Methodological considerations for the assessment of postural stability and lower limb bilateral asymmetry

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Abstract

Purpose: There are currently no agreed methods for the assessment of postural stability using centre of pressure (CoP) analysis of quiet standing nor assessment of lower limb bilateral asymmetry measured during a countermovement jump (CMJ). Much of the existing literature surrounding both of these biomechanical assessments are varied and inconclusive in the determination of a criterion methodology. There is also a dearth of information regarding the reliability of both measures or expected outcomes. Therefore, the purpose of the current study was twofold. Firstly, to assess the methodology and reliability of postural stability measures obtained from a force-platform. Secondly, to investigate the methodology and reliability of measuring lower limb bilateral asymmetry using a dual-force-platform set-up.

Methodology: Using a repeated measures design of test-retest reliability, postural stability and CMJ performance was assessed for male (n = 10, age = 32.7 ± 9.5 yrs., height = 1.797 ± 0.060 m, mass = 88.2 ± 14.4 kg) and female (n = 9, age = 32.4 ± 8.7 yrs., height = 1.662 ± 0.055 m, mass = 70.8 ± 13.5 kg) recreationally active individuals divided into three populations, female-only (FEM), male-only (MALE) and combined (ALL). For postural stability measurement, path length (L_p), sway area (A_s) and mean velocity (V_m) were reported from 8 trials for six epochs derived from 100 s of quiet standing. Four trials of each condition, were conducted on each of the two separate testing days. Reliability of bilateral CMJ performance was assessed from ten maximal CMJ trials using five kinetic and two temporal neuromuscular variables: peak force (F_{max}), impulse due to eccentric and concentric contraction (J_{ecc} and J_{con}), peak instantaneous mechanical power (PPO), take-off velocity (V_{to}), percentage of jump duration that changeover from eccentric to concentric phases occurs (t_{ecN}) and percentage of jump duration that peak force occurs (t_{FmaxN}). Lower limb bilateral asymmetry was then calculated for F_{max}, J_{ecc}, J_{con} and t_{FmaxN} using two SIs; sided, left leg vs right leg (LvsR) and un-sided, higher vs lower limb value (HvsL) to give asymmetry irrespective of limb side. Differences between conditions and SI methods were identified using paired-samples t-tests and test-retest reliability was assessed using ICC and Bland and Altman (B&A) plot analysis.

Postural Stability Results: L_p and V_m were significantly higher ($p \leq 0.05$) in the EC condition for all epochs. A_s demonstrated differences between the two conditions, however, not always significant; in all cases of significance, A_s was greater in the EC condition. Absolute ICC values for L_p and V_m were indicative of excellent reliability (>0.90) however, 95% CI ranged from poor (< 0.50) to excellent. A_s was the least reliable variable with regards to ICC 95% CIs, although absolute ICC values were still good (>0.75) to excellent across conditions and epochs. B&A plot analysis showed A_s was the most variable. In general, results showed that EC had the higher test-retest reliability, however differences between ICC values and the magnitude of the bias and LOA between conditions were small. It was not clear which epoch provided the most, or least, reliable results for L_p or V_m. For A_s, the 1st 30 s had the most variability, while for all variables, 90 s was one of the most reliable epochs. Cumulative moving average analysis showed a trend toward increased precision as number of repetitions increased for all epochs.

Bilateral Asymmetry Results: Kinematic variables derived from analysis of a CMJ resulted in high test-retest reliability and agreement (ICC > 0.9) for F_{max}, J_{ecc}, J_{con}, PPO, V_{to} and t_{ecN}. LvsR and HvsL methods of SI calculation were significantly different ($p \leq 0.05$) for 3/4 variables. F_{max} (ALL: LvsR - 0.72 ± 6.96%, HvsL 6.22 ±

3.89%), J_{ecc} (ALL: LvsR 4.99 ± 23.47%, HvsL 25.29 ± 12.33%), J_{con} (ALL: LvsR -6.93 ± 26.87%, HvsL 25.47 ± 13.80%), t_{FmaxN} (ALL: LvsR -2.10 ± 7.87%, HvsL 5.67 ± 6.83%). Overall, the absolute ICCs of J_{ecc} and t_{FmaxN} ranged from poor to excellent, while F_{max} and J_{con} showed better agreement, although 95% CI ranges and magnitude of B&A LOAs were still large, particularly in LvsR (e.g., F_{max}, ALL LvsR: Bias = 32%, LOA = 352%). B&A plot analysis demonstrated far smaller bias and LOA in HvsL than LvsR for all variables and populations.

In both postural stability and bilateral asymmetry, there was no substantial differences noted between the reliability of FEM and MALE populations. For both cases, greater reliability could be seen for the majority of variables when FEM and MALE were combined in the ALL population.

Conclusions: L_p and V_m had better reliability and lower variability than A_s. A_s is not recommended as a reliable postural stability performance parameter. There was a significant difference between visual conditions, indicating the impact of visual acuity on human postural control. Both EO and EC showed good reliability for all epochs, although B&A plots revealed variability in the data that should be considered in future research. Although EC appeared to be the slightly more reliable condition, it cannot be recommended over EO as they are representative of different requirements of human postural control. Bilateral CMJ performance showed good test-retest reliability, however, normalised temporal variables should be used with caution; t_{FmaxN} was the least reliable variable. LvsR and HvsL methods of SI calculation were significantly different and have the ability to quantify very different inherent characteristics of bilateral CMJ performance. Results identify the importance of determining a suitable set of reference values and the consideration of the directionality of asymmetries on an individual basis. In a bilateral CMJ, the differences in the force generating capacity between limbs does not necessarily determine the variation in the magnitude of VGRF generated during the jump. Instead, variations in VGRF symmetry should be considered to represent bilateral variations in limb loading that stem from differing jumping and compensatory strategies adopted by individuals.

Author Declaration and statements

1. I, Jordan Groom, hereby declare that the work presented in this thesis has not previously been accepted in substance for any degree and is not being concurrently submitted in candidature for any degree.

2. I, Jordan Groom, declare that the thesis is the result of my own investigations, except where otherwise stated and that other sources are acknowledged by footnotes giving explicit references. A bibliography is appended.

3. I, Jordan Groom, hereby give consent for the thesis, if relevant and accepted, to be made available for photo copying and for inter-library loan, and for the title and summary to be made available to outside organisations.

4. I, Jordan Groom, declare that the University's ethical procedures have been followed and, where appropriate, that ethical approval has been granted.

Signed: [redacted]

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Acknowledgements

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Nomenclature and Abbreviations

ACL	Anterior Cruciate Ligament
A-D	Analogue to Digital Converter
ALL	The All-participants Study Population
A/P	Anterior/ Posterior
A_s	Sway Area
AS	Arm Swing
ASI	Asymmetry Index
B&A	Bland and Altman
BM	Body Mass
BW	Body Weight
CI	Confidence Interval
cm	Centimetres
$cm \cdot s^{-1}$	Centimetres per Second
CMJ	Countermovement Jump
CNS	Central Nervous System
CoG	Centre of Gravity
CoM	Centre of Mass
CoP	Centre of Pressure
COVID-19	Coronavirus Disease 19
CV	Coefficient of Variance
DIP	Double-Inverted Pendulum
EMG	Electromyography
EC	Eyes Closed
EO	Eyes Open
F	Female
F	Force
F_{max}	Peak Force
F_x	Medial-Lateral Component of Ground Reaction Force
F_y	Anterior-Posterior Component of Ground Reaction Force
F_z	Vertical Component of Ground Reaction Force
FEM	Female-only population

FP	Force Platform
FPT	Functional Performance Test
g	Gravity
G	Gravitational Constant
GRF	Ground Reaction Force
HvsL	High vs Low
Hz	Hertz
ICC	Intraclass Correlation Coefficient
IT	Ischial Tuberosities
J	Impulse
J_{con}	Concentric Impulse
J_{ecc}	Eccentric Impulse
JH	Jump Height
kg	Kilograms
kN	Kilonewton
L_p	Path Length
LOA	Limits of Agreement
LvsR	Left vs Right
m	Mass
m	Metres
M	Male
MALE	Male-only population
M/L	Medial-Lateral
mm	Millimetres
mm^2	Millimetres Squared
$m \cdot s^{-1}$	Metres per Second
N	newton
NAS	No Arm Swing
$N \cdot s$	newton Seconds
PAR-Q	Physical Activity Readiness Questionnaire
PS	Perturbed Standing
PS	Postural Stability
PSD	Power Spectral Density

P_x	x Component of Centre of Pressure Coordinates
QS	Quiet Standing/ Unperturbed
R	Resultant Force
RAMP	Raise, Activate, Mobilise, Potentiate
RE	Rowing Ergometer
RFD	Rate of Force Development
RMS	Root Mean Square
RoM	Range of Motion
s	Second(s)
SD	Standard Deviation
SEM	Standard Error of the Mean
SI	Symmetry Index
SIP	Single-Inverted Pendulum
SJ	Squat Jump
SSC	Stretch-Shortening Cycle
t_{ec}	Instant of Eccentric to Concentric Changeover
t_{ecN}	Instant of Changeover from Eccentric to Concentric Normalised to Jump Duration
t_{Fmax}	Instant that Peak Force Occurred
t_{FmaxN}	Instant of Peak Force Normalised to Jump Duration
t_i	Instant of Jump Initiation
t_j	Jump Duration
t_{to}	Instant of Take-off
TIP	Triple-Inverted Pendulum
V_m	Mean Velocity
V_{to}	Take-off Velocity
VGRF	Vertical Component of Ground Reaction Force
VOR	Vestibulo-Ocular Reflex
VSR	Vestibulo-Spinal Reflex
W	Watts
W	Weight
WHO	World Health Organisation
3-D	Three Dimensional

Impact of the COVID-19 pandemic

Declared a pandemic by WHO in March 2020 (WHO, 2020), the COVID-19 outbreak caused the sudden closure of campuses as a social distancing measure to reduce interpersonal contact and, thereby, minimise the community transmission that could develop in dense social networks such as university campuses (Weeden & Cornwell, 2020). As a consequence, there was a significant restriction on the number of participants able to be recruited for the current study and, further, the demographic of those participants. The initial design of the current study was to investigate postural stability and bilateral asymmetry within the context of rowing related performance, however, due to restrictions put in place as a result of the coronavirus pandemic, no data collection was able to be completed on campus, or in the field, for over a year. When lab usage was then able to resume, very late into the academic year, the restrictions still in place prevented the inclusion of any participants outside of the postgraduate and staff offices which, consequently, meant no members of the rowing population were able to be recruited. This forced the direction of the study to move to researching a general population whom participate in sport on a recreation basis, with the hope of providing insight and recommendations that could later be applied to the desired populations.

Chapter 1

Thesis Introduction

1.1 Introduction to postural stability

The term balance, or equilibrium, refers to the state of an object when the resultant forces or moments acting upon it are equal to zero (Bell, 1998, pp. 26). In order for an object to remain balanced, the centre of gravity (CoG), and thus its line of gravity, must intersect the plane of the base of support (Watkins, 2014, pp. 240). With human balance, when the line of gravity falls outside the base of support, the human body has the inherent ability to sense the threat to its stability and to counteract in order to prevent falling (Horak, 1987 in Pollock, Durward, Rowe & Paul, 2000). The body is, therefore, never in a position of zero net force, and, because the human body is not a rigid structure, it is continuously performing oscillations about a vertical axis during standing. Postural control within the body has been modelled as an inverted pendulum (Günther & Wagner, 2016) with intermittent control strategies (Loram, Gollee, Lakie & Gawthrop, 2011) and in order to maintain postural equilibrium, the body continuously adopts changing neural strategies and reweighting of the sensory inputs (Kabbaligere, Lee & Layne, 2017; Paillard, 2012; Peterka, 2002).

In order for postural stability to be assessed in any population, it needs to be measurable, and, therefore, quantifiable. The assessment of postural control is commonly undertaken in clinical settings for a wide range of pathologies, with performance in prescribed tasks linked to factors such as physical function and fall risk (Butler, Lord & Fitzpatrick, 2011; Kerr *et al*., 2010; Melzer, Benjuya & Kaplanski, 2004; Weerdesteijn, Niet, Van Duijnhoven & Geurts, 2008). The application of postural stability research to athletic populations has been more limited, as primarily attention has been focussed on the distinction of neurological or neuromuscular dysfunction in clinical research, although it has been used in athletic populations to assess injury risk (McGuine, Greene, Best & Lverson, 2000) and to determine performance variations between athletes and control groups, or between athletes of different sports (Hammami, Behm, Chtara, Othman & Chaouachi, 2014).

The interest of researchers in the assessment of postural stability has led to the development of a variety of techniques; such as functional assessments, sway magnetometry and force platform (FP) derived measures (Browne & O'Hare, 2001; Murray, Seireg & Sepic, 1975). Thus, measurement tools for assessing postural control can range from simple time-based assessments to more 'in-depth' analysis through the measurement of kinematics, kinetics or muscle activity (Horak & Nashner, 1986;

Horak, Henry & Shumway-Cook, 1997; Nashner & McCollum, 1985; Kuo & Zajac, 1993; Visser, Carpenter, van der Kooij & Bloem, 2008). With this, the most common method of quantification, in well-resourced research, is the use of centre of pressure (CoP) coordinate movement analysis, derived from the forces and moments measured by a fixed FP .

While the evaluation of CoP excursions is a commonly used method for postural stability assessment, there is a vast amount of heterogeneity in the methodologies used as a result of an absence of standardisation of the method, nor the processing and interpretation of outcomes. Hence, there is very little consensus on an optimal, or criterion method, nor is there any definitive agreement on the most valid or reliable CoP parameters for the accurate understanding and quantification of postural stability (Caballero, Barbado & Moreno, 2015). Consequently, the ability to compare study outcomes becomes increasingly difficult on account of varied populations, protocols and analytical approaches. There are, however, studies that have investigated the reliability and suitability of individual methodological components in order to advise future research. For example, with respect to protocol structure, recommendations have been made for decisions such as sampling duration (Carpenter, Frank, Winter & Peysar, 2001; van der Kooij, Campbell & Carpenter, 2011) and number of trial repetitions (Golriz, Herbert, Foreman & Walker, 2012; Lafond, Corriveau, Hébert & Prince, 2004), while other studies have looked at the determination of the most appropriate acquisition settings, such as sampling and filtering frequencies (Schmid, Conforto, Camomilla, Cappozzo & D'alessio, 2002). Yet, because these methodological considerations and investigations are made in isolation from one another, the combination of variables yielding the most reliable or valid results is unknown, as a study investigating one factor, may do so using a methodology later deemed by another researcher to be inappropriate. Ruhe, Fejer and Walker (2010) and Hérbert-losier and Murray (2020) provide systematic reviews for the generation of future recommendations, however, even then, these are the summation of extensively variable literature.

1.2. Introduction to lower limb bilateral asymmetry

The performance of an athlete can be assessed in a number of qualitative and quantitative ways, such as the quality of movement and techniques, along with numerical values such as time to complete an activity or the load and/or volume of an exercise or activity an athlete can complete. A large amount of sports specific research involves neuromuscular performance, as it offers a unique ability to integrate biomechanics, muscle physiology and neurophysiology (Komi, 1984). The measurement of neuromuscular performance has a number of purposes within sport, such as quantification of training status, youth talent identification, strength and power diagnosis and injury identification and prevention (Owen, Watkins, Kilduff, Bevan & Bennett, 2014). The neuromuscular performance of an individual is underpinned by a number of anatomical and neuromuscular factors; these variables are commonly assessed by researchers in an attempt to understand their relationship with performance, as well as being used to define performance itself. Performance in vertical jumping is a widely used metric in assessing sporting performance indicators and training status by both recreational and professional athletes (Davis, Briscoe, Markowski, Saville & Taylor, 2003; Ziv & Lidor, 2010) and its use as a method of neuromuscular assessment in athletes is valuable as vertical jumping contributes in varying degrees to performance in most sports (Harman, Rosenstein, Frykman & Rosenstein, 1990). Using measures derived from a FP, i.e. force-time histories, has become the criterion, or 'gold standard', for the determination of neuromuscular performance variables such as peak force, velocity of the body's CoG, rate of force development and mechanical power from a countermovement jump (CMJ) (Davies & Rennie, 1968; Harman, Rosenstein, Frykman, Rosenstein & Kraemer, 1991; Hatze, 1998; Kibele, 1998; Sayers, Harackiewicz, Harman, Frykman & Rosenstein, 1999; Vanrenterghem, De Clercq & Cleven, 2001; Canavan & Vescovi, 2004; Owen *et al.*, 2014).

A more recent field of interest in neuromuscular performance assessment within both clinical and sports settings is bilateral asymmetry (Afonso *et al.,* 2020). This refers to the deviation from mirror symmetry across the coronal axis (Maloney, 2019) and is most often used to describe muscular imbalance. From a biomechanical perspective, kinetic asymmetry would refer to deviations in force production or rate of force development between mirrored limbs. Lateral preference, previous injury, or specific sport demands, have been suggested as possible reasons for the development of bilateral muscle strength imbalances, with several studies showing side-to-side strength imbalances to be present in well-trained athletes (Gioftsidou *et al.,* 2008; Impellizzeri, Rampinini, Maffiuletti & Marcora, 2007; Newton *et al.,* 2006).

The most common method of kinetic asymmetry assessment of the lower limbs is the CMJ, where neuromuscular variables are derived from a force-time history, measured using a force platform (FP). However, critical analyses of their effectiveness for measuring inter-limb differences and clear guidelines for their implementation are sparse (Bishop, Turner, Jarvis, Chavda & Read, 2017).

As with postural stability assessment, there has been little consistency within the literature regarding methodological consideration, in regards to both applied protocols and analytical approach. For example, in the calculation of lower limb asymmetries, a variety of approaches have been used, defining limb differences, or symmetry index (SI), in terms of dominance, strength, preference, or simply a right or left distinction (Bishop, Read, Chavda & Turner, 2016). The suitability of any one of these methods is, however, unclear as a consequence of the dependency of valid results on the chosen 'reference value' (Zifchock, Davis, Higginson & Royer, 2008). In addition, research has highlighted poor levels of agreement between the limb identified as generating the greatest value/ performance, and perceived limb dominance (Fort-Vanmeerhaeghe, Gual, Romero-Rodriguez & Unnitha, 2016). Menzel *et al.,* (2013) was a novel study that used more than one method of SI determination and, in doing so, have provided a valuable example of how different methods produce variable results; 'sided' analysis comparing left to right was noted to produce a mean SI of -0.69% while 'absolute' asymmetry, where the direction of the asymmetry is removed, produced a much larger value of 5.58% SI. It is clear from the existing research that an appropriate data acquisition protocol needs to be identified in order to improve the confidence that can be had in the validity, and reliability, of study outcomes. Furthermore, research into the consequence of SI calculation choices will be fundamental in identifying the most appropriate methodology for future application, both sporting and clinical.

1.3 Aims of the current study

There are currently no agreed methods for the assessment of postural stability using CoP analysis of quiet standing nor assessment of lower limb bilateral asymmetry measured during a countermovement jump (CMJ). Much of the existing literature surrounding both of these biomechanical assessments are varied and inconclusive in the determination of a criterion methodology. There is also a dearth of information regarding the reliability of both measures or expected outcomes. Therefore, the purpose of the current study was twofold. Firstly, to assess the methodology and reliability of postural stability measures obtained from a force platform. Secondly, to investigate the methodology and reliability of measuring lower limb bilateral asymmetry using a dual-force-platform set-up. With that, the aims of the current study were:

- Determine the test-retest reliability of commonly used variables derived from CoP measurements using a methodology reported from the recommendations of previous research.
- Assess the test-retest reliability of neuromuscular performance variables when using the criterion method for the determination of peak mechanical power output in a CMJ.
- Assess the contribution of left and right legs to the execution of a bilateral CMJ using a dual-platform methodology.
- Determine the reliability and validity of both 'sided' and 'un-sided' methods of symmetry index (SI) calculation.
- Identify differences in reliability between study populations of different sexes, and groups combined for both postural stability and asymmetry assessments.
- Provide recommendations for the directions of future research.

1.4 Hypotheses

1. There is no criterion method for the determination of postural stability performance, however, a methodology devised from the recommendations of previous research will produce valid and reliable results.
2. As sampling duration increases, so will the test-retest reliability of postural stability parameters derived from CoP analysis.
3. As the number of repetitions increases, as will the precision and validity of postural control outcome measures.
4. Visual condition will have a significant impact on postural stability performance. Further, CoP excursion measures will be significantly higher when visual feedback is removed.
5. Bilateral CMJ performance, defined using the criterion method for the determination of neuromuscular variables, will show excellent test-retest reliability and day-to-day agreement.
6. In the execution of a bilateral CMJ, there will be differences in the contribution of each individual limb to overall performance. These variations in the magnitude of vertical component of the ground reaction force are the result of differences in the force generating capacity between limbs.
7. 'Sided' asymmetry calculations will produce statistically significantly different results to 'un-sided'. Additionally, 'sided' methods of asymmetry calculation will have lower reliability and validity than 'un-sided' methods quantifying absolute asymmetry, due to the consequences of nullification when producing group means.
8. Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population.

1.5 Thesis Structure

The structure and content of this thesis is two-fold. Each chapter has been divided into the two topics of interest, postural stability and lower limb bilateral asymmetry. Chapter 1 has introduced the context of the study and has identified the purpose and aims. Chapter 2 will review the biomechanical principles applied and existing literature for both research topics, focusing on the methodological considerations for their assessment. Chapter 3 will present how the research was conducted with a focus on detailing all methodological steps in a way that would allow the research to be replicated. Chapter 4 will state the current study's findings and provide additional tables and figures to form a detailed analysis of postural stability parameters as well as the assessment of countermovement jump performance and the individual lower limb contributions. Chapter 5 will critically analyse the findings of the study while providing comparisons, where possible, to existing research to discuss the contributions this thesis makes to the literature. Limitations of the study will also be discussed. Finally, Chapter 6 will state the conclusions of the research and provide some recommendations for the direction of future research in order to improve the understanding of postural stability and lower limb asymmetry analysis and interpretation.

