

Participants were instructed to remain in an upright position on a cycle ergometer (Lode – Corival, Netherlands). Participants began with a period of tidal breathing, followed by a sub-maximal incremental exercise challenge. The cycle challenge began at 50 W and increased by 30 W every minute. Breathing pattern data was recorded using the OEP system at rest, during low, moderate, and high intensity exercise, and during a period of recovery post-exercise, following the method described in Chapter 3. Rate of perceived exertion (RPE)

scale was used to define the exercise intensities with 11, 13/14, and 17/18 RPE values representing low, moderate, and high intensity exercise, respectively (Muyor, 2013).

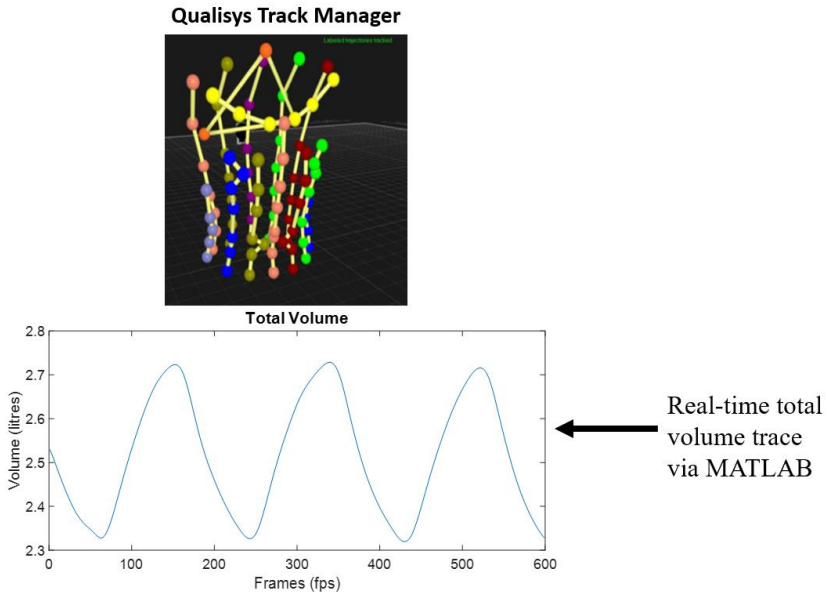
Breathing Training

Once recovered from the exercise challenge, both the control and experimental group were given the same set of breathing training instructions (Table 7.2). The control group used a real-time OEP plot of their total volume trace (Figure 7.1a). The experimental group used an additional real-time plot of the Konno-Mead loop associated with RcAbPhase, with the added instruction to aim to follow a straight 45-degree diagonal line within the Konno-Mead plot (Figure 7.1b). The real-time OEP feedback system was developed using QTM Connect for MATLAB (Qualisys, Goteborg, Sweden) and a custom-built MATLAB script which allowed for the real-time streaming of the OEP data which was projected onto a board in front of the cycle ergometer. Participants were allowed time to practice implementing the feedback and using the real-time plot at rest for as long as required. Once comfortable with the instructions and feedback plots, participants then repeated the exercise challenge with the real-time OEP plots. OEP data was again recorded at rest, low, moderate and high intensity exercise, and recovery post-exercise with the exercise intensities being matched to their pre-breathing training exercise challenge.

Table 7.2 Breathing Training Instructions

Instructions
• Aim to move the ribcage and abdomen together • Avoid initiating the breath from the shoulders or abdomen • Focus on using the diaphragm and intercostal muscles to initiate the movement of the lower ribcage laterally • Aim for smooth inhalation with maximum movement of the ribcage • Using the real-time OEP plot, get the breath trace as smooth as possible

a) Control Group



b) Experimental Group

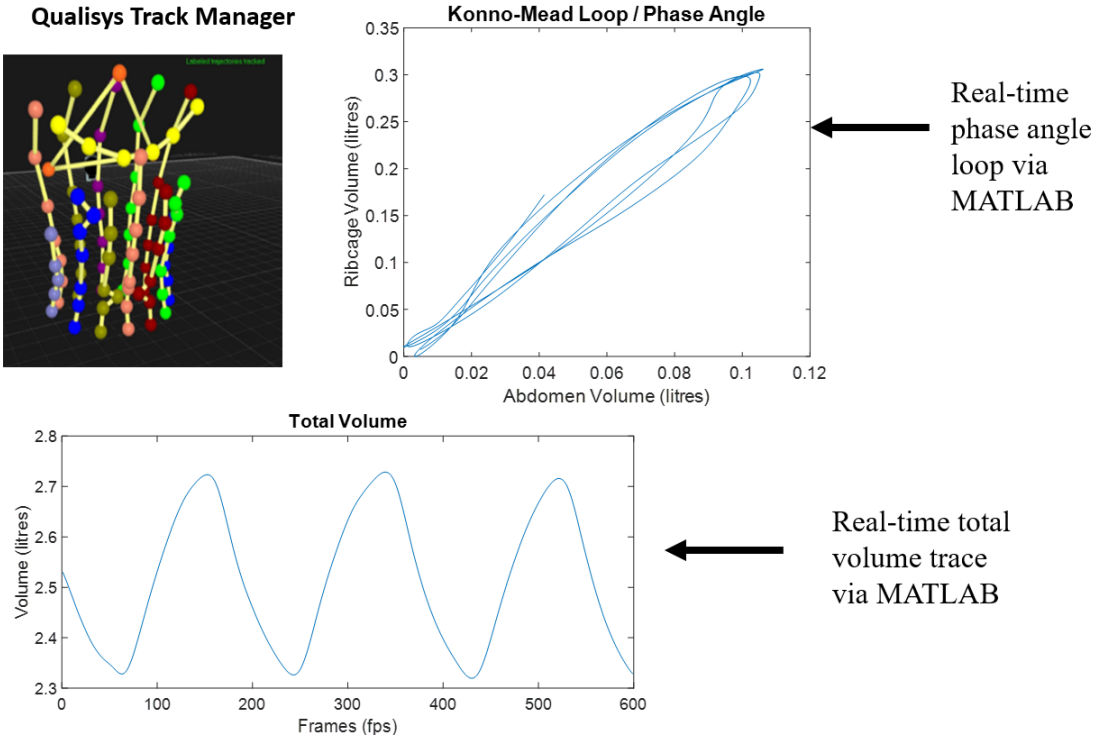


Figure 7.1 OEP real-time feedback system a) for the control group with Qualisys Track Manager attaching the AIM model to the markers and real-time total volume plot streamed through MATLAB and b) for the experimental group with the addition of real-time phase angle plot (abdomen vs ribcage) streamed through MATLAB.

Data Analysis

The OEP trials were gap filled for up to 10 frames using Qualisys Track Manager v2019.3. All of the OEP-derived breathing parameters (see Table 3.1) were calculated as described in Chapter 3 using custom-built MATLAB scripts. A mixed model repeated measures ANOVA was used to determine if there was an interaction effect between the intervention time points i.e., pre and post intervention and group. Follow-up paired sample t-tests were performed to determine if there were significant differences within a group pre and post intervention.

7.3 Results

Table 7.3, 7.4, 7.5, 7.6, and 7.7 display the comparison between the experimental and control group during rest, low, moderate, high intensity exercise, and recovery post exercise, respectively, for the collection of breathing parameters. During rest, there was no significant interaction effects for any of the breathing pattern parameters. However, during low, moderate, high intensity exercise, recovery post exercise there was a significant interaction effect for RcAbPhase, RCpAbPhase, and AbSphase. More specifically, the experimental group displayed a significant decrease, in the positive direction, post-intervention in RcAbPhase value of 0.6, 2.0, 1.5, and 1.6 degrees during low, moderate, high intensity exercise, and recovery, respectively (see Figure 7.2a). This indicates a reduction in asynchrony between the ribcage and abdomen post-intervention during exercise with the abdomen initiating the movement during moderate and high intensity only. While for the control group, RcAbPhase increased in value, in the negative direction, by 0.8, 1.0, 0.7, and 0.4 degrees during low, moderate, high intensity exercise, and recovery, respectively. This demonstrates an increase in asynchrony post intervention between these compartments with the abdomen initiating the movement during all conditions.

RCpAbPhase values decreased for the experimental group post-intervention by 0.4, 1.6, 1.5, and 0.5 for the low, moderate, high intensity exercise, and recovery conditions, respectively. This demonstrates that the asynchrony between the upper ribcage and combined upper ribcage and abdomen was reduced post-intervention with the abdomen and lower ribcage initiating the movement during high intensity exercise only. While for low, moderate, high intensity exercise, and recovery for the control group, the RCpAbPhase values increased post-intervention by 0.2, 0.7, 0.6, and 0.3, respectively. This indicates that the asynchrony between these compartments increased post-intervention with the upper ribcage and the abdomen moving first (see Figure 7.2b). For AbSPhase (see Figure 7.2c), the experimental group values decreased post-intervention by 0.1, 1.6, 1.9, and 1.1 degrees during low, moderate, high intensity exercise, and recovery post-exercise, respectively. This demonstrates that the asynchrony between the abdomen and shoulders reduced post-intervention with the abdomen initiating the movement during high intensity exercise only. For the control group, AbSPhase increased in value during low, moderate, high intensity exercise, and recovery by 0.2, 1.4, 1.0, and 0.5 degrees, respectively. This demonstrates that the asynchrony between these compartments became more asynchronous during exercise post-intervention with the abdomen initiating the movement across all conditions.

Table 7.3 Comparison of breathing parameters (mean and standard deviation (SD)) between the experimental and control group at rest pre and post breathing pattern acute intervention. * represents a significant interaction effect between time point (pre and post) and group with $p \leq 0.05$. † represents a significant change pre and post within a given group with $p \leq 0.05$.

Rest	**Experimental Group**				**Control Group**				**Interaction**
	Pre		**Post**		**Pre**		**Post**		**Effect**
	Mean	**SD**	**Mean**	**SD**	**Mean**	**SD**	**Mean**	**SD**	***p* - value**
RR (brpm)	14.63	3.37	11.39	2.55 †	15.32	3.22	12.80	4.38	0.73
Vt (litres)	0.79	0.44	1.33	0.35 †	0.68	0.30	1.48	1.09 †	0.47
VE (lit/min)	11.49	7.06	15.10	4.93	10.16	3.07	16.43	8.82	0.44
tI (s)	1.76	0.38	2.44	0.77	1.74	0.48	2.02	0.53	0.19
tE (s)	2.26	0.46	2.88	0.78 †	2.27	0.77	2.88	1.12 †	0.97
tTot (s)	4.07	0.82	5.36	1.51	4.10	1.24	1.89	1.62	0.48
tIERatio	0.82	0.16	0.85	0.12	0.76	0.09	0.74	0.14	0.53
tITotRatio	0.45	0.05	0.46	0.04	0.43	0.03	0.42	0.05	0.45
tPTEFRatio	0.40	0.15	0.33	0.10	0.33	0.09	0.34	0.13	0.38
tPTIFRatio	0.48	0.08	0.39	0.08 †	0.53	0.07	0.43	0.11 †	0.90
IE50	1.23	0.15	1.23	0.18	1.41	0.46	1.38	0.31	0.91
RCpCT (%)	56.43	4.71	56.39	4.68	56.28	5.41	55.80	5.45	0.32
RCaCT (%)	11.42	1.43	11.43	1.60	11.75	2.10	11.86	2.01	0.47
RcCT (%)	67.85	3.55	67.82	3.46	68.04	3.88	67.68	3.86	0.41
AbCT (%)	32.15	3.55	32.18	3.46	31.96	3.88	32.32	3.86	0.41
RCaAbCT (%)	43.57	4.71	43.61	4.68	43.72	5.41	44.20	5.45	0.32
RcAbIndex	2.15	0.38	2.14	037	2.17	0.42	2.14	0.40	0.39
HTA (deg)	-0.10	0.07	-0.13	0.18	0.04	0.08	-0.01	0.22	0.80
RcHTA (deg)	-0.08	0.09	-0.08	0.15	0.05	0.07	-0.04	0.27	0.45
AbHTA (deg)	-0.07	0.19	0.06	0.28	-0.01	0.20	0.03	0.27	0.48
RCpHTA (deg)	-0.08	0.10	-0.05	0.15	0.04	0.07	0.01	0.23	0.97
RCaHTA (deg)	-0.09	0.26	-0.22	0.65	0.12	0.16	-0.14	0.35	0.66
RcAbPhase (deg)	-0.05	0.37	-0.08	0.58	-0.34	0.70	-1.39	2.21	0.11
RCpAbPhase (deg)	-0.02	0.37	-0.01	0.52	-0.33	0.54	-1.14	1.94	0.14
RCpRCaPhase (deg)	0.06	0.30	0.17	0.41	-0.03	0.14	-0.43	1.09	0.17
RCaSPhase (deg)	0.12	0.27	0.20	0.67	0.01	0.14	-0.54	1.26	0.19
RCpSPhase (deg)	0.22	0.23	0.10	0.46	0.11	0.19	0.02	0.47	0.89
AbSPhase (deg)	0.04	0.40	-0.18	0.81	-0.47	0.78	-1.57	2.49	0.07
Inhale % Deviation (%)	-3.75	9.60	-10.69	17.51	6.07	11.90	11.23	14.04	0.13
Exhale % Deviation (%)	-6.11	12.95	-7.72	13.70	-10.63	14.85	-7.93	13.02	0.64

Table 7.4 Comparison of breathing parameters (mean and standard deviation (SD)) between the experimental and control group during low intensity exercise pre and post breathing pattern acute intervention. * represents a significant interaction effect between time point (pre and post) and group with $p \leq 0.05$. † represents a significant change pre and post within a given group with $p \leq 0.05$.

Low Intensity	Experimental Group				Control Group				Interaction
	Pre		Post		Pre		Post		Effect
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	*p* - value
RR (brpm)	22.66	5.41	12.64	2.59 †	22.88	4.34	15.40	4.05 †	0.30
Vt (litres)	1.81	0.55	2.88	0.77 †	1.68	0.67	2.30	0.82 †	0.10
VE (lit/min)	39.29	13.55	40.00	13.97	37.62	13.13	36.14	11.43 †	0.85
tI (s)	1.20	0.16	2.15	0.48 †	1.17	0.20	1.79	0.31 †	0.15
tE (s)	1.37	0.25	2.47	0.52 †	1.31	0.52	2.10	0.45 †	0.23
tTot (s)	2.57	0.40	4.60	0.95 †	2.47	0.45	3.76	0.85 †	0.13
tIERatio	0.88	0.08	0.88	0.10	0.91	0.05	0.90	0.17	0.79
tITotRatio	0.47	0.02	0.47	0.03	0.48	0.02	0.47	0.05	0.64
tPTEFRatio	0.49	0.07	0.42	0.08	0.49	0.06	0.42	0.10	0.88
tPTIFRatio	0.47	0.04	0.41	0.08	0.47	0.04	0.44	0.09	0.53
IE50	1.18	0.19	1.14	0.17	1.15	0.13	1.22	0.34	0.73
RCpCT (%)	56.71	4.75	56.61	4.78	56.42	6.00	56.44	6.16	0.77
RCaCT (%)	11.60	1.37	11.61	1.35	12.10	2.05	12.00	2.11	0.21
RcCT (%)	68.32	3.70	68.21	3.67	68.53	4.73	68.47	4.70	0.87
AbCT (%)	31.68	3.70	31.79	3.67	31.47	4.73	31.53	4.70	0.87
RCaAbCT (%)	43.29	4.75	43.39	4.78	43.59	6.00	43.56	6.16	0.65
RcAbIndex	2.20	0.41	2.19	0.40	2.25	0.55	2.24	0.53	0.96
HTA (deg)	0.07	0.39	-0.10	0.55	0.03	0.42	0.04	0.31	0.36
RcHTA (deg)	0.01	0.30	-0.02	0.32	0.02	0.35	0.00	0.18	0.91
AbHTA (deg)	-0.45	0.98	-0.13	0.80	-0.31	0.66	-0.23	0.77	0.37
RCpHTA (deg)	0.03	0.29	-0.10	0.43	-0.02	0.33	0.03	0.39	0.50
RCaHTA (deg)	-0.06	0.57	0.20	0.82	-0.21	0.38	-0.35	0.43	0.78
RcAbPhase (deg)	-1.32	0.76	0.68	1.75	-1.49	0.82	-2.27	3.01	0.02 *
RCpAbPhase (deg)	-1.01	0.46	0.61	1.38 †	-1.03	0.68	-1.26	1.13	< 0.01 *
RCpRCaPhase (deg)	0.11	0.40	0.52	0.55 †	-0.08	0.5	-0.03	0.92	0.20
RCaSPhase (deg)	0.30	0.53	0.92	1.23 †	0.10	0.61	0.08	0.82	0.13
RCpSPhase (deg)	0.35	0.43	0.69	0.83	0.30	0.41	0.29	0.49	0.23
AbSPhase (deg)	-0.99	0.88	0.85	2.09 †	-1.26	1.13	-1.42	1.06	< 0.01 *
Inhale % Deviation (%)	9.37	7.52	-1.39	14.10 †	1.41	15.98	18.68	12.44	< 0.01*
Exhale % Deviation (%)	-8.54	10.28	2.22	9.65	-12.40	10.95	-0.70	15.32 †	0.88

Table 7.5 Comparison of breathing parameters (mean and standard deviation (SD)) between the experimental and control group during moderate intensity exercise pre and post breathing pattern acute intervention. * represents a significant interaction effect between time point (pre and post) and group with $p \leq 0.05$. † represents a significant change pre and post within a given group with $p \leq 0.05$.

Moderate Intensity Exercise	Experimental Group				Control Group				Interaction Effect
	Pre		Post		Pre		Post		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	*p*-value
RR (brpm)	25.74	6.17	18.48	6.06 †	27.13	5.68	17.53	1.70 †	0.44
Vt (litres)	2.45	0.72	2.90	1.03 †	2.10	0.88	2.67	0.86 †	0.48
VE (lit/min)	61.48	19.60	52.53	21.64 †	51.90	14.67	44.98	12.36	0.67
tI (s)	1.13	0.46	1.63	0.49 †	1.03	0.21	1.53	0.27 †	1.00
tE (s)	1.19	0.19	1.79	0.62 †	1.15	0.28	1.72	0.38 †	0.92
tTot (s)	2.31	0.33	3.42	1.10 †	2.17	0.48	3.21	0.48 †	0.86
tIERatio	0.94	0.07	0.92	0.11	0.90	0.08	0.90	0.16	0.65
tITotRatio	0.48	0.02	0.48	0.03	0.47	0.02	0.47	0.05	0.81
tPTEFRatio	0.49	0.04	0.47	0.08	0.48	0.06	0.45	0.11	0.59
tPTIFRatio	0.47	0.04	0.44	0.05	0.48	0.03	0.45	0.05 †	0.88
IE50	1.12	0.19	1.11	0.16	1.16	0.15	1.23	0.47	0.54
RCpCT (%)	56.49	4.81	56.50	4.84	56.45	6.03	56.51	5.98	0.84
RCaCT (%)	11.64	1.30	11.66	1.34	12.13	2.11	12.08	2.01	0.56
RcCT (%)	68.15	3.70	68.19	3.70	68.60	4.74	68.67	4.64	0.93
AbCT (%)	31.85	3.70	31.81	3.70	31.40	4.74	31.33	4.64	0.93
RCaAbCT (%)	43.51	4.81	43.50	4.84	43.55	6.03	43.49	5.98	0.84
RcAbIndex	2.18	0.40	2.19	0.40	2.26	0.56	2.26	0.53	0.89
HTA (deg)	0.28	0.38	0.27	0.40	0.09	0.39	-0.02	0.30	0.63
RcHTA (deg)	0.10	0.21	0.11	0.36	-0.08	0.22	0.00	0.33	0.66
AbHTA (deg)	0.44	0.83	0.05	0.75	0.31	1.00	0.24	0.82	0.51
RCpHTA (deg)	0.10	0.32	0.10	0.35	-0.09	0.19	0.07	0.35	0.46
RCaHTA (deg)	0.04	0.33	-0.15	0.35	0.13	0.63	0.11	0.76	0.33
RcAbPhase (deg)	-2.21	0.66	-0.18	1.43 †	-2.04	1.10	-3.07	3.08	< 0.01 *
RCpAbPhase (deg)	-1.62	0.30	0.03†	0.82 †	-1.42	0.98	-2.10	2.63	< 0.01 *
RCpRCaPhase (deg)	0.06	0.70	0.56	0.78 †	0.27	0.68	-0.02	1.03	0.06
RCaSPhase (deg)	0.63	0.64	0.85	1.02	0.54	0.78	0.26	1.59	0.41
RCpSPhase (deg)	0.50	0.44	0.75	0.85	0.53	0.48	0.45	0.35	0.23
AbSPhase (deg)	-1.65	0.58	0.08	1.57 †	-1.66	1.17	-3.05	3.32	0.02 *
Inhale % Deviation (%)	11.54	7.64	4.66	13.08	4.63	23.92	5.81	50.49	0.44
Exhale % Deviation (%)	-10.17	9.88	0.70	12.12 †	-12.55	13.05	-7.39	18.12	0.26

Table 7.6 Comparison of breathing parameters (mean and standard deviations (SD)) between the experimental and control group during high intensity exercise pre and post breathing pattern acute intervention. * represents a significant interaction effect between time point (pre and post) and group with $p \leq 0.05$. † represents a significant change pre and post within a given group with $p \leq 0.05$.

High Intensity	Experimental Group				Control Group				Interaction
	Pre		Post		Pre		Post		Effect
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	*p* - value
RR (brpm)	32.90	7.79	25.51	8.27 †	36.41	5.42	26.27	4.97 †	0.40
Vt (litres)	2.98	0.91	3.24	0.85	2.59	1.06	2.75	1.03	0.64
VE (lit/min)	100.42	44.53	82.33	35.67 †	88.89	32.84	69.03	26.74 †	0.82
tI (s)	0.89	0.12	1.18	0.42 †	0.79	0.12	1.05	0.17 †	0.85
tE (s)	0.94	0.16	1.36	0.59 †	0.83	0.13	1.16	0.25 †	0.60
tTot (s)	1.83	0.27	2.51	0.93 †	1.63	0.25	2.22	0.37 †	0.76
tIERatio	0.95	0.05	0.92	0.09	0.95	0.03	0.93	0.12	0.84
tITotRatio	0.49	0.02	0.48	0.03	0.49	0.01	0.48	0.04	0.94
tPTEFRatio	0.49	0.03	0.47	0.04	0.50	0.02	0.47	0.08	0.87
tPTIFRatio	0.50	0.01	0.47	0.03	0.50	0.01	0.48	0.04	0.88
IE50	1.04	0.10	1.09	0.18	1.04	0.06	1.12	0.32	0.79
RCpCT (%)	56.45	4.69	56.49	4.86	56.61	6.09	56.79	6.24	0.31
RCaCT (%)	11.87	1.34	11.82	1.45	12.23	2.20	12.16	2.17	0.14
RcCT (%)	68.36	3.50	68.34	3.58	68.86	4.79	68.99	4.79	0.53
AbCT (%)	31.64	3.50	31.66	3.58	31.14	4.79	31.02	4.79	0.41
RCaAbCT (%)	43.55	4.69	43.51	4.86	43.40	6.09	43.21	6.24	0.54
RcAbIndex	2.20	0.38	2.20	0.39	2.29	0.58	2.30	0.56	0.69
HTA (deg)	-0.04	0.28	-0.08	0.38	0.20	0.37	-0.01	0.41	0.41
RcHTA (deg)	0.01	0.23	-0.13	0.26	0.10	0.21	-0.08	0.40	0.80
AbHTA (deg)	-0.22	0.50	-0.14	0.96	0.31	0.65	-0.10	1.37	0.32
RCpHTA (deg)	0.07	0.31	-0.12	0.40	0.14	0.29	-0.07	0.48	0.91
RCaHTA (deg)	0.12	0.58	0.32	0.59	0.28	0.76	-0.17	0.93	0.10
RcAbPhase (deg)	-2.89	1.20	-1.39	1.54 †	-2.97	1.98	-3.65	2.48	< 0.01 *
RCpAbPhase (deg)	-2.00	0.93	-0.50	1.31 †	-1.96	1.46	-2.54	1.93	< 0.01 *
RCpRCaPhase (deg)	0.57	0.59	0.71	0.85	0.27	0.87	0.23	0.86	0.42
RCaSPhase (deg)	0.84	0.70	1.14	1.37	0.49	1.09	0.31	1.08	0.29
RCpSPhase (deg)	0.57	0.63	0.79	0.86	0.58	0.55	0.75	0.81	0.87
AbSPhase (deg)	-2.60	1.35	-0.72	1.97 †	-2.39	2.34	-3.38	2.61	< 0.01 *
Inhale % Deviation (%)	10.35	4.91	6.86	10.85	5.27	26.04	16.90	14.34	0.11
Exhale % Deviation (%)	-14.79	11.27	-6.93	19.36	-19.64	9.62	-10.83	8.37 †	0.88

Table 7.7 Comparison of breathing parameters (mean and standard deviation (SD)) between the experimental and control group during recovery post exercise pre and post breathing pattern acute intervention. * represents a significant interaction effect between time point (pre and post) and group with $p \leq 0.05$. † represents a significant change pre and post within a given group with $p \leq 0.05$.