Chapter 2

Review of Biomechanical Foundations and Literature

2.1 Reliability, validity, accuracy, precision and measurement error

2.1.1 Reliability

Reliability is defined as 'the consistency of measurements, or of an individual's performance, on a test' (Atkinson & Nevill, 1998). Reliability can, therefore, also be described as the repeatability of a test or the ability of a test, or individual, to produce similar results on separate occasions. An example of human reliability would be an athlete performing the same event on two different occasions (under the same conditions) and producing very comparable times. Due to the external environment and any additional control variables, such as time of day or nutrition for example, being maintained consistent between the two events, the athlete that has produced the two sprint times can be considered as a reliable runner. Equally, if an athlete performs two sprints under the same conditions but produced significantly different times, they could be considered as unreliable. Reliability is also applicable to the instruments used for assessment; whether an instrument produces the same results on repeated occasions is indicative of its reliability. Terms that have been used interchangeably with 'reliability', in the literature, are 'repeatability', 'reproducibility', 'consistency', 'agreement', 'concordance' and 'stability' (Atkinson & Nevill, 1998).

2.1.2 Validity

Validity in scientific testing is, generally, the ability of the measurement tool to reflect what it is intended to measure (Atkinson & Nevill, 1998) or the extent to which a test compares to the criterion, or 'gold standard'. Although there are different types of validity (logical, content, criterion, and construct), in the presence of a 'gold standard' measure it is particularly useful to establish criterion validity, which assesses the extent to which scores on a test are related to some recognised standard (Thomas, Nelson, & Silverman, 2005).

An example of this would be in the measurement of an individual's height. It could be suggested that an estimation of height can be made based on shoe size; the greater the shoe size, the taller the individual. However, the testing of this theory identifies that shoe size has limited correlation to height. Therefore, estimating height based on shoe size would not be a valid measure and would not be appropriate. A stadiometer, however, is the criterion method for measuring height as height, or stature, is measured in normal units of lengths. This criterion method measures what is needed to be measured in the correct units. With this, a tape measure could also be considered a valid measure; it may not be the criterion method but it is a direct measurement that is taken in the correct units. In the same sense, a stopwatch is a valid measure for timing sprints, however, it can have limitations with respect to accuracy, precision and reliability. Therefore, the reliability of a person completing a sprint on two separate occasions will rely (1) on the athlete's ability to reproduce a consistent time and (2) on the equipment that will be measuring those times.

2.1.3 Accuracy and precision

Accuracy and precision refer to the degree to which a test, or device, measures the true value of what it attempts to measure. Accuracy, also defined as 'the conformity to a recognised standard or as a degree of approximation to the true value, to a desired or required result' (Strauss *et al*., 2006) is expressed in the spread of measurements when they are repeated under different external circumstances. The differences between precision and accuracy mostly arise from systematic errors or from changes in external influences (Strauss *et al*., 2006). With this, accuracy relates to the quality of a result and is, therefore, not equal to precision; precision focuses on the stability of the measuring device or it's reading during the measuring process itself (Strauss *et al*., 2006).

Accuracy and precision can be demonstrated using a stopwatch. If the stopwatch was stopped at real-time values of 9.9 s or 10.1 s, the stopwatch would display a time of 10 s (which would be precise as that is the internal accuracy of the device) however, it would not be accurate as it's not a true representation of the true value. The important factor to consider is the desired, or required, level of accuracy for the test being conducted; if it is necessary to get a result to ± 1 s of the true value then this stopwatch is sufficient and is considered to be both precise and accurate. If, however, the period of time needed to be measured is within ± 0.1 s or smaller, the stopwatch is no longer suitable as although it may be precise, it does not have the required level of accuracy.

The accuracy of equipment can be improved by using more finely incremented tools that require less estimation. A stopwatch measuring in 1 s increments inherently has a lot of estimation. For example, a reading of 10 s would be representative of a true value between 9.50 s and 10.49 s. If the precision of the stopwatch used was increased to increments of 0.1 s it would be possible to display a value closer to the true value, as a display value of 10.0 s would now be representative of an estimation between 9.950 s and 10.049 s. This effect of increasing precision on accuracy is demonstrated in Figure 2.1.

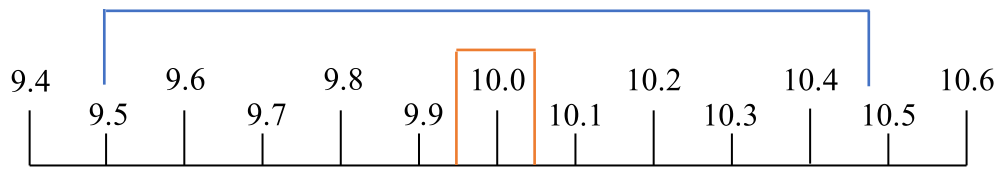

Figure 2.1 *The effect of increasing the precision of a stopwatch on the device's ability to measure the true value; whereby 10.0 seconds = true value, blue area = stopwatch that measures to 1 whole second, orange area = stopwatch that measures to 0.1 seconds.*

For research or coaching purposes, it is important to consider the level of accuracy and precision needed. For example, when monitoring an elite athlete who produces a consistent and reliable result in a given test, it may be necessary to increase the level of accuracy in measurement techniques in order to more clearly identify differences in their performance. A 100 m race, for example, could not be measured with a timer of accuracy = ± 0.1 s as the difference between one athlete and the next could easily be less than 0.1 s; in the Tokyo 2020 Olympics, for example, the difference between 1st and 3rd place in the men's final was < 0.1 s (9.80 s and 9.89 s, respectively) (Olympics, n.d.).

2.1.4 Systematic bias and random error

Measurement error is the amount of error inherent in a device. As a result, it cannot, therefore, be controlled, however, it can potentially be accounted for. Using the example of a stopwatch again, measurement error in the device could cause the time on the display to show 10.00 s, when in fact the true, real-time value is 9.75 s. This is termed systematic bias and is the result of an error in the device itself which, unless it can be identified and corrected, becomes uncontrollable.

Reliability and validity are considered the foundations of measurement because they represent attempts to reduce measurement error. Although it is impossible to eliminate all errors, it is possible to use the understanding of measurement error in designing research, analysing and interpreting data, and acknowledging limitations (Viswanathan, 2005, pg. xiii).

2.2 Mass, force, gravity and weight

2.2.1 Mass

The quantity of matter of which a body is composed is called its mass and is a direct measure of the inertia that the body possesses. The term inertia refers to the reluctance of a body to change its state of motion (Hay, 1993). The SI unit for mass is kilograms (kg).

2.2.2 Force

As described by Hay (1993, pg. 60 - 69), a body's state of rest or motion can be changed by the action of another body. The pushing or pulling action that causes the change is termed a force. The unit of force is the newton (N) and is defined in terms of the acceleration it produces and the effects of force can be described by Newton's three laws of linear motion. (1) *Every body continues in its state of rest or motion in a straight line unless compelled to change by external forces exerted upon it.* (2) *The rate of change of momentum of a body is proportional to the force causing it and the change takes place in the direction in which the force acts.* (3) *For every force that is exerted by one body on another there is an equal and opposite force exerted by the second body on the first.*

2.2.3 Newton's Law of Universal Gravitation

Newton's Law of Gravitation states that 'any two particles of matter attract one another with a force directly proportional to the product of their masses and inversely proportional to the square of the distance between their centres' (Hay, 1993). Equation 2.1 shows the algebraic expression of the universal law of gravitation with Figure 2.2 illustrating the principle.

$$F = G\frac{m_1 m_2}{r^2} \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 2.1.$$

where: F = attractive force between two bodies

G = gravitational constant (6.674 x 10^{-11} $m^3.kg^{-1}.s^{-2}$)

m_1 = mass of body 1

m_2 = mass of body 2

r^2 = square of the distance between the centre of masses of the two bodies

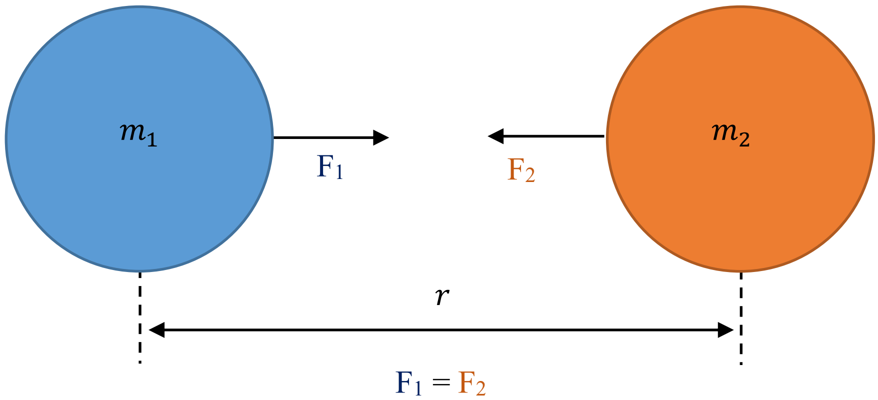

***Figure 2.2** Universal law of gravitation. Every point mass attracts every single other point mass by a force acting along the line intersecting both points. This force is proportional to the product of the two masses and inversely proportional to the square of the distance between their centres.*

In most cases between two objects, the total of all the attractive forces that particles of one body exert on the particles of any other body is generally so small that its effect is undetectable and the forces can be disregarded. However, one body whose effect cannot be disregarded, and makes Newton's law of gravitation of some significance, is the Earth. The attraction that the Earth has for all other bodies is known as gravity, and, as indicated by Equation 2.1, varies directly with the mass of the body involved and inversely with its distance from the centre of the Earth.

The gravitational constant, G, is a key quantity in Newton's law of universal gravitation and should not be confused with the local gravitational field of Earth, g. The two quantities are, however, closely related by:

$$g = \frac{GM_e}{{r_e}^2} \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 2.2.$$

where: g = local gravitational field

G = gravitational constant

M_e = mass of the Earth

r_e = radius of the Earth

The gravity of Earth, g, is the net acceleration that is imparted to objects due to the combined effect of gravitation (from mass distribution within Earth) and centrifugal force (from the Earth's rotation) (Boynton, 2001; Hofmann-Wellenhof & Moritz, 2006). The SI units for the measurement of acceleration due to gravity is metres per second squared ($m \cdot s^{-2}$) or equivalently newtons per kilogram ($N \cdot kg^{-1}$).

Both Equation 2.1 and 2.2 express the same principle that the acceleration due to gravity experienced by a body will vary depending on the distance between the centre of the Earth and that body. The nominal "average" value, referred to as standard gravity, is, by definition, 9.80665 $m \cdot s^{-2}$ (Thompson & Taylor, 2008) however, there is a variation in the magnitude of gravity at different altitudes- or rather different distances from the Earth's centre (varied r). For example, the gravity on the Earth's surface can vary by around 0.71% from 9.76392 $m \cdot s^{-2}$ on the summit of the Nevado Huascarán mountain in Peru (6768 m above sea level) to 9.83366 $m \cdot s^{-2}$ at the surface of the Arctic Ocean. In a study by Hirt *et al.* (2013) these two points were identified as candidate locations for Earth's minimum and maximum gravity acceleration.

2.2.4 Weight

The weight of a body on the Earth's surface (W) is simply defined as the gravitational force that the Earth exerts on a body. It can therefore be expressed as:

$$W = m \times g \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 2.3.$$

where: W = weight of a body measured in newtons (N)

m = mass of the body measured in kilograms (kg)

g = acceleration due to gravity measured in newtons per kilogram ($N.kg^{-1}$)

Thus, an individual who experiences a gravitational force of 700 N is said to have a weight of 700 N; which, when the standard value for gravity is used, means that the person has a mass of 71.4 kg. Weight, then, is merely the name given to a particular force and, just as g stands for a specific acceleration due to gravity, W stands for a particular force (Hay, 1973). The difference between weight and mass, two quantities very often misunderstood and wrongly used synonymously, is that weight can vary slightly depending on its location/ altitude as a result of the consideration of gravity, while mass remains constant irrespective of location due to the mass of a body only referring to the quantity of matter that composes it and not the product of the body's environment.

2.3 Force transducers and force platforms

In order for a force exerted by the body on an external body or load to be measured, a suitable force-measuring device is required. A force transducer is a device that gives an electrical signal proportional to the applied force and work on the principle that applied force causes a particular amount of strain within the transducer (Winter, 2009, pg. 117).

A force platform (FP) is a metal or composite platform instrumented with four force transducers, one in each of the four corners which deform with loading (Figure 2.3). There are three main types of force transducer commonly used in the construction of FP's: piezoelectric, strain gauge and Hall effect sensors, however, we are primarily interested in a piezoelectric FP, as that is the form used in this research. Piezoelectric transducers are manufactured from naturally occurring quartz crystal which is designed such that any applied force acts directly on the crystal; the crystal responds with a small electric charge value and a high output resistance (Şerban, Sîrbu, Roşca, & Drugă, 2019). This change in resistance is measured by the change in voltage (Figure 2.4) and the resulting charge is proportional to the applied force (Owen, 2008).

Signals of forces measured by the transducers in a FP are not directly compatible with a digital computer and as a result additional signal conditioning is necessary in order to achieve an appropriate interface (Pohlmann, 2010). Each signal produced by a transducer is first converted into a voltage proportional to the original signal and then converted into a digital signal, via an analogue to digital (A-D) converter. Figure 2.5 demonstrates this process. Once the signal is in a digital form it can be processed, displayed and recorded using specialised software.

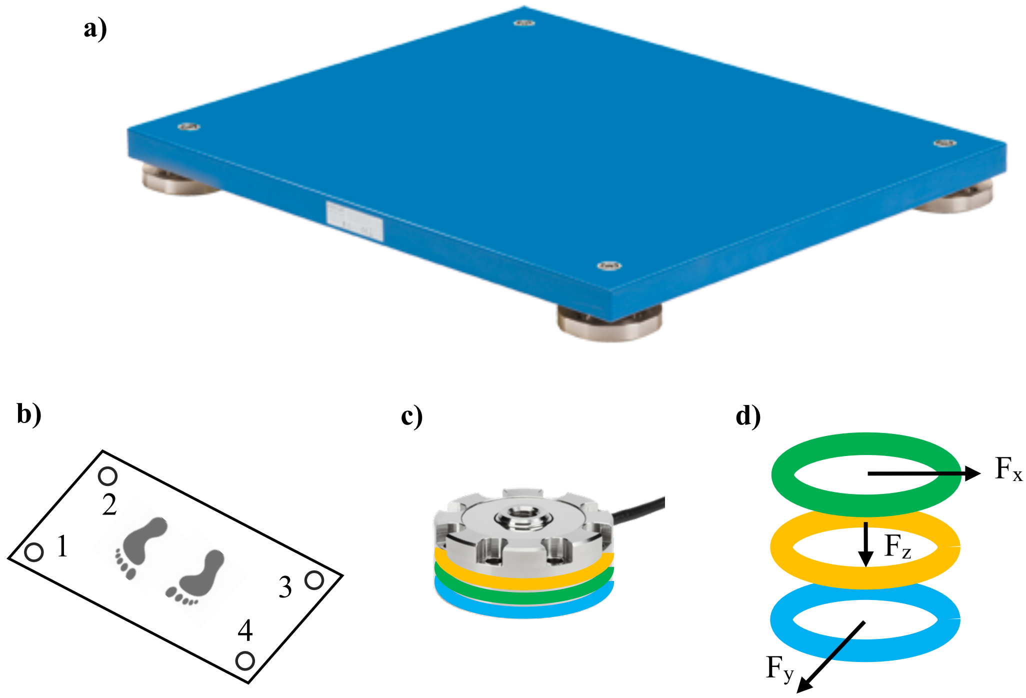

Figure 2.3 *Multi-component piezoelectric sensors for measuring the reaction forces on the x, y, and z axis. a) A type 9260AA Kistler 3D portable force platform. Sourced Kistler UK. b) Position of each of the four strain gauges. c) Single strain gauge showing discs representing piezoelectrical material. d) Discs of piezoelectrical material sensitive to compression forces (F_z) and shear forces (F_x, F_y). Adapted from Şerban, Sîrbu, Roşca, & Drugă (2019).*

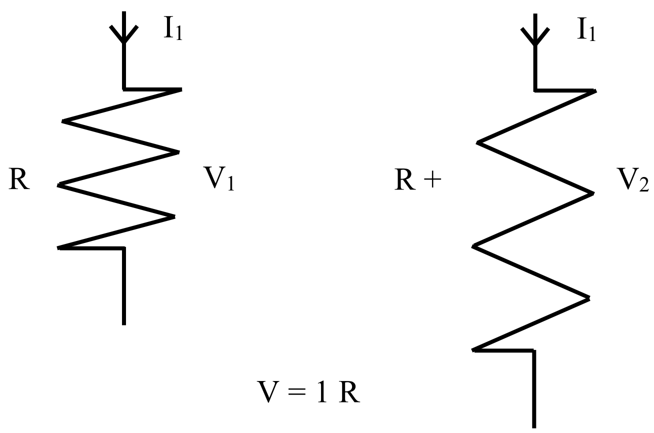

Figure 2.4 *Deformation of strain gauges results in change in resistance. Change in resistance is measured by change in voltage.*

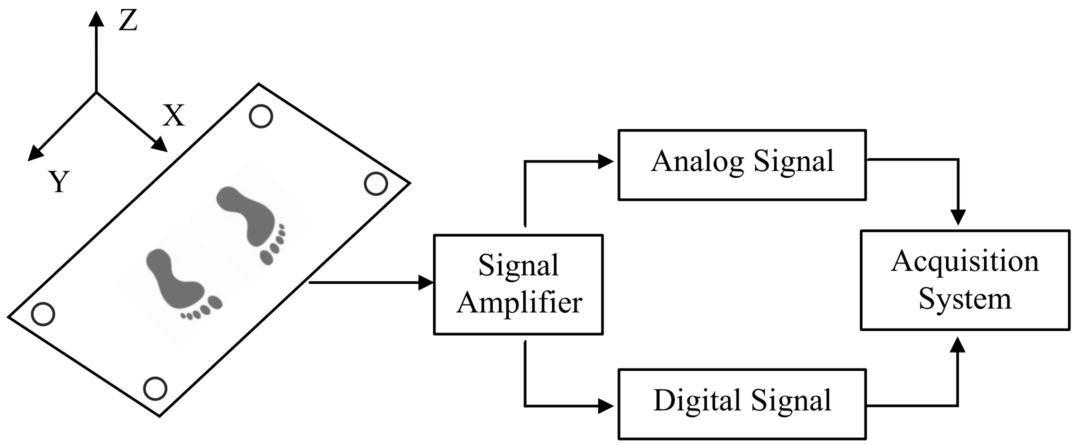

Figure 2.5 *Set-up of the force platform to acquisition system series.*

Despite FP's having a very large dynamic range, there are limitations within the A-D converters that restrict the overall resolution of the system. The range and resolution of a FP work in combination, for example a ± 1 kN range would have a higher resolution than a ± 10 kN, but would be limited to a low maximum force. It is for this reason a reasonable compromise between the two must be found that enables a sufficient force range to be measured, while maintaining an optimal resolution. Analogue signals are represented digitally as a series of discrete values; this means that there are only certain values available to represent the corresponding analogue signal. A digital signal is made up of a binary number; a series of 0's and 1's, or bits. The length of this binary number determines the number of discrete levels that can be represented (Pohlmann, 2010). For example, a 2-bit binary number can represent 4 discrete values (2^2), whereas a 10-bit binary number can represent 1024 discrete values (2^{10}). The most common resolutions used in research are 12-bit (e.g., AMTI) and 16-bit (e.g., Kistler and Bertec). A 12-bit A-D converter is capable of representing an analogue signal as 4096 (2^{12}) discrete steps and, as such, a range of 10 kN would be represented in discrete steps of 2.5 N, whereas a 16-bit A-D converter would be capable of representing that same signal in 65536 (2^{16}) discrete steps, that is a resolution of 0.15 N. As a result, it is reasonable to expect a 16-bit A-D converter to better represent an analogue signal and should, therefore, be used when high force ranges are used, such as those in a countermovement jump (CMJ). With this, because of their interaction, when displaying the resolution of a FP system in a study's methodology, the absolute range should also be clearly stated (Owen *et al.*, 2014).

The product of the data collected from a FP is a ground reaction force-time history in three orthogonal dimensions which produce three components: the vertical component (F_z), anterior-posterior component (F_y) and medial-lateral component (F_x). In a FP the total vertical component of ground reaction force (VGRF) is the arithmetic sum of the output of the four individual vertical transducers. This summation of the VGRF would be calculated within the computer software as all transducer signals are input individually, as standard.

2.4 Balance and stability

2.4.1 Centre of mass and centre of gravity

The centre of mass (CoM) of the body is defined as the net location of the CoM in three-dimensional (3D) space and is the weighted average of the CoM of each segment (Winter, 2009, pg. 127). The centre of gravity (CoG) of an object is the point at which the whole weight of the object can be considered to act (Watkins, 2014, pg. 237). The weight of an object acts vertically downward from its CoG along what is referred to as a line of action (hereafter referred to as line of gravity). The principle of CoG applies to all bodies, both inanimate and animate.

Very often the two terms, CoM and CoG, are used interchangeably, however this is not always appropriate, as CoM should only be used when there is no influence of gravity. With that, when referring to the human body in an upright position, it is appropriate to use CoG as we are concerned with the effect of the line of gravity through the CoG. It should be noted, however, that when referring to the CoG for a human body it can, on occasion, mean the CoM but the umbrella term of CoG will be used henceforth.

The position of an object's CoG depends on the distribution of the weight of the object; for a shape of uniform density, such as a cube or sphere, the CoG is located at the object's geometric centre as illustrated in Figure 2.6a For objects that are irregular in shape, the CoG could be positioned both inside or outside, depending on the distribution of weight and its shape. An example of this is shown in Figure 2.6b.

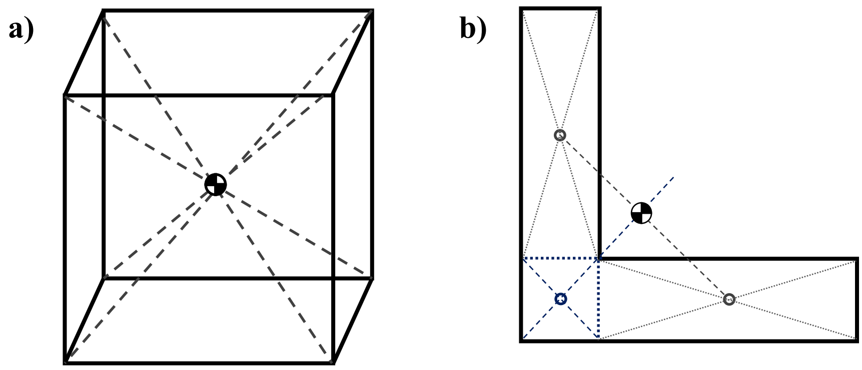

Figure 2.6 *a) The centre of gravity of a uniform density shape (such as a cube) is located at the geometric centre. b) The centre of gravity of an irregular shaped object (such as an L-shape) can lie outside the object.* *= Centre of Gravity*

2.4.2 Centre of gravity and the human body

The human body is a highly irregular shape which can be considered to consist of a number of segments (Hay, 1973). In addition to the morphology and anthropometry, the orientation of these segments determines the location of the body's CoG. When standing upright, the CoG of an adult is located within the body close to the level of the navel (54.0 ± 2.8% of the stature for women and 57.1 ± 2.3% of stature for men) and midway between the front and back of the body (Watkins, 2000). When body segments, and therefore weight, is redistributed the CoG moves as a consequence. As with inanimate objects, the CoG of a human can fall outside the body when a significant percentage of mass is moved, for example during full flexion of the trunk. The movement of the body's CoG in response to mass distribution is illustrated in Figure 2.7.

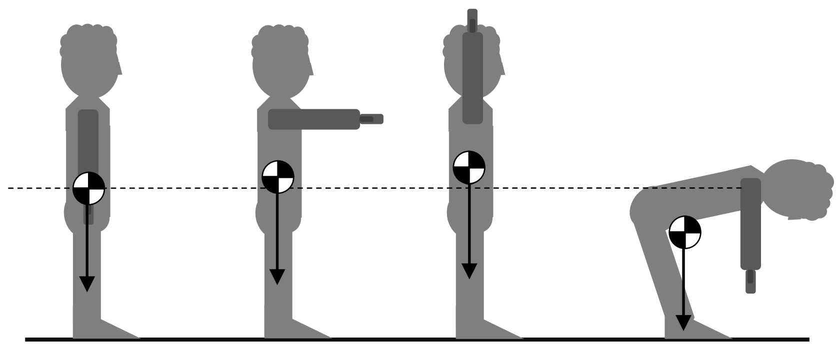

Figure 2.7 *Movement of the centre of gravity location in the human body resulting from changes in segment/ mass distribution.*

2.4.3 Measurement of centre of gravity in the human body

The location of an individual's CoG can be measured and/or estimated in a number of ways, however, there are two approaches to determining its position; the direct (whole-body) approach, in which the body is considered as a whole and the indirect (segmental) approach. In both approaches, the position of the CoG is determined from the intersection of three non-parallel planes that contain the CoG (Watkins, 2014, pg. 351). However, the location of the CoG of the whole body, a body segment, or other objects that interact with the human body can be a demanding task due to their complex geometry and heterogeneity of the materials that compose them (Oliveira, Roriz, Marques & Frazão, 2018).

Theoretically, there are three direct methods of determining CoG positioning within the body: (1) suspension method: a method involving suspending the body from at least two points and noting the point of intersection of the lines of gravity in the different positions. This method is, however, highly impractical when applied to humans as maintaining the required posture and positioning while suspended would be incredibly difficult. Dempster (1955) applied the suspension method to individual body segments, however, this was through the use of cadavers and, therefore, utilised the ability to measure dismembered regions and segments. (2) balancing; where CoG is estimated by balancing the body on top a fulcrum in three orientations (Figure 2.8).

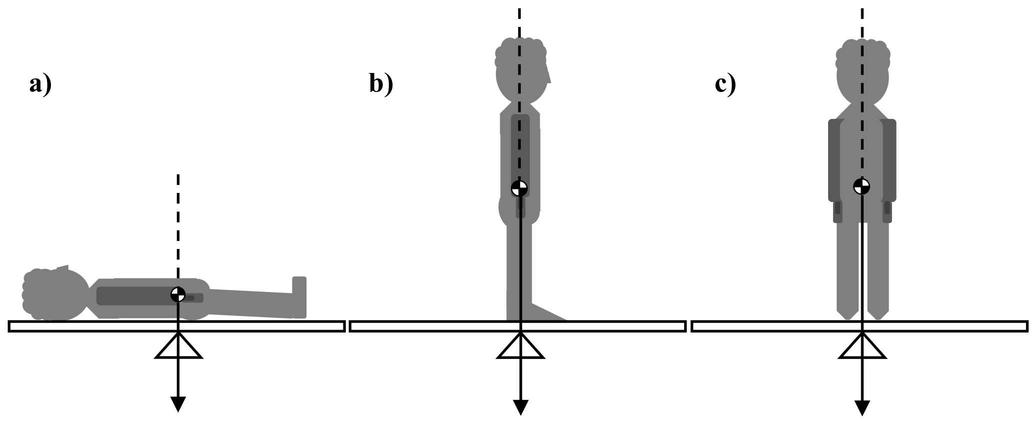

Figure 2.8 *Balancing method for determination of centre of gravity.*

The third method is the use of a reaction board. For this method, the principles of static equilibrium are applied to calculate the distance of the body's CoG from the pivot point, or location relative to a reference point, provided that location is known (Oliveira *et al.*, 2018). Static equilibrium refers to the state of body when the sum of all the forces and moments acting on it is zero (Ünal, Akkuş & Marcus, 2016). This method, however, only calculates CoG in one dimension along the transverse plane, but could be extended to two-dimensions by use of a rectangular or triangular reaction board with three points of support and two sets of weighing scales which enable the location of the CoG of the body in two planes to be determined in one step (Hay, 1993, pg. 136).

Figure 2.9a, b and c illustrate a standard reaction board set-up and and free body diagrams associated with a reaction board both with and without a person lying on it, respectively.

CoG can also be calculated from body segment parameters. By determining the position of the CoG and weight of each segment, the coordinates of the whole-body CoG can be determined by application of the principle of moments, i.e., the sum of the moments of the weights of the segments about a particular axis is equal to the moment of the total weight (Watkins, 2014, pg. 356).

Conventionally, body segment parameters are measured by a water displacement/ immersion method (Clauser, McConnville & Young, 1969; Dempster, 1955), geometric modelling method (Matsui, 1956) or photogrammetric method (Jensen, 1978). These parameters are normally derived from predictive equations based on data from cadavers or living subjects. However, applying these predictive equations to populations other than that from which they are derived is likely to cause large errors in estimation (Durkin & Dowling, 2003). These errors might have a tendency to be larger in groups such as children, athletes, and obese subjects which are not likely to have segmental shapes and body proportions similar to those of the general population. A further limitation to the segmental approach is that due to the continuous connection of segments through joints, accurate measurements such as the mass or weight of a single segment could be a difficult task as segment ends or boundaries are not clearly distinguishable. With that, depending on the assumptions made to define segmental landmarks, the coordinates of the CoG location could easily change (Hinrichs, 1990).

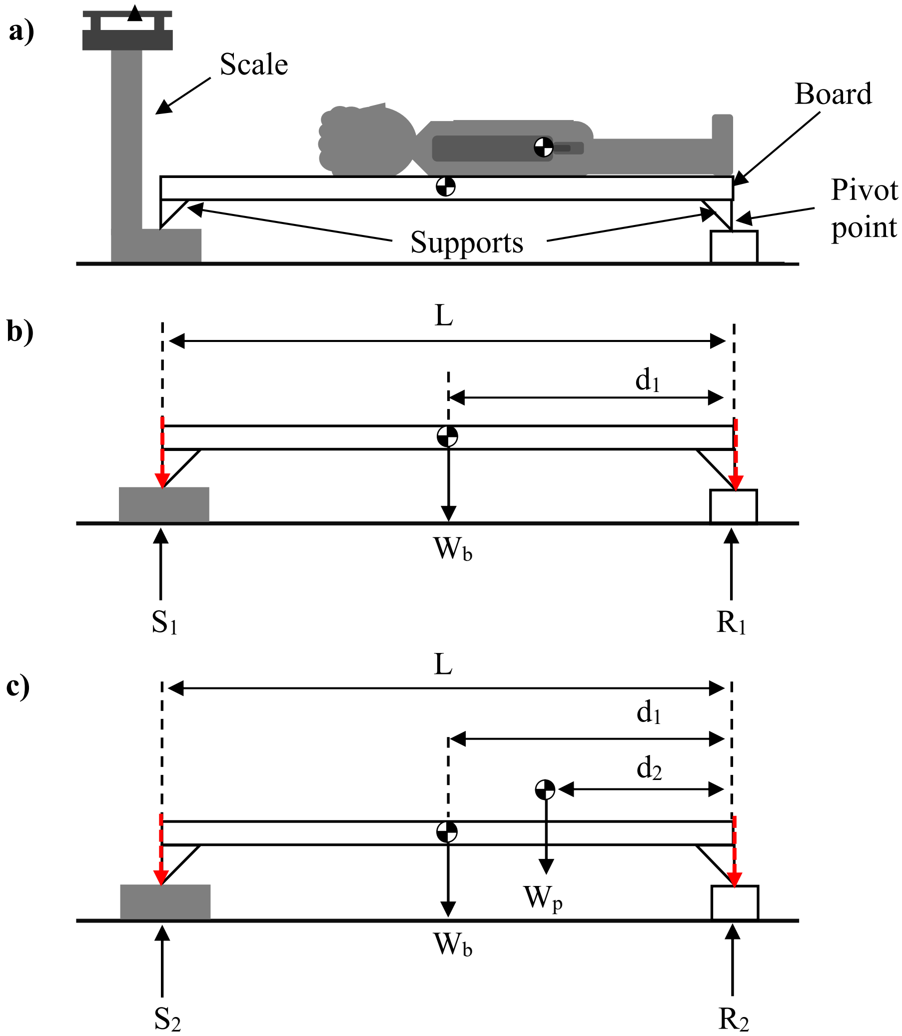

Figure 2.9 *Reaction board method for determination of centre of gravity. a) Reaction board set-up with person lying on it. b) Free body diagram of board. c) Free body diagram of board with person lying on it.*

S_1, vertical force exerted by scales on support without person; S_2, vertical force exerted by scales on support with person lying on it; R_1, vertical force exerted by ground/ block on pivot point without person; R_2, vertical force exerted by ground/ block on pivot point with person lying on it; W_b, weight of board; W_p, weight of person; L, horizontal distance between supports; d_1, horizontal distance between support and vertical plane containing centre of gravity of board; d_2, horizontal distance between support and vertical plane containing centre of gravity of person.

= Corresponding equal and opposite reaction force exerted by the reaction board on the supports which maintains the static equilibrium.

2.4.4 Centre of pressure

Centre of pressure (CoP) is the point on the supporting surface, between the soles of the feet, at which the resultant vertical force vector would be considered to act if it were concentrated to a single point of application (Ruhe *et al*., 2010). CoP is a principle closely linked to human CoG and is often referred to as an approximation of the CoG under static or slow-moving conditions (Winter, 1995).

Approximations of CoM and the term CoP are often used interchangeably, however, the CoP moves to react to movement of the CoM and the difference between them (CoP-CoM) is proportional to the horizontal acceleration of the CoM.

The trajectory of the CoP is totally independent of the CoM, and is the location of the vertical ground reaction force (VGRF) vector from a single FP, assuming that all body contact points are on that platform. The VGRF is a weighted average of the location of all downward (action) forces acting on the FP. These forces depend on the foot placement and the motor control of the ankle muscles. Thus, the CoP is the neuromuscular response to the imbalances of the body's CoM. With this, it is apparent that the CoP must be continuously moving anterior and posterior of the CoM and therefore the dynamic range of the CoP must be somewhat greater than that of the CoM (Winter, 2009, pg.127-129). Evidence of this variation between CoM and CoP can be seen in Figure 2.10. CoP amplitude exceeds that of the CoM, and changes in the direction of the CoM are a result of 'overshoots' of the CoP signal.

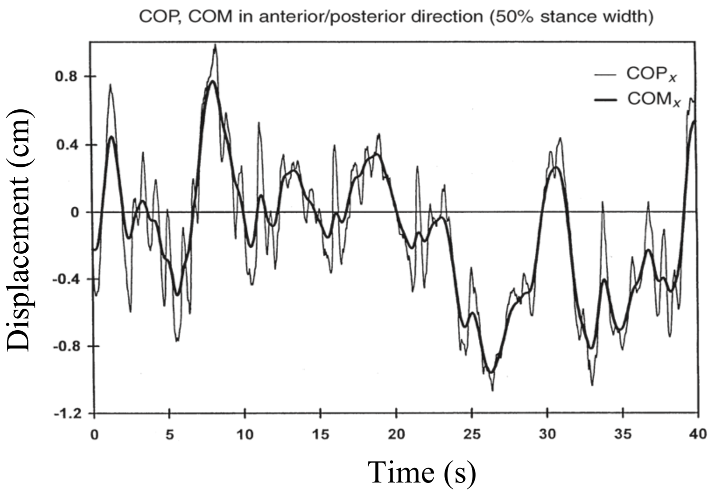

Figure 2.10 *Sourced from Winter (2009, pg. 129). Typical 40-s record of the total body centre of mass (CoM_x) and centre of pressure (CoP_x) in the anterior/posterior direction during quiet standing.*

2.4.5 Balance and stability in inanimate objects

The term balance, or equilibrium, refers to the state of an object when the resultant forces or moments acting upon it are equal to zero (Bell, 1998, pp. 26). In order for an object to remain balanced, the CoG, and thus its line of gravity, must intersect the plane of the base of support (Watkins, 2014, pp. 240). In light of this, the ability to balance in a static situation is closely related to stability; the stability of an object, with respect to its base of support, is dependent on both the area of the base of support and position/ height of the objects centre of gravity above this base (Watkins, 2014, pp. 241). When referring to stability the greater the displacement of the line of gravity before an object becomes unbalanced, the greater the stability of that object (Pollock *et al.*, 2000). Therefore, it follows that stability increases when the ratio of the height of the centre of gravity to length of the base of support (with respect to each possible tilt axis) is lowered (Pollock *et al.,* 2000; Watkins, 2014, pp. 241). An illustration of this is demonstrated in Figure 2.11.

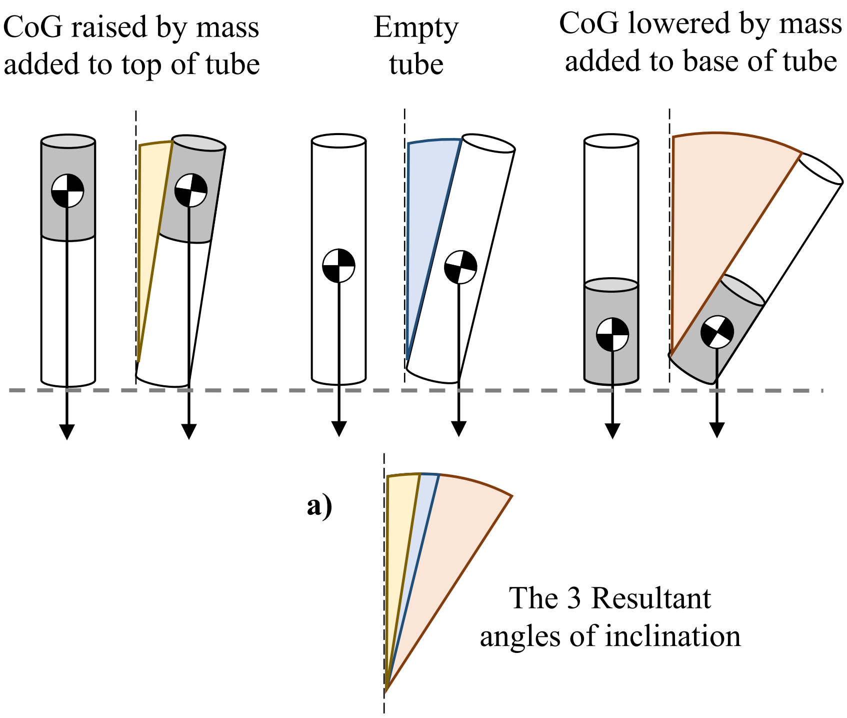

Figure 2.11 *The effect height of centre of gravity to length of base of support ratio has on stability. Although tubes are identical in size and volume, the tube with a lower centre of gravity can tilt further before the line of gravity falls outside the base of support and is, therefore, more stable. a) Illustrates the three angles of inclination when stacked ontop of each other in order to demonstarte the magnitude of the differences between them.*

2.4.6 Balance and stability in humans

With regard to human balance, the terms balance and stability are often used synonymously; the behaviour of the human body does, however, vary from an inanimate object. Balance in humans is defined as the dynamic control of stability. This differs from Bell's definition in inanimate objects as humans are never in a position of zero net force due to the dynamic nature of their balancing, which is most commonly referred to as postural control. For an inanimate object, if the line of gravity falls out of the base of support, the force of gravity dictates that the object will fall (or move). However, when, in a human, the line of gravity falls outside the base of support the human body has the inherent ability to sense the threat to stability and to counteract the force of gravity in order to prevent falling (Horak, 1987 in Pollock *et al.,* 2000). Because the body is not a rigid structure, it is continuously performing oscillations about a vertical axis during standing. In the case of human balance, perfectly stable posture is when the body's line of gravity passes though the CoP.

2.4.6.1 Neurological and physiological control mechanisms

In order to maintain balance, the human body has a complex system of neurological and physiological control mechanisms that each allow the body to interpret and recognise its position in space. These control mechanisms are the vestibular, vision, proprioceptive and somatosensory systems and work collectively to control balance via the neuromuscular system. Sensory information from each of the systems is received by the central nervous system (CNS) via sensory feedback and then based on an estimate of body kinematics, appropriate control plans are selected and corresponding motor commands are produced as joint torques (Kim, Horak, Carlson-Kuhta & Park, 2009). This process of sensory input, feedback and motor command is demonstrated in Figure 2.15.

2.4.6.2 Vestibular

Sensory information about motion, equilibrium and spatial orientation is provided by the vestibular apparatus (the balance organs in the inner ear as illustrated in Figure

2.12a.). The two main components of the vestibular system are (1) the peripheral component (Figure 2.12b) and (2) the central component.

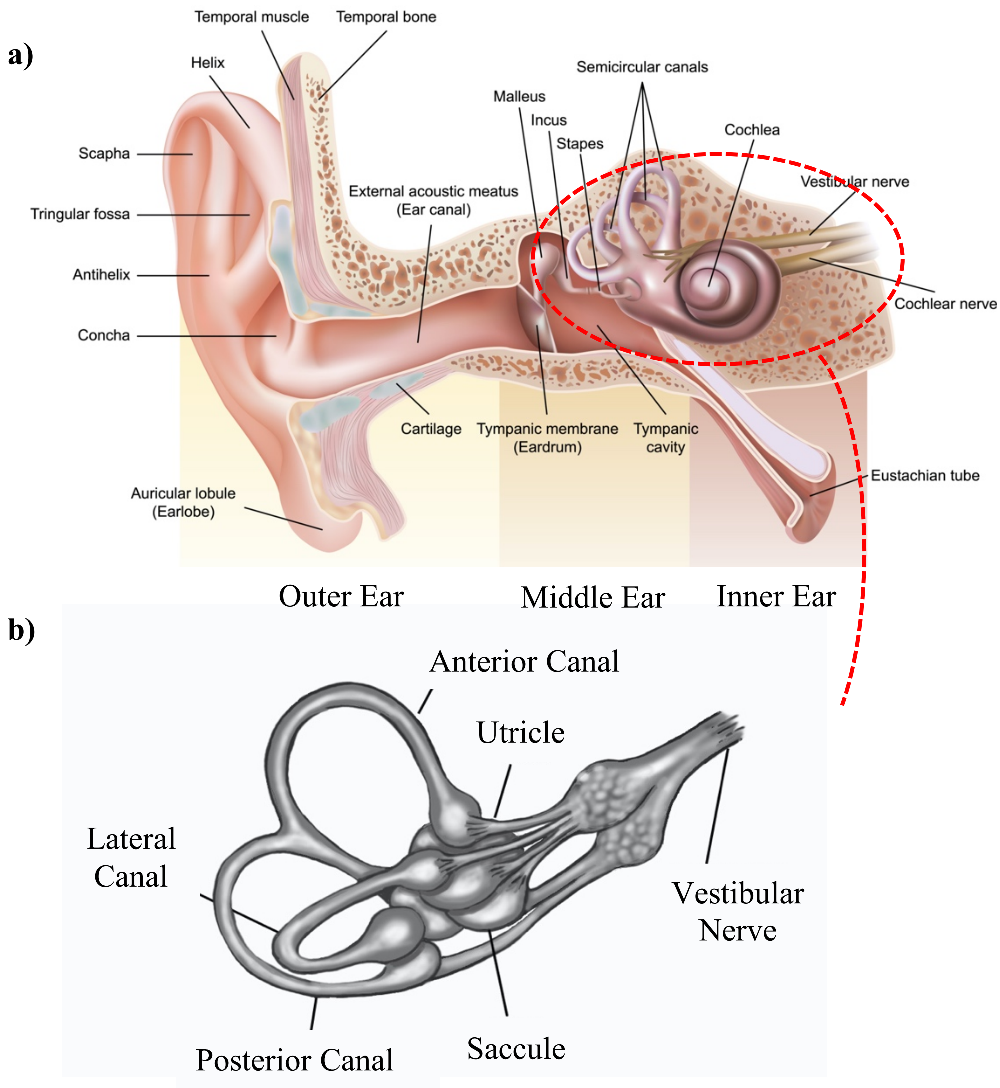

Figure 2.12 *a) Anatomy of the ear with outer, middle and inner regional separation. (Source: bestcare.org). b) Peripheral component of vestibular system. (Source: Vestibular disorder association).*

The peripheral component is composed of three semi-circular canals located at right angles to each other which are filled with a fluid called endolymph providing information of orientation in three directions of space; these are x, y and z which refer to anterior-posterior (A/P), medial-lateral (M/L), and vertical, respectively. When a canal senses rotation, the endolymphatic fluid within it lags behind, because of inertia, and exerts pressure against the canal’s sensory receptor (Watson & Black, 2008). The receptor then sends impulses to the brain about movement from the specific canal that is stimulated. In the event of rotation, the number of signals received from one ear or the other changes (for example when the head turns to the left, the number of impulses from the left ear increases while the number from the right ear decreases). This difference in impulses controls eye movements and stabilizes the gaze during active head movements (Watson & Black, 2008).