Recovery	Experimental Group				Control Group				Interaction
	Pre		Post		Pre		Post		Effect
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	*p* - value
RR (brpm)	23.19	7.51	17.19	4.36 †	23.52	5.35	17.76	2.97 †	0.95
Vt (litres)	2.13	0.77	2.26	0.62	1.74	0.80	1.89	0.88	0.95
VE (lit/min)	49.10	23.42	39.53	18.95	41.77	23.79	32.09	11.74	0.99
tI (s)	1.25	0.37	1.70	0.57 †	1.17	0.23	1.54	0.20 †	0.74
tE (s)	1.39	0.35	2.01	0.69 †	1.36	03.3	1.88	0.41 †	0.70
tTot (s)	2.64	0.70	3.68	1.21 †	2.53	0.55	3.42	0.55 †	0.75
tIERatio	0.91	0.10	0.86	0.11	0.86	0.09	0.82	0.12	0.75
tITotRatio	0.48	0.03	0.46	0.03	0.46	0.03	0.45	0.04	0.84
tPTEFRatio	0.44	0.09	0.40	0.10 †	0.43	0.08	0.38	0.13	0.83
tPTIFRatio	0.47	0.04	0.43	0.09	0.49	0.03	0.45	0.06	0.98
IE50	1.13	0.15	1.20	0.17	1.17	0.13	1.23	0.22	0.91
RCpCT (%)	56.65	4.95	56.59	4.88	56.59	5.98	56.57	6.13	0.84
RCaCT (%)	11.47	1.44	11.44	1.59	11.87	2.30	11.87	2.04	0.89
RcCT (%)	68.13	3.76	68.03	3.58	68.47	4.50	68.46	4.60	0.81
AbCT (%)	31.87	3.76	31.97	3.58	31.53	4.50	31.54	4.60	0.81
RCaAbCT (%)	43.35	4.95	43.41	4.88	43.42	5.98	43.43	6.13	0.84
RcAbIndex	2.18	0.42	2.17	0.40	2.24	0.51	2.24	0.50	0.77
HTA (deg)	0.05	0.11	0.03	0.22	0.09	0.10	0.00	0.17	0.54
RcHTA (deg)	0.06	0.16	0.07	0.24	0.05	0.13	-0.07	0.22	0.16
AbHTA (deg)	0.02	0.41	-0.04	0.28	0.06	0.17	0.02	0.38	0.92
RCpHTA (deg)	0.04	0.17	0.02	0.25	0.06	0.11	-0.04	0.19	0.43
RCaHTA (deg)	0.22	0.37	0.22	0.46	0.05	0.34	-0.07	0.38	0.36
RcAbPhase (deg)	-1.16	1.21	0.06	1.16 †	-0.96	1.15	-1.38	1.95	0.02 *
RCpAbPhase (deg)	-0.73	1.15	0.26	0.86 †	-0.56	0.94	-0.86	1.42	< 0.01 *
RCpRCaPhase (deg)	0.46	0.82	0.59	0.66	0.41	0.72	0.31	0.65	0.44
RCaSPhase (deg)	0.46	1.05	0.65	0.83	0.57	0.91	0.33	1.11	0.24
RCpSPhase (deg)	0.41	0.64	0.50	0.64	0.37	0.52	0.36	0.87	0.67
AbSPhase (deg)	-1.17	1.61	0.11	1.08 †	-0.78	1.26	-1.31	1.52	< 0.01 *
Inhale % Deviation (%)	8.27	11.33	-1.68	13.20 †	9.00	13.88	13.88	18.08	0.04*
Exhale % Deviation (%)	-6.38	12.58	2.19	12.84	-6.97	16.66	-9.48	18.67	0.22

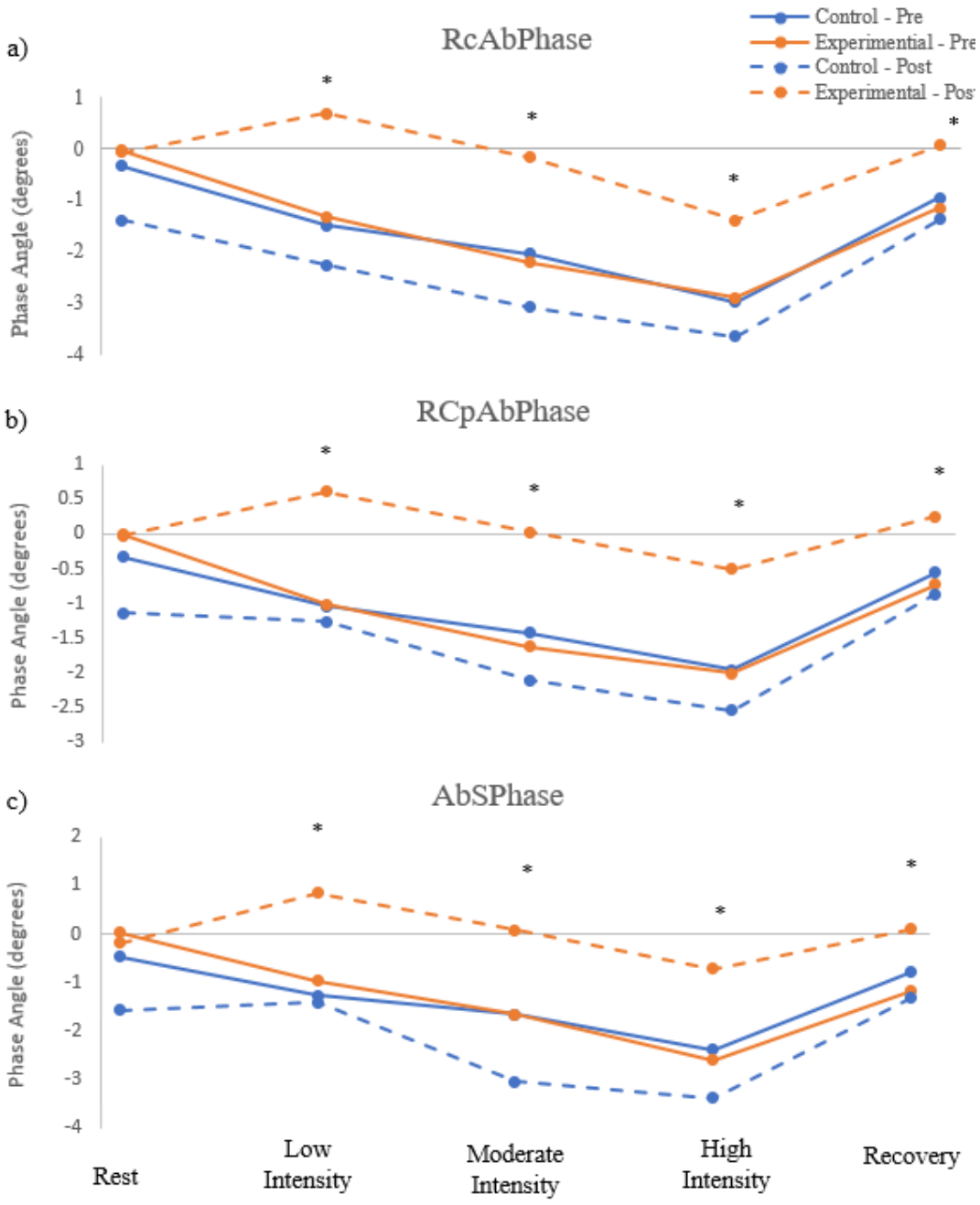


Figure 7.2 Illustration of the phase angle between the ribcage and abdomen (RcAbPhase), the phase angle between the upper ribcage and the combined lower ribcage and abdomen (RCpAbPhase), and the phase angle between the abdomen and the shoulders (AbSPhase) exercise response pre (solid lines) and post (dashed lines) breathing pattern intervention for the experimental and control group. A phase angle of zero degrees represents perfect synchrony between the compartments and values above or below this indicating that asynchrony is present with the larger the value the more asynchrony present. * significant with $p \leq 0.05$.

7.4 Discussion

For the first time, this study used OEP as a real-time feedback system for breathing pattern training in athletes with dysfunctional breathing at rest and during exercise as part of an acute intervention. The findings demonstrate that real-time OEP phase angle feedback can reduce the values of the phase angles between the ribcage and abdomen (RcAbPhase), the upper ribcage and combined lower ribcage and abdomen (RCpAbPhase), and the abdomen and shoulders (AbSPhase) during high intensity exercise. However, this improvement did not change the coordination of the compartment movements of athletes with dysfunctional breathing during high intensity exercise.

The respiratory rate (RR) decreased significantly post-intervention for both groups during exercise and recovery (Tables 7.1 – 7.5). The timing variables (tI, tE, tTot) responded as expected with the reduction in RR post-intervention with significant decreases in time during exercise and recovery. This indicates that both forms of feedback aided the control of respiratory rate. Dysfunctional breathing may not only present as compartment asynchrony but in some individuals, it may present as hyperventilation (Boulding *et al.,* 2016). In the previous chapter, when comparing the respiratory rate of healthy and dysfunctional breathing patterns there were no significant differences found at rest and during exercise (Chapter 6). However, reducing respiratory rate is a common goal of breathing retraining (Bott, 2009) as it improves control and reduces respiratory alkalosis and symptoms such as dizziness (Courtney *et al.,* 2009). Previously, diaphragmatic, inspiratory sighs, sustained maximal inspiration, and intercostal breathing have all been shown to significantly reduced OEP-derived respiratory rate in healthy individuals (Vieira *et al.,* 2014). Similarly, Tomich and colleagues (2010) demonstrated that volume-incentive spirometry reduced respiratory rate obtained from RIP in obese patients when compared to flow-incentive spirometry. Breathing exercises have also been shown to reduce symptoms of hyperventilation when assessed using the Nijmegen Questionnaire, improving breath control in individual with asthma (Courtney *et al.,* 2019) and dysfunctional breathing (Hagman *et al.,* 2011). This study indicates that OEP feedback, both phase angle and volume, can aid in reducing respiratory rate in individuals with dysfunctional breathing during exercise.

The regional contribution parameters did not significantly change pre and post intervention between the two groups across each condition. For both groups, any changes between pre and post intervention were less than 2% indicating that using both forms of OEP feedback did not alter the regional contributions during a breath for all conditions measured. Other

types of breathing exercises have been shown to alter regional contribution, for example diaphragmatic breathing significantly increases OEP-derived abdominal contribution in healthy individuals, while inspiratory sighs, sustained maximal inspiration and intercostal breathing significantly reduce abdominal contribution (Vieira *et al.*, 2014). In obese individuals, the ribcage contribution has been shown to significantly increase during diaphragmatic breathing, and flow and volume incentive spirometry using RIP (Tomich *et al.*, 2010). Interestingly, when comparing healthy and dysfunctional breathing patterns in Chapter 6, ribcage and abdomen contribution did significantly differ at rest. The values of both groups pre and post intervention in this chapter are comparable to the dysfunctional breathing pattern group in Chapter 6, therefore, this acute use of OEP real-time feedback had no impact on dysfunctional regional contributions.

In the experimental group, the phase angle between the ribcage and abdomen (RcAbPhase) displayed significant reduction in phase angle value i.e., less asynchrony, during exercise (low, moderate, and high) and recovery post-exercise post intervention when compared to the control group (Figure 7.2). At rest, RcAbPhase did not significantly differ between the groups (Table 7.3). The control group demonstrated an increase in phase angle values post-intervention indicating more asynchrony between these thoracic compartments. This supports our hypothesis, as RcAbPhase is the phase angle that was displayed in real-time, we would expect to see improvements in ribcage and abdomen synchrony if this phase angle feedback was useful to athletes with dysfunctional breathing during breathing retraining. For example, during high intensity exercise the experimental group pre-intervention RcAbPhase value is -2.89 degrees, with this decreasing to -1.39 degrees post-intervention returning this phase angle value closer to that seen in a healthy population (95% confidence interval [-2.2, -1.6]) (see Table 7.6). This phase angle is a measure of thoracoabdominal asynchrony which is a commonly reported sub-type of dysfunctional breathing (Boulding *et al.*, 2016). It has previously been reported that breathing exercises including diaphragmatic and intercostal breathing significantly increases RcAbPhase in healthy individuals using OEP (Vieira *et al.*, 2014). Similarly, in obese patients, flow and volume incentive spirometry increases RcAbPhase using RIP (Tomich *et al.*, 2010). Our findings suggest that using this phase angle feedback, athletes were able to reduce the amount of asynchrony in the breathing pattern between the ribcage and abdomen, therefore, reverting towards a healthy breathing pattern.

Similar to RcAbPhase, the phase angle between the upper ribcage and abdomen (RCpAbPhase) demonstrated phase angle values significantly closer to zero in the

experimental group post intervention during all of the exercise intensities (Tables 7.4 – 7.6) and post-exercise recovery (Table 7.7). This indicates that participants were able to reduce the asynchrony between these two compartments with the abdomen initiating the movement during high intensity exercise. This is expected as RCpAbPhase and RcAbPhase are similar except RCpAbPhase excludes the lower ribcage (RCa). Since, the upper ribcage (RCp) contributes more to the ribcage than the lower ribcage, the upper ribcage has more of an influence on ribcage (RCp + RCa) movement, therefore, we would expect to see a similar response in both phase angles. OEP derived RCpAbPhase has previous been reported to increase during flow and volume incentive spirometry from tidal breathing, although not significantly, in healthy adults (de Moraes Paisani *et al.,* 2013).

Continuing this trend, the phase angle between the shoulders and abdomen (AbSPhase) displayed significant changes post intervention in the experimental group, with values closer to zero during exercise (Tables 7.4 – 7.6) and recovery (Table 7.7). During rest, the shoulders moved before the abdomen pre-intervention, however, post-intervention the abdomen moved prior to the shoulders (Table 7). During high intensity exercise, the abdomen moved first for both pre and post intervention conditions (Table 7.6). This phase angle is novel to this thesis therefore, we cannot make comparisons with previous literature. This phase angle was developed in an attempt to quantify thoracic dominant, also referred to as apical breathing. Thoracic dominant breathing has been reported as a sub-type within dysfunctional breathing with the upper thorax primarily used during a breath with minimal lateral movement of the ribcage (Boulding *et al.,* 2016). Individuals with thoracic dominant breathing score highly on the Nijmegen questionnaire and may be exacerbated with increased ventilation demand such as during exercise (Boulding *et al.,* 2016). Previous research has used breathing exercises such as diaphragmatic breathing to successfully improve thoracic dominant breathing in COPD patients (Gosselink, 2004). The reduction in AbSPhase values post-intervention and the switch from shoulders to abdomen as the leading compartment in this study suggests that the athletes were able to use the real-time phase angle feedback to alter their breathing pattern with more abdominal movement and less thoracic excursion.

All other phase angles assessed in this study did not change significantly in the experimental group post-intervention when compared to the control group (Tables 7.1 – 7.5). The non-significant phase angles (RCpRCaPhase, RCaSPhase, and RCpSPhase) do not include the abdomen compartment, while the three phase angles that displayed significant changes do.

This suggests that the participants used the real-time phase angle feedback to alter the movement of their thoracic compartments and reducing the abdomen asynchrony with the shoulders and ribcage. Interestingly, the regional contributions did not differ post-intervention, this indicates that using real-time OEP phase angle feedback alters the timing of the movement of the thoracic compartments but does not alter the contribution of each compartment to the total thorax volume.

Inhale and exhale percentage deviations are novel measures of the deviation of the phase angle trace from the 'perfect' straight 45 degree diagonal line of the breath loop during inhale and exhale. Inhale percentage deviation was reduced post intervention during exercise and recovery for the experimental group when compared to pre intervention. While the control group displayed an increase in percentage deviation during all conditions. However, inhale percentage deviation was only significant post-intervention during low intensity when comparing the groups (Table 7.4). Both groups displayed a reduction in exhale percentage deviation during exercise. In Chapter 6, the dysfunctional breathing pattern group displayed significantly greater inhale percentage deviation values when compared to the healthy breathing pattern group during rest and exercise. The findings of this study indicate that during the inhale, the experimental group were able to reduce the deviation of their Konno-Mead/breath loop to become closer to the straight diagonal line that would be expected in a healthy breathing pattern. This is in line with the reduction in RcAbPhase in this study indicating athletes where able to use the phase angle feedback to reduce the asynchrony between the ribcage and the abdomen in real-time.

Although significant reductions in three of the phase angles and deviation parameters were seen in the experimental group post-intervention, these improvements did not alter the overall breathing pattern coordination of the experimental group during high intensity exercise. Both the phase angle and control group displayed a breathing pattern which consisted of the abdomen initiating the breath, followed by the shoulders, the upper ribcage (RCp), and finally the lower ribcage (RCa) pre intervention during high intensity exercise. This order of compartment movement did not change during high intensity exercise post intervention for both groups. In the previous chapters of this thesis, it was discussed that individuals with a healthy breathing pattern initiate their breath from the lower ribcage (RCa), while athletes with a dysfunctional breathing pattern display a delay in this compartment movement. This phase angle feedback system reduced the asynchrony in the experimental group's breathing pattern but did not alter the overall breathing pattern

coordination. Therefore, the athletes in this study still displayed a dysfunctional breathing pattern post-intervention. This is most likely due to the fact that this was an acute intervention with a small sample size. It would not be expected that any individual would be able to completely alter their breathing pattern from a dysfunctional to a healthy breathing pattern in one session particularly during exercise. Previous interventions which demonstrated improvements from breathing training/exercises have spanned from a few weeks (Courtney *et al.,* 2019) to a 5 year follow up (Hagman *et al.,* 2011). Although this study establishes OEP real-time phase angle feedback as a promising breathing training tool, further investigation is required to assess if this system would elicit chronic improvements in breathing pattern if used longitudinally in a larger sample of individuals with dysfunctional breathing. This would not only allow breathing pattern changes to be tracked but also quality of life and symptom management.

Traditional breathing pattern retraining methods, including Buteyko breathing (Bowler *et al.,* 1998), diaphragmatic breathing (Hagman *et al.,* 2011), Papworth method (Holloway & West, 2007), and yoga breathing (Singh *et al.,* 1990), focus on the inhalation phase of the breath cycle. Additionally, feedback mechanisms such as inspiratory muscle training devices (Nepomuceno *et al.,* 2016) also focus on the inhalation phase. Breathing is dual cycle and as such future work should include the exhalation phase as part of breathing pattern retraining. The novel feedback system presented in this chapter offers the opportunity to do so as the phase angle plot includes both the inhale and exhale phase of the breath. Additionally, the newly developed exhale percentage deviation parameter provides an objective and quantifiable measure specific to exhale phase of the breath.

In conclusion, this study demonstrates that real-time OEP phase angle feedback results in altered values for the following parameters that are closer to those seen in healthy adults RcAbPhase, RCpAbPhase, AbSPhase, and inhale percentage deviation values during exercise in athletes with dysfunctional breathing post-intervention. However, this improvement did not change the regional contribution parameters or the overall breathing pattern coordination of athletes with dysfunctional breathing during high intensity exercise. This study provided some interesting insight into using real-time phase angle feedback, in order to inform breathing pattern retraining in a clinical practice, future work should include larger sample size and longitudinal interventions.

Chapter 8. OEP phase angle differences in simulated dysfunctional breathing patterns using the POWERbreathe device

8.1 Introduction

A normal breathing pattern can be described within the literature as the ribcage and the abdomen moving in synchronisation with each other (Bradley & Esformes, 2014), this is further supported by the findings in Chapter 4. Individuals with dysfunctional breathing differ from this normal pattern with chronic breathing pattern changes which may be classified into various subtypes such as hyperventilation, thoracic dominant/apical breathing, and thoracoabdominal asynchrony (Boulding *et al.,* 2016). Thoracic dominant/apical breathing can be described when there is predominant use of the upper thorax with minimal lateral costal expansion or abdomen movement. Thoracoabdominal asynchrony is a delay between the movement of the abdomen and ribcage which may result in paradoxical breathing (Boulding *et al.,* 2016). Previously in Chapter 6, OEP derived phase angle parameters were found to be significantly different between the dysfunctional and normal breathing patterns during high intensity exercise. The primary treatment for dysfunctional breathing is breathing retraining through breathing exercises (Bott, 2009). In Chapter 7, an acute intervention using real-time OEP phase angle feedback improved phase angle values, with the values returning towards a phase angle of zero i.e., perfect synchrony, for individuals with dysfunctional breathing during high intensity exercise.

Respiratory muscle training aims to improve the strength of respiratory muscles (Lorca-Santiago *et al.,* 2020). A form of respiratory muscle training is inspiratory muscle (IMT) training which refers to just the inspiratory phase of the breath. Previously, there has been some evidence which suggests that IMT may be beneficial for athletes with improvements in maximal inspiratory pressure (Nepomuceno *et al.,* 2016) and performance in some sports such as cycling (Johnson *et al.,* 2007) and rowing performance (Griffiths & McConnell, 2007). Similarly, some promising benefits, such as reduced symptoms and improved quality of life scores, for individuals with chronic obstructive pulmonary disease (Geddes *et al.,* 2008), exercise-induced laryngeal obstruction (Sandnes *et al.,* 2018), asthma (Turner *et al.,* 2011), inspiratory stridor (Dickinson *et al.,* 2007), adults on mechanical ventilation (Moodie *et al.,* 2011), and haemodialysis patients (Medeiros *et al.,* 2019) have been demonstrated

with respiratory muscle training when used with traditional treatment methods. However, there is no consensus on the use of respiratory muscle training for these population groups (Silva *et al.,* 2013) and further evidence is required to establish its use within clinical practice.

One inspiratory muscle training device on the market is the POWERbreathe breathing trainer. This device is a linear load resistor model which creates resistance through a spring-loaded or electronic valve system (Nepomuceno *et al.*, 2016). The POWERbreathe device has a number of uses as it is simple to use and portable. Previously, POWERbreathe has been used to deliver IMT to individuals with COPD (Beckerman *et al.,* 2005), asthmatics (Turner *et al.,* 2011), and athletes (Volianitis *et al.,* 2001), with respiratory muscle strength measured by maximal inspiratory pressure showing improvements. It has also been used for inspiratory muscle endurance tests for patients with COPD (Dias *et al.,* 2013) and post-myocardial infarction (Neves *et al.,* 2012; 2014). OEP has assessed the impact of inspiratory loads via the POWERbreathe device on OEP timing, volume, compartment contribution breathing parameters in healthy individuals (Da Gama *et al.,* 2013; Fonseca *et al.,* 2019). The addition of inspiratory load of 20% maximal inspiratory pressure was found to significantly increase tidal volumes when compared to rest and recovery (Fonseca *et al.,* 2019) however, tidal volume did not significantly differ from 20% maximal inspiratory pressure to 30% (Nobre *et al.,* 2007) or 40% maximal inspiratory pressure (Fonseca *et al.,* 2019).