Therefore, the two primary functions of the vestibular system are to stabilize the eyes during movements of the head and to stabilize the body during movements of the head (Kinne, 2015). The motor control signals sent via the nervous system to the muscles of the eyes are done so with an automatic function called the vestibulo-ocular reflex (VOR) and the body is primarily stabilised through a mechanism known as the vestibulo-spinal reflex (VSR).

2.4.6.3 Visual

Vision helps to assist with balance by reporting where the head and body are in relation to the world around us and to sense motion between the body and the environment. The visual system enables a person to see the consequence of any movement by looking at horizontal lines with the eyes, which can then be processed in order to identify whether the body is stable. Like with the vestibular system, the visual system can be broken down into a central and peripheral processing system. The central system is used mainly for clarity in order to identify details of an object while the peripheral system is used to initiate spatial localisation and to process movement (Davis, 2016). Dizziness and disequilibrium are often the result of a vestibulo-ocular reflex (VOR) dysfunction (a reflex which coordinates eye and head movement) and an unstable binocular (how well the eyes work together) system (Cohen, 2013).

2.4.6.4 Proprioception

Proprioception, the somatosensory system used specifically for awareness of position and body parts, refers to the detection of sensory stimuli such as stretch and pressure in the muscles, tendons and joints in order to detect movements such as joint displacement (Tyldesley & Grieve, 1996 in Lephart, Pincivero & Rozzi, 1998). The proprioceptive system does not include senses of external stimuli such as touch or temperature.

Sensory input from the peripheral articular and musculotendinous receptors, in addition to information regarding joint position and motion, enable the brain to interpret how the feet and legs are positioned in comparison to the ground and how the head is positioned in relation to the chest and shoulders (Hoffman, 2010). The proprioceptive system is also assisted by afferent nerves, also referred to as mechanoreceptors, located within the skin, in the musculotendinous unit and within the bone, joint ligaments and joint capsule (Kennedy, Alexander & Hayes, 1982; Nyland, Brosky, Currier, Nitz & Caborn, 1994: Tyldesley & Grieve, 1996 in Lephart *et al*., 1998). Proprioception and accompanying neuromuscular feedback mechanisms provide an important component for the establishment and maintenance of functional joint stability (Lephart *et al*., 1998).

During quiet standing the proprioceptive system detects A/P and M/L movement, or sway, and feeds-back to the CNS. The CNS responds through motor output that results in increased pressure and force production onto the floor against the direction of travel. This causes the body to move the opposite way in order to correct; this process links closely to CoP as the point source of the restoring GRF is the location of the CoP.

2.4.7 Postural control mechanism

Postural control within the body can be modelled as an inverted pendulum model (Figure 2.13 and 2.14) (Günther & Wagner, 2016) with intermittent control strategies (Loram *et al*., 2011) and is maintained through neuromuscular control based on appropriated and integrated sensory information from somatosensory/ proprioceptive, visual and vestibular systems (Chiba, Takakusaki, Ota, Yozu & Haga, 2016; Peterka, 2002, 2018; Windhorst, 2007). In order to maintain postural equilibrium, the body continuously adopts changing neural strategies and reweighting of the sensory inputs; according to the sensory weighting hypothesis, motor performance of balance is based on the "weighted sum" of all sensory inputs (Kabbaligere *et al*., 2017; Paillard, 2012; Peterka, 2002). The process of reweighting refers to the inputs of these three sensory systems being organised and the correct signals selected. Postural control also encompasses reactive (compensatory), predictive (anticipatory) or feedforward adjustments, context-dependent sensorimotor modulations and integration of posture and movements (Ivanenko & Gurfinkel, 2018; Maki & McIlroy, 1997; Massion, 1994). Integration of the sensorimotor system refers to the integration of the afferent, efferent, and CNS (Lin *et al*., 2019). The basic integration of the postural control systems is displayed in Figure 2.15.

Postural control is a complex process and is, therefore, very difficult to conceptualise and explain fully. As a result, simplified models are developed to allow postural control to be understood more easily. Typically, when referring to postural control as an inverted pendulum, there are three levels used; single- (SIP), double- (DIP) and triple-inverted pendulum (TIP). Figure 2.13 shows an exaggerated model of the three levels. A SIP portrays the body as one rigid segment oscillating about the ankles while DIP accounts for motion at both the ankle and hip joints. The TIP model takes this one step further and accounts for additional movement at the knee joints. As we move up the levels of the pendulum model it is possible to see how movement at the joints enables the displacement of the body's CoG to be reduced.

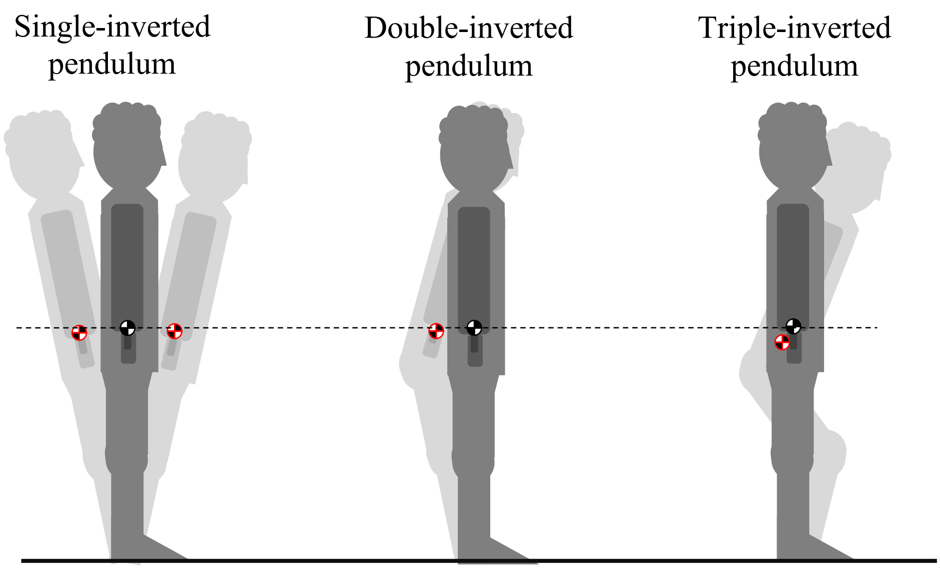

Figure 2.13 *Simplified model of postural control as an inverted pendulum. a) Single-inverted pendulum (SIP) - body is one rigid segment oscillating about the ankles; b) Double-inverted pendulum (DIP)- accounts for motion at both ankle and hip joints: c) Triple-inverted pendulum (TIP)- accounts for additional movement at knee joints.*

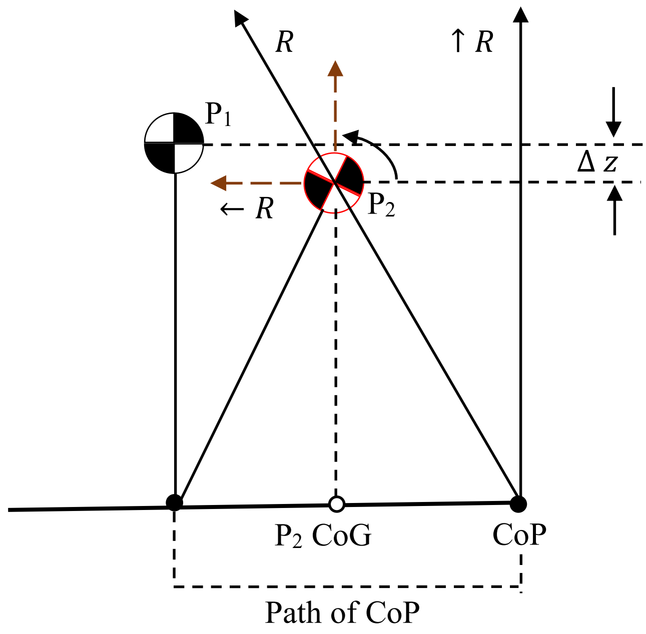

Figure 2.14 *Simplified model of postural control based on a single-inverted pendulum model with one free joint. P_1, CoG in upright position; P_2, CoG in forward leaning position; R, resultant force vector from CoP to return CoG to upright position; ↑ R, vertical component of force vector R; ←R, horizontal component of force vector R; Δz, change in displacement of the body's CoG in the vertical direction; P_2 CoG, interception of P_2 line of gravity with supporting surface .*

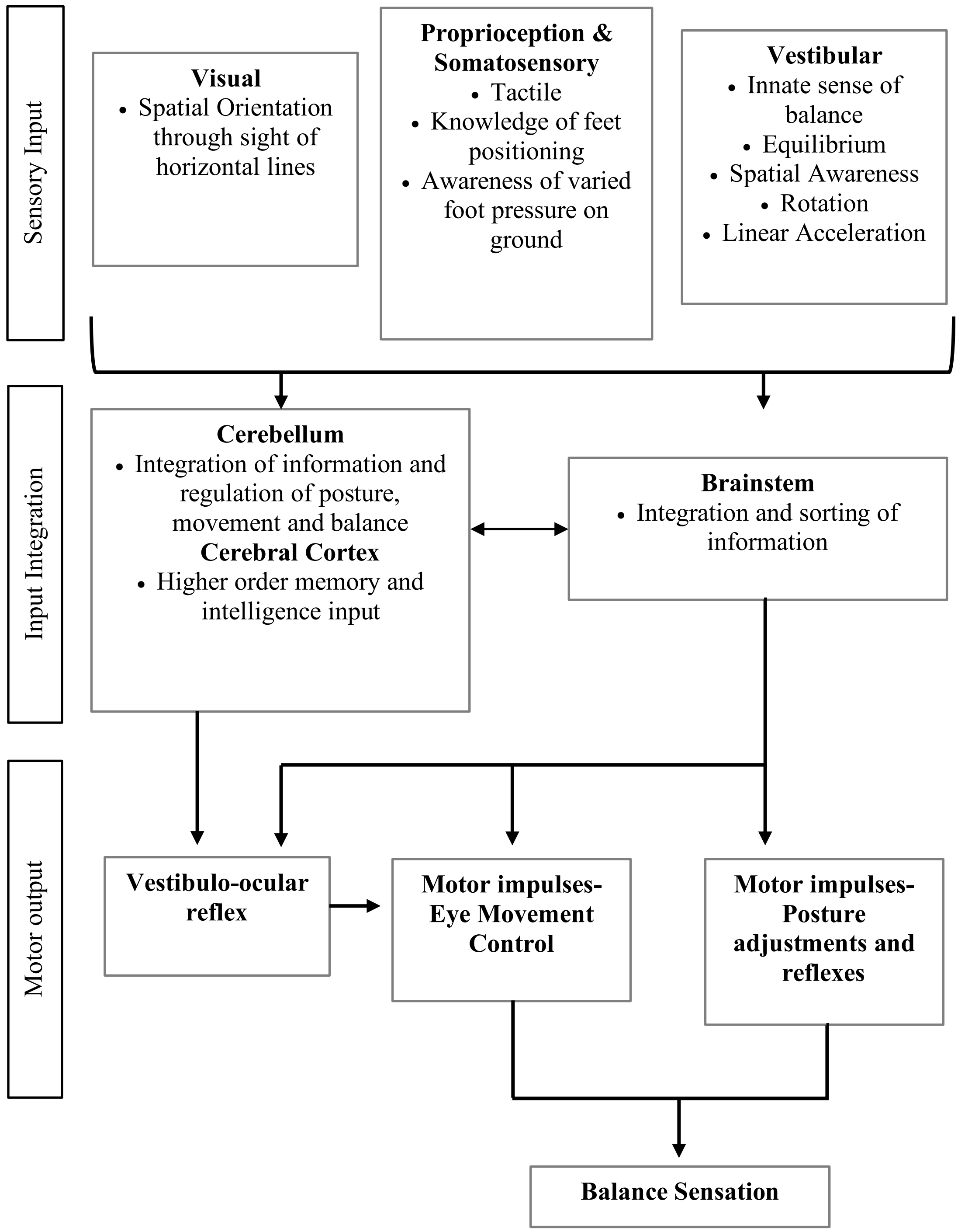

Figure 2.15 *The human balance control system. Adapted from Vestibular Disorder Association 2008.*

2.4.8 Measurement of human balance control

In order for balance control to be assessed in any population, it needs to be measurable, and, therefore, quantifiable. In previous studies various methodologies have been used with the goal of identifying balance mechanisms, in addition to using balance as a wider tool for analysis of varied populations, both clinical and sporting. The interest of researchers in the assessment of standing balance has led to the development of a variety of techniques; such as functional assessments, sway magnetometry and FP derived measures (Browne & O'Hare, 2001). In addition to this, more 'in-depth' analysis looking towards balance control strategies are often identified by measuring kinematics, kinetics or muscle activity (Horak & Nashner, 1986; Horak *et al.,* 1997; Nashner & McCollum, 1985; Kuo & Zajac, 1993).

The earliest research of human standing balance was conducted by Romberg (1853) during the assessment of diseases of the CNS. In this research Romberg assessed the sway of an individual with their eyes closed. Since its introduction, developments are consistently being made in human balance research in order to establish the most effective and trusted methods. Two of the most commonly reported methods of measuring balance are under two conditions; perturbed (PS) or unperturbed/ quiet standing (QS) (Bardy, Oullier, Lagarde & Stoffregen, 2007). With this, the most common method of quantification, in well-resourced research, is the use of CoP analysis in one or both of these conditions.

2.5 Postural stability assessment

2.5.1 Centre of pressure analysis of quiet standing

Centre of pressure (CoP) analysis has become a widely used method of balance quantification. There is consistency within the literature regarding how to measure the position of CoP, which still needs to be established, however, there are inconsistencies in terms of the methodologies used to quantify balance using the measurements of CoP and time histories.

2.5.2 Identifying and recording CoP location using a force platform

The measurement of CoP by means of FP's is relatively simple; however, it is limited in its use. Analysis of FP data exclusively does not provide insight into which part of the body is in motion at a given time. Despite this, FP analysis does allow for a reliable quantitative measure of CoP location and tracking.

A point along the line of action of the applied force can be determined using the forces and moments measured by a fixed FP (Giacomozzi & Macellari, 1997) (Figure 2.16). The quantification of CoP uses a coordinate system in which the distance of the line of action-supporting surface interaction from an origin (the centre of the FP) is calculated. When a vertical force F_z is applied a distance x_{CoP} and y_{CoP} from the centre of the FP coordinate system, the FP measures the force F_z and the associated M_x and M_y moments that are generated about the platform's x and y axes, respectively (Figure 2.16d).

To obtain the components of forces and moments, the following equations can be written:

$$M_x = \left[\left(-F_{z_1} - F_{z_2}\right) + \left(F_{z_3} + F_{z_4}\right)\right] \cdot b,$$

$$M_y = \left[\left(-F_{z_1} + F_{z_2}\right) + \left(F_{z_3} - F_{z_4}\right)\right] \cdot a,$$

$$M_z = \left[\left(F_{x_1} + F_{x_2}\right) - \left(F_{x_3} + F_{x_4}\right)\right] \cdot b + \left[\left(F_{y_1} + F_{y_4}\right) - \left(F_{y_2} + F_{y_3}\right)\right] \cdot a \quad \cdots\cdots 2.4.$$

where: F_x = medial-lateral reaction force (N)

F_y = antero-posterior reaction force (N)

F_z = vertical reaction force (N)

$F_{x_{1-4}}$ = components of the medial-lateral reaction forces for each sensor (N)

$F_{y_{1-4}}$ = components of the antero-posterior reaction forces for each sensor (N)

$F_{z_{1-4}}$ = components of the vertical reaction forces for each sensor (N)

M_x, M_y, M_z = moments of the force platform around the axes O_x, O_y, O_z (Nm)

a, b = location of the geometric centre of force platform (m)

With this information, the x and y locations of the CoP are computed as follows:

$$x_{CoP} = \frac{-M_y}{F_z} \qquad y_{CoP} = \frac{M_x}{F_z} \qquad \text{2.5.}$$

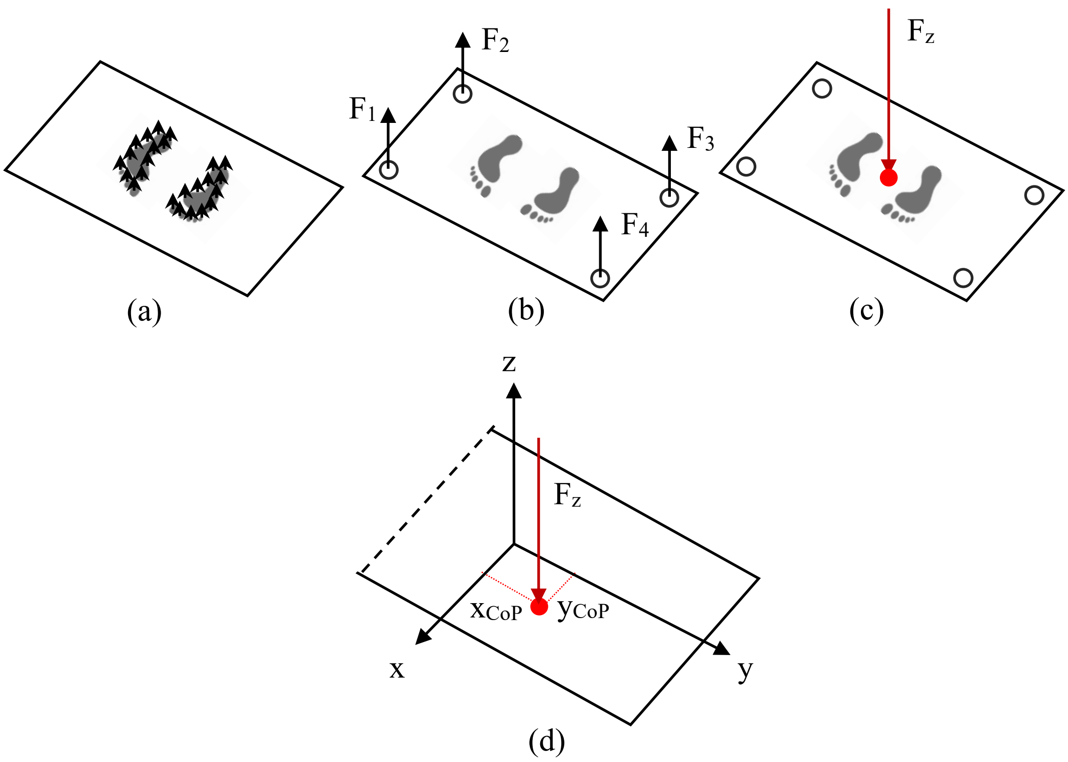

Figure 2.16 *Schematic of coordinate and force platform calculation of vertical force and CoP. a) pressure distribution of feet in quiet standing, b) forces captured from force platform sensors, c) Vertical force vector and CoP location, and d) CoP determined using measured forces and moments.*

The measurement accuracy is closely related to the reliability of the force platform signal. Quagliarella, Sasanelli and Monaco (2008) outline that the accuracy and reliability of a signal are affected by factors such as: (1) Systematic errors (bias): which can be related to the use of the device or its age. It can, however, be overcome by static or dynamic calibration procedures. (2) Low-frequency noise: its typical behaviour presents a power spectral density (PSD) inversely related to frequency, and it mainly affects static or quasi-static events (Quagliarella *et al*, 2008). Browne and O'Hare (2000) assessed low-frequency noise affecting descriptive parameter accuracy and demonstrated that filtering can be effective in noise reduction. (3) Drift: defined as an undesirable change in output signal over time, which is not a function of the measured variable (The Institute of Measurement and Control, 1998; Kistler, n.d.). Drift can be assessed by applying a known dead weight to the platform; the components and point of application should not be time dependent so consequently the platform output should have a vertical component of GRF (F_z) with a constant value equal to the dead weight and horizontal components of GRF equal to zero, i.e., $F_x = F_y = 0$ N. Any movement of values away from these expected outputs are indicative of platform and electronics limitations (drift).

2.5.3 Methodologies used to quantify postural control using measurements of CoP

Although the measurement of CoP location appears consistent across research studies, the methodology used by these studies to obtain their data and the ways in which CoP location is processed and interpreted can be varied. Table 2.1 and Table 2.2 contain only a small proportion of the existing research where CoP has been used as a method of quantifying postural control. A large proportion of balance research has been conducted in general and clinical settings or populations (Table 2.1), with quite limited reports related to sport- specific balance (Table 2.2).

There are a number of important variables to be considered when devising and developing a methodology for postural control research; many of which are included in Tables 2.1 and 2.2.

2.5.3.1 CoP parameters

The first consideration of a CoP research study should be the descriptive measures, or parameters, used to evaluate an individual's postural stability. There are a vast number of parameters that can be obtained from CoP data, however, those selected for analysis varies from one researcher to the next. Evaluation parameters of CoP movement are often categorised into 7 domains: distance, centre average, distribution of the amplitude, area, velocity, power spectrum and body sway vector (Demura, Yamaji, Noda, Kitabayashi & Nagasawa, 2001; Mitzuta, 1993 in Kitabayashi, Demura & Noda, 2003). The selectivity of parameters within these domains and the number of parameters used can vary significantly; for example, Kitabayashi *et al*., (2003) assessed 34 parameters from 6 domains (distance, area, velocity, distribution of amplitude, power spectrum and vector) in order to determine reliable parameters for CoP evaluation. The results of that study identified high reliability coefficients (ICC) in a number of parameters; with almost all values above 0.8 with the exception of some power spectrum parameters. Correlation coefficients between parameters in each domain were generally significant and very high, and the coefficients between parameters representing domains except the power spectrum were significant.

Despite many studies opting for a wide range of parameters, it can be said that many parameters show considerable overlap; for instance, both root mean square (RMS) amplitude and mean rectified amplitude give a measure of absolute displacement from the mean position, while RMS velocity, mean rectified velocity and sway path are all measures of the average absolute displacement in time (Hufschmidt, Dichgans, Mauritz & Hufschmidt, 1980; Kapteyn *et al*., 1983). Additionally, the necessity of using numerous parameters is reliant on the depth and/ or objective of the study. In studies looking to evaluate the effect of changing variables on parameter results, or those assessing the parameters themselves, the use of so many descriptive measures can be justified. For studies looking to simply quantify postural control, for the purpose of comparing it to other factors, the use of fewer parameters with high reliability and validity could prove to be more effective and ensure results are concise while characterising different aspects of postural steadiness.

The most common variables in CoP analysis are total path length, area of 95% confidence interval ellipse and CoP velocity (Blosch, Schäfer, de Marées & Platen, 2019). Path length refers to the quantification of the magnitude of the two-dimensional displacement based on the total distance travelled by the CoP over the course of the trial. It is considered to be a valid outcome measure and conclusions are based upon the idea that the smaller the path length, the better an individual's postural stability (Donath, Roth, Zahner & Faude, 2012). Path length has also been identified as one of the most reliable parameters in a systematic review by Hébert-losier and Murray (2020) along with the area of 95% CI ellipse; this refers to the quantification of 95% of the total area covered in the AP and ML direction using the smallest ellipse containing 95% of all CoP data points (Figure 2.17). The ellipse area is considered to be an index of overall postural performance. Asseman, Caron and Crémieux (2004) suggested that the smaller the ellipse area, the better the "performance".

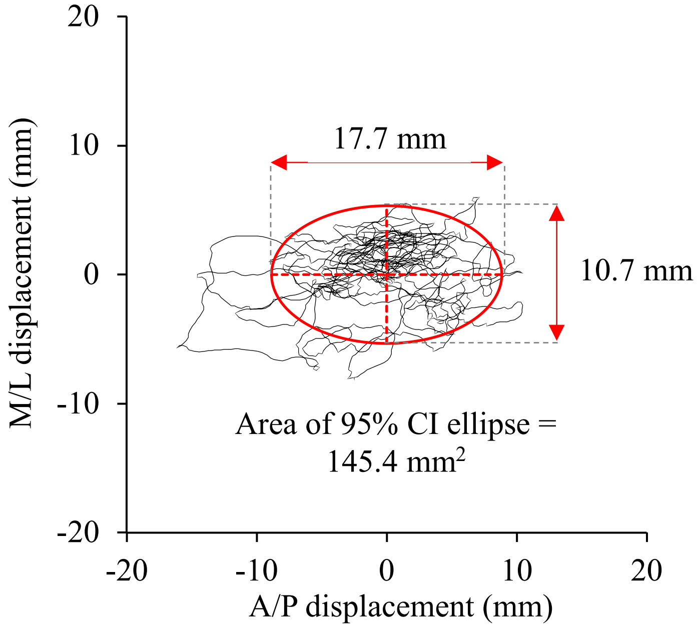

Figure 2.17 *90 s CoP trace with corresponding 95% CI ellipse.*
*Note: Magnitude of the 95% CI in the A/P and M/L directions are calculated as 1.96 * SD of displacement in the x and y directions, respectively.*

Parameters that employ minimal and/ or maximal readings, such as maximum amplitude, are often avoided because they only use one or two data points from the entire data set; they are, therefore, subject to much greater variances and are thus less reliable. The process of averaging data decreases the effect of extreme individual readings, so CoP summary measures, such as mean velocity are favoured instead (Ruhe *et al.*, 2010). CoP velocity refers to the CoP excursion divided by the trial time and represents the efficiency of the postural control system (Paillard & Noe, 2015). A study by Raymakers, Samson and Verhaar (2005), which looked to assess choice of stability parameters, concluded that mean displacement velocity seemed to be the most informative parameter and has been confirmed to have high reliability and validity (Salavati *et al.*, 2009). A systematic review by Ruhe *et al.*, (2010) also considered mean velocity to be the most reliable traditional CoP parameter. In addition to this, this study also advised that parameter selection "should include both distance (e.g., area) as well as time-distance (e.g., mean velocity) based parameters to gain a diverse description of the CoP excursion".

The reliability of traditional measures such as area and mean velocity has previously been questioned; for example, by Doyle, Newton and Burnett (2005) who noted that reliability coefficients were low ($ICC_{2,1}$ 0.05-0.71). However, this variation in conclusions between researchers could be put down to study design, for example, Doyle *et al.* (2005) used only 10 s of data compared to 60 s by others.

2.5.3.2 Acquisition settings

2.5.3.2.1 Sample duration

Time-frequency analysis of CoP motion has revealed that balance is non-stationary with time varying properties (Schumann, Redfern, Furman, El-Jaroudi & Chaparro, 1995). However, despite the numerous studies throughout the literature, a standard for trial duration in CoP analysis has not yet been developed, nor have there been many recommendations. As a consequence, the sampling durations of studies across the literature are grossly varied, as can be seen in Tables 2.1 and 2.2. A number of earlier studies have suggested that reliable data can be obtained when sampling durations are within the region of 10-60 s (Goldie, Bach & Evans, 1989 in Ruhe *et al*., 2010; Letz & Gerr, 1995; Le Clair & Riach, 1996). However, later studies have concluded that durations of at least 60 s and up to 90 s and 120 s are necessary for most CoP parameters to be confidently reliable (Carpenter *et al*., 2001; Corriveau, Hébert, Prince & Raîche, 2000; Doyle, Hsiao-Wecksler, Ragan & Rosengren, 2007; Harringe, Halvorsen, Renström & Werner, 2008). It is important to note, however, that the methodology of these studies all vary from one another, along with their statistical analysis models, and with that, interpretations of reliability are based upon the measurement and comparison of different parameters and summary measures; this heterogeneity makes the combination of reliability data and inter-study comparison of results and conclusions very difficult. A systematic review by Ruhe *et al* (2010) suggested that a positive trend towards increased reliability follows with increasing trial duration. Further, despite the data used in their analysis deriving from varying statistical models, values for mean velocity, root mean square in both A/P and M/L directions and CoP area have shown a positive relationship between trial duration and reliability coefficient, although greater variability exists in CoP area between time intervals. The recommendation of this systematic review following the inclusion of 32 papers was a trial duration of 90 s. Van der Kooij *et al.* (2011) conducted a study with the sole purpose of assessing sampling duration effects on CoP descriptive measures using a duration that far exceeded those used in previous research (600 s trial which was divided into 60 s increments i.e., 0-60 s, 0-120 s… 0-600). The results of this study confirmed prior recommendations to sample CoP measures for at least 60 s when using eyes open (Carpenter *et al*, 2001). For eyes closed, however, they identified that significantly greater sampling durations are required to achieve stable standard deviations, suggesting that the removal of vision introduces larger amplitude displacements in the CoP signal that may only emerge after longer periods of stance.

Despite this, longer trial durations have their limitations. The practicality and validity of longer durations must be considered with regard to the capabilities of both the participant and equipment; for example, the confounding effects of fatigue (Collins & De Luca, 1993 in van der Kooij *et al*, 2011) in addition to the restrictions of a force platform. Data collected for the current study will be using a piezoelectric force platform; because the platform uses charge as opposed to voltage, it is not recommended for extended periods of time due to charge dissipation resulting in consequential drift. Quagliarella *et al*. (2008) identified drift in a piezoelectric FP to be a linear trend with time and predominantly affected the F_z component of GRF and x component of CoP coordinates (P_x).

2.5.3.2.2 Time slicing

An additional variable, associated with trial duration, is whether or not a study has 'time sliced' their data. This refers to the removal of data from the beginning and/or end of the data set. There are very limited reports in the literature regarding the impact of data removal, and, with that, there are very few studies that do so. There are, however, some studies that have implemented time slicing prior to the interpretation of their results. Benvenuti *et al.* (1999) retained the remaining data following a 25 s accommodation period, however, their research was with geriatric populations with normal, moderate or severe levels of equilibrium, so there will have been a larger accommodation period than needed for younger, healthier participants. Raymakers *et al.* (2005) made the decision in their study to systematically ignore the first 10 s to avoid disturbance from delayed stabilisation of the recording equipment after the person stepped on the platform. It has been outlined in Scoppa, Capra, Gallamini and Shiffer (2013) that any task requires an "adaptation phase" and will be affected by fatigue or lack of attention; with this it was suggested that 5 s would be an appropriate time slice. The removal of data from either end of a sample would allow for disturbances from trial initiation and anticipation of trial completion to be ignored and ensure that such disturbances do not factor into the results and negatively impact the interpretation.

2.5.3.2.3 Sampling frequency

The sampling frequency refers to the number of data samples the FP measures each second (measured in the S.I. unit hertz (Hz)); these measurements can only be taken at regular pre-determined intervals, as opposed to continuously, so it is important to determine the optimal frequency for the data being collected which is high enough to provide an accurate force-time history.

As demonstrated in Tables 2.1 and 2.2, sampling frequency, and following filtering frequency, are acquisition settings that are also yet to be standardised and are, therefore, considerably varied across the literature. Scoppa *et al*. (2013) suggested a minimum sampling rate of 50 Hz as classical parameters, such as sway path, area and confidence ellipse area, are reasonably steady and have acceptable reliability at this rate. However, both oscillations and sway density parameters require a higher sampling frequency and, therefore, a sampling rate of 100 Hz is recommended. Schmid *et al.* (2002) suggested that an acquisition sampling rate of 100 Hz both allows for the limited selectivity of real filters and the exploitation of the improvements on signal resolution allowed by over-sampling technique. It has also been suggested that increasing sampling rate above what is necessary has the potential to add "noise" rather than beneficial "information" to the signal (Scoppa *et al*., 2013).

When establishing sample design, aliasing should be considered. Aliasing refers to when an input signal frequency is faster than half the sampling frequency; as a result, the sampled result will appear to be a low-frequency wave. This principle is demonstrated in Figure 2.18. The Nyquist sampling theorem (Nyquist, 1928) suggests that to avoid aliasing the slowest possible sampling rate should be two times the highest frequency of the input signal.

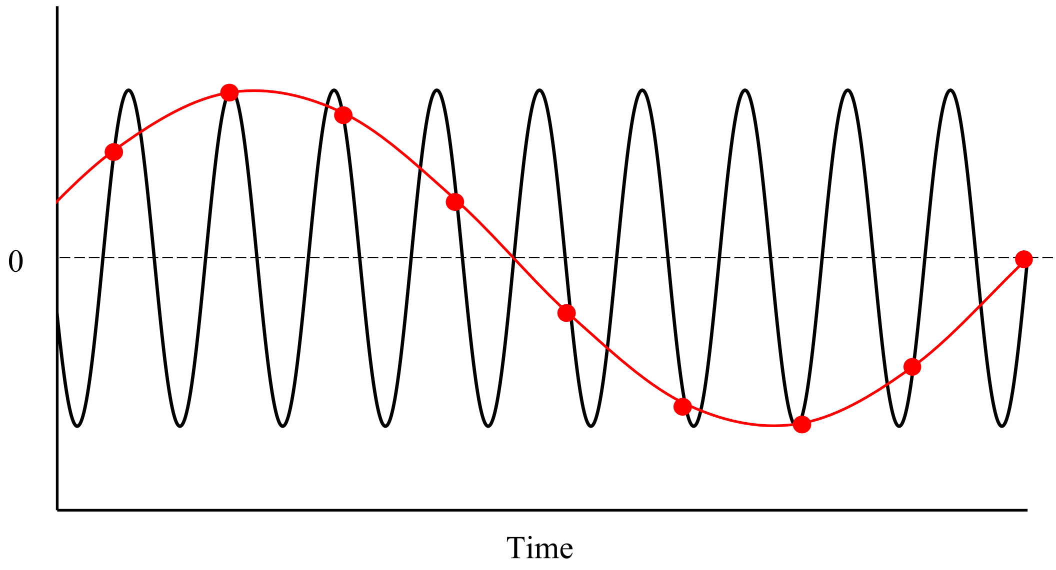

Figure 2.18 *Aliasing as a result of a sampling rate less than half the input frequency. The black trace represents the input waveform and the red markers indicate a potential set of sampling points that produce the appearance of a slower waveform.*

2.5.3.2.4 Filtering frequency

Filtering of signals is the selective rejection, or attenuation, of certain frequencies in order to remove undesirable noise and movement artifacts (Winter, 2009, pg. 27 & 35). The filtering process in CoP measurement is the implementation of a 'low-pass' filter; low-pass, as opposed to high-pass, is a filter that passes signals with a frequency lower than a selected cut-off frequency and attenuates signals with frequencies higher than the cut-off. The choice of the filtering frequency used in CoP acquisition is shown to have a significant effect on the variability of results (Schmid *et al*., 2002). However, this variation is dependent on the nature of the parameter; the effect of filtering frequency is greater for the parameters that contain a derivative operation (such as mean velocity and sway area). This increased sensitivity is due to the operation of filtering emphasising the effect of higher frequencies. As a result, finding an 'optimum' cut-off frequency can be difficult. Throughout the literature a wide range of cut-off frequencies can be seen, mostly ranging from 5 Hz (Carpenter *et al*., 2001; Ferdjallah, Harris & Wertsch, 1999) to 30 Hz (Geurts, Nienhuis and Mulder, 1993). However, there are surprisingly a number of papers showing no information regarding the filtering of their data at all (Chapman, Needham, Allison, Lay & Edwards, 2008; Chiari, Cappello, Lenzi & Della Croce, 2000; Corriveau *et al*., 2000; Demura, Noda, Kitabayashi & Aoki, 2008; Doyle *et al.*, 2005; Doyle, Ragan, Rajendran, Rosengren & Hsiao-Wecksler, 2008; Hill, Carroll, Kalogeropoulos & Schwarz, 1995; Kitabayashi *et al.,* 2003; Le Clair & Riach, 1996; Letz & Gerr, 1995; Mononen, Konttinen, Viitasalo & Era, 2007; Noé & Paillard, 2005; Paillard, Costes-Salon, Lafont & Dupui, 2002; Paillard & Noé, 2006; Pinsault & Vuillerme, 2009; Raymakers *et al.,* 2005; Riley, Brenda, Gill-Body & Krebs, 1995; Rogind, Simonsen, Era & Bliddal, 2003; Samson & Crowe, 1996; Schmit, Regis & Riley, 2005; van der Kooij *et al.*, 2011).

Despite the wide range, authors of these papers rarely provide explanation for their processing choices. Schmid *et al*. (2002) provided an analysis of the sensitivity of CoP parameters to acquisition settings and identified that a standard for filtering frequency is difficult to determine as ICC values prevent it from being easily detected. Despite this, they suggested that if the cut-off frequency were higher than 10 Hz, its variations would not affect parameter estimation and further, although the frequency content of CoP displacement is typically considered as extending up to 2 Hz (Soames & Atha, 1982), their analysis demonstrates there is high variability of parameters around this value and, therefore, propose 10 Hz as an appropriate low-pass cut-off frequency for CoP analysis. A recommended filtering frequency of 10 Hz is also supported by Ruhe *et al.* (2010) and Salavati *et al*. (2009).

2.5.3.3 Number of trial repetitions

Further to the duration of a trial, the number of repetitions required to gain acceptable reliability varies depending on the CoP parameter being investigated. For example, measures relating to CoP velocity have demonstrated good to excellent test-retest reliability when two trials or more were performed (Lafond *et al*., 2004). Other parameters, however, such as RMS and CoP range only demonstrate similar levels of reliability after four trials (Lafond *et al*., 2004). Similar results were shown by Golriz *et al.* (2012) who, when researching repeated measures reliability, identified an average of two measures allowed for reliable measurements of CoP velocity, the average of five measures was required to obtain acceptable reliability for sway area and overall higher measurement precision values were seen by averaging four or five repetitions for all variables. The systematic review by Ruhe *et al*., (2010) suggested that, when comparing results of similar set-ups, the trend for increased data reliability with increased trial repetitions is apparent. An enhanced reliability in mean rather than singular performance measures is expected given the tendency of measures to regress towards the mean when repeated, resulting in a reduced random variation or noise (Beach & Baron, 2014).

2.5.3.4 Protocol design

Protocol design refers to the details of experimental setup and conditions of testing, such as instructions given to participants, body orientation, foot positioning and the visual conditions tested.

2.5.3.4.1 Instructions given

With regards to instructions given to participants, Zok, Mazzà and Cappozzo (2008) reported that participants instructed to "stand as still as possible" demonstrated higher consistency in their CoP displacements than participants instructed to "stand quietly", suggesting the former is more adequate. In this study, most of the investigated parameters showed variations of 8-71% between the two instructions/ statements. With such a small variation in instruction impacting results, it would be reasonable to suggest that those studies who give little to no instruction at all would show less consistency and reliability. Further, if the consistency of instruction between trials of the same study differed, for example instructions were given for one trial but not another, then it could be said that the average of those two trials is not a truly valid measure, nor are they reliably comparable to one another.

2.5.3.4.2 Body orientation

Body orientation, in posturography, refers to the positioning of body segments, in particular the arms. There have been a few variations throughout the literature, some of which can be seen in Tables 2.1 and 2.2. The most common positions are arms folded across the chest and hanging at the sides of the body. As suggested by Ruhe *et al*., (2010), from a biomechanical point of view, arms hanging at the sides of the body is a more appropriate position as it is more likely to keep the CoP in a natural position than if the arms were in front (Benvenuti *et al*., 1999) or behind (Geurts *et al*., 1993) the body. Rogind *et al*., (2003) had participants position their hands on their hips which, despite preventing segment mass from being forward or behind the body, would raise the CoG of the individual higher than its natural position due to segment mass being raised (a concept introduced/ demonstrated in "Centre of Gravity and the Human Body" (Figure 2.7)).

2.5.3.4.3 Foot placement

The foot placement of participants is an additional factor that has the potential to affect reliability. In a number of studies, foot placement has been standardised either by tracing or implementing a pre-determined foot width or angle; these practices ensure replication between trials and participants, and are likely to improve the reliability of measures. The pre-determined positions, however, have been shown to vary largely from one study to the next (Ruhe *et al*., 2010). In addition to this, the effect foot positioning has on reliability of results has conflicting results; this is likely to be a consequence of variations in other factors of the methodology and protocol. The decision to use particular orientations is not always justified, however, Samson and Crowe (1996) chose the feet together stance because it minimised the base of support area on the ground. When the feet are apart or at an angle, the base of support is increased and there is a greater scope to voluntarily shift the weight distribution and, therefore, influence the CoP pathway. Studies looking to compare a 'normative' stance against a narrow stance have found that narrow stance measurements can sometimes lead to lower overall reliability (Hill *et al.,* 1995), but often both reach acceptable levels of reliability independently. Uimonen, Laitakari, Sorri, Bloigu & Palva (1992) investigated the effect of positioning of the feet in posturography and concluded that positioning is not crucial in posturographic measurements provided the distance between the heels is determined. Although non-significant, the study also identified the position chosen by the participant to be more stable than with a specified 30^{o} angle.

Using pre-determined and set positions may ensure repeatability and thus have the potential for higher reliability, but these positions don't reflect the habitual foot placement of different individuals (Herbert-losier and Murray, 2020). A dissertation study conducted at Swansea University assessed the repeatability and consistency when participants self-selected their foot placement and found that participants naturally placed their feet at close to the same distance and angle each time; this suggests that the need to trace foot placement may not be as important as first thought (N. Owen, personal communication, October 13, 2020). Additionally, allowing participants to self-select their foot position would ensure that results were representative of each individual.

2.5.3.4.4 Visual conditions

The two conditions examined in posturography are eyes open and eyes closed. There have been conflicting reports about the reliability of CoP parameters in each condition; Bauer, Gröger, Rupprecht and Gaßmann (2008) recommends eyes closed when assessing static balance following findings of higher reliability in that condition, meanwhile, Santos, Delisle, Larivière, Plamondon and Imbeau (2008) found ICCs were generally higher with eyes open (mean: 0.46, range: 0.03 – 0.76) than with eyes closed (mean: 0.41, range: 0.02 – 0.72) across all summary measures. The methodology of these studies, along with all others reporting reliability of visual condition, vary greatly which makes it more difficult to compare and contrast. With this, a systematic review by Hérbert-losier and Murray (2020) established that, based on studies with 'better-quality' methodology, researchers can be confident that the relative reliability of sway area and path length measures from balance tests is good in both eyes closed and eyes open conditions. In addition to this, van der Kooij *et al.* (2011) showed that differences between eyes closed and eyes open only become obvious when considering their entire 600 s samples. An important factor in eyes open conditions, however, is the presence of a fixation point; as can be seen in Tables 2.1 and 2.2, many studies that use eyes open include a point of visual fixation on the wall in front of the participant. The location of this point varies with respect to its distance and diameter, however, as expected, it is always positioned at eye level. The use of a fixation point ensures that CoP movement isn't affected by distractions or variations in eye movements across the surroundings and, therefore, the condition becomes repeatable and consistent between participants.

2.5.4 Limitations/ details to address

As previously mentioned, there is a vast amount of variability in study design and statistical models from author to author. Since there are so many variations, the measurement techniques and protocols chosen cannot be regarded as representative and influences the extent to which all study conclusions can be used to make judgements about favourable protocols. Despite this, recommendations have been made and, based on the trend of results across the literature, some favourable methods can be identified.

Further to variations in methodology, an additional limitation within a number of studies is the lack of information given regarding the equipment being used. Although studies identify that a FP was used, many do not specify the make and/or model, therefore, further reducing our ability to compare results. Drift is an intrinsic aspect of the piezoelectric FP; but it can differently affect each system because of the variability of both components and the system architectures adopted by each supplier (Quagliarella *et al*., 2008). There are also considered to be some requirements of the technical performance of the measuring device used for posturographic measurements. Scoppa *et al.* (2013) list that the CoP sway signal should be produced by a FP with accuracy better than 0.1 mm, precision better than 0.05 mm, resolution higher than 0.05 mm and linearity better than 90% over the whole range of measurement parameters.

In addition to limited equipment information, as demonstrated in Tables 2.1 and 2.2, only a minority of papers have presented information of the A-D converter resolution within their studies, and of those that have, some limitations can be identified. For example, Ball (2003) found that 12-bit A-D conversion data collection was not accurate enough for CoP measurement in their population (shooters) and so, as a result, we could begin to question the reliability of results from studies using an A-D conversion of 12-bit or lower. Of those studies that did provide A-D converter resolution, it can also be seen that no papers, to my knowledge, have provided further information regarding the 'range'.

The statistical models used by the majority of studies is also considered to be a confounding limitation. Many studies have selected to use ICC models to determine reliability, which may not be the most effective method when used in isolation, as ICC doesn't allow quantification of the error. Blosch *et al*., (2019), however, included the use of Bland and Altman plots and linear regression graphs to illustrate the reliability of their measures, which are methods considered to be more favourable due to the ability to illustrate and quantify deviations and error (Bland & Altman, 1986,1999, 2003).