Dysfunctional breathing may result in muscular imbalances (Bradley & Esformes, 2014) which result in chronic changes in breathing pattern typically characterised into thoracoabdominal asynchrony or thoracic dominant (Boulding *et al.,* 2016). The addition of inspiratory load has been shown to significantly increase the activity of sternocleidomastoid and scalene muscles in healthy individuals (Fonseca *et al.,* 2019). IMT may be a potential breathing retraining aid for individuals with dysfunctional breathing. IMT promotes increases in muscle activity including the sternocleidomastoid, intercostals, and the diaphragm (Walterspacher *et al.,* 2018). Inspiratory pressure threshold loading, via the POWERbreathe device, has been shown to result in the highest muscle activation of the diaphragm when compared to other respiratory muscle training modalities such as inspiratory flow resistive loading and voluntary isocapnic hyperpnea (Walterspacher *et al.,* 2018). Typically, the POWERbreathe device is used as part of an IMT intervention with other measures such as maximal inspiratory pressure used to monitor the impact of the IMT

through measuring improvements in inspiratory muscle strength (ATS/ERS, 2002; Laveneziana *et al.*, 2019). However, the POWERbreathe device also has the capability of providing real-time breathing parameter output traces such as volume, pressure, and power. To our knowledge there have not been any studies which assess the POWERbreathe real-time output parameters under different conditions or as potential respiratory muscle training aids.

Therefore, purpose of this study was twofold: firstly, to investigate if OEP-derived phase angles increase as expected with simulated dysfunctional breathing patterns (thoracic dominant/apical and thoracoabdominal asynchrony) when compared to a normal breathing pattern, and secondly, to determine if the output traces of the POWERbreathe device differ significantly between these three breathing pattern conditions.

8.2 Methods

Participants

Sixteen healthy active males (n = 9; age 24.7 ± 3.0 years; height 1.8 ± 0.1 m; body mass 66.6 ± 11.4 kg) and females (n = 7; age 23.4 ± 3.6 years; height 1.7 ± 0.1 m; body mass 59.1 ± 7.7 kg) gave written informed consent to participate in this study, which was approved by the School of Sport & Exercise Sciences Research Ethics Advisory Group at the University of Kent (Prop 92_2018_2019). Participants were required to have a consistent training regime with a minimum of 5 hours of physical activity per week and no history of chronic respiratory diseases/disorders.

Protocol

Each participant performed three different types of breathing patterns using the POWERbreathe device (KH2, POWERbreathe International Ltd., Southam, UK) with OEP data simultaneously collected (see Chapter 3 for full OEP set-up details). The three breathing patterns were: 1) normal breathing pattern (NBP), 2) apical/thoracic dominant breathing pattern (TBP) with the breath generated from upwards shoulder and upper chest movement with minimal abdominal movement, and 3) asynchronous breathing pattern (ABP) with the ribcage and abdomen moving asynchronously initiated by the abdomen. Participants were given instructions on how to achieve the simulated breathing patterns as described above and performed familiarisation trials until comfortable with the patterns. Instructions for performing the NBP included moving the ribcage and abdomen together, and focusing on the diaphragm and intercostal muscles to move the lower ribcage laterally. Instructions to perform the TBP included initiating the breath using the shoulders, and avoid moving the abdomen. Instructions to perform the ABP included initiating the breath using the abdomen followed closely by the ribcage to achieve a ‘double inhale’. The POWERbreathe device required participants to maximally inhale as quickly and deeply as possible and exhale in a slow and controlled manner. Each participant then performed each pattern in a randomised order unloaded (3 cmH_2O) with the POWERbreathe device, with periods of rest between each pattern. Each trial consisted of 15 breaths with the last 10 breaths used for analysis.

Data Analysis

Timing, compartment contribution and phase angle parameters were calculated from the OEP volume trace (see Table 3.1 for full list and description) using a custom-built MATLAB script. Additionally, the percentage deviation of the inhale (inhale % deviation) and exhale (exhale % deviation) were obtained from the RcAbPhase Konno-Mead loops (Chapter 6).

The POWERbreathe device outputs, via Breathe-Link Medic software (POWERbreathe, K-Series, version 3.3.2a), a number of variables including mouth inspiratory pressure, inspiratory power, and volume for each breath. These variables were outputted as a CSV file and imported into MATLAB. Each variable was time normalised (0-100% of inhale) and normalised between 0 and 1 based on minimum and maximum values. If a participant was unsuccessful in performing the simulated breathing patterns, which was determined through the visual inspection of the Konno-Mead breath loops obtained from the OEP data, that breath was excluded. All parameters obtain from the OEP system and the POWERbreathe system were averaged across 10 breaths for each participant.

Statistical Analysis

Using SPSS (IBM SPSS Statistics 25), Pearson's correlation coefficients and intraclass correlation coefficients were calculated between volume obtained from OEP and POWERbreathe systems for each simulated breathing pattern.

A repeated measures ANOVA was used to analyse the differences between the simulated breathing patterns for each of the OEP parameters.

Statistical parametric mapping (Pataky, 2010) was used to statistically compare the POWERbreathe outputs between the simulated breathing patterns. All statistical parametric mapping (SPM) analysis was conducted using MATLAB code adapted from open-source spm1d package (v.M0.1, www.spm1d.org). A one-way repeated measured ANOVA was used to compare the three normalised simulated breathing patterns ($\alpha = 0.05$). Prior to SPM analysis, normality tests were conducted to determine if parametric or non-parametric testing was required. The scalar output statistic (SPM{t} or SnPM{t}) was calculated at each time point in the normalised time-series. A critical threshold was calculated so that only α % (5%) of smooth random curves would be expected to transverse based on Random Field Theory (Adler & Taylor, 2007). If the SPM{t} or SnPM{t} values surpassed this threshold, then the POWERbreathe output variable is considered significantly different at that specific percentage of inhale. Multiple adjacent SPM output points often surpass the threshold due

to inter-dependence of neighbouring points and waveform smoothness, which are identified as ‘supra-threshold clusters. Cluster specific p-values are obtained using Random Field Theory (Adler &Taylor, 2007). Post-hoc tests were performed if significance was found, with a Bonferroni correction applied to α. This process was conducted for each POWERbreathe output variable.

8.3 Results

The mean volume inhaled obtained from both measurement systems significantly correlated for each simulated breathing pattern (Table 8.1).

Table 8.1 Mean and standard deviation (SD) of OEP and POWERbreathe volume for each simulated breathing pattern with Pearson's correlation coefficient (r) and intraclass correlation coefficients (ICC) comparing the two measurement systems. * represent significance of $p \leq 0.05$

	OEP Volume (litres)		**POWERbreathe Volume (litres)**		**Correlation (r)**	**ICC**
	Mean	**SD**	**Mean**	**SD**		
Normal Breathing Pattern	2.84	1.16	2.10	0.95	0.863*	0.943*
Thoracic Dominant Breathing Pattern	2.75	1.08	1.96	0.86	0.904*	0.751*
Asynchronous Breathing Pattern	2.90	1.07	1.95	0.84	0.918*	0.800*

Optoelectronic Plethysmography Data

Table 8.2 displays the mean and standard deviation of the OEP-derived phase angles from each simulated breathing pattern. The shoulder breathing pattern displayed greater phase angle values for each parameter compared to the normal breathing pattern, with significant differences for RCpRCaPhase, RCaSPhase, and RCpSPhase. The asynchronous breathing pattern demonstrated greater RcAbPhase, RCpAbPhase, RCpRCaPhase, and RCaSPhase values compared to the normal breathing pattern, with the differences being significant for RcAbPhase. Asynchronous breathing pattern also demonstrated significant differences compared to the shoulder breathing pattern for RcAbPhase, RCpRCaPhase, and AbSPhase.

Both the shoulder and asynchronous breathing pattern demonstrated greater inhale % deviation compared to the normal breathing pattern, although not significant. The asynchronous breathing pattern displayed the lowest exhale % deviation values when compared to the other two breathing patterns.

Table 8.2 Mean and standard deviation (SD) for the OEP derived-phase angle breathing parameters for each simulated breathing patterns. Significance represented by N*, T*, and A* as a mean significant ($p \leq 0.05$) difference from normal, thoracic dominant, and asynchronous breathing patterns, respectively.

Parameters	**Normal Breathing Pattern**			**Thoracic Dominant Breathing Pattern**			**Asynchronous Breathing Pattern**		
	Mean	**SD**		**Mean**	**SD**		**Mean**	**SD**	
RcAbPhase (deg)	0.81	1.55	A*	1.09	2.00	A*	-1.46	1.70	N*, T*
RCpAbPhase (deg)	0.73	1.34		0.76	1.78		0.76	1.70	
RCpRCaPhase (deg)	0.28	0.79	T*	-1.84	3.58	N*, A*	0.40	1.44	T*
RCaSPhase (deg)	0.36	1.68	T*	-3.07	3.78	N*	-0.75	2.99	
RCpSPhase (deg)	0.48	1.20	T*	-1.74	2.60	N*	-0.26	1.82	
AbSPhase (deg)	0.82	1.88		2.38	1.50	A*	-0.30	2.57	T*
Inhale % Deviation	-6.76	22.13		-32.32	39.82		-25.80	62.96	
Exhale % Deviation	1.85	13.89		-1.42	18.03		-0.29	15.31	

POWERbreathe Data

The one-way repeated-measures ANOVA demonstrated that both the normalised volume ($F^* = 4.471$, $p = 0.002$) and the normalised cumulative volume ($F^* = 5.004$, $p = 0.004$) were significantly different between the simulated breathing patterns, with the normal breathing pattern displaying the largest values. Post-Hoc tests (Figure 8.1) revealed that for both variables there was a significant difference between the normal and asynchronous breathing patterns only. The asynchronous breathing pattern displayed significantly lower volume (Figure 8.1b) and cumulative volume (Figure 8.1d) at ~10 – 60% and ~10 – 100% of the inhale phase, respectively, when compared to the normal breathing pattern.

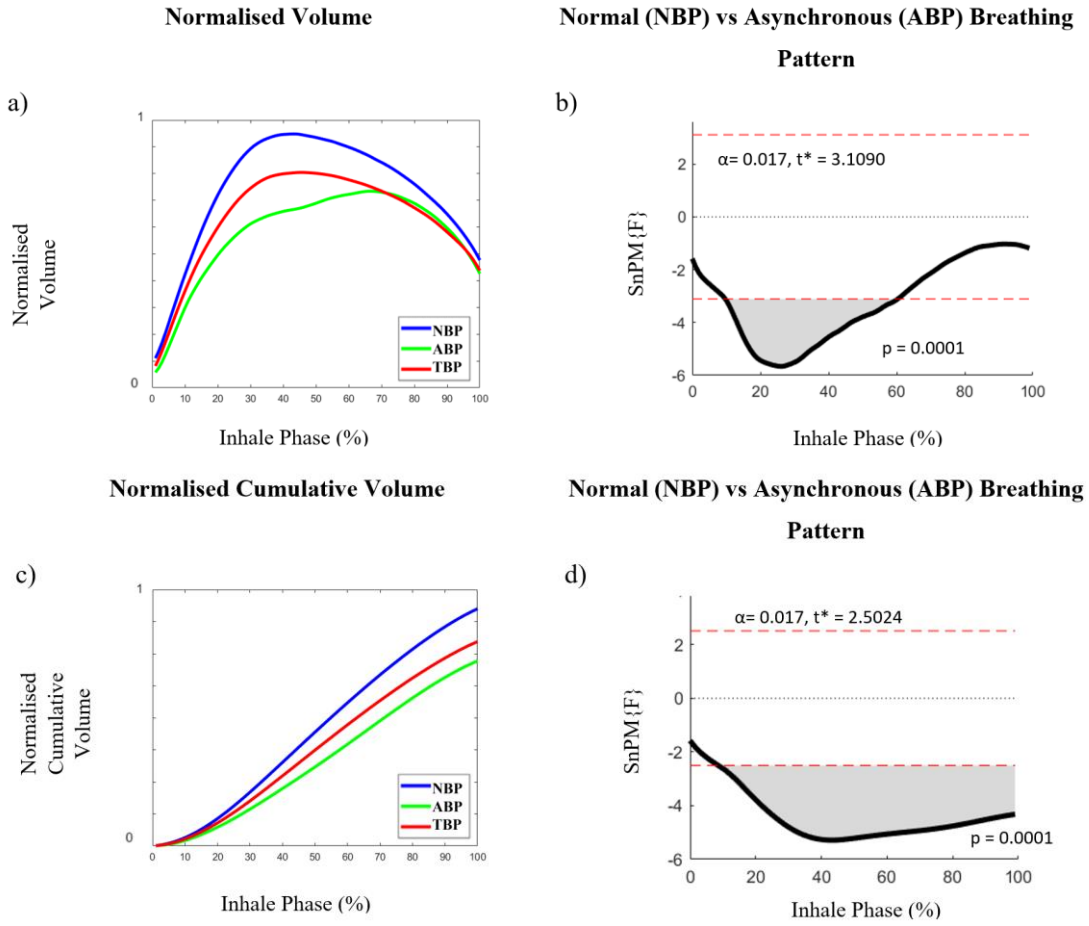


Figure 8.1 Mean normalised volume (a) and cumulative volume (c) for each breathing pattern of the inhale phase from the POWERbreathe device. The SnPM{t} (post-hoc tests) statistic with critical threshold (t*) of 3.1090 and 2.5024 for volume (b) and cumulative volume (d), respectively.

Normalised pressure was significantly different ($F^* = 6.068$, $p = 0.019$) between the simulated breathing patterns (Figure 8.2b). Post-Hoc tests revealed that pressure was a significant difference between the shoulder and asynchronous breathing patterns only. The asynchronous breathing pattern displayed significantly greater pressure (Figure 8.2h) at ~50 – 75% of the inhale phase when compared to the shoulder breathing pattern.

The normalised power output was significantly different ($F^* = 5.927$, $p = 0.007$) between the simulated breathing patterns (Figure 8.3b). Power was significantly different between the normal and asynchronous breathing patterns (Figure 8.3f) at ~18 – 25 % of the inhale phase, with the asynchronous breathing pattern displaying lower power values during this phase. Power was also significantly different between the thoracic dominant and asynchronous breathing patterns (Figure 8.3h) at ~60 – 80% of the inhale phase, with the asynchronous breathing pattern displaying higher power values during this inhale phase.

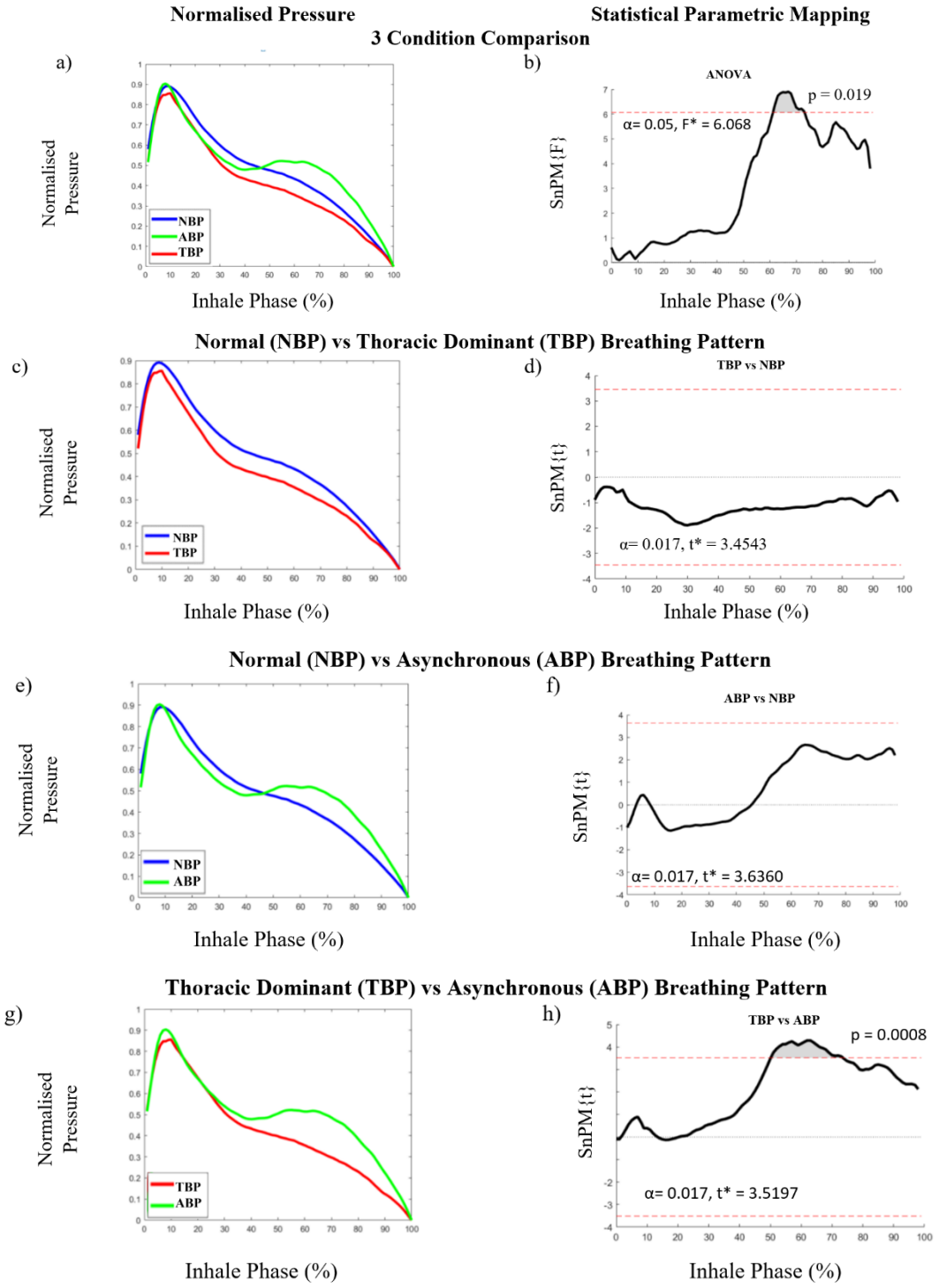


Figure 8.2 Mean normalised pressure for each breathing pattern (a, c, e, g) of the inhale phase from the POWERbreathe device. b) illustrates the SnPM{t} (one-way repeated measured ANOVA) statistic with critical threshold (F*) of 6.068. SnPM{t} of post-hoc tests with their respective critical thresholds (t*) illustrated between which breathing patterns the significant differences occurred (d, f, h).

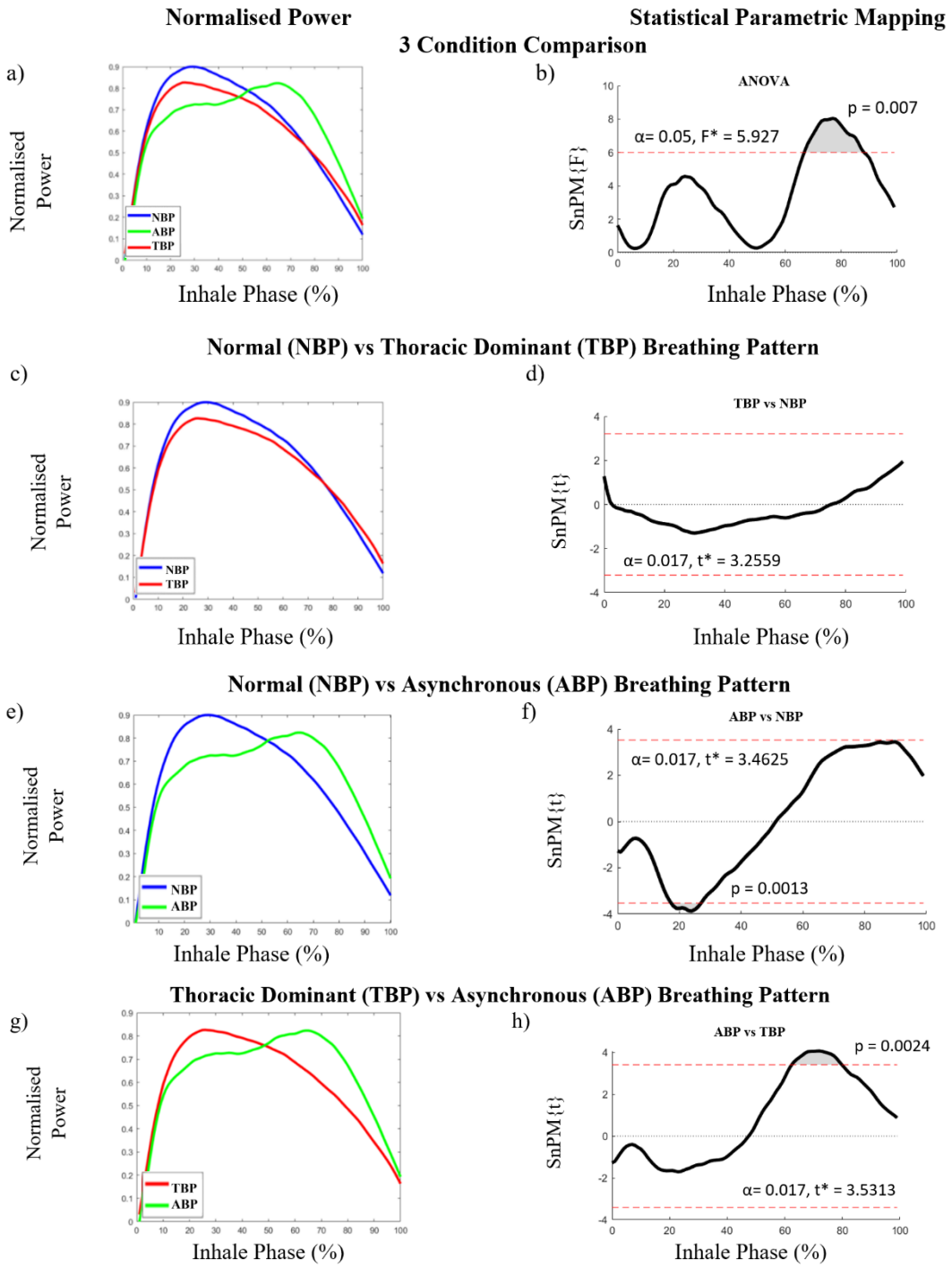


Figure 8.3 Mean normalised power for each breathing pattern (a, c, e, g) of the inhale phase from the POWERbreathe device. b) illustrates the SnPM{t} (one-way repeated measured ANOVA) statistic with critical threshold (F*) of 5.927. SnPM{t} of post-hoc tests with their respective critical thresholds (t*) illustrated between which breathing patterns the significant differences occurred (d, f, h).

8.4 Discussion

The findings of this study firstly indicate that the OEP-derived phase angle parameters RcAbPhase, RCpRCaPhase, RCaSPhase, RCpSPhase, and AbSPhase alter as expected between the simulated breathing patterns (Chapter 6). Secondly, for the first time, this study demonstrates the output traces of the POWERbreathe device differ significantly between simulated breathing pattern conditions. The POWERbreathe outputs volume and cumulative volume demonstrated significantly lower values in the asynchronous breathing pattern compared to the normal breathing pattern. The POWERbreathe pressure trace for the asynchronous breathing pattern was significantly different to the thoracic dominant pressure trace during the second half of the inhale. Finally, the POWERbreathe power trace for the asynchronous breathing pattern was significantly different to both the normal and thoracic dominant breathing patterns during the first and second half of the inhale, respectively. This identifies the POWERbreathe device as potential method of distinguishing between normal and dysfunctional breathing patterns.

The simulated breathing patterns were used to assess if OEP-derived phase angles altered as expected with asynchronous and thoracic dominant breathing patterns. The phase angle between the abdomen and ribcage (RcAbPhase) was significantly greater for the ABP when compared to both the normal and thoracic dominant breathing pattern (Table 8.2). The abdomen moved first for the asynchronous breathing pattern, while the ribcage moved first for both the normal and thoracic dominant breathing pattern. A larger phase angle represents more thoracoabdominal asynchrony, therefore, it would be expected the simulated asynchronous breathing pattern would display the largest RcAbPhase value. Thoracoabdominal asynchrony has previously been used to identify dysfunctional breathing in asthmatics (Pyris-Picard *et al.,* 2004) and patients with COPD (Motamedi-Fakhr *et al.,* 2016) using respiratory inductive plethysmography and structured light plethysmography. Similarly, in Chapter 6, it was established that individuals with dysfunctional breathing exhibit significantly larger RcAbPhase values during high intensity exercise when compared to a healthy population using OEP. Using simulated breathing patterns, this study further validates OEP as a measurement system capable of identifying thoracoabdominal asynchrony, one of the characteristics of dysfunctional breathing.

The phase angles between the shoulders and the abdominal ribcage (RCaSPhase), and the pulmonary ribcage (RCpSPhase) values were both significantly greater in the thoracic dominant breathing pattern when compared to the normal breathing pattern (Table 8.2). Similarly, the phase angle between the shoulders and the abdomen (AbSPhase) was significantly greater in the thoracic dominant breathing pattern when compared to the asynchronous breathing pattern (Table 8.2). The thoracic dominant breathing pattern displayed the largest phase angle values for each of the shoulder phase angles. This is an expected response as thoracic dominant/apical breathing is generated from the respiratory accessory muscles including the upper trapezius, the sternocleidomastoid, and scalene muscles with minimal abdomen movement and poor use of the diaphragm (Bradley & Esformes, 2014), resulting in an increased phase angle. Previously in this thesis, RCpSPhase, RCaSPhase, and AbSPhase were found to be parameters which distinguished dysfunctional breathing pattern from normal and exercise-induced bronchoconstriction breathing patterns (Chapter 6). The asynchronous breathing pattern was the only breathing pattern to display movement of the abdomen before the shoulders (AbSPhase). This indicates that the simulated breathing patterns were performed correctly as the simulated asynchronous breathing pattern aimed to move the abdomen asynchronously with the ribcage with the abdomen as the leading compartment.

The inhale and exhale percentage deviation parameters measure the deviation of the phase angle trace from the 'perfect' synchronous breath loop using OEP. The differences between each of the breathing patterns were non-significant for both deviation parameters (Table 8.2). The normal breathing pattern displayed the lowest inhale % deviation value (-6.8%) when compared to both the thoracic dominant (-32.3%) and the asynchronous (-25.8%) breathing patterns. This is in line with RcAbPhase as the normal breathing pattern displayed the lowest phase angle value. These deviation parameters give greater insight into where in the breath cycle asynchrony occurs. As anticipated, the thoracic and asynchronous breathing patterns displayed larger deviation values during the inhalation. In contrast, the exhale % deviation values ranged between -1.4 to 1.9% across the three breathing patterns. In Chapter 6, individuals with a dysfunctional breathing pattern displayed a significant increase in exhale % deviation during high intensity exercise when compared to both their resting state and individuals with a normal breathing pattern. Therefore, we would expect dysfunctional breathing patterns to elicit an increase in exhale % deviation. However, the participants in this study were not asked to alter their breathing patterns during the exhale. Participants were instructed to exhale with a slow and controlled breath as per the POWERbreathe device

instructions. Although dysfunctional breathing patterns may increase exhale % deviation, it was not anticipated that this parameter would change with these particular simulated breathing patterns. Overall, OEP parameters altered as anticipated for both simulated breathing patterns. This provides further validation for the use of OEP as a potential diagnostic tool for dysfunctional breathing.

The inhale volumes obtained from the both the OEP system and POWERbreathe device significantly correlated for each given simulated breathing pattern (Table 8.1). Although there is a correlation in the volumes, the POWERbreathe device seems to underestimate the volume with a mean difference of 0.83 litres across all breathing patterns when compared to the OEP system. The larger OEP volumes may be due to the fact that OEP calculates the total volume of the torso while the POWERbreathe records the inhaled volume. OEP has previously been shown to slightly overestimate tidal volume when compared to spirometry (Massaroni *et al.,* 2017a). The simulated breathing patterns were also used to assess if different dysfunctional breathing patterns would elicit changes in the POWERbreathe device outputs, potentially as an alternative to OEP for identifying dysfunctional breathing patterns. For both the volume and cumulative volume, asynchronous breathing pattern was significantly different from the normal breathing pattern (Figure 8.1). The asynchronous breathing pattern produced the lowest volumes with the normal breathing pattern producing the largest volumes (Table 8.1). The significance occurred after the first 10% of the inhale. Dysfunctional breathing patterns arise from the poor recruitment pattern of the respiratory muscles, predominately the inspiratory muscles and the diaphragm, which cause alterations in the breathing mechanics (Pereira *et al.,* 2017). The incorrect use of the respiratory muscles in the simulated dysfunctional breathing patterns limited the volume inhaled in a single breath. While the normal breathing pattern with the respiratory muscles and diaphragm functioning most effectively, allowed for the largest volume to be inhaled (Figure 8.1).

The POWERbreathe pressure output was significantly different between the asynchronous and thoracic dominant breathing patterns (Figure 8.2h). The asynchronous breathing pattern displayed a second distinct peak unlike the other two breathing patterns (Figure 8.2a). The significant difference between asynchronous and thoracic dominant breathing patterns occurs during this second peak at approximately 50 to 75% of the inhale. For the asynchronous breathing pattern, the initial movement comes from abdomen which is then followed by the ribcage. This distinct asynchrony between the two compartments is likely to have caused the double peak in the pressure output. In contrast, the other breathing

patterns do not involve alternating compartment movements with the normal breathing pattern, abdomen and ribcage are moving together and with the thoracic dominant breathing pattern, abdomen movement is minimal with most motion occurring in the upper thorax. This demonstrates that the use of different thoracic compartments during an inhale, alter the pressure output of the POWERbreathe device.

The POWERbreathe power output was significantly different between the asynchronous breathing pattern and both the normal and thoracic dominant breathing patterns but at different stages of the inhale (Figure 8.3). Both the normal and the thoracic dominant breathing pattern peaked during the first half of the inhale, with the normal breathing pattern generating higher power values (Figure 8.3c). In contrast, the asynchronous breathing pattern peaks in the second half of the breath, generating lower peak power values when compared to the normal breathing pattern. This delayed peak in the asynchronous breathing pattern caused a significant difference between the asynchronous and normal breathing patterns during the first half of the inhale (~18 – 25%) when the normal breathing pattern peak occurs (Figure 8.3f). When comparing the asynchronous and thoracic dominant breathing patterns, the significant difference occurs in the second half of the inhale (~60 – 80%), when the asynchronous breathing patten peak occurs (Figure 8.3h). The ribcage compartment is larger than the abdomen, therefore, it would be expected that ribcage compartment movement may generate larger power values compared to the abdomen in isolation. Therefore, this delayed peak in the asynchronous breathing pattern is likely to be caused by the delayed movement of the ribcage.

Overall, the POWERbreathe outputs seem to be able to identify aspects of the simulated asynchronous breathing pattern, however, there were no significant differences between the normal and the thoracic dominant breathing patterns. Although the participants were able to successfully perform the simulated breathing patterns correctly, which is evident by the OEP-derived phase angles which altered as expected. In order to determine if the POWERbreathe device could be a potential tool to identify dysfunctional breathing patterns, it should be assessed with individuals diagnosed with dysfunctional breathing and not just simulated dysfunctional breathing patterns. This preliminary data begins to establish normal and dysfunctional breathing pattern POWERbreathe output traces. However, a more substantive data set would be required to establish clear trace characteristics and cut-off criteria of dysfunctional breathing patterns. Following this, the POWERbreathe outputs, may potentially be a useful breathing retraining tool for individuals with dysfunctional breathing.

In conclusion, this study further validates OEP-derived phase angle parameters, including RcAbPhase, RCpRCaPhase, RCaSPhase, RCpSPhase, and AbSPhase, as each responded as anticipated to the altered mechanics of the simulated dysfunctional breathing patterns. This study for the first time, evaluates POWERbreathe outputs as potential measures to identify differences between different simulated breathing patterns. The outputs including volume, cumulative volume, pressure, and power were significantly different for the asynchronous breathing pattern when compared to the normal and thoracic dominant breathing patterns. Our findings suggest that the POWERbreathe output demonstrates differences between normal and simulated breathing patterns, providing a basis for further investigation into the POWERbreathe device as a potential portable diagnostic and breathing retraining device.

Chapter 9. Comparison of OEP and impedance pneumography as respiratory measurement techniques

9.1 Introduction

The previous chapters of this thesis have evaluated the capabilities of optoelectronic plethysmography (OEP). It has been demonstrated as a useful breathing pattern measurement technique in assessing a variety of populations including exercise-induced bronchoconstriction and dysfunctional breathing, as well as a potential training tool for individuals with these conditions. However, OEP has several limitations including being confined to the laboratory setting, it is expensive, requires specialised expertise, and is time consuming.

Wearable technology is becoming increasingly popular in recent years in both healthcare and sporting applications. There have been a number of wearable systems developed to allow the monitoring of breathing function continuously overtime in any/natural conditions and providing personalised real-time feedback (Aliverti, 2017). The ideal wearable system for breathing pattern measurement would provide similar breathing pattern parameters as OEP while overcoming the laboratory-based limitations of OEP. The development of such a system would objectively support clinicians in the diagnosis of respiratory diseases/disorders such as dysfunctional breathing and allow athletes to monitor and train their breathing pattern overtime. Some wearable technology systems have had some success in measuring respiratory rate and tidal volume, such as photoplethysmography (Meredith *et al.,* 2012), accelerometers (Fekr *et al.,* 2014), strain gauges (Al-Halhouli *et al.,* 2020), respiratory inductive plethysmography (Wu *et al.,* 2009) and impedance pneumography (Mlynczak & Cybulski, 2012) to name a few.

Impedance pneumography (IP) is a non-invasive method of monitoring respiratory function. It measures changes in thoracic electrical impedance through two or more skin electrodes. It involves passing a low-amplitude high frequency electrical current into the torso and recording the voltage gradient between the electrodes. IP is a system which allows for the portable, non-invasive measurement of respiration (Ansari *et al.,* 2017). Recent developments in sensor configurations and signal processing have allowed IP to be validated against pneumotachography i.e. direct breath-by-breath (BbB) analysis (Seppa *et al.,* 2016). Previous studies have used IP to measure respiratory rate and tidal volume in healthy adults

in a variety of conditions such as different body postures (Mlynczak *et al.,* 2015) and walking (Jeyhani *et al.,* 2016). Currently, IP is mostly used in a clinical setting (Seppa *et al.,* 2016) to measure respiratory rate in patients across a wide range of conditions including asthma (Seppa *et al.,* 2013c), sleep disorders (Poupard *et al.,* 2012; Rotariu *et al.,* 2014), chronic obstructive pulmonary disease (Sim *et al.,* 2013), and breathing disorders (Seppa *et al.,* 2013c). It has also been used on populations which other respiratory systems may have difficulty in monitoring such as infants (Malmberg *et al.,* 2016).

Although IP is widely used as a method of measuring respiratory rate, unlike other non-invasive methods, changes in IP signal is predominately caused by changes in lung tissue conductivity during breathing rather than solely from the movement of the chest wall (Seppa *et al.,* 2013a). This mechanism may mean that IP signal may be distorted by abnormal/dysfunctional breathing patterns resulting in a reduction in IP signal accuracy. However, Seppa and colleagues (2013a) used expiratory loading to induce alterations in breathing pattern and demonstrated that IP was capable of accurately measuring tidal breathing flow-volume curves despite the changes in pattern when compared to BbB analysis. IP has also been used to track methacholine-induced bronchoconstriction in children with breathing difficulties, demonstrating agreement with direct BbB analysis (Seppa *et al.,* 2013c).

Therefore, the purpose of this study was to compare the impedance pneumography system to the OEP system and traditional breath-by-breath analysis, to assess the agreement between these systems in terms of respiratory rate and tidal volume at rest and during exercise in a non-pathological population.

9.2 Methods

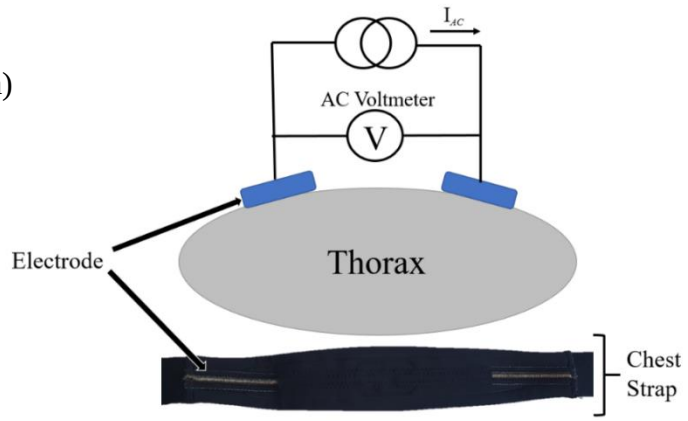
Participants

Thirty-one physically active healthy males (n = 19; age 31.7 ± 7.5 years; height 1.8 ± 0.1 m; body mass 77.7 ± 11.2 kg) and females (n = 12; age 26.6 ± 7.4 years; height 1.7 ± 0.1 m; body mass 61.1 ± 5.9 kg) gave written informed consent to participate in this study and performed a sub-maximal incremental cycling exercise challenge. Participants were required to have a consistent training regime with a minimum of 5 hours of physical activity per week and no history of chronic respiratory diseases/disorders. This study was approved by the School of Sport & Exercise Sciences Research Ethics Advisory Group at the University of Kent (Prop 44_2017_18).

Equipment

Respiratory data was collected using three measurement methods simultaneously, namely OEP, IP, and a BbB analyser (CORTEX Biophysik MetaLyzer 3B-R2 stationary CPX system). The OEP data was collected following the procedure outlined in Chapter 3.

a)



b)



Figure 9.1 a) Illustration of electrode arrangement for impedance pneumography and b) positioning of the impedance pneumography chest strap with the OEP marker set.

The IP system consisted of a chest strap with two embedded electrodes. The electrodes were textile sensors consisting of woven silver yarn. The chest strap was positioned directly below the OEP marker at the xiphoid process with the strap positioned between the row of markers at this level. This allowed the electrodes to be positioned on the lower ribcage (Figure 9.1b). The electrodes were dampened with water prior to being placed on the participants to increase conductivity and thus, signal quality. The IP signal, sampling at 125 Hz, was processed and sent to an iOS device via Bluetooth.

Protocol

The BbB analyser flow sensor and face mask were fitted to the participant. Participants were positioned upright on a cycle ergometer (Lode – Corival, Netherlands). The IP system and the BbB analyser recorded impedance data continuously throughout the test. The BbB analyser recorded respiratory data throughout the test, namely respiratory rate (RR), tidal volume (Vt), and respiratory exchange ratio (RER). The OEP system recorded data for approximately 30 – 60 seconds at each increment, namely rest pre-exercise, low intensity, moderate intensity, high intensity and recovery post exercise. Participants were instructed to breathe quietly during the rest trial, starting with two maximum inhalations and exhalations to act as reference points during synchronisation of the measures. Participants then began cycling at 50 W, this was then increased by 30 W every minute. The exercise intensities were defined using the RER obtained from the BbB analyser, with low, moderate, and high intensity being defined as a RER value of 0.8, 1.0, >1.1, respectively (Deuster & Heled, 2008).

Data Analysis

The OEP trials were gap filled for up to 10 frames where necessary using Qualisys Track Manager v2.17. The BbB analyser data was recorded and exported using CORTEX Biophysik MetaSoft CPX testing software. The BbB analyser and the IP signals were synchronised using the two maximal breaths at the beginning of the rest trial. The trials were then cut to match the timings of the OEP data trials.

OEP respiratory rate and tidal volume were obtained following the methods in chapter 3. IP respiratory rate was obtained by counting the number of peaks within the impedance signal and change in impedance obtained by calculating the difference between end-expiratory and end-inspiratory impedance for each breath all using a custom written script in MATLAB.

Statistical Analysis

The breathing rates and tidal volumes obtained simultaneously from each measure were averaged over 30 seconds for each trial. Using SPSS (IBM SPSS Statistics 25), Pearson's correlation coefficient and intraclass correlation coefficients were calculated between the respiratory rates and tidal volumes/change in impedance obtained from the IP system and the OEP system, and between the BbB analyser and the IP system. Bland-Altman plots (Bland & Altman, 1999) were generated to assess the level of agreement between the OEP system and the BbB analyser to the IP system for both respiratory rate and tidal volume. The plots display the mean difference between the methods and the upper and lower limits of agreement calculated as the mean difference ± 1.96 the standard deviation. The level of significance was set to $p \leq 0.05$.

9.3 Results

Table 9.1 displays the Pearson's correlation coefficient and intraclass correlation coefficient values comparing the respiratory rates obtained from the IP system with the OEP system and the BbB analyser. The values demonstrate that the IP respiratory rates significantly correlated to both OEP and BbB respiratory rates at rest, during exercise, and recovery post exercise. All Pearson's correlation coefficients were above 0.4 and ICC values were all above 0.5, with the highest correlation between the systems was found at rest.

Table 9.1 Mean and standard deviation (SD) respiratory rates obtained from the IP system, the OEP system and the BbB analyser during rest and various exercise intensities with Pearson's correlation coefficient (r) and intraclass correlation coefficients (ICC) comparing the IP respiratory rate to the other two systems (* represents $p \leq 0.05$).

	IP		**OEP**				**BbB Analyser**			
Activity	**Mean**	**SD**	**Mean**	**SD**	**r**	**ICC**	**Mean**	**SD**	**r**	**ICC**
Rest	15.94	4.88	15.36	3.51	0.927 *	0.962 *	15.55	3.80	0.910 *	0.948 *
Low	19.29	6.65	21.83	4.80	0.621 *	0.730 *	22.59	5.08	0.556 *	0.683 *
Moderate	20.48	7.61	24.84	5.01	0.408 *	0.582 *	25.64	4.97	0.412 *	0.609 *
High	30.13	10.96	33.15	6.59	0.402 *	0.564 *	33.56	6.27	0.431 *	0.593 *
Recovery	24.52	9.29	23.37	4.69	0.774 *	0.869 *	24.08	4.15	0.729 *	0.839 *

Bland-Altman plots have been used to assess the agreement in respiratory rate obtained from the IP system and both the OEP system (Figure 9.2) and the BbB analyser (Figure 9.3). The plots displayed agreement between the IP system and both of the other systems, however, they also revealed a limited number of outliers at each exercise intensity, with no distinct trend. The rest condition had the lowest mean differences between IP and the other two systems, OEP (0.71 breaths) and BbB (0.48 breaths). During the exercise and recovery phases the mean differences increased for both comparisons.

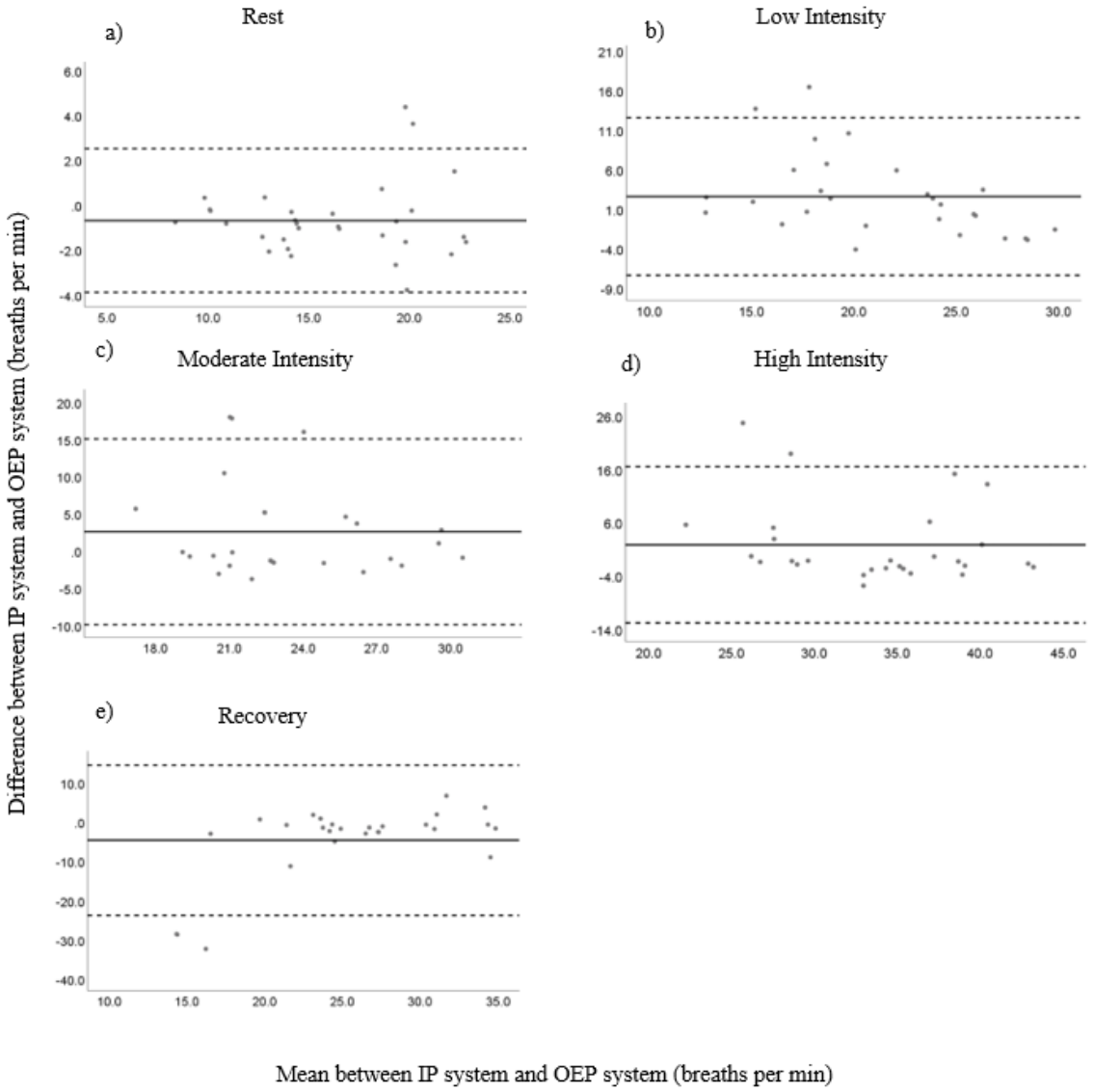


Figure 9.2 Bland-Altman plots representing the agreement between the respiratory rate obtained from the IP system and from the OEP analyser. Black solid line: mean difference between both methods, Black dashed lines: upper and lower limits of agreements (mean ±1.98 SD).