Table 2.1 *Sample of existing literature that adopted CoP measurements from a force platform in general and clinical populations.*

Variables / Author	Eyes Open /No vision	Fixation point?	Instruction for staying still	Bipedal Foot Position	Arm Position	Duration & Repetitions	Time Slice	A-D converter resolution	Sampling Frequency (Hz)	Filtering Frequency (Hz)
Geurts *et al*. (1993)	Both	No info	Stand as still and symmetrically as possible	Against frame- (heels 8.4cm apart; toeing-out angle 9°)	Hands clasped lightly behind back	20 s x 3 30 s x 2 x 5	No	14-bit & 12-bit	20 & 60	10 & 30
Hill *et al*. (1995)	Eyes open	23 cm square picture 2 m at eye level.	No info	Narrow & Normal stance (12 cm apart) & Replicated	No info	25 s 9 x 3 Rest: 10 s	No	No info	100	No info
Letz & Gerr (1995)	Both	No info	No info	Feet together & heels together 30º	No info	60 s	No	No info	10	No info
Riley *et al*. (1995)	Both	No info	'Stand as still as possible'	Wide: feet parallel 30 cm apart at midheel Narrow: feet together 1cm apart	No info	7 s x 2	0.5 s either end	No info	153	No info
Le Clair & Riach (1996)	Both	No info	No info	Normal stance	Arms resting at their sides	10, 20, 30, 40, 50, 60 x 2	No	No info	50	No info
Samson & Crowe (1996)	Both	Look ahead with head erect.	No info	Feet together	Jendrassik manoeuvre & Hanging loosely, palms resting on thighs	60 s x 10 Rest: 3 mins x 2 protocols	No	No info	25	No info
Takala *et al*. (1997)	Both	Facing wall 1.5m away	'Stand quiet and as stable as possible'	Bar with width 4 cm separated the feet	Arms crossed	30 s 2 x 2	No	12-bit	40	-3 dB at 10 Hz
Benvenuti *et al*. (1999)	Both	No info	'as still as possible'	Large-positioned under anterior superior iliac spine. Narrow – half large (Position recorded & recreated)	Folded and loosely secured to the chest wall	40 s (2 attempts allowed) x 3 Rest: 2 mins	Last 15 s recorded	No info	50	No info
Ferdjallah *et al*. (1999)	Both	No info	Instructed to maintain a quiet comfortable stance	Feet together	Arms to the sides	90 s x 2 Rest: 120 s	No info	No info	20	5
Mientjes & Frank (1999)	Both	No info	No info	Feet no more than one-foot length apart. (Recorded & replicated)	Crossed in front of torso	No info x 3	No info	No info	20	4
Corriveau *et al* (2000)	Eyes open	instructed to look straight ahead with their head erect	Instructed to maintain balance	Feet at pelvis width. Tracings were taken of foot placement	Hanging at their sides	120 s x 9 Rest: 5 mins	No info	No info	20	No info

Chiari *et al*. (2000)	Both	Look at circular achromatic target, diameter of 3 cm, placed at eye height, about 3 m away.	No info	Stand in a comfortable stance.	Arms at the side	50 s x 10	No info	No info	20	No info
Carpenter *et al*. (2001)	Eyes open	Target located approximately 2 m away	Stand quietly	Box defined by dimensions equal to their foot length.	Hanging at their sides	120 s (8 x 15, 4 x 30, 2 x 60) x 3 Rest: 120 s	Trials broken down into segments.	16-bit	20	5
Chiari *et al*. (2002)	Both	gaze straight ahead at a 2 m far achromatic target (a 5 cm diameter circle)	Stand quietly	Recreated by tracing on squared paper. Foot anthropometry assessed	arms at their sides	50 s x 2 Only first valid trial used	No info	No info	200	20 8
Kitabayashi *et al* (2003)	Eyes open	Circular achromatic target placed at eye level	No info	'Closed feet' Romberg posture	Held comfortably	60 s x 3 Rest: 1 min	No	No info	20	No info
Rogind *et al*. (2003)	Both	Black cross on a white wall located 2m away at eye level	No info	Parallel feet approx. shoulder width apart, or tandem Romberg.	Hands placed on hips	25 s x 4	No info	No info	100	No info
Doyle *et al*. (2005)	Both	Look straight ahead	No info	Adjusted according to the patient's height. Heels aligned.	Arms by their sides	10 s x 3	No info	No info	100	No info
Raymakers *et al*. (2005)	Eyes open	Black spot of 10 cm diameter on wall 150 cm away at eye height	Ask to stand as still as possible	Foot parallel on both sides of a 4cm broad T-shaped separator	No info	Recorded for at least 60 s x 2	First 10 s ignored	No info	10	No info
Amoud *et al*. (2007)	Eyes open	Target of a 10-cm cross fixed on the wall two meters in front of the force-plate	No info	No constraint given over foot position	Arms by their sides	Up to 30 s x 4 Includes step onto and off force plate	No info	16-bit	100	10
Doyle *et al*. (2007)	Both	Look at a picture placed 5 m in front at eye level.	Instructed to stand quietly	No info	Arms at their side	90 s x 2 x 10	No info	No info	100	5
Bauer *et al*. (2008)	Eyes open	told to look straight ahead to a point 90cm in front of them	'Stood quietly'	heel distance 2cm & 30° angle & narrow stance (ankles & toes touching).	arms resting at their sides	30 s x 3 Rest: 2 min	Recorded for last 25.6 s	12-bit	40	25
Demura *et al*. (2008)	Eyes open	Fixed eyes on a point in front of them`	No info	Feet together	Arms relaxed at their sides	60 s x 2 Rest: 1 min	No info	No info	20	No info

Doyle *et al*. (2008)	Both	asked to focus on a picture placed 5 m in front of them at eye level	instructed to stand quietly	No info	Arms at their side	90 s 10 x 2 Rest: up to 60 s	Broken down into segments	No info	100	No info
Hadian *et al*. (2008)	Both	'looking forward'	Instructed to stand relaxed	'feet together'	Let their arms hang at their sides	30 s x 3 Rest: 5 mins 1 to 3 familiarisation trials	No info	No info	200	10
Lin *et al*. (2008)	Eyes closed	Head facing straight ahead	Stood as still as possible	'feet together' Placement replicated by outlining feet on poster board	arms at their sides	75 s 2 x 3 Rest: ~ 1 min	Initial 10 s & final 5 s removed	No info	100	5
Santos *et al*. (2008)	Both	stationary target (at approximately eye-level) located 2 m from the centre of the force plate.	instructed to stand quietly	Feet parallel on both sides of a 5.1 cm T-shaped separator	arms hanging to their sides	60 s x 8 (4:4)	No info	16-bit	100	10
Zok *et al*. (2008)	Eyes open	flat screen monitor placed at a distance of 4 m and at eye level. 10 cm diameter circle	"stand quietly" & "stand as still as possible"	Marked and the base of support area was computed	Along the sides of their body,	60 s x 1	No info	No info	100	10
Salavati *et al*. (2009)	Both (Inc. blindfold)	look at a wall approximately 3.8 m in front of their faces	Instructed to stand relaxed	Feet pressed together	Arms resting at their sides	30 s x 3 Rest: 1 min	No info	No info	200	10
Pinsault & Vuillerme (2009)	Eyes closed	n/a	Stand as still as possible	Feet abducted at 30^0, heels separated by 3 cm - Feet outlined for repeatability	Arms hanging loosely by their sides	30 s x 10 Rest: 1 min	No info	No info	64	No info
Van der Kooij *et al*. (2011)	Both	Stationary target at eye level, approx. 2m away.	Stand quietly	Positioned comfortably within square defined by dimensions of foot length (traced for consistency)	Arms hanging at their sides	600 s x 2 Rest: >4 mins	Divided into intervals with 60 s increments	16-bit	20	No info
Donath *et al*. (2012)	Both	Marked circle at the nearby wall (distance: 1.5 m; height: 1.75 m)	Stand as still as possible	Nearly parallel, approx. shoulder width	Hands on hips	30 s x 3 Rest: 1 min	10 s interval between 5^{th} and 15^{th} s	No info	40	10

Table 2.2 *Sample of existing literature that adopted CoP measurements from a force platform in athletic populations.*

Variables / Author	Eyes Open /No vision	Fixation point?	Instruction for staying still?	Bipedal Foot Position	Arm Position	Duration & Repetitions	Time Slice	A-D converter resolution	Sampling Frequency (Hz)	Filtering Frequency (Hz)
Vuillerme *et al*. (2001)	Both	White cross (20x25 cm) 1.20 m away at eye level	Yes	20º relative with heels 4cm apart	'Loosely hanging by their sides'	10 s x 3	No	12-bit	100	10
Paillard *et al*. (2002)	Both	No info	'Stand as still as possible'	No info	'arms hanging by his side'	51.2 s	No	No info	40	No info
Ball *et al*. (2003)	Not specified	No info	No info	No info	Holding rifle	5 s, 3 s, 1 s before shot	No	16-bit	128	4
Schmit *et al*. (2005)	Both	No info	Directed not to speak, gesture or make large voluntary movements	shoulder-width stance	Suspend arms by sides	30 s	No	No info	100	No info
Noe & Paillard (2005)	Both	No info	'remain as still as possible'	No info	No info	51.2 s	No	12-bit	40	No info
Paillard & Noe (2006)	Both	fixed-level target (1 cm) at a distance of 2m.	'stand as still as possible'	30º relative with inter-malleolar distance of 5 cm	'arms hanging along the body'	51.2 s	No	12-bit	40	No info
Paillard *et al*. (2006)	Both	'Gaze in a straight-ahead direction'	'stand as still as possible'	n/a	Arms at sides	51.2 s	No	12-bit	40	6
Mononen *et al*. (2007)	Not specified	No info	No info	No info	No info	3 s to shot	No	12-bit	50	No info
Harringe *et al*. (2008)	Both	Target at eye level 2 m in front	Instructed to stand comfortable	feet close together in platform centre	hanging along the side of the body	120 s x 2	No	16-bit	50	10
Asseman *et al*. (2008)	Both	Gaze horizontal, fixed on a 3 m distanced wall	Yes	Width freely chosen-marked & replicated	'arms are relaxed'	34 s x 5 Rest- 30 s	First & last s	No info	120	5
Calavelle *et al*. (2008)	Both	Eye level target approx. 2 m away	'as immobile as possible'	30º relative with heels 3 cm apart	'hanging freely at their sides'	60 s x 1	No	10-bit	100	20
Chapman *et al*. (2008)	Both	gazing in a natural forward direction at nothing in particular	Attempt to stand as still as possible	'foot position was standardised'	No info	30 s	No	No info	100	No info

2.6 The assessment of athlete performance

The performance of an athlete can be assessed in a number of qualitative and quantitative ways, such as the quality of movement and techniques, along with numerical values such as time to complete an activity or the load and/or volume an athlete can complete. The desired variable can change depending on the goal or outcome, however, in many sports and movements in particular, improvements in qualitative factors can be associated with improvements in quantitative outcomes. For example, it could be said that improvements in a sprinter's technique and the way they carry or drive their arms (i.e., their running economy) could lead to an improvement in running velocity and, therefore, the time it takes them to complete a 100 m race. For this reason, quantitative assessment can be an effective tool and provide a useful insight. A common method of assessing athletic performance is the assessment of neuromuscular performance.

2.6.1 The assessment of neuromuscular performance

The neuromuscular system can be defined as the interaction between the nervous system and muscular systems in the control of joint movements (Watkins, 2014, pg. 112) and the performance of this system can be described as the force-generating capacity of the muscle (Place, Yamada, Bruton & Westerblad, 2010). The basic functional unit in the neuromuscular system that allows the production of force and movement is the motor unit (Duchateau & Enoka, 2011 in Hunter, Pereira & Keenan, 2016) and with this, the force exerted by a muscle-tendon unit during voluntary contractions is the result of the concurrent recruitment of motor units and modulation of the rate at which they discharge action potentials. With regard to control of joint movements, muscles perform a number of different roles, including stabiliser, agonist, prime mover, antagonist, synergist and neutraliser; each muscle that contributes to a particular movement may have more than one role and the relative importance of each of those roles may change throughout a movement (Watkins, 2014, pg. 139).

The measurement of neuromuscular performance has a number of purposes within sport, such as quantification of training status, youth talent identification, strength and power diagnosis and injury prevention (Owen *et al.,* 2014). The neuromuscular performance of an individual is underpinned by a number of anatomical and neuromuscular factors; these variables are commonly assessed by researchers in an attempt to understand their relationship with performance, as well as being used to define performance itself.

The importance of individual variables is disputed, as it is understood that there are numerous determinants for any one area of performance. For example, as addressed in Trezise, Collier and Blazevich (2016), torque produced during a maximum voluntary contraction is not only dependent upon muscle size, the architectural arrangement of its fibres or the level of voluntary activation of those muscles, but is also influenced by several other anatomical and neurological variables such as moment arm distance and activation (or inhibition) of the antagonist muscles.

Further, the combined effects of 'clusters' of neuromuscular variables may be more important, and of more value, to study than any one variable. Despite this, the extent to which multiple factors interact and influence a given variable or performance measure is yet to be fully determined as it is not often that they are measured simultaneously.

Further to exploring the underpinning factors/ determinants of an individual neuromuscular variable, research also chooses to use some variables more broadly in order to generate performance measures that are representative, and of which can form a point of inter- and intra-subject comparison. Meaning that researchers can break down an individual's performance by assessing fewer 'key' neuromuscular variables, such as peak force or mechanical power, that are representative of a larger field of factors/ variables. For example, mechanical power could be a generalised variable representative of factors such as velocity, impulse and force which would be further representative of the speed and strength of muscular contractions or, furthermore, the degree of energy system utilisation, whether that be phosphagen-splitting or oxidative (Margaria, Aghemo & Rovelli, 1966).

In its simplest form, the greater the performance of the neuromuscular system, the greater the ability to exert force and maintain movement efficiency and, with that, a greater ability to produce better values of neuromuscular variables. It, therefore, follows that increased values in neuromuscular performance variables are representative of increased, or greater, athletic performance.

2.6.2 The use of vertical jumping to assess neuromuscular variables

Performance in vertical jumping is a widely used metric in assessing sporting performance indicators and training status by both recreational and professional athletes. It is extensively used for the assessment of neuromuscular variables considered key to successful sporting performance; for example, force, work, displacement, velocity, power and time. Its use as a method of neuromuscular assessment in athletes is valuable, as vertical jumping contributes in varying degrees to performance in most sports (Harman *et al.,* 1990).

There are different types of vertical jumps, for example, running, weighted and unilateral equivalents; however, the most common of these used in performance analysis is a standing vertical jump, where a participant is initially at rest and then jumps in a vertical direction. The two forms of standing vertical jump are the countermovement jump (CMJ) and squat jump (SJ).

2.6.2.1 Countermovement jump

2.6.2.1.1 Kinematics of the vertical motion of the whole-body centre of gravity in a countermovement jump

A countermovement refers to an initial movement in the opposite direction to that of the final movement and, therefore, involves two phases. Watkins (2014, pg. 150) describes a countermovement jump (CMJ) in the following way: In the first phase, the body, or body segments, develop a speed of movement in the opposite direction to that of the final movement, but before the final movement can be initiated this opposite movement must be 'arrested'. Consequently, in the second phase of a countermovement, the muscles of the lower limbs contract; in doing so, the hip, knee and ankle extensor musculotendinous units are forcibly stretched and, as such, act eccentrically. This eccentric phase is then immediately followed by a concentric phase to produce the final single continuous movement in the vertical direction as the participant commences the propulsive phase of the jump. This concentric phase involves the coordinated and more-or-less simultaneous extension of the hips and knees and plantar flexion of the ankles generating sufficient upward speed of the whole-body CoG in order to drive the body vertically upwards while the trunk extends

(Bartlett, 2014, pg. 29-30). Figure 2.19 shows a sequence of key positions for a typical CMJ and the corresponding velocity-time and displacement-time histories.

When the upward direction is considered to be positive, the whole-body CoG during eccentric phase of the jump (A-B) has negative displacement and consequently, negative velocity. Therefore, during the concentric phase (B-D) and upward flight phase (D-E) the CoG has positive displacement and velocity with maximum velocity being reached just before take-off (t_{to}). At point B (the changeover from eccentric to concentric contraction) and point E (maximum height) the vertical velocity of the CoG is momentarily zero. Following t_{to}, the body becomes a projectile, resulting in the CoG continuing with upward displacement, but decreasing velocity, until it reaches maximum height (E), where velocity is zero, after which it falls back to the ground with acceleration due to gravity. The displacement of the CoG at landing can be lower than that of t_{to}, or even the starting/ standing position, due to differences in joint angle- whether this be the ankle, knee, hip or a combination. These differences are exaggerated in point G of Figure 2.19a. Given the relatively low magnitude of velocities achieved in a CMJ ($v < \sim 3\ m \cdot s^{-1}$), the effect of air resistance is usually ignored.

2.6.2.1.2 Kinetics of the vertical motion of the whole-body centre of gravity in a countermovement jump

The velocity and consequent displacement of the CoG are influenced by forces acting on the body due to gravity and coordinated muscle activity. Figure 2.19b shows the corresponding force-time, velocity-time and displacement-time histories from a typical CMJ.

Just prior to the initiation of a jump the participant is stationary and, therefore, the resultant force, R, acting on the participant must be zero. A force-time history at this point would reflect a vertical component of ground reaction force (VGRF), F_y, equal in magnitude, but opposite in direction, to the body weight, BW, of the participant. i.e., $R = BW + F_y = 0$ N (Owen, 2008). Any reduction in F_y would result in downward resultant force and, consequently, downward acceleration of the CoG, i.e., BW > F; over time, this would produce a negative impulse and would consequently result in a downward velocity of the CoG. In the event that $F_y > BW$ there would be an upward resultant force and, consequently, downward deceleration (eccentric) or upward acceleration (concentric) of the CoG; the result of this would be a decrease in downward velocity or increase in upward velocity, respectively.

In order for the CoG to begin upward vertical motion, following the initial negative impulse (representative of the first unweighting phase), the downward velocity must be reduced to zero, i.e., there must be an equal magnitude, but opposite in direction, impulse. This first section of positive impulse is referred to as the first weighting phase (these first two phases are shown by the shaded areas in Figure 2.19b). When balanced, the succeeding positive impulse is responsible for the generation of upward velocity of the CoG. Maximum positive velocity is achieved just prior to t_{to}, as F_y passes through BW (between positions C and D in Figure 2.19). When the participant is no longer able to maintain a ground reaction force greater than their body weight, just prior to t_{to}, there is a small negative impulse; this is termed the second unweighting phase and the consequence is a small decrease in vertical velocity, hence, the peak of vertical velocity is not at the instant of t_{to} (as demonstrated in Figure 2.19).

When airborne, the only force acting on the CoG is force due to gravity. The region between points D and E in Figure 2.19 marks the ascent of the flight phase, whereby the CoG is moving upward with decreasing velocity due to the effect of gravity. When maximum height is reached, the body falls back to the ground, under the influence of gravity, and lands.

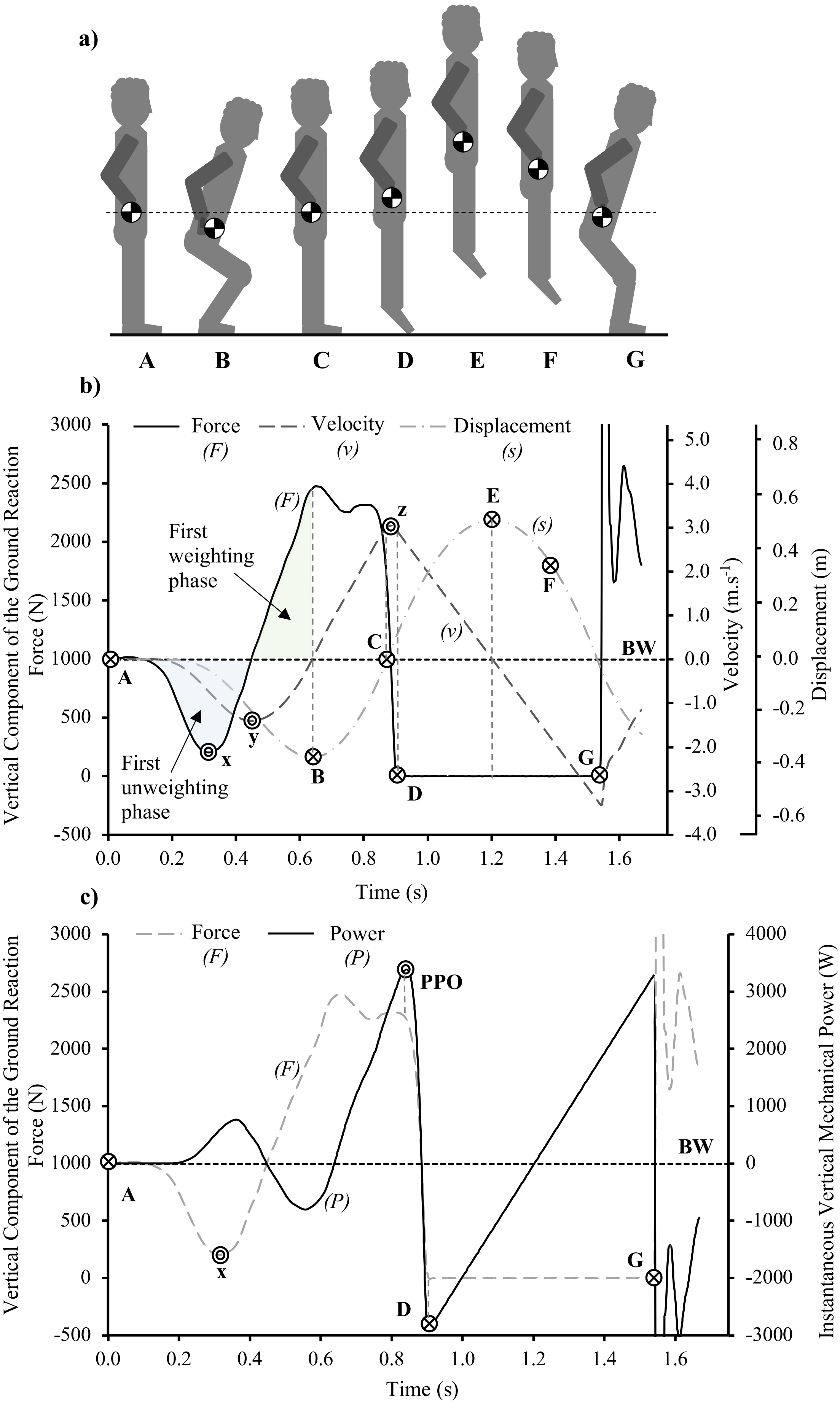
a)
A
B
C
D
E
F
G
b)
Force (F)
Velocity (v)
Displacement (s)
First weighting phase
First unweighting phase
(F)
(v)
(s)
x
y
z
BW
Vertical Component of the Ground Reaction Force (N)
Velocity (m.s-1)
Displacement (m)
Time (s)
c)
Force (F)
Power (P)
PPO
(P)
Instantaneous Vertical Mechanical Power (W)

Figure 2.19 *a) Sequence of actions in a typical countermovement jump. b) Corresponding force-time, velocity-time and displacement-time histories of the whole-body centre of gravity. c) Corresponding force-time and Instantaneous power-time histories.*
A= *start of jump and start of eccentric phase*
B= *limit of downward motion and end of eccentric phase of jump and change over to the concentric phase; velocity is zero*
C= *position in jump where participant's CoG is at the same vertical displacement as the start of the jump*
D= instant of take-off and end of concentric phase NB just after peak velocity
E= *peak vertical displacement of CoG; velocity is zero*
F= *arbitrary point after maximum height during the downward phase/ falling*
G= *instant of landing; peak downward velocity*
x = *maximum downward acceleration*
y = *maximum downward velocity*
z = *peak upward velocity*
PPO = *Peak Instantaneous Mechanical Power*

2.6.3 Utilisation of the stretch-shorting cycle

The countermovement jump (CMJ) utilises the stretch-shortening cycle (SSC), and differences in CMJ performance can be related to the effective use of the SSC. This refers to the pattern of eccentric action followed without a pause by concentric action (Komi, 1992).