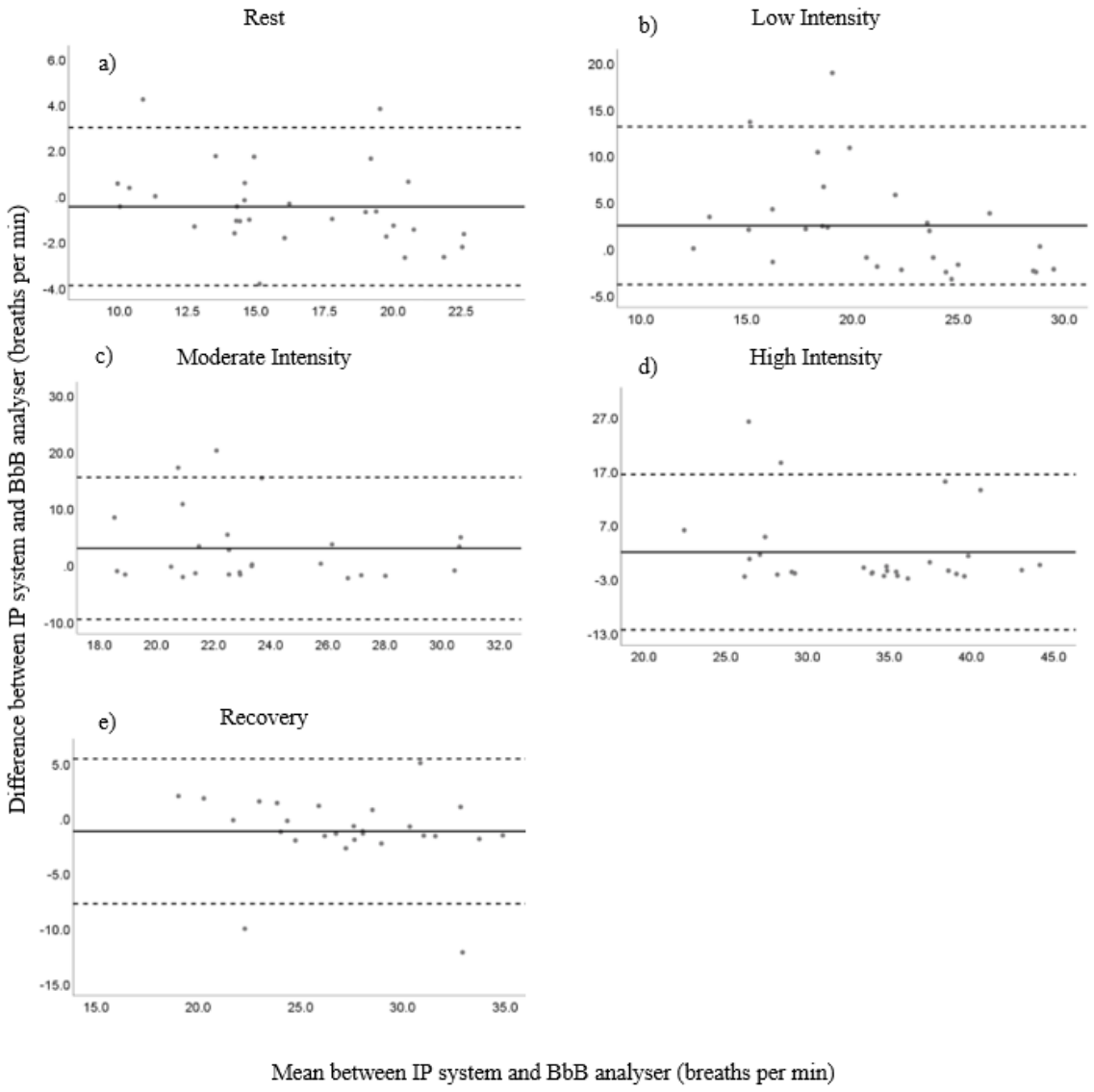


Figure 9.3 Bland-Altman plots representing the agreement between the respiratory rate obtained from the IP system and from the BbB analyser. Black solid line: mean difference between both methods, black dashed lines: upper and lower limits of agreements (mean ±1.98 SD).

Table 9.2 displays the mean Pearson's correlation coefficient and intraclass correlation coefficient values comparing the change in impedance measured by the IP system to the tidal volumes obtained from the OEP system and the BbB analyser for each participant. The IP system significantly correlated to both of the other systems during rest only ($r > 0.8$). The correlation values reduced as exercise intensity increased. Intra-class correlation coefficient values revealed very poor reliability between the IP system and the other two systems when comparing the change in impedance signal and the tidal volumes.

Table 9.2 Mean and standard deviation (SD) change in impedance obtained from the IP system, and tidal volumes obtained from the OEP system and the BbB analyser during rest and various exercise intensities with Pearson's correlation coefficient (r) and intraclass correction coefficients (ICC) comparing the IP change in impedance to the tidal volumes obtained from the other two systems (* represents $p \leq 0.05$).

	IP (Change in Impedance)		**OEP (Tidal Volume)**				**BbB Analyser (Tidal Volume)**			
Activity	**Mean**	**SD**	**Mean**	**SD**	**r**	**ICC**	**Mean**	**SD**	**r**	**ICC**
	(Volts)		**(litres)**				**(litres)**			
Rest	1.08×10^{-4}	2.16×10^{-4}	0.75	0.33	0.810 *	0.003	0.60	0.29	0.830 *	0.003
Low	1.73×10^{-4}	3.11×10^{-4}	1.36	0.29	0.610	0.001	1.16	0.23	0.671	0.002
Moderate	1.78×10^{-4}	1.47×10^{-4}	2.09	0.67	0.468	0.001	1.72	0.49	0.560	0.001
High	2.41×10^{-4}	3.23×10^{-4}	2.64	0.86	0.414	0.001	2.13	0.69	0.538	0.001
Recovery	1.38×10^{-4}	0.70×10^{-4}	1.83	0.53	0.568	0.001	1.56	0.44	0.620	0.001

9.4 Discussion

This study demonstrates that IP respiratory rate correlated with both OEP and BbB analysis at rest and across various exercise intensities, with higher correlation values at rest. This study also uses unique fabric-based sensors embedded into clothing, exploring the potential use of IP as a discrete wearable system to measure the movement of the chest wall during breathing. Change in impedance signal demonstrated poor reliability when compared to tidal volumes obtained from OEP and BbB analysis. Factors such as sensor movement, sensor type, and participant characteristics impact the quality of this IP system's signal which seemed to vary between participants as there were several outliers with low correlation coefficient values in this data set.

The respiratory rate obtained from the IP system significantly correlated to those obtained from the BbB analyser. However, the mean correlation values are slightly lower during exercise ($r < 0.600$; ICC: < 0.700) when compared to the rest ($r = 0.910$; ICC: 0.948, $p \leq 0.05$) and recovery phase (0.729, $p \leq 0.05$). This supports the previously established IP validation against direct pneumotachography i.e., BbB analysis in healthy adults during loaded breathing (Seppa *et al.,* 2013a). Similarly, when comparing the IP system with the OEP system, the respiratory rate obtained displayed strong correlations ($r > 0.700$, $p \leq 0.05$) during rest and recovery and moderate to weak correlations during exercise ($r < 0.600$). Further investigation using Bland-Altman plots revealed that there were a number of outliers when comparing IP to both systems (Figure 9.2, Figure 9.3). These outliers could be due to the quality of the impedance signal obtained from the IP system.

Two main factors influence the changes in electrical resistance of the lungs during breathing. Firstly, the gas volume in relation to the fluid volume within the lungs as an increase in gas volume during inspiration reduces the conductivity. Secondly, the length of the conductance paths which increases during inspiration due to the expansion of the chest wall. Both of these factors combined cause an increase in the electrical impedance during inspiration (Gupta, 2011). However, other factors may influence quality of the signal (Vuorela *et al.,* 2010) such as sensor type, sensor position, sensor movement, sweat, body composition (Seppa *et al.*, 2009), sex, fit of garment, and body hair. Traditional IP systems use electrodes attached to the skin with a skin preparation protocol which involves shaving the area and cleaning the skin with alcohol to remove hair and dead skin cells. For this study, the fabric electrodes were dampened to promote conductivity but the skin was not prepared in any way as per the manufacturer's instructions, making this study more ecologically valid.

As the correlation between the breathing rate obtained from the IP system and the other two systems were lower during the exercise conditions, this may suggest that the movement of the upper body during cycling has an effect on the quality of the IP signal for certain participants. The movement may have caused mechanical disturbances between the sensor and the skin or caused the sensors to move position during the exercise. This disturbance may produce unpredictable errors in IP signals which is a commonly known issue with IP (Seppa *et al.,* 2010; Khambete *et al.,* 2000). However, from the Bland-Altman plots (Figure 9.2 and Figure 9.3) this data set had a limited number of outliers therefore, suggesting that these additional factors only affected a limited number of the participants in this study. The fabric sensors used in this study may not provide the quality of the signal typically seen with traditional electrodes particularly during exercise. However, fabric sensors are necessary for wearable technology to be integrated discretely into clothing and to avoid hindering or constraining the user.

Overall, the respiratory rate from the IP system correlated well with both the OEP system and BbB analyser. One of the issues with using wearable technology, is the lack of available information regarding algorithms used within wearable devices. Companies and manufactures do not often disclose the processes behind their algorithms. For this reason, custom-built MATLAB scripts were used in this study which counted the number of breaths in a trial. Understanding how the IP signal behaves across a range of breathing frequencies and lung volumes is important (Seppa *et al.,* 2013b). In a previous investigation, using the same IP system as in this study, it was found differences in breathing frequency accuracy between algorithms, with time- and frequency-domain algorithms displaying more accuracy during exercise and rest, respectively (Smyth *et al.,* 2018). Therefore, the type of signal processing will impact the accuracy of the outputs. Although there have been major advancements in IP signal processing (Ansari *et al.,* 2017; Seppa *et al.,* 2011), caution should be taken when using manufacture's algorithms as the accuracy may change across breathing frequency values i.e., rest versus exercise.

The change in impedance obtained from the IP system correlated with change in volume obtained from both the OEP and BbB systems significantly during rest only ($r > 0.8$, $p \leq 0.05$), with weaker correlations seen during exercise ($r < 0.7$) (Table 9.2). However, the IP system displayed poor reliability when comparing the change in impedance signal and the other two systems (ICC < 0.1) at rest and during exercise. The reduction in r values during high intensity exercise may be explained by IP signal quality and the factors which affect it

as discussed above. In order to obtain accurate absolute volume values, IP calibration procedures with calibration coefficients determined from a BbB analyser is required. However, the calibration seems to be dependent on factors such as body mass index (Seppa *et al.,* 2009), electrode configuration (Logic *et al.,* 1967; Seppa *et al.,* 2013b) and body position (Houtveen *et al.,* 2005; Mlynczak *et al.,* 2015). It has been suggested that participant- and position-specific calibration procedure should be performed before IP-derived breathing parameters are calculated (Houtveen *et al.,* 2005; Mlynczak & Cybulski, 2017). Others have not used calibration coefficients (Malmbery *et al.,* 2017), but have ensured linearity between the IP signal and BbB analyser signal within a specific posture (Seppa *et al*., 2016). Although calibration may be possible within a laboratory or clinical setting, it is not feasible for a user to calibrate an IP system in their own environment, for example, an athlete in their training environment due to the lack of accessible equipment. This potentially limits the use of an IP system for measuring volume parameters outside of a laboratory setting.

The sensor configuration of the IP system seems to be the most important factor in IP signal quality (Mlynczak *et al.,* 2015; Seppa *et al.,* 2013b). The two-sensor configuration used in the IP system is the most commonly used sensor configuration for impedance pneumography due to its simplicity (Gupta, 2011; Jeyhani *et al.,* 2017). However, more sensors can be used such as a four-sensor configuration. The four-sensor configuration produces a more accurate measurement as the location sites of the injected current and the voltage recording can be physically separated (Roe & Sudarshan, 2015). In contrast, the potential difference measured between the two-sensor configuration contains non-linear voltages at the interface of the sensor and chest, and may introduce some unpredictable errors in the measurement (Gupta, 2011). Therefore, using a configuration with more than two sensors may be able to overcome any possible errors caused by body movement during exercise (Khambete *et al.,* 2000). Future studies should investigate the use of additional IP sensors to further improve the impedance signal quality at rest and during exercise.

In order for IP to obtain all of the OEP phase angle metrics, it would have to have the ability of measuring the movement of the abdomen. This would involve placing sensors on lower torso positions not typically used with an IP system. Seppa and colleagues (2013b) investigated electrode configurations for IP and found that the electrodes placed only on the lower chest resulted in non-linear impedance change to lung volume change. Using IP to measure abdomen movement may present with some similar challenges as organs in the

abdomen have different electrical conductivity when compared to the lung tissue (Gabriel *et al.*, 1996; Seppa *et al.*, 2013b). Therefore, further investigation would be required to determine the quality of the IP signal when sensors are placed on the abdomen. Additionally, the regional and phase angle OEP-derived breathing parameters require multiple regional compartment volumes to be measured simultaneously. To our knowledge, this has never been attempted before with IP as it would require using multiple separate IP electrode configurations, for example, an IP configuration to measure the ribcage volume and a separate configuration to measure the abdomen volume. However, passing two separate low-amplitude high frequency electrical currents into the torso simultaneously would result in signal interference and cross-over. A potential way of overcoming this issue would be to develop an algorithm which would allow for coordinated electrical input and sampling which would alternate between configurations. This may require substantial computational power and, depending on the sampling frequency, may result in reduced samples per configuration, potentially losing important information. To obtain the OEP-derived phase angles, four separate compartments volumes namely, the shoulders, pulmonary ribcage, abdominal ribcage, and abdomen, are essential and therefore, four coordinated IP configurations would be required. Even if this coordinated IP system was possible, it would not have the ability to measure any of the hemi-thoracic (left versus right compartment) phase angles.

In conclusion, this study demonstrated the potential use of an IP garment-based system in the form of two embedded fabric sensors for measuring respiratory rate at rest, exercise, and recovery post-exercise. However, further investigation is required, potentially into the incorporation of additional sensors and algorithm development, to expand the capabilities of IP and potentially allow similar parameters to those from OEP to be obtained. Unlike current measurement techniques, this would allow for the measurement of these breathing parameters over longer periods of time in a variety of environments. This discrete continuous monitoring may reveal significant diagnostic information for athletes with respiratory disorders including dysfunctional breathing and may be implemented as a training aid with real-time biofeedback.

Chapter 10. General Discussion

10.1 Main Findings

The primary aim of this thesis was to improve our understanding of breathing mechanics in athletes with respiratory disorders, namely exercise-induced bronchoconstriction (EIB) and dysfunctional breathing, by characterising normal and pathological breathing patterns using OEP. OEP is already a well-established respiratory measurement technique. The findings of this thesis add to its validity and potential applications.

Key Experimental Chapter Findings

- Chapter 4 established normative ranges for OEP-derived breathing pattern parameters for a non-pathological population at rest and during exercise, and identified candidate parameters which differed significantly between the rest and exercise conditions. This included novel parameters AbSPhase (shoulders versus abdomen), RCpSPhase (upper ribcage versus shoulders), and RCaSPhase (lower ribcage versus shoulders). The normative ranges provide the basis for future studies to differentiate other population groups against.
- Chapter 5 identified OEP parameters including V_E, RCpRCaPhase (upper versus lower ribcage), RCaSPhase (lower ribcage versus shoulders), and RCpSPhase (upper ribcage versus shoulders) which differed between rest and bronchoconstriction induced by eucapnic voluntary hyperpnoea test in athletes with exercise-induced bronchoconstriction, and established how OEP-derived breathing parameters differed in this population from a healthy population at rest.
- Chapter 6 presented breathing pattern for athletes with a dysfunctional breathing pattern during rest and exercise, and identified significant differences in OEP breathing parameters between athletes with and without a dysfunctional breathing pattern at rest and during exercise, with more compartment asynchrony seen in those with a dysfunctional breathing pattern. This involved the development of novel parameters including inhale and exhale percentage deviations which significantly differed between healthy and dysfunctional breathing patterns.

- Chapter 7 demonstrated the potential use of OEP as a novel real-time feedback aid for breathing retraining. As part of an acute breathing intervention, real-time OEP phase angle feedback reduced the phase angle values of RcAbPhase (ribcage versus abdomen), RCpAbPhase (upper ribcage versus lower ribcage and abdomen), and AbSPhase (abdomen versus shoulders) in athletes with dysfunctional breathing during high intensity exercise.
- Chapter 8 demonstrated that OEP-derived phase angle parameters altered as expected with simulated dysfunctional breathing patterns. The POWERbreathe volume, cumulative volume, pressure, and power outputs were shown for the first time to significantly differ between simulated asynchronous breathing pattern to both normal and simulated thoracic dominant breathing patterns, identifying the potential of the POWERbreathe inspiratory muscle training device for distinguishing between different breathing patterns.
- Chapter 9 established that respiratory rate and change in impedance signal correlated with the OEP system and traditional breath-by-breath analysis at rest and during exercise. However, impedance pneumography signal quality seemed to be reduced during exercise.

Characterising Breathing Patterns

OEP has previously been used to characterise the breathing patterns of various populations including healthy individuals (Layton *et al.,* 2012), COPD patients (Aliverti *et al.,* 2009), intensive care patients (Aliverti *et al.,* 2000) and new-borns (Dellaca *et al.,* 2010) to name a few. However, when characterising these populations, previous studies have used a select number of the potential OEP breathing parameters. Using a limited number of parameters may potentially result in important information being lost and potential breathing pattern characteristics to be missed. This thesis utilised a large number of existing OEP breathing parameters along with newly developed breathing parameters, including the shoulder phase angles RCaSPhase (shoulders versus lower ribcage), RCpSPhase (shoulders versus lower ribcage), and AbSPhase (shoulders versus abdomen) and percentage deviation parameters (inhale and exhale), to characterise the breathing patterns of healthy individuals (Chapter 4), athletes with EIB (Chapter 5), and athletes with dysfunctional breathing (Chapter 6). Prior to this thesis, OEP has not been used to assess the breathing patterns of athletes with EIB or dysfunctional breathing.

Initially in Chapter 4, the full list of breathing parameters were calculated on a healthy population at rest, during a sub-maximal incremental exercise test, and during recovery post-exercise. This identified parameters, including novel parameters (RcAbIndex and AbSPhase), which differed between the exercise conditions along with timing and phase angle parameters, as discussed in Chapter 4. The value ranges for a non-pathological population group in this thesis were comparable to previous literature for the timing (Parreira *et al.,* 2010), compartment contribution (Priori *et al.,* 2013), and phase angle parameters (Pereira *et al.,* 2017). This chapter not only provided a non-pathological data set to compare the pathological populations against but also identified candidate parameters which may also be capable of differing between populations.

This thesis for the first time objectively characterises the breathing patterns of athletes with EIB and dysfunctional breathing using OEP. Since these population groups have not been assessed before, the full list of OEP breathing parameters were calculated for both groups as to not miss any potentially important parameter changes. The parameters minute ventilation, and phase angles including RCpRCaPhase (upper versus lower ribcage), RCaSPhase (lower ribcage versus shoulders), and RCpSPhase (upper ribcage versus shoulders) significantly altered within the EIB group when comparing rest and post-bronchoconstriction (Chapter 5). These alterations indicate that individuals with EIB display more compartment asynchrony during bronchoconstriction when compared to rest. During bronchoconstriction, there is an increase in airway obstruction which has been shown to result in greater retention of air volume within the chest (Feitosa *et al.,* 2018). This is likely causing the breathing pattern adjustments reported in this thesis experienced by individuals with EIB during the eucapnic voluntary hyperpnea test.

Within the dysfunctional breathing pattern group (Chapter 6), the OEP-derived timing parameters significantly altered from rest to high intensity exercise as an expected typical exercise response, for example increased respiratory rate (see Table 6.2). The compartment contributions significantly changed from rest to exercise with increased ribcage (RcCT) and reduced abdomen (AbCT) contribution during exercise (see Table 6.3). From rest to high intensity exercise individuals with dysfunctional breathing demonstrated an increase in compartment asynchrony with increases in a number of phase angles including RCpHTA (right versus left upper ribcage), RcAbPhase (ribcage versus abdomen), RCpAbPhase (upper ribcage versus combined lower ribcage and abdomen), RCpRCaPhase (upper versus lower ribcage), RCpSPhase (shoulders versus upper ribcage), and AbSPhase (shoulders versus

abdomen), see Table 6.4 for full details. In addition, the newly developed inhale and exhale percentage deviation parameters were both significantly greater during high intensity exercise. These deviation parameters indicate the amount of Konno-Mead breath loop deviation from the straight 45 degree diagonal line which is expected for a normal synchronous breathing pattern. As the deviation parameters split the breath into the inhale and exhale, they also offer more information as to where in the breath asynchrony occurs. Generally, as exercise intensity increased, compartment asynchrony increased for individuals with dysfunctional breathing. This increase is likely due to the poor recruitment pattern of the respiratory muscles which alters in the breathing mechanics during the increased ventilatory demands of exercise (Pereira *et al.,* 2017).

Previously breathing patterns have been used to differentiate pathological and non-pathological populations, such as chronic obstructive pulmonary disease (Motamedi-Fakhr *et al.,* 2016) and asthma (Hmeidi *et al.,* 2017). When comparing both of the pathological (EIB and dysfunctional) groups to the healthy breathing pattern group, only three parameters significantly altered for both pathological groups during their evoked state. These parameters included RCpRCaPhase (upper versus lower ribcage), RCaSPhase (lower ribcage versus shoulders), and RCpSPhase (upper ribcage versus shoulders) (Table 10.1). These phase angle parameters each displayed significantly greater phase angle values for both the EIB and dysfunctional breathing pattern group when compared to the healthy breathing pattern group. This indicates that both of the pathological groups displayed greater amounts of compartment asynchrony compared to the non-pathological group. Outside of these parameters, the pathological groups significantly altered the remaining breathing parameters differently. Table 10.1 summaries the parameters which significantly differed from both the EIB and dysfunctional breathing pattern group when compared to the healthy breathing pattern group, indicating whether the parameter value was significantly larger or smaller than the healthy population dataset.

During bronchoconstriction, the EIB group (Chapter 5) displayed significantly larger phase angle values for four of the phase angle parameters when compared to the healthy group at rest (Table 10.1). As already stated, three of these phase angles were also significantly greater for the dysfunctional breathing pattern group, however, the EIB group also displayed a significant increase in RCaHTA (right versus left lower ribcage). This indicates that athletes with EIB displayed an increase in asynchrony within their breathing pattern during bronchoconstriction. These alterations, when compared to the healthy population, are again

likely to be a compensatory mechanism in response to the limited expiration caused by bronchoconstriction.

While for the dysfunctional breathing pattern group at rest, the compartment contribution parameters were significantly different to the healthy breathing pattern group along with the phase angle HTA and the inhale percentage deviation (Table 10.1). However, apart from RCaCT and inhale percentage deviation, these parameters were not significantly different to the healthy breathing pattern group during high intensity exercise, instead timing (tI, tE, tTot) and phase angle parameters were significantly different. In addition to the common three phase angles which were also significantly different for the EIB group, the dysfunctional breathing pattern group displayed significantly larger RCpHTA, RcAbPhase, inhale and exhale percentage deviation values in comparison to the healthy breathing pattern group during exercise. These alterations indicate that the dysfunctional breathing pattern group took significantly shorter breaths with more asynchrony between the compartments. Previously there have been attempts to categorise different types of dysfunctional breathing based on subjective measures and anecdotal evidence (Boulding *et al.*, 2016). Most participants recruited in this thesis did not overtly display, from the OEP data, distinct types of dysfunctional breathing that are currently reported with the respiratory literature e.g., thoracic dominant or thoracoabdominal asynchrony (Chapter 5). This may be due to the severity of the disorder and perhaps, similar to other respiratory disorders like asthma, dysfunctional breathing may be graded from mild to severe. It may be difficult to determine which type of breathing pattern an individual with 'mild' dysfunctional breathing is displaying. However, OEP allows for a wide range of parameters to be obtained simultaneously, providing detailed information on an individuals' breathing pattern. Additionally, participants may display combination breathing patterns, for example, a participant does not display thoracoabdominal asynchrony alone but also some thoracic dominant breathing pattern as well. The dysfunctional breathing pattern group recorded in this thesis was not categorised into different types of dysfunctional breathing patterns, however, using OEP it would be possible to further categorise individuals with dysfunctional breathing patterns including individuals with a mild dysfunctional breathing pattern (Chapter 5 and 6).

When comparing the EIB and dysfunctional breathing group directly during their respective evoked states in Chapter 6, the timing variables were found to be significantly different (see Table 6.7). These timing differences are likely due to the different ventilatory demands

required for high intensity exercise which involves an increase respiratory rate, tidal volume, and minute ventilation, versus bronchoconstriction which limits expiration. In addition, the phase angles RCpHTA, RcAbPhase, RCpAbPhase, and AbSPhase also demonstrated significant differences between the groups, indicating that the dysfunctional breathing pattern group displayed greater amounts of asynchrony between the related compartments compared to the EIB group (see Table 6.7). These differences are likely caused by the mechanisms underlying these conditions. Although EIB and dysfunctional breathing patterns can present with similar symptoms, the mechanisms behind the disorders are not the same. The mechanism underlying dysfunctional breathing is not fully understood but it is characterised by chronic changes in breathing pattern which can be caused by psychological or behavioural factors such as depression or anxiety (Vidotto *et al.,* 2019), or can occur alongside other respiratory disorders such as asthma or chronic obstructive pulmonary disease (Boulding *et al.,* 2016). In contrast, EIB is the temporary narrowing of the airways occurring during exercise that limits expiration (Hull *et al.,* 2016).

It is clear from this thesis that there is not one singular OEP-derived breathing parameter that could be used to distinguish between pathological and non-pathological breathing patterns. The established and novel parameters developed in this thesis have proven to be useful in identifying differences between conditions and between population groups. OEP has the advantage over other respiratory monitoring systems in that it can measure a wide range of parameters simultaneously. By using a combination of these parameters, OEP provides a detailed characterisation of pathological and non-pathological breathing patterns. Understanding the characteristics and alterations in breathing patterns within pathological groups may aid in the diagnosis of and differentiation between respiratory disorders, and potentially provide bespoke breathing retraining programmes.

Table 10.1 Breathing parameters which demonstrate significant differences in the dysfunctional breathing pattern (DBP) group and the exercise-induced bronchoconstriction (EIB) group at a) rest, b) an evoked state (high intensity exercise or post-bronchoconstriction), and c) recovery when compared to the corresponding healthy breathing pattern group conditions,



Abbreviations: RR – respiratory rate, Vt – tidal volume, tI – inspiratory time, tE- expiratory time, tTot – total breath time, RCaCT – abdominal ribcage volume contribution, RcCT – ribcage volume contribution, AbCT – abdominal volume contribution, RcAbIndex – ratio between ribcage and abdomen volume, HTA – hemi-thoracic (left and right) asynchrony phase angle, RCpHTA – phase angle between the right and left pulmonary ribcage, RCaHTA – phase angle between the right and left abdominal ribcage, RcAbPhase – phase angle between the ribcage and abdomen, RCpRCaPhase – phase angle between the pulmonary and abdominal ribcage, RCaSPhase – phase angle between the abdominal ribcage and shoulders, RCpSPhase – phase angle between the pulmonary ribcage and shoulders, Inhale % Deviation – percentage deviation of the phase angle (ribcage vs abdomen) during inhalation, Exhale % Deviation – percentage deviation of the phase angle (ribcage vs abdomen) during exhalation.

So far, the alterations in phase angle values between populations have been discussed however, the sign of the phase angle value indicates the order in which the compartments movement (see Table 3.2). By using all of the phase angle value signs, it is possible to determine the order in which all of the compartments move in relation to each other during a breath, providing insight into the overall coordination of the compartments. Within the previous literature, a normal symptom free breathing pattern can be characterised by the thoracic compartments moving in synchronisation (Bradley & Esformes, 2014). The findings of thesis (Chapter 4) supports this, as the healthy group displayed small phase angle values indicating that the thoracic compartments moved close to perfect synchrony. Although the phase angle values were close to zero, there were slight differences in the timing of the compartment movements for this healthy population at rest. During exercise, these asynchronies increased with significant increases in phase angle values for RCaHTA (left versus right upper ribcage), RcAbPhase (ribcage versus abdomen), RCpAbPhase (upper ribcage versus lower ribcage and abdomen), and RCpRCaPhase (upper versus lower ribcage) (see Table 4.4). During rest, the abdomen seems to be the leading compartment, followed by the shoulders, the lower ribcage, and finally the upper ribcage. This order changes during high intensity exercise, with the lower ribcage becoming the leading compartment followed by the abdomen, shoulders, and upper ribcage. This alteration is likely due to the increased ventilatory demand experienced during high intensity exercise. This compartment coordination reverts back to the resting order during recovery. Figure 10.1 illustrates the compartment coordination for the healthy group at rest, high intensity exercise, and recovery. Although there are slight differences in the initial movement of the compartments for this non-pathological population, it is most likely that these differences are not overtly apparent to the human eye and still result in a smooth breath due to the lower phase angle values.

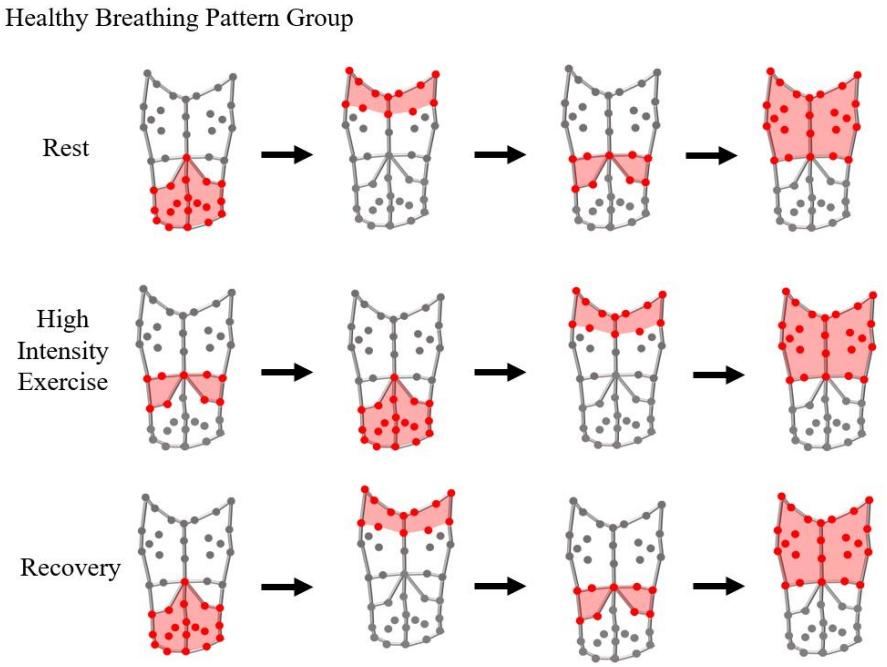


Figure 10. 1 Illustration of the compartment movement order, from left to right, for the normal breathing pattern group during rest, high intensity exercise, and recovery post-exercise.

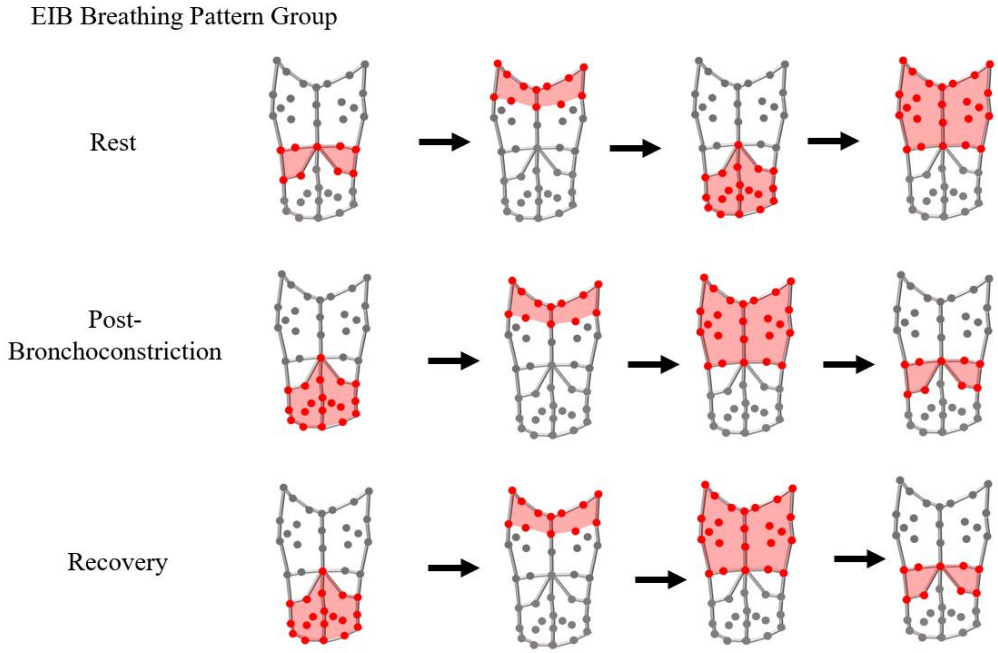


Figure 10.2 Illustration of the compartment movement order, from left to right, for the exercise-induced bronchoconstriction breathing pattern group during rest, post-bronchoconstriction, and recovery post-exercise.

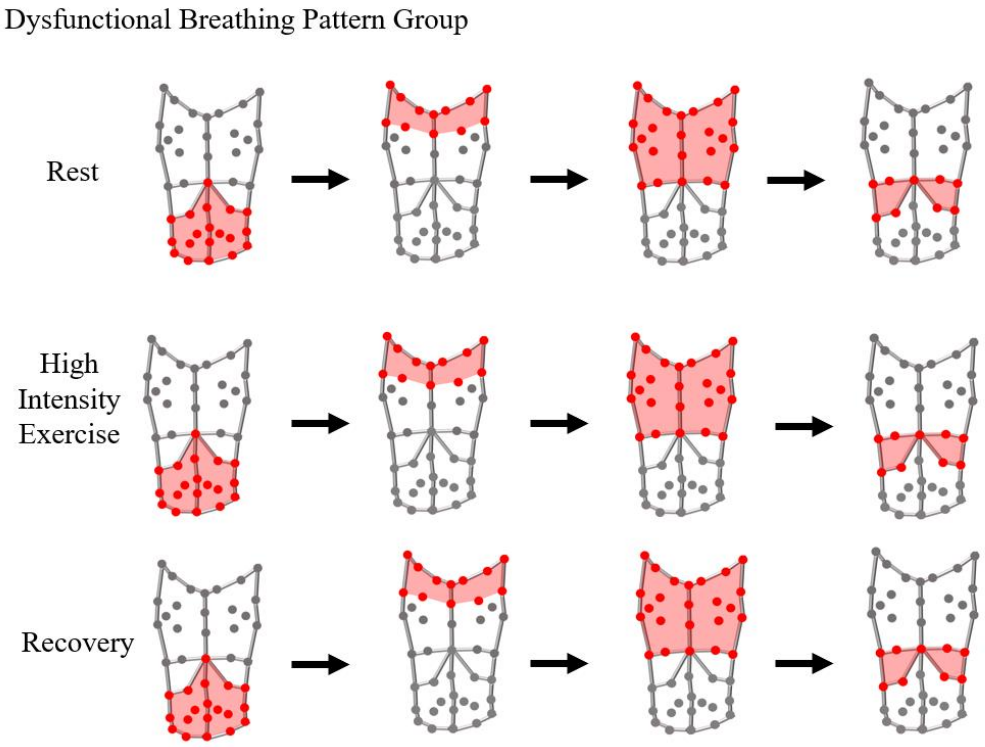


Figure 10.3 Illustration of the compartment movement order, from left to right, for the dysfunctional breathing pattern group during rest, high intensity exercise, and recovery post-exercise.

The breathing pattern coordination for both the EIB group (Figure 10.2) and the dysfunctional breathing group (Figure 10.3) did not match the normal breathing pattern coordination (Figure 10.1). For the EIB group at rest, the leading compartment was the lower ribcage, followed by the shoulders, the abdomen, and the upper ribcage. However, post-bronchoconstriction the abdomen leads the breath, followed by the shoulders, upper ribcage, and lower ribcage. During recovery, the coordination did not revert back to the resting coordination. Although the values of the breathing pattern parameters reverted back towards the resting values (see Table 5.5), this compartment movement order potentially indicates that full recovery was not yet complete.

The compartment coordination for the dysfunctional breathing pattern group did not change across any of the conditions (Figure 10.3). At rest, the abdomen leads the breath followed by the shoulders, upper ribcage, and lower ribcage. Although the phase angle values of RCpHTA (left versus right upper ribcage), RcAbPhase (ribcage versus abdomen), RCpAbPhase (upper ribcage versus lower ribcage and abdomen), RCpRCaPhase (upper versus lower ribcage), RCpSPhase (upper ribcage versus shoulders), and AbSPhase (abdomen versus shoulders) significantly increased during high intensity exercise, the order

of the compartment coordination did not change but the amount of asynchrony significantly increased. Dysfunctional breathing patterns arise from the poor recruitment pattern of the respiratory muscles, predominately the inspiratory muscles and the diaphragm (Pereira *et al.*, 2017), which may have caused these alterations in breathing mechanics.

When the ventilatory demands were increased for the healthy group during high intensity exercise, the lower ribcage became the leading compartment. In contrast, when the ventilatory demands were increased for both the EIB and dysfunctional breathing pattern group, the lower ribcage was the last compartment to move within the breathing pattern compartment movement order (Figures 10.1 – 10.3). The leading compartment of the breath may be a key characteristic of identifying a healthy and pathological breathing patterns. There have been previous attempts to categorise the different types of dysfunctional breathing such as thoracoabdominal asynchrony, thoracic dominant breathing, and hyperventilation, however, these have been categorised using subjective measures such as questionnaires and the manual assessment of respiratory motion (Boulding *et al.*, 2016). Both the thoracoabdominal asynchrony and thoracic dominant dysfunctional breathing patterns are currently defined by the leading compartment of the breath (Boulding *et al.*, 2016). The findings of this thesis suggest that both the size of the phase angle values i.e., larger the value the more asynchrony between compartments, and the sign of the phase angle values i.e., the compartment movement order, are both important for the objective characterisation of pathological breathing patterns and their distinction from normal, non-pathological breathing patterns. From this thesis, it is suggested that the phase angle parameters should be used as a group of parameters, rather than selecting a single phase angle, in order to obtain a comprehensive overview of an individual's breathing pattern.

OEP and Breathing Retraining

In Chapters 4, 5 and 6, it was established that OEP-derived breathing parameters could identify differences within a population under different conditions, for example different exercise intensities. A large number of these OEP-derived breathing parameters including the timing (see Table 4.2), regional contribution (see Table 4.3), and phase angle (see Table 4.4) parameters displayed significant differences between the resting and recovery conditions for the healthy population group, indicating that the participants had not fully recovered from the high intensity exercise during this recovery phase. Similarly, for the dysfunctional breathing pattern group all of the timing parameters (see Table 6.2) demonstrated significant differences between rest and recovery, again indicating that these metrics had not completely reverted to their resting values. In contrast to the healthy group, none of the regional contribution parameters displayed significant differences between rest and recovery, which may demonstrate that the regional contribution parameters are some of the first parameters to revert back to a rested state for the dysfunctional breathing pattern group (see Table 6.3). From the phase angle parameters, only the phase angle between the upper ribcage and shoulders (RCpSPhase) and the inhale percentage deviation parameters were significantly different during recovery compared to rest (see Table 6.4). Therefore, OEP shows promise in being able to monitor the recovery of individuals with a healthy and dysfunctional breathing pattern post sub-maximal exercise.

In contrast to the healthy and dysfunctional breathing pattern groups, the EIB group in Chapter 5 did not demonstrate any significant differences between the rest and post-inhaler recovery conditions across all of the OEP-derived breathing pattern parameters. However, the compartment movement order for the EIB changed during bronchoconstriction but it did not revert back to its original resting order during recovery post-inhaler (Figure 10.2). This may indicate that the breathing parameter values reverted back to or close to their resting state but the breathing pattern compartment order had not yet returned to the resting order. Previously within the literature, SLP has been used to assess the impact of a bronchodilator on the breathing pattern of children with asthma. It has been demonstrated that SLP-derived IE50 is a promising metric for monitoring bronchodilator effects in asthmatics (Hmeidi *et al.,* 2017). This thesis demonstrates that OEP, similar to SLP, may be a useful tool in assessing the effect of bronchodilators in athletes with exercise-induced bronchoconstriction by tracking when the breathing parameters return to their resting state and potentially could allow for the refinement of medication dosage.

Along with tracking the effects of bronchodilators in EIB, OEP may also be a potential tool to track the effect of breathing retraining interventions. Previously, volume and flow incentive spirometry feedback have been used as part of breathing retraining interventions in both pathological and non-pathological population groups, with volume based feedback promoting more breathing control and alterations in thoracic compartment synchrony (de Moraes Paisani *et al.,* 2013; Tomich *et al.,* 2010). The differences found between volume and flow incentive spirometry demonstrate that different forms of breathing feedback can have different effects on the breathing pattern retraining. In Chapter 7, a novel OEP feedback system was developed which had the capability of displaying volume and phase angle (ribcage versus abdomen) feedback in real-time. When used as part of an acute breathing retraining intervention for athletes with dysfunctional breathing, the OEP combined phase angle and volume feedback and the volume only feedback produced a similar reduction in respiratory rate during exercise. However, the phase angles between the ribcage and abdomen (RcAbPhase), between the upper ribcage and abdomen (RCpAbPhase) and between the abdomen and shoulders (AbSPhase) altered differently between the different feedback groups, with the combined phase angle and volume feedback resulting in a reduction in these phase angle values during exercise post-intervention. This demonstrates that OEP can be utilised not only as a real-time breathing retraining feedback system but also track improvements in breathing pattern caused by breathing retraining. This application of OEP is particularly useful for athletes with respiratory disorders, as OEP is one of the few breathing pattern measurement systems capable of accurately tracking breathing pattern during exercise.

Other Potential Methods for Measuring Breathing Patterns

In Chapter 4, we established agreement between the OEP system against traditional breath-by-breath analysis at rest and during exercise. This has been done many times before in previous studies (Cala *et al.,* 1996; Layton *et al.,* 2012), but is a necessary step to establish the correctness of the custom MATLAB scripts used throughout this thesis. In Chapter 8, the OEP system was further validated by assessing if the OEP phase angle parameters altered as expected with simulated dysfunctional breathing patterns including thoracic dominant and thoracoabdominal asynchrony breathing patterns. The OEP-derived phase angles altered as anticipated (see Table 8.2), for example, the phase angle between the ribcage and the abdomen was significantly greater for the thoracoabdominal asynchrony breathing pattern when compared to a normal and thoracic dominant breathing pattern.

Although OEP is a valid and powerful objective breathing pattern measurement system and this thesis has begun to expand its full capabilities for physically active individuals with respiratory disorders, OEP is still restricted to the laboratory setting, requires expensive equipment and expertise. In order to make the findings within this thesis accessible to clinicians and athletes, these parameters would ideally be obtained outside of the laboratory. Since validated, OEP may act as a gold standard system to compare future breathing pattern systems against, particularly for parameters such as compartment contribution and phase angles not obtainable by traditional gold standard systems such as breath-by-breath analysis. Therefore, the final two experimental chapters of this thesis explored two potential alternative measurement systems including an inspiratory muscle training device (Chapter 8) and an impedance pneumography system (Chapter 9), to determine whether further development of these alternative systems could allow for similar breathing parameters as OEP to be obtained outside of the laboratory setting.

The portable inspiratory muscle training device, POWERbreathe, has previously been used for inspiratory muscle training on individuals with respiratory disorders (Beckerman *et al.,* 2005; Turner *et al.,* 2011) and athletes (Volianitis *et al.,* 2001). However, the outputs produced by the POWERbreathe device have never been used to identify differences between respiratory disorders or inform the user on the quality of their breathing pattern. In Chapter 8, the POWERbreathe outputs including volume, pressure, and power significantly differed between the simulated dysfunctional breathing patterns and normal breathing. More specifically, the simulated thoracoabdominal asynchrony breathing pattern significantly differed from the normal and simulated thoracic dominant breathing patterns. This

demonstrates that the thoracoabdominal asynchrony breathing pattern produced significantly different POWERbreathe output traces indicating that these output traces, with further investigation, may be used to identify thoracoabdominal asynchrony. The thoracoabdominal asynchrony breathing pattern resulted in the lowest volume values out of the three breathing patterns indicating that the asynchronous breathing pattern limits the amount of volume inhaled (see Table 8.1). Similarly, the thoracoabdominal asynchrony pattern displayed a delay in peak power achieved as discussed in Chapter 8. By continuing this work, cut-off criteria for the POWERbreathe trace outputs may be established for different breathing patterns, potentially developing the device as a tool to aid in the diagnosis of dysfunctional breathing outside of the laboratory. By establishing clear POWERbreathe trace characteristics for a normal breathing pattern, it may be possible to create visual or verbal prompts to a user with thoracoabdominal asynchrony to aid in the retraining of their breathing pattern towards a normal breathing pattern. With further development, the POWERbreathe device may be a useful training tool with an integrated feedback mechanism for individuals with dysfunctional breathing with the capability of tracking improvements in breathing pattern caused by retraining.

Although, the POWERbreathe outputs differed significantly between the simulated dysfunctional breathing patterns i.e., thoracoabdominal asynchrony and thoracic dominant, it was not successful in identifying differences between the thoracic dominant and normal breathing pattern. Therefore, if an individual with dysfunctional breathing presented with a thoracic dominant breathing pattern, the POWERbreathe device firstly, may not be capable of identifying the thoracic dominant breathing pattern and secondly, will likely be unable to track any alterations in thoracic dominant breathing pattern towards a normal breathing pattern with breathing retraining. Therefore, some individuals with dysfunctional breathing may find the current POWERbreathe device useful depending on how their dysfunctional breathing presents, while others may not.

Impedance pneumography is a respiratory monitoring system that has been used to monitor respiratory rate in patients across a wide range of conditions including asthmatics (Seppa *et al.,* 2013a) and healthy individuals in different conditions such as varying postures (Mlynczak *et al.,* 2015) and walking (Jeyhani *et al.,* 2016). It is often used over other systems as it is portable, non-invasive, and simple to use (Ansari *et al.,* 2017). In Chapter 9, a garment-based impedance pneumography system with two embedded fabric sensors was assessed against the OEP system. Respiratory rate obtained from the impedance

pneumography system correlated with both the OEP system and traditional breath-by-breath analysis. However, the impedance pneumography signal quality was reduced due to noise during exercise which was likely caused by sensor movement, sensor type, and participant characteristics. This noise reduced the agreement between the impedance pneumography system and both the OEP and breath-by-breath analysis system for the respiratory rate (see Table 9.1) and the change in impedance signal (see Table 9.2). The change in impedance signal demonstrated poor intra-class correlation coefficient values when compared to both OEP and BbB analysis (see Table 9.2). This may limit the application of impedance pneumography to static measurement conditions, making it less useful for athletes with respiratory disorders. In recent years, there have been considerable developments with impedance pneumography sensor configurations and signal processing with it being validated against breath-by-breath analysis (Seppa *et al.,* 2016). However, in order to obtain parameters similar to the OEP-derived breathing parameters, the impedance pneumography system would be required to measure multiple thoracic compartment volumes simultaneously. As discussed in Chapter 9, this has never been done before and would require significant development and testing.

Both of these alterative systems show some promise for further development as breathing pattern assessment systems. However, both demonstrate some critical issues and disadvantages, requiring considerable development. The solution to a portable breathing pattern measurement system may lie outside of the alternative systems investigated in this thesis. In recent years there has been progress with using fiber bragg gratings in monitoring respiration in both the clinical and sport settings (Presti *et al.,* 2020). Fiber bragg gratings have been integrated into garment based systems and has had some success in measuring respiratory rate and tidal volume and may provide a potential solution to monitoring breathing pattern outside of the laboratory setting (Ciocchetti *et al.,* 2015; Massaroni *et al.,* 2016; Presti *et al.,* 2017). However, like the alternative systems investigated in this thesis, fiber bragg grating based systems requires further development and extensive investigation into its accuracy and validity.