2.6.3.1 Storage and utilisation of energy in the stretch-shortening cycle

The concentric contraction of a muscle results in energy expenditure in creating tension and work by pulling skeletal attachments closer together. The amount of work done by a muscle in a concentric contraction is the product of the distance over which the musculotendinous unit shortens and the muscle force (Watkins, 2014, pg. 151). Eccentric muscle contraction also expends energy in creating tension, however, unlike a concentric contraction, the work is done on the musculotendinous unit through lengthening by an external load. In the case of a CMJ, the amount of work done on the musculotendinous unit by the external load is the product of the muscle force and the distance over which the unit is lengthened (as opposed to shortened in concentric contractions).

When energy is expended in stretching a musculotendinous unit, the energy is absorbed in the form of strain energy. The extent to which this strain energy can be used in return for movement is dependent on the speed of the changeover from eccentric to concentric muscle action within the SSC. In a SJ, for example, strain energy is absorbed and stored as the individual lowers into the squat, similar to the CMJ, but the pause at the bottom of the squat reduces the speed of eccentric to concentric changeover and, as a result, energy is dissipated as heat within the muscle; therefore, reducing the utilisation of the SSC.

Effective utilisation of the SSC has been shown to elicit greater neuromuscular performance values with resulting muscular force, work and power during an SSC achieving up to 50% higher values, compared to an isolated concentric contraction (Cavagna, Dusman & Margaria, 1968 and Bosco *et al*., 1987 in Gillen *et al*., 2020). As addressed in Gillen *et al*. (2020), the increase in neuromuscular performance during the SSC is, however, not completely understood as it can be attributed to a number of factors and will also be influenced by the speed of the SSC. The speed refers to the differentiation between slow (longer than 250 ms, such as a CMJ) and fast SSC (shorter than 250 ms, such as drop jumps).

In the initial design of this research, the sport of interest was rowing, so it is important to consider the movement pattern that it requires; since the leg muscles are first stretched (during the 'recovery' phase) and then immediately contracted (during the 'drive' phase) it can be assumed that the entire musculotendinous complex performs an SSC during every stroke cycle (Held, Siebert & Donath, 2020). This assumption is confirmed by eccentric muscle activity during the latter stage of the recovery, 'catch', and start of the 'drive' phase (Fleming, Donne & Mahony, 2014). With regard to the speed of the SSC, Held *et al*., (2020) have identified, using assessment of electromyographic activity, that rowing is more attributable to a slow SSC. As a result, the use of a CMJ to assess neuromuscular variables can be considered as an appropriate measure, and implies that the CMJ reflects a discipline specific muscle action. Therefore, when discussing vertical jumping hereafter, it refers to a CMJ, unless stated otherwise.

2.6.4 Instrumental methods for assessing countermovement jump performance

Traditionally, the most common form of measuring a countermovement jump (CMJ) has been the Sargent jump, also referred to as the jump and reach (Sargent, 1924). In this test the participant performs a static reach as high as possible, marking this point on a vertical board or wall. They then perform a CMJ marking their highest point. The vertical distance between these two marks is recorded as the jump height (JH) and used as an indirect measure of leg power. A recent variation of this test, the Vertec^{Tm}, utilises plastic swivel vanes attached to a telescopic vertical stand; this allows for the adjustment to individual static reach height and consequently jump height. Although highly practical and low cost, these methods, however, have their limitations. In the first instance, both of these forms of jump assessment require the contribution of an arm swing; the use of arm swing has been shown to enhance momentum (Lees, Vanrenterghem & De Clercq, 2004) and is thought to affect results. In addition to this, the reliability and validity of results from these tests rely on the individual's arm position being correct and making contact with the wall or measurement device at the highest point of their jump, as well as avoiding any additional unilateral reach as a result of rotation when trying to reach as high as possible with their dominant hand/ arm. As a result of these factors, and many others, there is likely to be variability seen in performance and, therefore, the jump height may not reflect actual lower limb neuromuscular performance. The reliability and validity of the Vertec has been reported in previous research with notable reports of under-reporting vertical jump performance (Buckthorpe, Morris & Folland, 2012).

Since the development of the jump and reach, Davies and Rennie (1968) proposed a method of measuring the instantaneous vertical mechanical power output of a CMJ by means of a FP. Due to the costs associated, and its lack of provision in field settings, it is not universally used by all researchers, however, the method has become the criterion, or 'gold standard', for the determination of neuromuscular performance variables such as peak force (F_{max}), velocity, rate of force development (RFD), jump height (JH) and mechanical power from a CMJ (Harman *et al.*, 1991; Hatze, 1998; Kibele, 1998; Sayers *et al.,* 1999; Vanrenterghem *et al*., 2001; Canavan & Vescovi, 2004; Owen *et al.*, 2014). Neuromuscular variables such as peak mechanical power output (PPO), JH and velocity are derived mathematically from the VGRF-time history (as illustrated in Figure 2.19).

Performance in a CMJ will be affected by the type of CMJ performed; typically, there are two ways to perform a CMJ, that is with or without arm swing. The use of arm swing in research should be dependent on the aim of the study and, more specifically, the movement pattern of the sport of interest should be considered where possible. For example, if the aim is to assess jump performance of a volleyball player, it may be relevant to include arm swing as this is a fundamental part of the movement pattern of that sport. If, however, the objective is to assess the performance of the lower limbs it would be reasonable to assume that the lower limbs should be isolated, as far as possible, through the removal of arm swing and, consequently, the CMJ should be performed with arms akimbo (Aragon-Vargas & Gross, 1997; Hatze, 1998; Komi & Bosco, 1978).

2.6.4.1 Force platform method

The force platform (FP) method, first introduced by Davies and Rennie (1968) was devised for the measurement of mechanical power in vertical jumping. This method has since become the criterion method due to its ability to directly measure force and, subsequently, more advanced neuromuscular variables. The FP method refers to the collection of the ground reaction force-time history during a vertical jump which is comprised of the three orthogonal components: the vertical component (VGRF), anterior-posterior component (F_y) and medial-lateral component (F_x). Often in the assessment of vertical jump performance we are only concerned with the vertical component and, as such, the other two components are removed during analysis. An example of the resultant force-time history of a CMJ is demonstrated in Figure 2.19. Neuromuscular variables such as displacement, velocity and power are derived mathematically from the VGRF-time history through the process of integration.

For the determination of instantaneous velocity of the whole-body centre of gravity (CoG), the impulse momentum relationship is applied, at the sample rate of the FP, to the net vertical force. Following that, instantaneous power throughout the jump is then calculated from the product of instantaneous vertical ground reaction force and instantaneous vertical velocity (Davies & Rennie, 1968; Winter, 2005) as shown in Equation 2.6.

$$P_i = F_i . v_i \qquad \cdots\cdots 2.6.$$

where: Pi = instantaneous power at time t = t_i (W)

F = instantaneous vertical ground reaction force (VGRF) at time t = t_i (N)

v = instantaneous vertical velocity of CoG at time t = t_i ($m.s^{-1}$)

This approach of instantaneous power calculation was used by Davies and Rennie (1968) with the experimental data measured to an arbitrary scale of intervals of 0.02 s (50 Hz). However, their results were expressed in terms of horse power (h.p.), as opposed to the more commonly used SI units, watts (W). Since the work of Davies and Rennie (1968), advances in the methodology have identified limitations. Factors that contribute to random and systematic error being accumulated during a jump have been identified as the sampling frequency, integration frequency, resolution of the FP, selection of the vertical force range, chosen method of measuring body weight and identification of the start of the jump and start of integration (Hatze, 1998; Kibele, 1998; Owen *et al*., 2014; Street, McMillan, Board, Rasmussen & Heneghan, 2001; Vanrentghem, De Clerq & Van Cleven, 2001). With the exception of sampling frequency, none of these factors were considered in Davies and Rennie (1968), however, this study provided the foundation for the later research that culminated at an accepted criterion method by Owen *et al*. (2014) which, as of March 2022, has been cited 179 times (Google Scholar, 2022).

2.6.4.2 Estimating mechanical power in a countermovement jump using prediction equations

In addition to the mathematical derivation of mechanical power from force-time histories, other methods have been developed as a way to estimate power output. One of these such methods is regression equations.

Due to the force platform method being costly and highly impractical for field-testing, attempts have been made to devise field tests that are capable of predicting neuromuscular variables, namely leg power, from other, more easily measured variables. Typically, the estimation of peak and average mechanical power has been derived from regression equations using body mass (BM), standing height and JH (Harman *et al*., 1991; Johnson & Bahamonde, 1996; Sayers *et al.,* 1999, Shetty, 2002;

Canavan & Vescovi, 2004, Lara, Abián, Alegre, Jiménez & Aguado, 2006) as well as different explanatory variables such as body composition (Güçlüöver & Gülü, 2020). The key point to note regarding regression equations is that, although they may provide a reasonable estimate for a group, they are not appropriate for individual assessment due to the level of error they allow (Tessier, Basset, Simoneau & Teasdale, 2013) and due to the lack of detail provided in the methodology of the developmental studies, the validity of regression equations becomes unclear. Therefore, the use of prediction and estimation equations for power output in vertical jumping are not appropriate and should be avoided; methods of direct force measurement, such as a FP are favourable, provided the appropriate settings are selected and considered, as defined by the accepted criterion method Owen *et al*. (2014).

2.6.4.3 Criterion method of determining neuromuscular variables in a countermovement jump using a force platform

Unlike research into postural stability, a criterion method has been developed for the determination of peak mechanical power (PPO) and other processed variables in a countermovement jump (CMJ) (Owen *et al*, 2014). This criterion method refers to the key variables that must be controlled and specified to achieve maximum validity and reliability for the determination of PPO in a CMJ. In addition, because mechanical power is derived from the product of force and velocity, and force is directly measured, this method also provides a criterion for determination of peak velocity.

The criterion method of Owen *et al.* (2014) specifies the following: (1) the selection of the vertical force range, (2) selection of the resolution, (3) selection of the sampling frequency, (4) the integration frequency, (5) method of numerical integration, (6) the determination of body weight and (7) the identification of the initiation of the countermovement. There has been previous research investigating these specifications (Hatze, 1998; Kibele, 1998; Street *et al*., 2001; Vanrenterghem *et al*., 2001), however, they have each been investigated in isolation; Owen *et al*. (2014) was the first to investigate the combination of all specifications in order to develop a criterion. The criterion method specification devised by Owen *et al*. (2014) is given in Table 2.3

Table 2.3 *Criterion method specification for the measurement of neuromuscular variables in a countermovement jump by the criterion force platform method, as specified by Owen et al. (2014).*

Variable	Criterion method specification
Vertical force range and resolution	5.6 · BW or higher at 16-bit resolution
Sampling frequency	1000 Hz
Integration frequency	1000 Hz
Method of integration	Simpson's rule or trapezoidal rule
Determination of body weight	Mean ground reaction force measured for 1 second of the stationary stance phase immediately before the signal to jump
Determination of initiation of jump	(The instant that BW ± 5 SD is exceeded after the signal to jump has been given) minus 30 ms

2.6.4.3.1 Selection of vertical force range and resolution

Accurate determination of what vertical force range to use is crucial in producing an accurate force-time history; lower ranges would typically be used for balance and gait measurements, whereas higher ranges are typically used for impact and jumping measurements. The total force measured by a platform is the arithmetic sum of each of the four individual corner force transducers; consequently, it is necessary to consider the maximum force transmitted through an individual transducer because, unless the applied force is at the geometrical centre of the platform, the VGRF is going to vary between individual corner transducers (Owen, 2008; Owen *et al*., 2014). Kibele (1998) reported maximum VGRF during a CMJ was in the region of 3 – 3.5 times body weight (BW), however, in this study the component vertical loads were not reported and, as such, this may have led to errors as a result of individual force transducers being exceeded. Owen *et al*. (2014) demonstrated the level of error this vertical range selection would produce if applied to their data set; one or more component transducers would have been exceeded in 47% of the jumps.

The error produced from an overloaded sensor would not initially be obvious from the resultant force profile, as it would either produce a seemingly correct force-time history, but out of the calibrated range or, if the absolute maximum of the transducer had been reached, a plateaued force-time history would be produced (Owen, 2008). The range for the study conducted by Owen *et al*. (2014) was defined as the mean maximum vertical component force plus 3 standard deviations (SD). The corresponding resultant maximum vertical force range for the FP was then calculated at 5.7 times BW. It was concluded that setting a FP's range to this value, or higher, would reduce the probability of it being exceeded to $p \leq 0.003$.

With regard to the selected resolution, despite FPs having a very large dynamic range, there are limitations within the analogue to digital (A-D) converters that restrict the overall resolution of the system. As the range and resolution work in combination, it's important to establish a reasonable compromise that enables a sufficient force range to be measured, while maintaining an appropriate resolution. The most common resolutions used in research are 12-bit (e.g., AMTI) and 16-bit (e.g., Kistler and Bertec) of which are capable of representing an analog signal in 2^{12} and 2^{16} discrete steps, respectively. As a result, it is reasonable to expect a 16-bit A-D converter to better represent an analogue signal and should, therefore, be used when high force ranges are used, such as those in a CMJ.

2.6.4.3.2 Selection of sampling and integration frequency

Generally, when sampling a signal, the higher the sampling frequency the greater the fidelity of the representation of the original signal; Nyquist's sampling theorem (Nyquist, 1928) states that, in order for none of the original signal to be lost during the sampling process, a sampling frequency of double the highest frequency contained in the signal is necessary. This also prevents aliasing, which is demonstrated in Figure 2.18. Usually, the highest frequency within a signal is determined using Fourier's analysis, however, a force-time history of a CMJ is non-cyclical and, as such, cannot be represented by a function. Consequently, this force-time history profile is not suitable for this type of analysis.

With that in mind, several authors have reported a consistent recommendation of 1000 Hz (Kibele, 1998; Linthorne, 2001; Owen *et al*., 2014; Street *et al*., 2001; Vanrenterghem *et al*., 2001). In the criterion method study of Owen *et al*. (2014) 1000 Hz was chosen arbitrarily as, although FP systems have the capability to sample up to 2000 Hz (Hatze, 1998), there would be no need to sample this high because a sampling frequency of 1000 Hz would achieve a precision of <1%. It is also noted that a sampling frequency of 500 Hz would likely produce a similar level of precision, however, 1000 Hz produces more accurate results and has a convenient sampling interval of milliseconds. In order to maintain this level of accuracy when deriving instantaneous velocity and power it is necessary to also integrate at this same frequency so that no data is lost.

2.6.4.3.3 Method of integration

The method of integration refers to the form of numerical integration used to determine the physical quantities other than force, such as velocity, acceleration or displacement. In order to determine a participant's vertical velocity, it is necessary to integrate the expression, $(F_z - m \cdot g)/\, m$, numerically (Kibele, 1998). Integration itself refers to the process of calculating the area between the graph and the x axis. If the graph can be described by an algebraic equation, then often standard integrals can be used to evaluate the area under the graph (Owen, 2008). However, when this isn't possible, other methods must be employed; the most common of which is numerical integration.

There are two methods of numerical integration often used; these are the trapezoidal rule and Simpson's rule (Kibele, 1998). Owen *et al*. (2014) integrated using both Simpson's rule and the trapezoidal rule, at the sampling frequency, to determine the velocity-time data and, hence, mechanical vertical power output. They found that the maximum error, ΔP, in the determination of PPO between the two methods would be, $\Delta P \leq 0.13\%$ (CI = 95%), however, because it was unclear which method of integration gave a more correct result, the best estimate of the correct value of PPO could be taken as the mean of the two results. Therefore, it was concluded that if a maximum error of 0.13% is acceptable, then the two methods can be used interchangeably and, consequently, their criterion method is defined as such.

2.6.4.3.4 Determination of body weight

The criterion method for determining body weight (BW), as defined by Owen *et al*. (2014), is by taking the mean VGRF value, as measured by the FP, for 1 s of the stance phase immediately prior to the signal to jump being given. The determination of BW is critically important due to its use to determine net force and body mass (BM), and subsequently velocity, displacement and power. The impulse-momentum method is highly sensitive to the correct BW determination as an input variable (Cormie, McBride & McCaulley, 2007; Vanrenterghem *et al*., 2001) and, as a result, any errors in BW will result in drifting and errors in those variables dependent on it. The correct determination of BW is also crucial for an accurate identification of jump initiation (t_i) as it is used to define a threshold which, when exceeded, determines t_i. The importance of determining BW from a period of data, as opposed to a single point, is to allow for variation in force as a result of system noise, slight vertical oscillation of the whole-body CoG due to breathing and pendular swing of the CoG over the feet as a result of postural control.

2.6.4.3.5 Determination of initiation of jump and, hence, start of integration

There appears to be no agreed method for the determination of t_i, with some researchers qualitatively assessing where the jump has started through manual inspection of the force trace (Hanson, Leigh & Mynark, 2007), while others defining initiation as the moment BW changes beyond normal variation. In order for this moment to be identified, a relative threshold level of normal variation has to be established. Some researchers have used a relative threshold such as 5% BW during the stance phase (Cormack, Newton, McGuigan & Doyle, 2008; Hori *et al*., 2009; Sheppard, Doyle & Taylor, 2008). However, although reliable, a 5% reduction may not be sensitive enough to highlight the exact moment of t_i and, therefore, may not retain the entire jump signal. An additional method cited numerous times by a variety of authors (Eagles, Sayers, Bousson & Lovell, 2015) defines the start of a jump as the point when the VGRF exceeds a set quiet standing value (typically 10 N). This method is hugely limited as it is likely to produce a false start, especially in participants of a higher BW; for example, in a participant weighing 1000 N, a threshold of 10 N would only represent 1% of their overall BW and would be far too sensitive to expected BW changes within quiet standing. A limitation of all of these methods is their inability to consider that the VGRF at t_i may increase; as demonstrated by Owen *et al*. (2014), not accounting for this factor would identify an incorrect start time for 50% of participants.

Owen *et al*. (2014) sought to identify an instant, such that the entire jump signal was retained whilst ignoring/ removing any data from the stance phase. With that, to account for normal variation of BW, the instant of ± 5 standard deviation (SD) was selected to reduce the probability of erroneous initiation (Owen, 2008) with the extension of subtracting 30 ms from this point (Owen *et al*., 2014).

The validity of velocity and displacement data is a consequence of the integration used to achieve it. Kibele (1998) reported that an error of 5-10 ms in the identification of t_i would result in only a 0.1% error as the rate of change of force would be low at this time; however, he presented no supporting evidence for this claim. The method of numerical integration only allows us to determine the change in velocity and change in displacement at each interval, so in order to know the velocity and position of the CoG at a given point, through the accumulation of instantaneous values, the assumption is made that the velocity of the CoG is zero before the initiation of the jump (during the period of weight measurement). Because the velocity of the participant's CoG is taken to be zero at t_i, it is important that an accurate and reliable starting point is obtained in order for all following calculations and integration values to be valid and valuable; if t_i is identified at the wrong point, this will result in a drifting of accumulated values which results in erroneous derived velocity variable and, subsequently, other variables such as mechanical power (Owen *et al*., 2014).

2.7 Lower limb bilateral asymmetry

Symmetry can be defined as the ability of an object to demonstrate an exact correspondence to size, shape and form across its two halves when split along a given axis (Maloney, 2019). When referring to the human body, mirror symmetry is typically considered along the coronal axis (frontal plane) which allows us to partition the body into left and right halves. Deviation from mirror symmetry across the coronal axis is termed bilateral asymmetry (Maloney, 2019). Asymmetry can be used to refer to deviations from symmetry in a number of ways; for example, anthropometric asymmetry may refer to differences in muscle size or leg length, while fluctuating asymmetry describes a characteristic that would be expected to develop symmetrically but deviates from this path (Van Valen, 1962; Watson & Thornhill, 1994) such as ear size or nostril width. From a biomechanical perspective, kinematic asymmetries could describe such differences as variations in the magnitude of joint angle or angular velocity between left and right limbs during a movement, and kinetic asymmetry would refer to deviations in force production or rate of force development between the mirrored limbs (i.e., in a CMJ the left leg may produce a higher peak force than the right).

The term bilateral asymmetry should not be confused with bilateral deficit, which is used to describe the phenomenon of a reduction in performance during synchronous bilateral movements when compared to the sum of identical unilateral movements (Hay, De Souza & Fukashiro, 2006; Howard & Enoka, 1991; Ohtsuki, 1983; Sale, 1992; Škarabot, Cronin, Strojnik & Avela, 2016). For example, bilateral deficit in a CMJ refers to the maximum force elicited by a two-legged (bilateral) jump being smaller in magnitude than the arithmetic sum of the forces of each leg individually (two maximal unilateral jumps).