10.2 Limitations

The limitations of this thesis have been discussed within each experimental chapter. Throughout this thesis, all of the exercise conditions consisted of cycling. Cycling was chosen as it allows for optimal OEP data to be collected. However, breathing patterns may change during different forms of exercise, due to the various postures associated with other types of exercise such as rowing or running. Additionally, since OEP must be performed in a laboratory setting, it does not fully replicate field settings and participants breathing patterns may have been altered due to their awareness of respiration and being recorded. Additionally, OEP has been shown to slightly overestimate tidal volume when compared to breath-by-breath analysis so caution should be taken when interpreting the volume results or making comparisons to other systems, particularly if assessing symptoms such as hyperinflation which has been associated with EIB.

Participants were placed into the dysfunctional breathing pattern group based on self-reported symptoms via a questionnaire and exclusion of EIB via the EVH test. Although this ruled out these individuals having EIB, it may not rule out other conditions potentially causing respiratory symptoms during exercise. Therefore, it was assumed that, once the EVH test was negative, respiratory symptoms during exercise were caused by dysfunctional breathing and the participants were not exhaustively screened for other respiratory conditions.

The inter-rater and intra-rater reliability of the OEP system was not assessed within this thesis. Previously there has only been one known study to assess the inter- and intra-rater reliability in a healthy population at rest and during exercise which found intra-class correlation coefficient above 0.70 and a coefficient of variation below 11% (Vieira *et al.*, 2013). However, Vieira and colleagues (2013) only assessed timing and compartment contribution parameters for a healthy population. Before OEP can be utilised as a potential diagnostic tool for EIB and dysfunctional breathing within a clinical setting, its reliability needs to be established.

10.3 Future Directions

The findings of this thesis demonstrate the capabilities of OEP for athletes with respiratory disorders. However, many of the studies within this thesis involve the use of novel OEP-derived parameters and/or novel applications of OEP. Therefore, further research is required to expand on the present findings.

Significant differences between populations and conditions (rest versus exercise) are discussed throughout this thesis. However, it is important to understand that this does not necessarily mean that these differences are clinically significance or important. Within the literature, minimal clinical important differences have not yet been established for the breathing pattern parameters used within this thesis (Courtney *et al.,* 2019). Clinical significance is dependent on the implications of the results on existing practice and establishes the size and scope of an effect (Ranganathan *et al.*, 2015). For example, in Chapter 7, significant improvements (more thoracic compartment synchrony) in dysfunctional breathing patterns are reported post an acute breathing retraining intervention. In order to determine whether these improvements are clinically significant, further work is required to establish how long participants can retain these improvements and if the changes in breathing pattern elicit improvements in the participants lives, for example, reduced symptoms. Other factors should also be taken into account when considering clinical significance including reproducibility, cost-effectiveness and ease of implementation into clinical practice.

One of the main aims within this thesis was to identify breathing parameters that differed between pathological and non-pathological groups. In Chapter 5 and 6, candidate parameters were identified which were capable of identifying differences between healthy, EIB, and dysfunctional breathing patterns. However, to establish OEP as a diagnostic tool for either EIB or dysfunctional breathing, further investigation is required to identify clear cut-off values for these candidate breathing parameters to extend the use of OEP into clinical diagnostic practice. One option to obtain cut-offs would be by constructing receiver operating characteristics taking into account sensitivity and specificity. In order to obtain adequate sensitivity and specificity, larger sample sizes would be required.

This thesis investigated the use of OEP as a method of identification (Chapter 6) and potential treatment (Chapter 7) for individuals with a dysfunctional breathing pattern for the first time. OEP was used as part of an acute breathing pattern retraining intervention, future work should include longitudinal intervention in order to determine if OEP real-time

feedback could result in chronic improvements in breathing pattern. However, there are various other populations and therapies which have never been assessed with OEP. For example, speech and language therapy interventions, which involve breathing pattern retraining exercise, have been shown to improve cough symptoms for refractory chronic cough (Mitchell *et al.,* 2016b). OEP may provide important information into the impact of speech and language therapy interventions on breathing pattern alterations, providing objective insight into how and when breathing pattern changes occur. In recent months there has been a lot of interest in treatment for individuals post- COVID-19, with approximately 10% of individuals who have tested positive remaining with symptoms weeks or even months after diagnosis (Greenhalgh *et al.,* 2020). Initial literature has suggested that breathing exercises can help individuals with delayed COVID-19 recovery by reducing symptoms such as breathlessness (Greenhalgh *et al.,* 2020; Zhao *et al.,* 2020). Once the OEP markers have been placed on by the practitioner, all OEP assessments can be performed from a distance, and due to the contagious nature of COVID-19, OEP may be a potential safe method of assessing the breathing pattern and breathing retraining of patients with or post COVID-19.

This thesis explored two potential portable alternatives to OEP, however, it was determined that both systems would require substantial development to potentially be a portable breathing pattern measurement system. With this development, the findings of this thesis indicate that both systems would have considerable limitations for example, identifying between pathological breathing patterns and signal quality during exercise. Therefore, the solution to a portable breathing pattern measurement system may not involve these systems. Further investigation is required into other potential wearable technology solutions such as fiber bragg grating (Presti *et al.,* 2020) or other non-invasive portable optical methods such as smartphone cameras (Reyes *et al.,* 2016).

Before an alternative method is developed, it may be beneficial to attempt to reduce the marker set required of OEP to improve its use and efficiency. There have been previous attempts to simplify and reduce the marker set (Massaroni *et al.,* 2018), however, care must be taken so markers necessary for the calculation of the full breathing parameter list are not removed, potentially losing important information. By reducing the marker set, it may make OEP more clinically viable through the reduction in set-up time.

10.4 Conclusion

The findings of this thesis demonstrate that athletes with respiratory disorders, namely exercise-induced bronchoconstriction and dysfunctional breathing, present with different breathing pattern characteristics during periods of increased ventilatory demand when compared to each other and to a normal breathing pattern data set. This thesis has demonstrated the potential capabilities of OEP by using previously established and novel breathing pattern parameters, with the finding suggesting that a selection of parameters should be used to fully characterise breathing patterns. These novel metrics include AbSPhase (abdomen versus shoulder phase angle), RCaSPhase (lower ribcage versus shoulders phase angle), RCpSPhase (upper ribcage versus shoulders phase angle), inhale percentage deviation and exhale percentage deviation. Through the use of these parameters, this thesis provides the first objective characterisation of the breathing pattern of athletes with dysfunctional breathing and exercise-induced bronchoconstriction.

This thesis presented a novel real-time OEP feedback system and as part of an acute intervention demonstrating that athletes with dysfunctional breathing could utilise the feedback to significantly improve their breathing pattern with breathing pattern parameters shifting towards those of a normal breathing pattern, more specifically real-time phase angle feedback reduced phase angle (RcAbPhase - ribcage versus abdomen) values. Further investigation is required to establish if such a feedback system would elicit long term breathing pattern changes.

This thesis identifies OEP as a powerful tool for potentially diagnosing and treating respiratory disorders in athletes, however, it is still confined to the laboratory setting. Finally, two potential portable alternatives to OEP were explored however, it was determined that both systems would require considerable development and validation in order to act as a portable breathing pattern measurement system.

Overall, this thesis provides the basis for future work which should involve establishing cut-off points for the breathing pattern parameters for each of the populations assessed. This could then be expanded to other population groups and age ranges. Additionally, further work is required to determine if using OEP as part of a breathing retraining intervention can elicit chronic changes in breathing pattern. The solution to a portable breathing pattern measurement system may involve other wearable technology systems.

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Appendices

Appendix 1: Ethical Approval Forms

Appendix 1

Ethical Approval Forms

University of Kent

School of Sport & Exercise Sciences
Research Ethics and Advisory Group (REAG)
University of Kent at Medway
Chatham Maritime
Kent
ME4 4AG

Ethics Reference:

Prop 44_2017_18

Date: 15th December 2017

Dear Carol Smyth,

Re: Respiratory function measurements during exercise.

I am delighted to confirm that SSES REAG has approved your research study (REF No. Prop 44_2017_18) and you are now permitted to recruit participants and commence your research.

If you need to amend any aspect of your research, please ensure you inform SSES REAG by completing a request for amendment form and submitting all revised paperwork (e.g. participant information sheet, questionnaires).

If there should happen to be any adverse event during your study, please also ensure SSES REAG is kept informed.

I hope your study is successful.

With kind regards,

Louis Passfield

(Chair SSES REAG)

School of Sport & Exercise Sciences
Research Ethics and Advisory Group (REAG)
University of Kent at Medway
Chatham Maritime
Kent
ME4 4AG

Ethics Reference:

Prop 21_2018_19

Date: 15th November 2018

Dear Carol Smyth,

Re: Respiratory function measurements during exercise

I am delighted to confirm that SSES REAG has approved your research study (REF No. Prop 21_2018_19) and you are now permitted to recruit participants and commence your research.

If you need to amend any aspect of your research, please ensure you inform SSES REAG by completing a request for amendment form and submitting all revised paperwork (e.g. participant information sheet, questionnaires).

If there should happen to be any adverse event during your study, please also ensure SSES REAG is kept informed.

I hope your study is successful.

With kind regards,

Louis Passfield

(Chair SSES REAG)

School of Sport & Exercise Sciences
Research Ethics and Advisory Group (REAG)
University of Kent at Medway
Chatham Maritime
Kent
ME4 4AG

Ethics Reference:

Prop 92_2018_19

Date: 15th August 2019

Dear Carol Smyth,

Re: Respiratory function measurements during exercise

I am delighted to confirm that SSES REAG has approved your research study (REF No. Prop 92_2018_19) and you are now permitted to recruit participants and commence your research.

If you need to amend any aspect of your research, please ensure you inform SSES REAG by completing a request for amendment form and submitting all revised paperwork (e.g. participant information sheet, questionnaires).

If there should happen to be any adverse event during your study, please also ensure SSES REAG is kept informed.

I hope your study is successful.

With kind regards,

Louis Passfield

(Chair SSES REAG)

Appendix 2

Study Related Documents

Health Questionnaire

HEALTH QUESTIONNAIRE

Code Number:

Age:

Please answer these questions truthfully and completely. The sole purpose of this questionnaire is to ensure that you are in a fit and healthy state to complete an exercise test.
ANY INFORMATION CONTAINED HEREIN WILL BE TREATED AS CONFIDENTIAL.

SECTION 1: GENERAL HEALTH QUESTIONS

Please read the 10 questions below carefully and answer each one honestly: check YES or NO.

	YES	NO
1. Has your doctor ever said that you have a heart condition or high blood pressure?	☐	☐
2. Do you feel pain in your chest at rest, during your daily activities of living, or when you do physical activity?	☐	☐
3. Do you lose balance because of dizziness or have you lost consciousness in the last 12 months? (Please answer NO if your dizziness was associated with over-breathing including vigorous exercise).	☐	☐
4. Have you ever been diagnosed with another chronic medical condition (other than heart disease or high blood pressure)?	☐	☐
If yes, please list condition(s) here:		
5. Are you currently taking prescribed medications for a chronic medical condition?	☐	☐
If yes, please list condition(s) and medications here:		
6. Do you currently have (or have you had within the past 12 months) a bone, joint or soft tissue (muscle, ligament, or tendon) problem that could be made worse by becoming more physically active? Please answer NO if you had a problem in the past but it *does not limit your ability* to be physically active.	☐	☐
If yes, please list condition(s) here:		
7. Has your doctor ever said that you should only do medically supervised physical activity?	☐	☐
8. Are you, or is there any chance you could be, pregnant?	☐	☐
9. Are you currently taking any nutritional supplement? **If 'YES' please inform the researcher what supplements are being taken**	☐	☐
10. Are you involved in any other research project? **If 'YES' please inform the researcher about details of the project**	☐	☐

Please provide brief details of your current weekly levels of physical activity (sport, physical fitness or conditioning activities), using the following classification for exertion level:
L = light (slightly breathless)
M = moderate (breathless)
V = vigorous (very breathless)

	Activity	Duration (mins)	Level (L/M/V)
Monday			
Tuesday			
Wednesday			
Thursday			
Friday			
Saturday			
Sunday			

If you answered NO to all of the questions above, you are cleared to take part in the exercise test



You do not need to complete section 2. Please sign relevant declaration on the consent form.



If you answered YES to one or more of the questions in Section 1 - PLEASE GO TO SECTION 2.

SECTION 2: CHRONIC MEDICAL CONDITIONS

Please read the questions below carefully and answer each one honestly: check YES or NO.

		YES	NO
1.	**Do you have arthritis, osteoporosis, or back problems?** If YES answer questions 1a-1c. If NO go to Question 2.	☐	☐
1a.	Do you have difficulty controlling your condition with medications? or other physician-prescribed therapies? (Answer NO if you are not currently taking any medications or other treatments).	☐	☐
1b.	Do you have joint problems causing pain, a recent fracture or fracture caused by osteoporosis or cancer, displaced vertebrae (e.g. spondylolisthesis), and/or spondyloysis/pars defect (a crack in the bony ring on the back of the spinal column)?	☐	☐
1c.	Have you had steroid injections or taken steroid tablets regularly for more than 3 months?	☐	☐
2.	**Do you have cancer of any kind?** If YES answer questions 2a-2b. If NO, go to Question 3.	☐	☐
2a.	Does your cancer diagnosis include any of the following types: lung/bronchogenic, multiple myeloma (cancer of plasma cells), head and neck?	☐	☐
2b.	Are you currently receiving cancer therapy (such as chemotherapy or radiotherapy)?	☐	☐
3.	**Do you have heart disease or cardiovascular disease? This includes coronary artery disease, high blood pressure, heart failure, diagnosed abnormality or heart rhythm.** If YES answer questions 3a-3e. If NO go to Question 4.	☐	☐
3a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking any medications or other treatments).	☐	☐
3b.	Do you have an irregular heartbeat that requires medical management? (e.g. atrial fibrillation, premature ventricular contraction)	☐	☐
3c.	Do you have chronic heart failure?	☐	☐
3d.	Do you have a resting blood pressure equal to or greater than 160/90mmHg with or without medication? Answer YES if you do not know your resting blood pressure.	☐	☐
3e.	Do you have diagnosed coronary artery (cardiovascular) disease and have not participated in regular physical activity in the last 2 months?	☐	☐

		YES	NO
4.	**Do you have any metabolic conditions? This includes Type 1 Diabetes, Type 2 Diabetes and Pre-Diabetes.** If YES answer questions 4a-4c. If NO, go to Question 5.	☐	☐
4a.	Is your blood sugar often above 13mmol/L? (Answer YES if you are not sure).	☐	☐
4b.	Do you have any signs or symptoms of diabetes complications such as heart or vascular disease and/or complications affecting your eyes, kidneys, OR the sensation in your toes and feet?	☐	☐
4c.	Do you have other metabolic conditions (such as thyroid disorders, current pregnancy related diabetes, chronic kidney disease, or liver problems)?	☐	☐
5.	**Do you have any mental health problems or learning difficulties?** This includes Alzheimer's, dementia, depression, anxiety disorder, eating disorder, psychotic disorder, intellectual disability and down syndrome. If YES answer questions 5a-5b. If NO go to Question 6.	☐	☐
5a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking any medications or other treatments).	☐	☐
5b.	Do you also have back problems affecting nerves or muscles?	☐	☐
6.	**Do you have a respiratory disease?** This includes chronic obstructive pulmonary disease, asthma, pulmonary high blood pressure. If YES answer questions 6a-6d. If NO, go to Question 7.	☐	☐
6a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking any medications or other treatments).	☐	☐
6b.	Has your doctor ever said you blood oxygen level is low at rest or during exercise and/or that you require supplemental oxygen therapy?	☐	☐
6c.	If asthmatic, do you currently have symptoms of chest tightness, wheezing, labored breathing, consistent cough (more than 2 days/week), or have you used your rescue medication more than twice in the last week?	☐	☐
6d.	Has your doctor ever said you have high blood pressure in the blood vessels of your lungs?	☐	☐
7.	**Do you have a spinal cord injury?** This includes tetraplegia and paraplegia. If YES answer questions 7a-7c. If NO, go to Question 8.	☐	☐
7a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking any medications or other treatments).	☐	☐
7b.	Do you commonly exhibit low resting blood pressure significant enough to cause dizziness, light-headedness, and/or fainting?	☐	☐
7c.	Has your physician indicated that you exhibit sudden bouts of high blood pressure (known as autonomic dysreflexia)?	☐	☐

		YES	NO
8.	**Have you had a stroke?** This includes transient ischemic attack (TIA) or cerebrovascular event. If YES answer questions 8a-8c. If NO go to Question 9.	☐	☐
8a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking any medications or other treatments).	☐	☐
8b.	Do you have any impairment in walking or mobility?	☐	☐
8c.	Have you experienced a stroke or impairment in nerves or muscles in the past 6 months?	☐	☐
9.	**Do you have any other medical condition which is not listed above or do you have two or more medical conditions?** If you have other medical conditions, answer questions 9a-9c. If NO go to Question 10.	☐	☐
9a.	Have you experienced a blackout, fainted, or lost consciousness as a result of a head injury within the last 12 months OR have you had a diagnosed concussion within the last 12 months?	☐	☐
9b.	Do you have a medical condition that is not listed (such as epilepsy, neurological conditions, and kidney problems)?	☐	☐
9c.	Do you currently live with two or more medical conditions?	☐	☐
	Please list your medical condition(s) and any related medications here:		
10.	**Have you had a viral infection in the last 2 weeks (cough, cold, sore throat, etc.)?** If YES please provide details below:	☐	☐
11.	**Is there any other reason why you cannot take part in this exercise test?** If YES please provide details below:	☐	☐

Please see below for recommendations for your current medical condition and sign this document:



If you answered NO to all of the follow-up questions about your medical condition, you are cleared to take part in the exercise test.



If you answered YES to one or more of the follow-up questions about your medical condition it is strongly advised that you should seek further advice from a medical professional before taking part in the exercise test.

This health questionnaire is based around the PAR-Q+, which was developed by the Canadian Society for Exercise Physiology www.csep.ca

Respiratory Questionnaire

EVH ATHLETE QUESTIONNAIRE

NAME							
SPORT							
DATE		**D.O.B.**		**HEIGHT**	**cm**	**WEIGHT**	**KG**

1) Do you currently smoke? Yes/No **If yes, how many cigarettes per day?_____**

2) During or after training or competition do you experience any of the following?
Please tick as many as appropriate

Coughing_____ Wheezing breathing in _____ Wheezing breathing out _____ Chest tightness breathing in_____
Chest tightness breathing out_____ Difficulty in Breathing (Dyspnoea)_____ Excess Mucus Production_____

3) If you have reported experiencing any of the symptoms in question 2, which area of your body would you say is the most effected from the following…

Nose_____ Throat_____ Chest_____

4) During training or competition what environmental conditions seem to make your breathing worse?
Please tick as many as appropriate

Cold Climate_____ Hot Air_____ Dry Air_____ Humid Air_____ High Pollen Content_____

High Pollution_____ Altitude _____ Other_____

5) Do you have problems breathing through your nose? **Yes/No** **(Circle as appropriate)**

6) Do you use any Nitrate Supplements? **Yes/No**

7) Have you ever suffered from asthma? **Yes/No**

8) Are you currently diagnosed asthmatic and use medication? **Yes/No**

9) Do you suffer from Exercised Induced Asthma and use medication? **Yes/No**

If you have answered ***YES*** to either or both of questions 8 and 9 please complete table below.
Asthma Medication Table: Example

Type of Drug	Drug Name	Dose	Dose Frequency	Year Started
Reliever	Ventolin	200 mcg	3 times a day	2016
Preventer	Beclomethasone	250mcg	Twice a day	2017
Other	Serevent	150 mcg	2 times a week	2018

Type of Drug	Drug Name	Dose	Dose Frequency	Year Started
Reliever	1.			
	2.			
Preventor	1.			
	2.			
Other	1.			
	2.			

10) Have you taken any of the above in the past 72 hours? **Yes/No** **If yes, how long ago? _____**

11) Have you consumed any caffeine within the past 4 hours? **Yes/No**

Study 1: Respiratory Function Measurements during Exercise

We would like to invite you to take part in our research study. Before you decide, we want you to understand why the research is being done, and what we will ask you to do. It is a good idea to talk to your friends and family about whether you should take part. One of our team will go through the information sheet with you and answer any questions you have.

What is the purpose of the study?

In this study, we would like to investigate different measurements of breathing function at rest, during and following a sub-maximal exercise challenge.

Who can take part?

We would like to recruit female and male participants aged over the age of 18 years, who take part in regular exercise (at least 5 hours per week)

You will not be able to participate if you have had a chest infection in the 4 weeks or any other illnesses within 2 weeks prior to the test, have any cardiovascular problems (e.g., high blood pressure or any kind of heart problems), metabolic diseases (e.g., diabetes), neurological conditions (e.g., stroke, epilepsy), if you are pregnant, injured or if you have any conditions that limit mobility. You will be asked to complete a screening questionnaire that will check for any of these conditions.

Do I have to take part?

No. It is up to you to decide whether or not to take part. If you agree to be included but later change your mind, you can withdraw from this study at any time without explanation. A decision to withdraw, or a decision not to take part, will result in all of your data collected to that point being deleted.

What will happen to me if I take part?

If you decide to take part, you will be asked to attend the Biomechanics Laboratory at the School of Sport and Exercise Sciences, University of Kent, Medway Campus on a single occasion.

a) For <u>healthy, physically active individuals;</u>

The testing session involve investigating your respiratory function through a breathing assessment and an exercise challenge test using three different measurement methods.

The visit will last approximately 3 hours in total.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and your sport or physical activity.

- **Part 2.** A respiratory assessment that includes:

  Spirometry: A test to measure the maximum amount of air that can be inhaled and exhaled from your lungs. From a standing position, you will breathe in until your lungs are full. You will then put a mouthpiece in your mouth and breathe out as hard and as fast as possible until your lungs are empty (approximately 6 seconds).

- **Part 3.** Exercise Challenge:

  You will undergo a staged incremental exercise challenge to the point where you cannot continue to match the resistance of the bike. The exercise begins at a relatively light intensity (50 W) and increases every minute until you cannot keep going.
  During the test you will be required to wear a facemask to allow us to measure the gas you are breathing out, a Smartlife garment in the form of a chest strap for measurement of a number of physiological parameters including heart rate and breathing rate, and removeable reflective markers placed on your torso (chest, back and sides) for measurement of chest and abdomen wall movement. A motion capture system will capture the position of these reflective markers only.

  The test will begin with a few minutes of rest which will involve breathing in and out normally during quiet sitting, to obtain a baseline for all measurements. Following the period of rest, you will then complete a warm up and then the exercise challenge. The intensity of the exercise challenge will be increased in controlled increments until you cannot match the resistance of the bike. The exercise challenge should not last longer than approximately 20 minutes. The exercise challenge will then be followed by another period of recovery in a seated position. All measurements will be obtained prior, during and after the exercise challenge simultaneously.

What are the possible benefits of taking part?

You will get a free assessment of your respiratory function during rest and/or exercise. With your assistance, we are hoping to gain a better understanding of respiratory function measurement techniques during rest and/or exercise, which could help develop new, more efficient methods of measuring respiratory function for future athletes and patients to improve their performance, and in the future. You will also receive feedback on your breathing technique with the aim to improve your breathing pattern.

Will my taking part in this study be kept confidential?

Yes. All the information about your participation in this study will be kept confidential. The information collected will be stored on a password protected computer with all identifiable information removed. Only the principal researcher and supervisors will have access to your personally identifiable data.

If I take part in this study, how will you protect my privacy?

The findings may be published in scientific journals or presented at conferences, but this would not include any information that would let others know who you are.

What are the risks and possible discomforts from being in this study?

- The risks associated with the exercise test are fatigue, muscle soreness, irregular heartbeat, chest pain and sudden heart attack. To minimize these risks, you will undergo a warm-up exercise prior to the starting of testing. You will have your heart rate and the rating of perceived exertion monitored continuously throughout the test. The test will be stopped at any point if we think that you are either (a) having difficulty breathing or (b) incurring any other form of distress. Also, the test will be terminated if the participant feels that either of the above conditions is occurring. Immediate first aid will be provided by qualified individuals in the unlikely event that something adverse occurs.

How long does the study participation last for?

In total, your involvement will be no more than 3 hours over one testing session.

Prior to arriving for the testing session please observe the following restrictions

1. No high intensity exercise 24 hours before each visit.
2. No food within 2 hours of testing
3. No alcohol within 24 hours before the visit.
4. No sports drinks or caffeine intake 4 hours prior to the test.
5. Free from illness or infection in the two weeks before the test.
6. Drinking water is permitted.

Who is organising and funding the research?

This research study is being organised by the School of Sport and Exercise Sciences at the University of Kent and Smartlifeinc Limited, and is funded by Innovate UK.

The research will be conducted with an industrial partner within a Knowledge Transfer Partnership and thus, the research is in part funded by a commercial company (Smartlifeinc Limited) and is intended to validate their product. The company will not have any influence over the integrity of the research or the dissemination of the findings. In addition, the academic supervisors (Dr John Dickinson & Dr Samantha Winter) will not derive any personal financial benefit from the work or from the commercial product.

What will happen to the results of the research?

We will analyse all of the anonymous results to help us decide how to investigate our research question further. We may report group results in academic journals or conferences.

If I have questions or concerns about this study, who can I contact?

You can contact the principal researcher listed below for questions specifically related to this study.

Carol Smyth at the School of Sport and Exercise Sciences, University of Kent (email:carol.smyth@smartlifeinc.com)

What do I do now?

If you understand the information given in this form and are happy to participant in the research study, then please fill in the attached consent form.

Study 2: Respiratory Function Measurements during Exercise

We would like to invite you to take part in our research study. Before you decide, we want you to understand why the research is being done, and what we will ask you to do. It is a good idea to talk to your friends and family about whether you should take part. One of our team will go through the information sheet with you and answer any questions you have.

What is the purpose of the study?

In this study, we would like to investigate different measurements of breathing function at rest, during and during and following a challenge (Eucapnic voluntary hyperventilation challenge and/or a sub-maximal exercise challenge).

Who can take part?

We would like to recruit female and male participants aged over the age of 18 years, in one of the following groups:

- Active individuals without any respiratory disease/disorders (at least 5 hours of exercise per week) **or**
- Active individuals with dysfunctional breathing (at least 5 hours of exercise per week) **or**
- Active individuals with asthma/exercise induced bronchoconstriction (at least 5 hours of exercise per week)

You will not be able to participate if you have had a chest infection in the 4 weeks or any other illnesses within 2 weeks prior to the test, have any cardiovascular problems (e.g., high blood pressure or any kind of heart problems), metabolic diseases (e.g., diabetes), neurological conditions (e.g., stroke, epilepsy), if you are pregnant, injured or if you have any conditions that limit mobility. You will be asked to complete a screening questionnaire that will check for any of these conditions.

Do I have to take part?

No. It is up to you to decide whether or not to take part. If you agree to be included but later change your mind, you can withdraw from this study at any time without explanation. A decision to withdraw, or a decision not to take part, will result in all of your data collected to that point being deleted.

What will happen to me if I take part?

If you decide to take part, you will be asked to attend the Biomechanics Laboratory at the School of Sport and Exercise Sciences, University of Kent, Medway Campus on a single occasion.

b) For <u>healthy, physically active individuals;</u>

The testing session involve investigating your respiratory function through a breathing assessment and an exercise challenge test using measurement methods.

The visit will last approximately 3 hours in total.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and your sport or physical activity.

- **Part 2.** Exercise Challenge:
  You will undergo a staged incremental exercise challenge to the point where you cannot continue to match the resistance of the bike. The exercise begins at a relatively light intensity (50 W) and increases every minute until you cannot keep going.
  The test will begin with a few minutes of rest which will involve breathing in and out normally during quiet sitting, to obtain a baseline for all measurements. Following the period of rest, you will then complete a warm up and then the exercise challenge. The intensity of the exercise challenge will be increased in controlled increments until you cannot match the resistance of the bike. The exercise challenge should not last longer than approximately 20 minutes. The exercise challenge will then be followed by another period of recovery in a seated position. All measurements will be obtained prior, during and after the exercise challenge simultaneously.

c) For individuals with dysfunctional breathing

The testing session involve investigating if you have asthma/exercise induced bronchoconstriction though an Eucapnic Voluntary Hyperventilation (EVH) challenge. If negative, your breathing function will then be assessed at rest and during an exercise challenge. You will then complete an acute intervention and repeat the same exercise challenge.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and physical activity.

- **Part 2.** EVH (Eucapnic Voluntary Hyperventilation) challenge: A test to assess the health of your airways at rest and following high intensity breathing. If your lung function reduces following the EVH test it is suggested that you may suffer from a form of asthma known as exercise induced bronchoconstriction. To complete the EVH challenge you will be asked to breathe through the mouthpiece at approximately 85% of your maximal breathing rate (similar to breathing during high intensity exercise) for 6 minutes. You will breathe a cold, dry gas mixture that contains 21% O_2, 5% CO_2, 74% N_2. After the 6-minute high intensity breathing, your maximal lung function will be measured at 3, 5, 7, 10 and 15 minutes. If you have had a 10% fall during this test, you will be advised to contact your GP. You will be provided with a report to allow you to discuss the results with them. If you do not demonstrate a 10% fall during this test, you will continue to part 3 (see below).

- **Part 3.** Exercise Challenge: You will then undergo a staged incremental exercise challenge to the point where you cannot continue to match the resistance of the bike. The exercise begins at a relatively light intensity (50 W) and increases every minute until you cannot keep going.

The test will begin with a few minutes of rest which will involve breathing in and out normally during quiet sitting, to obtain a baseline for all measurements. Following the period of rest, you will then complete a warm up and then the exercise challenge. The intensity of the exercise challenge will be increased in controlled increments until you cannot match the resistance of the bike. The exercise challenge should not last longer than approximately 20 minutes. The exercise challenge will then be followed by another period of recovery in a seated position. All measurements will be obtained prior, during and after the exercise challenge simultaneously.

c) For individuals with asthma, the testing session will involve investigating your respiratory function through an EVH challenge.

The visit will last approximately 1 hour in total.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and physical activity.

- **Part 2.** EVH (Eucapnic Voluntary Hyperventilation) challenge: A test to assess the health of your airways at rest and following high intensity breathing. If your lung function reduces following the EVH test it is suggested that you may suffer from a form of asthma known as exercise induced bronchoconstriction. To complete the EVH challenge you will be asked to breathe through the mouthpiece at approximately 85% of your maximal breathing rate (similar to breathing during high intensity exercise) for 6 minutes. You will breathe a cold, dry gas mixture that contains 21% O_2, 5% CO_2, 74% N_2. After the 6-minute high intensity breathing, your maximal lung function will be measured at 3, 5, 7, 10 and 15 minutes. If you have had a 10% fall during this test, you will be advised to contact your GP. You will be provided with a report to allow you to discuss the results with them.

All participants in each group will be required to wear a facemask to allow us to measure the gas you are breathing out. OEP (optoelectronic plethysmography) is a non-invasive 3D motion capture technique which measures the movement of the chest wall using 11 infrared cameras and 90 reflective markers placed on the chest, abdomen and back. The markers will be placed in a grid-like pattern onto the torso using hypoallergenic bi-adhesive tape. Participants will be required to be shirtless during the OEP data collection with females wearing a sports bra only. OEP data will be collected during each stage of the testing.

What are the possible benefits of taking part?

You will get a free assessment of your respiratory function during rest and/or exercise. With your assistance, we are hoping to gain a better understanding of respiratory function measurement techniques during rest and/or exercise, which could help develop new, more efficient methods of measuring respiratory function for future athletes and patients to improve their performance, and in the future. You will also receive feedback on your breathing technique with the aim to improve your breathing pattern.

Will my taking part in this study be kept confidential?

Yes. All the information about your participation in this study will be kept confidential. The information collected will be stored on a password protected computer with all identifiable information removed. Only the principal researcher and supervisors will have access to your personally identifiable data.

If I take part in this study, how will you protect my privacy?

The findings may be published in scientific journals or presented at conferences, but this would not include any information that would let others know who you are.

What are the risks and possible discomforts from being in this study?

- The risks associated with the exercise test are fatigue, muscle soreness, irregular heartbeat, chest pain and sudden heart attack. To minimize these risks, you will undergo a warm-up exercise prior to the starting of testing. You will have your heart rate and the rating of perceived exertion monitored continuously throughout the test. The test will be stopped at any point if we think that you are either (a) having difficulty breathing or (b) incurring any other form of distress. Also, the test will be terminated if the participant feels that either of the above conditions is occurring. Immediate first aid will be provided by qualified individuals in the unlikely event that something adverse occurs.

- EVH challenge has the potential to provoke severe narrowing of your airways. You will not be able to take part in the study if your resting lung function appears reduced (70% of predicted). Your lung function will be closely monitored during the EVH challenge. If your lung function falls below 10% of your baseline level you will be offered inhaled medication (salbutamol) that will reverse the reduction in lung function. You will be advised not to leave the laboratory until your lung function is within 10% of your resting value.

How long does the study participation last for?

In total, your involvement will be no more than 3 hours over one testing session.

Prior to arriving for the testing session please observe the following restrictions

7. No high intensity exercise 24 hours before each visit.
8. No food within 2 hours of testing
9. No alcohol within 24 hours before the visit.
10. No sports drinks or caffeine intake 4 hours prior to the test.
11. Free from illness or infection in the two weeks before the test.
12. Drinking water is permitted.

Who is organising and funding the research?

This research study is being organised by the School of Sport and Exercise Sciences at the University of Kent and Smartlifeinc Limited, and is funded by Innovate UK.

The research will be conducted with an industrial partner within a Knowledge Transfer Partnership and thus, the research is in part funded by a commercial company (Smartlifeinc Limited) and is intended to validate their product. The company will not have any influence over the integrity of the research or the dissemination of the findings. In addition, the academic supervisors (Dr John Dickinson & Dr Samantha Winter) will not derive any personal financial benefit from the work or from the commercial product.

What will happen to the results of the research?

We will analyse all of the anonymous results to help us decide how to investigate our research question further. We may report group results in academic journals or conferences.

If I have questions or concerns about this study, who can I contact?

You can contact the principal researcher listed below for questions specifically related to this study.

Carol Smyth at the School of Sport and Exercise Sciences, University of Kent (email:carol.smyth@smartlifeinc.com)

What do I do now?

If you understand the information given in this form and are happy to participant in the research study, then please fill in the attached consent form.

Athlete Information – Eucapnic Voluntary Hyperpnoea Challenge

Prior to Arriving for the Test

- If you are using any asthma medication you will need to stop using it prior to the testing as follows*:
    - Montelukasts 7 days prior to testing
    - Brown and Purple inhalers (Inhaled corticosteroid and combination inhalers) at least 72 hours prior to testing
    - Green inhalers (long acting β_2 -agonists) 42 hours prior to testing
    - Blue inhalers (short acting β_2 -agonists) 24 hours prior to testing
- Within 4 hours of the test avoid exercise that involves high intensity running, cycling, or swimming.
- Within 4 hours of the test avoid caffeine. This includes drinks and food such as: coffee, tea, energy drinks, fizzy, chocolate.
- Bring a drink of water with you

* In an emergency, you should take short acting Beta 2 Agonist (Salbutamol, Ventolin), terbutaline (Bricanyl). You MUST inform the tester of nay medication not conforming with this protocol.

Test Day

- You will be given questionnaires to complete which will ask name, age, height, weight, and past asthma history and respiratory symptoms.
- Your maximal lung function will be measured
- You will complete a 6 minute heavy breathing exercise called a Eucapnic Voluntary Hyperpnoea (EVH) challenge, which is explained below
- 3, 5, 7, 10, and 15 minutes after the EVH challenge your maximal lung function will be measured.
- If your lung function drops 10% or more following the EVH test, it is deemed a positive test. If this happens you will be offered a reliver inhaler (salbutamol) and have your maximal lung function measured 10 minutes later.
- You will be advised not to leave the lab until your function is within 10% of what it was when you started the test
- In total the test will last approximately 45 minutes

EVH challenge

- Maximal lung function is measured
- A target breathing rate is calculated based on your maximal lung function
- You will be asked to breath at this target breathing rate for 6 minutes. The breathing rate will be similar to the rate you breathe at during high intensity running.
- The air you breathe during the test will be cold and dry. It is a special gas designed to ensure to do not have any adverse effects during the test.
- After the 6 minutes of high intensity breathing your maximal lung function will be measured at 3, 5, 7, 10, and 15 minutes.

Study 3: Respiratory Function Measurements during Exercise with Acute Intervention

We would like to invite you to take part in our research study. Before you decide, we want you to understand why the research is being done, and what we will ask you to do. It is a good idea to talk to your friends and family about whether you should take part. One of our team will go through the information sheet with you and answer any questions you have.

What is the purpose of the study?

In this study, we would like to investigate different measurements of breathing function at rest, during and following a challenge (Eucapnic voluntary hyperventilation challenge and/or a sub-maximal exercise challenge). Additionally, we would like to investigate your breathing function during an acute breathing retraining intervention.

Who can take part?

We would like to recruit female and male participants aged over the age of 18 years, in one of the following groups:

a) Active individuals without any respiratory disease/disorder (at least 5 of exercise hours per week), **or**
b) Active individuals with dysfunctional breathing (at least 5 hours of exercise per week), **or**
c) Active individuals with asthma (at least 5 hours of exercise per week)

These three pathological groups were chosen as they all have similar respiratory symptoms which often lead to misdiagnosis. Most commonly, individuals with dysfunctional breathing are often misdiagnosed with asthma, particularly in athletes, due to the lack of specific respiratory diagnostic measures.

You will not be able to participate if you have had a chest infection in the 4 weeks or any other illnesses within 2 weeks prior to the test, have any cardiovascular problems (e.g., high blood pressure or any kind of heart problems), metabolic diseases (e.g., diabetes), neurological conditions (e.g., stroke, epilepsy), if you are pregnant, injured or if you have any conditions that limit mobility. You will be asked to complete a screening questionnaire that will check for any of these conditions.

Do I have to take part?

No. It is up to you to decide whether or not to take part. If you agree to be included but later change your mind, you can withdraw from this study at any time without explanation. A decision to withdraw, or a decision not to take part, will result in all of your data collected to that point being deleted.

What will happen to me if I take part?

If you decide to take part, you will be asked to attend the Biomechanics Laboratory at the School of Sport and Exercise Sciences, University of Kent, Medway Campus on a single occasion.

d) For healthy, physically active individuals;

The testing session involve investigating your respiratory function through a breathing assessment and an exercise challenge test using three different measurement methods.

The visit will last approximately 3 hours in total.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and your sport or physical activity.

- **Part 2.** Exercise Challenge: You will undergo a staged incremental exercise challenge to the point where you cannot continue to match the resistance of the bike. The exercise begins at a relatively light intensity (50 W) and increases every minute until you cannot keep going.

  The test will begin with a few minutes of rest which will involve breathing in and out normally during quiet sitting, to obtain a baseline for all measurements. Following the period of rest, you will then complete a warm up and then the exercise challenge. The intensity of the exercise challenge will be increased in controlled increments until you cannot match the resistance of the bike. The exercise challenge should not last longer than approximately 20 minutes. The exercise challenge will then be followed by another period of recovery in a seated position. All measurements will be obtained prior, during and after the exercise challenge simultaneously.

- **Part 3.** Powerbreathe Intervention:
  Once recovered from the exercise challenge participants will be trained how to use the Powerbreathe device and perform different breathing patterns. You will be asked to perform 30 breaths using a Powerbreathe device, you will receive verbal and visual feedback. You will then perform 6 randomised trials using the Powerbreathe device using 3 different breathing patterns, 3 trials without resistance and 3 trials with resistance. Powerbreathe guidelines will be followed at all times.

e) For individuals with dysfunctional breathing;

The testing session involve investigating if you have asthma/exercise induced bronchoconstriction though an Eucapnic Voluntary Hyperventilation (EVH) challenge. If negative, your breathing function will then be assessed at rest and during an exercise challenge. You will then complete an acute intervention and repeat the same exercise challenge.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and physical activity.

- **Part 2.** EVH (Eucapnic Voluntary Hyperventilation) challenge: A test to assess the health of your airways at rest and following high intensity breathing. If your lung function reduces following the EVH test it is suggested that you may suffer from a form of asthma known as exercise induced bronchoconstriction. To complete the EVH challenge you will be asked to breathe through the mouthpiece at approximately 85% of your maximal breathing rate (similar to breathing during high intensity exercise) for 6 minutes. You will breathe a cold, dry gas mixture that contains 21% O_2, 5% CO_2, 74% N_2. After the 6-minute high intensity breathing, your maximal lung function will be measured at 3, 5, 7, 10 and 15 minutes. If you have had a 10% fall during this test, you will be advised to contact your GP and you will be provided with a report to allow you to discuss the results with them). If you do not demonstrate a 10% fall during this test, you will continue to part 3 (see below).

- **Part 3.** Exercise Challenge: You will then undergo a staged incremental exercise challenge to the point where you cannot continue to match the resistance of the bike. The exercise begins at a relatively light intensity (50 W) and increases every minute until you cannot keep going.

  The test will begin with a few minutes of rest which will involve breathing in and out normally during quiet sitting, to obtain a baseline for all measurements. Following the period of rest, you will then complete a warm up and then the exercise challenge. The intensity of the exercise challenge will be increased in controlled increments until you cannot match the resistance of the bike. The exercise challenge should not last longer than approximately 20 minutes. The exercise challenge will then be followed by another period of recovery in a seated position. All measurements will be obtained prior, during and after the exercise challenge simultaneously.

- **Part 4.** Acute Intervention
  Once fully recovered from the exercise challenge, you will be given visual real-time biofeedback from the OEP system and verbal feedback to improve their breathing pattern, and will repeat the same exercise challenge.

For individuals with asthma, the testing session will involve investigating your respiratory function through an EVH challenge.

The visit will last approximately 1 hour in total.

- **Part 1.** You will be asked to fill out questionnaires prior to testing that asks questions about your general health and physical activity.

- **Part 2.** EVH (Eucapnic Voluntary Hyperventilation) challenge: A test to assess the health of your airways at rest and following high intensity breathing. If your lung function reduces following the EVH test it is suggested that you may suffer from a form of asthma known as exercise induced bronchoconstriction. To complete the EVH challenge you will be asked to breathe through the mouthpiece at approximately 85% of your maximal breathing rate (similar to breathing during high intensity exercise) for 6 minutes. You will breathe a cold, dry gas mixture that contains 21% O_2, 5% CO_2, 74% N_2. After the 6-minute high intensity breathing, your maximal lung function will be measured at 3, 5, 7, 10 and 15 minutes. If you have had a 10% fall during this test, your GP will be notified and we will recommend you to discuss the results with them.)

All participants in each group will be required to wear a facemask to allow us to measure the gas you are breathing out. OEP (optoelectronic plethysmography) is a non-invasive 3D motion capture technique which measures the movement of the chest wall using 11 infrared cameras and 90 reflective markers placed on the chest, abdomen and back. The markers will be placed in a grid-like pattern onto the torso using hypoallergenic bi-adhesive tape. Participants will be required to be shirtless during the OEP data collection with females wearing a sports bra only. OEP data will be collected during each stage of the testing.

What are the possible benefits of taking part?

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Will my taking part in this study be kept confidential?

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If I take part in this study, how will you protect my privacy?

The findings may be published in scientific journals or presented at conferences, but this would not include any information that would let others know who you are.

What are the risks and possible discomforts from being in this study?

- The risks associated with the exercise test are fatigue, muscle soreness, irregular heartbeat, chest pain and sudden heart attack. To minimize these risks, you will undergo a warm-up exercise prior to the starting of testing. You will have your heart rate and the rating of perceived exertion monitored continuously throughout the test. The test will be stopped at any point if we think that you are either (a) having difficulty breathing or (b) incurring any other form of distress. Also, the test will be terminated if the participant feels that either of the above conditions is occurring. Immediate first aid will be provided by qualified individuals in the unlikely event that something adverse occurs.

- EVH challenge has the potential to provoke severe narrowing of your airways. You will not be able to take part in the study if your resting lung function appears reduced (70% of predicted). Your lung function will be closely monitored during the EVH challenge. If your lung function falls below 10% of your baseline level you will be offered inhaled medication (salbutamol) that will reverse the reduction in lung function. You will be advised not to leave the laboratory until your lung function is within 10% of your resting value.

How long does the study participation last for?

In total, your involvement will be no more than 3 hours over one testing session.

Prior to arriving for the testing session please observe the following restrictions

13. No high intensity exercise 24 hours before each visit.
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15. No alcohol within 24 hours before the visit.
16. No sports drinks or caffeine intake 4 hours prior to the test.
17. Free from illness or infection in the two weeks before the test.
18. Drinking water is permitted.

Who is organising and funding the research?

This research study is being organised by the School of Sport and Exercise Sciences at the University of Kent.

What will happen to the results of the research?

We will analyse all of the anonymous results to help us decide how to investigate our research question further. We may report group results in academic journals or conferences.

If I have questions or concerns about this study, who can I contact?

You can contact the principal researcher listed below for questions specifically related to this study.

Carol Smyth at the School of Sport and Exercise Sciences, University of Kent (email:cmes2@kent.ac.uk)

What do I do now?
If you understand the information given in this form and are happy to participant in the research study, then please fill in the attached consent form.

Athlete Information – Eucapnic Voluntary Hyperpnoea Challenge

Prior to Arriving for the Test

- If you are using any asthma medication you will need to stop using it prior to the testing as follows*:
    - Montelukasts 7 days prior to testing
    - Brown and Purple inhalers (Inhaled corticosteroid and combination inhalers) at least 72 hours prior to testing
    - Green inhalers (long acting β_2 -agonists) 42 hours prior to testing
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- Bring a drink of water with you

* In an emergency, you should take short acting Beta 2 Agonist (Salbutamol, Ventolin), terbutaline (Bricanyl). You MUST inform the tester of nay medication not conforming with this protocol.

Test Day

- You will be given questionnaires to complete which will ask name, age, height, weight, and past asthma history and respiratory symptoms.
- Your maximal lung function will be measured
- You will complete a 6 minute heavy breathing exercise called a Eucapnic Voluntary Hyperpnoea (EVH) challenge, which is explained below
- 3, 5, 7, 10, and 15 minutes after the EVH challenge your maximal lung function will be measured.
- If your lung function drops 10% or more following the EVH test, it is deemed a positive test. If this happens you will be offered a reliver inhaler (salbutamol) and have your maximal lung function measured 10 minutes later.
- You will be advised not to leave the lab until your function is within 10% of what it was when you started the test
- In total the test will last approximately 45 minutes

EVH challenge

- Maximal lung function is measured
- A target breathing rate is calculated based on your maximal lung function
- You will be asked to breath at this target breathing rate for 6 minutes. The breathing rate will be similar to the rate you breathe at during high intensity running.
- The air you breathe during the test will be cold and dry. It is a special gas designed to ensure to do not have any adverse effects during the test.
- After the 6 minutes of high intensity breathing your maximal lung function will be measured at 3, 5, 7, 10, and 15 minutes.