2.7.1 Application of bilateral asymmetry assessment

The evaluation of bilateral asymmetry is carried out in both clinical and sports settings, either as between-subject (population vs. controls) or within-subject comparisons (inter-limb asymmetries). With regard to inter-limb asymmetries, a number of classifications of quantifying these differences have been established including left vs. right (Atkins, Bentley, Hurst, Sinclair & Hesketh, 2016; McLean & Tumilty, 1993;

Zifchock *et al.,* 2008), dominant vs. non-dominant (Newton *et al*., 2006; Rouissi *et al*., 2016; Valderrabano *et al*., 2007), stronger vs. weaker (Bailey, Sato, Burnett & Stone, 2015; Impellizzeri *et al.,* 2007) and, in rehabilitation research in particular, injured vs. non-injured limb (Hunt, Sanderson, Moffet & Inglis, 2004; Jordan, Aagaard & Herzog, 2015).

2.7.2 Factors that influence bilateral asymmetry

The presence of bilateral asymmetry may be the result of a number of factors; for example, anthropometric differences, such as leg length discrepancies (Blustein & D'Amico, 1985 in Lawson, Stephens, DeVoe & Reiser, 2006), previous injury (Ferber, Osternig, Woollacott, Wasielewski & Lee, 2004; Paterno, Ford, Myer, Heyl & Hewett, 2007) or repeated performance of a unilateral task or movement that requires the use of one side of the body differently from the other. Krawczyk, Sklad, Majle and Jackiewicz (1998) stated "a high degree of asymmetry observed in athletes is the result of long lasting, intense training and of employing sport-specific techniques and movements. In addition to this, although skill dominance (i.e., handedness) is likely to influence the direction of force asymmetries (i.e., whether the left or right limb elicit greater force production), the way this will manifest will be dependent on sport-specific demands. With that, the type of activity an athlete is engaged in, together with their volume of exposure to the sport, is likely to influence the magnitude of asymmetry (Hart *et al*., 2016; Maloney, 2019).

2.7.3 Methods of bilateral asymmetry assessment

Although it is in the interest of coaches and clinicians to assess bilateral differences between numerous different components of the human body, such as the upper limbs or musculature of the back or lower limbs, it is the targeted assessment of these different areas in isolation that is important to quantify. The focus of this assessment can be further narrowed to the assessment of individual neuromuscular variables, such as variation in muscular strength, peak force production or impulse, both at individuals time points, and during the phases of movement. Within the literature, injury risk and recovery is frequently reported when compared to the prevalence of physical or sports performance analysis, however, the methods of asymmetry analysis are consistent regardless of study objective.

A number of methods have been used for the assessment of bilateral strength asymmetry of the lower limbs, the most common of which being isokinetic and isometric assessments which quantify bilateral force generating asymmetry of specific, selected, muscle groups, i.e., the knee extensors and/or flexors. Despite isokinetic assessment being considered the most accurate method for evaluation of force generating ability (Molczyk, Thigpen, Eickhoff, Goldgar & Gallagher, 1991) and a "gold standard" screening tool for the assessment of hamstring and quadriceps strength (Harding *et al*., 2017; Santos, Pavão, Avila, Salvini & Rocha, 2013), it requires very expensive equipment and its nature (open-chain movement and isokinetic muscle action) is not specific to a lot of sporting activities that would otherwise be characterised by closed-chain movements and fast muscle actions involving the stretch-shortening cycle (Abernethy, Wilson & Logan, 1995). In addition to this, Clark (2001) have suggested that isokinetic quadriceps force generating ability demonstrates weak to moderate and often insignificant relationships with functional tasks. On this basis, unless the objective of a research study is to specifically isolate and assess individual muscle groups, a method that incorporates the SSC and allows for the evaluation of the coordinated movement of the entire lower limb should be favoured. In light of this, an alternative, highly used method in bilateral assessment is jumping.

2.7.3.1 Use of jumping to assess bilateral asymmetry

Unilateral jumping assessments, commonly termed a 'functional performance test (FPT)', are frequently reported in the literature. Typically, FPT require an individual to perform identical functional tests, e.g., hopping on each leg individually with the difference in performance between the two limbs being displayed using a symmetry index to maintain the directionality of the asymmetry (Impellizzeri *et al*., 2007). It is common to see unilateral jump tests in ACL rehabilitation studies (Barber, Noyes, Manginen & Hartman, 1990; Brosky, Nitz, Malone, Caborn & Rayens, 1999; Grindem *et al*., 2011; Juris *et al*., 1997; Barber & Mangine, 1991). Often in this population unilateral tests are the preferred type of FPT due to utilisation of the uninjured limb as a control. In addition to unilateral jumping assessments, more recently bilateral jumps have been used to assess neuromuscular variables of individual limbs while performing a two-legged jump.

2.7.3.2 Unilateral vs bilateral assessment methods

Research investigating the deficit between bilateral and unilateral force production has demonstrated the variability between the data obtained from bilateral and unilateral jumping methods and the factors that influence these resultant differences and, further, influence the reliability and validity of single-leg measures. Consequently, careful consideration should be taken when selecting a desired method for bilateral, inter-limb, asymmetry assessment.

It is important to first consider the differences due to skill effect. Single-legged jumps require an element of balance and coordination in order to counteract the weight and movement of the leg not in use. If an individual has a low level of balance, this is likely to cause increased movement, especially in the preparation phase. A potential consequence may be seen when using a FP to determine t_i ; this increased motion may result in an erroneous interpretation of jump start which will carry forward into the integration, thus, reducing the validity of the measurement. The influence of skill effect also extends to limb dominance, or "lateral preference" (Carpes, Mota & Faria, 2010) which refers to the choice of one side of the body to be used to perform a motor action. Swearingen *et al*. (2011) studied the correlations between single-leg vertical jumps and hop distance/ time and reported that 'participants frequently stated that they felt more confident jumping off their non-dominant lower extremity'. Data from the study also suggested that the non-dominant lower limb was capable of greater functional performance than the dominant limb for single-legged vertical jumps. The effect of lower limb dominance is, however, quite controversial, as throughout the literature there is little consistency on the definition of limb dominance itself, while also being considered as potentially task specific (Velotta, Weyer, Ramirez, Winstead & Bahamonde, 2011). McGrath *et al*. (2016) presented a systematic review of the effect of limb dominance on lower limb functional performance; however, their review is composed of studies with varying methodologies, including the use of isokinetic analysis and functional tests. As such, the results should be interpreted with caution.

Alongside skill effect is the differences in biomechanical demand between the two methods. Studies investigating bilateral deficit have shown that single-legged jumps increase muscle recruitment and, as such, are not representative of the same motion required for two-legged jumps. During a two-legged jump the body weight is equally distributed between the two legs, which results in the muscles of individual legs having a reduced active state in the initial equilibrium position. Further, due to the increased load on individual limbs in a unilateral jump, the push-off/ propulsion phase of the jump is longer compared to bilateral jumps (Bobbert, de Graaf, Jonk & Casius, 2006; Van Soest, Roebroeck, Bobbert, Huijing & van Ingen Schenau, 1985). As a result of these factors, assessing asymmetries between limbs using single-legged, unilateral, jumps may produce different results to single-legged assessments during a two-legged, bilateral, jump.

Due to the differences between unilateral and bilateral jumping it is important to consider the study objectives and the movement patterns of the sport being investigated. The current study looked to assess asymmetries in relation to the rowing motion; rowing inherently requires the equal application of force with both legs simultaneously. As a result, assessment of individual limb differences during a bilateral movement would be most appropriate.

2.7.3.3 Bilateral jumping for the kinetic assessment of lower limb asymmetry

It is widely accepted that analysis of the force-time history of a CMJ performed on a FP is the most valid and reliable methodology for assessing lower limb kinetic neuromuscular variables whilst utilising the SSC. In the assessment of lower limb bilateral asymmetry there are two methods of isolating individual limb contributions during a jump from FP analysis. Previously, Impellizzeri *et al*. (2007) developed an assessment to quantify asymmetries in peak VGRF, where participants jump with one foot placed on a FP and the other on a platform at the same level as the FP. An alternative method increasingly being used, proposed by Newton *et al*. (2006), is the dual-platform methodology in which each lower limb is positioned on an individual FP placed side-by-side. Synchronisation of these platforms allows for the conventional analysis of bilateral actions during the CMJ (Heishman *et al*., 2020), in addition to monitoring the force-time history of individual limbs, thus allowing the delineation of lower limb force contributions and asymmetries (Heishman *et al*., 2019). Table 2.4 provides a sample of the existing literature that have adopted the FP methodology during two-legged CMJs.

2.7.3.3.1 Single force platform methodology

Impellizzeri *et al*. (2007) conducted a study to establish the validity and reliability of a single platform methodology in assessing bilateral asymmetry. The results of their third study, the test-retest reliability of the vertical jump test determined by different numbers of jumps, identified increasing intraclass correlation coefficient (ICC) values and decreasing limits of agreement (LOA) as the number of jumps included in the mean calculation increased from one to five (0.58 [1] to 0.91[5] and ±15.2% [1] to ±6.5% [5], respectively). Further to this, the residuals in the Bland and Altman (B&A) plots showed no evidence of heteroscedasticity (R = 0.02) and the changes in the means between test and retest were not ‘substantial’, except for the bilateral strength asymmetry calculated from only one jump for each leg. It was concluded that the best test-retest reliability was obtained using the mean value of five jumps per leg. However, the results of ICC analysis are influenced by sample heterogeneity (Atkinson & Nevill, 1998) and, as such, do not provide the most effective assessment; the addition of the B&A 95% LOA (Bland & Altman, 1986) to assess the agreement between repeated measures improves the validity of, and confidence in, the conclusions made by a study.

Despite concluding a high reliability of repeated measures with their single platform methodology, the results of Impellizzeri *et al*. (2007) may have limited effectiveness and validity due to the requirement of two separate jump attempts in order to collect just one set of comparable data for individual limbs. A consequence of this would be the influence of changing jump strategies between attempts; this could be overcome by the simultaneous recording of both limbs in a single jump attempt.

This same limitation can be applied to the study conducted by Luk, Winter, O’Neill and Thompson (2014) who also adopted a single platform methodology in their examination of unilateral and bilateral force production differences in powerlifters and field jumpers in order to determine the existence of leg dominance. This study provided limited reports of reliability, the only given detail was that the test-retest reliability of the dependent variables showed ICCs of $R \geq 0.947$; this would suggest a very high level of test-retest reliability. However, details of the study protocol of Luk *et al.* (2014) are limited, consequently their results need to be interpreted with caution. Table 2.4, records that there are no details regarding the acquisition settings of the FP, so we cannot be certain of the accuracy and validity of the data. Also, given that mechanical power and velocity are variables requiring integration, it is important to understand how this data was derived, including the method of determining t_i. A further limitation to this study is their small sample size (11 M powerlifters, 8 M field jumpers). Loken & Gelman (2017) note that greater measurement error leads to greater variation in the measured effect sizes which, particularly for smaller sample sizes, means that some measured effect sizes overestimate the true effect size by chance.

2.7.3.3.2 Dual-platform methodology

Newton *et al*. (2006) proposed a method of determining the asymmetries in GRF generated by individual limbs by simultaneously recording information from two, side-by-side FPs. Consequently, the need to conduct two individual jumps for a single comparison between limbs becomes unnecessary, removing the limitation of changing jump strategies affecting inter-limb differences. There were a number of objectives within the Newton *et al*. (2006) study, however, the methodology referring to double-leg vertical jumping was composed of two maximal CMJs with the inclusion of arm swing (AS). AS can influence outcome variables in a CMJ (Lees *et al.,* 2004) and, as such, may reduce the validity of lower limb assessment.

While the impact of AS has been explored in relation to CMJ performance, it is only very recently that the influence of AS on the reliability of inter-limb asymmetries during a bilateral CMJ has been explored (Heishman *et al*., 2019; Heishman *et al*., 2020). Heishman *et al*. (2019) identified that in both conditions the majority of variables met their acceptable criterion of intersession and intrasession relative reliability, (ICC > 0.700), while fewer than half met standards established for absolute reliability (CV < 10%). These criteria were established ‘based upon recommendations from prior literature’ (Baumgartner & Chung, 2001; Cormack *et al*., 2008; Gathercole, Sporer, Stellingwerff & Sleivert, 2015). The results of the study must, however, be interpreted with caution due to its relatively small sample size (22 M), which can have an effect on generating a lack of statistical power. In addition to this, despite a detailed methodology, the only specification of the method reported was sampling frequency (1000 Hz). The force range, A-D resolution nor filtering of data specifications were reported. It is given in the methodology that the study defined t_i as an offset of 20 N from measured BW quantified before the jump. Despite there being no agreed method for the determination of t_i, methods that define it as the point when VGRF exceeds a set value are limited greatly as they are likely to produce an erroneous initiation point, especially in participants of higher bodyweight. Owen *et al.* (2014) identified that to account for normal variation of BW, the instant of ± 5 standard deviations (SD) minus 30 ms is an appropriate method and would improve the validity of velocity and displacement data determined by integration of the VGRF-time history.

The results of Heishman *et al.* (2019; 2020) state that both AS and no arm swing (NAS) CMJ conditions provide reliable information with respect to inter-limb asymmetries, however, the inclusion of AS has an influence on the variability of lower extremity inter-limb asymmetries. They reported that AS influenced CMJ symmetry during the loading phase of the CMJ, but improved asymmetry during propulsion. However, the study was conducted in a population of skilled jumpers and, as such, there would be a much higher level of familiarity to the arm swing protocol which may not translate to different study populations.

With the exception of the studies by Heishman *et al.* (2019; 2020) who tested both conditions, the remainder of dual-platform methodology studies considered (Benjanuvatra, Lay, Alderson & Blanksby, 2013; Jordan *et al*., 2015; Menzel *et al*., 2013) have utilised a NAS condition in order to isolate the lower limbs.

A typically measured variable in bilateral asymmetry assessment is peak vertical component of the ground reaction force (F_{max}) (Benjanuvatra *et al*., 2013; Heishman *et al*., 2019, 2020; Impellizzeri *et al*., 2007; Luk *et al*., 2014; Menzel *et al*., 2013; Newton *et al*., 2006). However, despite F_{max} being an important variable in assessing neuromuscular capacity, it is only representative of an instant in the CMJ.

Benjanuvatra *et al*. (2013), Jordan *et al*. (2015) and Menzel *et al*. (2013) are in agreement that phase specific analysis of a CMJ can also be beneficial. Jordan *et al*. (2015) report the use of phase-specific kinetic impulse, J, calculation based on pilot data of VGRF-time history that revealed directional asymmetries throughout CMJs, thus providing a rationale for this analytical approach. Consequently, as the propulsion and change of velocity of the CoG are mechanically determined by the impulse, and, as impulse measures the accelerative force required to change the momentum of the body, its inclusion in bilateral assessment should be considered. For these reasons, it is likely that measuring impulse is perhaps more suitable for assessing CMJ performance than F_{max} alone. Menzel *et al*. (2013) reported that impulse seemed to be a more sensitive variable for the identification of asymmetries and suggested that the higher number of identified asymmetries can be explained by the higher dispersion of impulse compared with the maximal force while the cut-off criteria value applied remained the same (15%). Consequently, it was proposed that further studies would be necessary to determine whether the same cut-off value could be adequately applied for all variables or whether the level of dispersion of the variable should also be taken into consideration and scaled as such. Therefore, considering the dispersion and consequently altering cut-off values for distinct variables, the consistency of the diagnostic information could be enhanced. Previous studies (Impellizzeri *et al*., 2007; Newton *et al*., 2006) did not include impulse in their investigation so did not allow for more in-depth and varied interpretations of force production during the CMJ.

Both Benjanuvatra *et al*. (2013) and Menzel *et al*. (2013) provide reports of repeatability in their studies. Benjanuvatra *et al*. (2013) used an ICC to determine a strong relationship (r = 0.99, 95% CI: 0.96-0.99) between two testing sessions for asymmetry index and impulse, while Menzel *et al*. (2013) identified, within a pilot study of 3 valid attempts on a single day, ICCs for bilateral differences of maximal force of 0.74 ($p < 0.05$), of impulse 0.71 ($p < 0.05$) and of maximal power 0.81 ($p < 0.05$). It should be noted, however, that although the primary study conducted by Benjanuvatra *et al*. (2013) had 58 participants (28M, 30F), only five male and female participants returned for the repeatability assessment, creating a very small sample size for this form of analysis. As a result, the small sample size, in addition to the aforementioned limitations of ICC assessments, reduces the validity of their repeatability conclusion. The same is true for Menzel *et al*. (2013) whom had a study population of 46, but only a sample size of 16 for the pilot study of which the repeatability was reported from. Jordan *et al*. (2015) provided no report of the test-retest reliability of their study, despite their analysis utilising the average of ten trials.

Despite all three of these studies using impulse as a key variable, there is very limited information regarding the numerical integration that is required for its calculation. Benjanuvatra *et al*. (2013) simply states in their methodology that the ‘left and right vertical impulses during the extension phase…were calculated’, providing no detail on the method used to identify t_i and subsequent integration. Menzel *et al*. (2013) identified the start of the CMJ, however, it was taken as ‘the moment when the GRF dropped below body weight'; as previously established, this is not considered an appropriate method as it is likely to produce an erroneous initiation time. As a result, the corresponding numerical integration will have been flawed, reducing both the validity and reliability of their results and subsequent analysis.

Eccentric and concentric phase impulse were the only variables assessed in the study by Jordan *et al* (2015), Table 2.4. However, there was a lack of detail provided in the methodology regarding the way they were derived. Given the significance of correctly and accurately calculating impulse on the study's outcome, we cannot be sure of the accuracy, reliability and validity of their results, consequently the study's conclusions should be interpreted with caution. The study has, however, provided more information than previous studies; the velocity of the body's CoG was obtained by time integration of the instantaneous acceleration signal calculated from F_z and from the velocity, the eccentric deceleration phase was defined as the time interval from the maximum negative velocity to zero velocity. The concentric phase was defined from this instant of zero CoG velocity to the instant of t_{to}. The study does not, however, provide detail of how t_i, or time point 0, was determined and, therefore, the numerical integration that they describe using could be considered unreliable as a consequence of an erroneous start time. Additionally, Jordan *et al.* (2015) provided no definition of t_{to} and, as such, the validity of the concentric phase impulse calculation cannot be guaranteed due to its dependence on phase duration determined from point of zero velocity and t_{to}.

Table 2.4 provides a summary for the preceding discussion.

Table 2.4 *Sample of existing literature that have adopted the force platform methodology during two-legged CMJ for the assessment of lower limb bilateral asymmetry.*

Author	Separation of limbs	One FP or Dual Platform	Acquisition settings	Number of jumps	Inclusion of arms	Outcome variables	Calculation of asymmetry
Newton *et al*. (2006)	Left vs Right & Dominant vs Non-Dominant	Dual Platform	No Info	2	Yes	Peak Force & Average Force	LvsR = $\frac{right\ leg\ score - left\ leg\ score}{right\ leg\ score} \times 100$ D vs ND = $\frac{strong\ leg - weak\ leg}{strong\ leg} \times 100$
Impellizzeri *et al*. (2007)	Left vs Right & Stronger vs Weaker	Single force plate next to level platform 1cm apart	500 Hz	10 (5 each leg)	No- hands on hips	Peak force	$\frac{stronger - weaker}{stronger} \times 100$ (-) assigned when left was the stronger limb & (+) when right leg was stronger
Benjanvatra *et al*. (2013)	Left vs Right	2 side-by-side force plates	2000 Hz, low-pass filter with cut-off frequency of 16 Hz	5	No- hands on hips	Duration of propulsion phase (PU-TIME), Impulse (I), Peak & mean force (GRF)	$\frac{(x_r - x_l)}{Max\ (x_r, x_l)} \times 100$ Where x_r is the right limb and x_l is the corresponding variable for the left limb. A positive value represents a greater magnitude recorded by the right limb.
Menzel *et al*. (2013)	Left vs Right	2 side-by-side mounted platforms	1000 Hz low-pass filtered at 50 Hz Jump initiation was moment GRF dropped below BW	3	No- hands fixed on hips	F_{max}, P_{max} & I	$\frac{value\ of\ the\ right\ limb - value\ of\ the\ left\ limb}{greater\ value\ of\ both\ limbs} \times 100$ Positive indicates higher values of the right leg, and a negative indicates higher values of the left leg 'Absolute LSI' removed the directionality of the asymmetry

Table 2.4 *Continued.*

Luk *et al*. (2014)	Dominant vs Non-Dominant	Single force plate next to platform	No info	3	No- hands on hips	Force, Power & Velocity	$1 - \frac{ND\ limb}{D\ limb} \times 100\%$
Jordan *et al*. (2015)	Left vs Right	Dual force plate system	500 Hz	10	No- hands on hips	Impulse -eccentric & concentric	$\frac{left\ impulse - right\ impulse}{Maximum\ of\ left\ and\ right\ impulse} \times 100$ $\frac{uninjured\ impulse - ACL-R\ impulse}{Maximum\ of\ left\ and\ right\ impulse} \times 100$
Heishman *et al*. (2019) & (2020)	Left vs Right	Dual force platforms	1000 Hz 20 N offset from bodyweight defined the start of movement.	3 AS 3 NAS	Both conditions	F_z at PPO, F_z at zero vel, take-off peak force Concentric: imp, mean & peak force Eccentric: Brak RFD, Dec RFD, mean & peak force.	$Lower - limb\ difference\ score = (Right - Left)$ Only the arithmetic difference with respect to each variable was presented, not the percentage difference.

2.7.4 Limitation of jumping as a method of asymmetry assessment

There are some limitations to the use of jumping for asymmetry assessment; jumping requires precise coordination of the muscles responsible for movement at the ankle, knee and hip which means that performance can be highly sensitive to changes in muscle activation timing patterns (Prokopow, Hay, Fukashiro & Himeno, 2005 in Hay *et al.,* 2006). This not only supports the argument that single-legged and two-legged jumps are not comparable, but also introduces the difficulties that could be encountered through differences in adopted strategies between individual attempts/ repetitions through means of micro-adjustments of posture or altered limb loading and contractile speeds (Latash & Latash, 1996).

2.8 Purpose of the current study

It is clear from the existing literature that despite numerous studies using CoP analysis of quiet standing for the assessment of postural stability, there is no agreed methodology. The reported heterogeneous protocols and FP acquisition settings have made it very challenging to compare and contrast the outcomes of studies that are beneficial for research progression. There have, however, been individual investigations into some components of the area of study, that provide recommendations for a more reliable and, or, valid methodology. Therefore, the purpose of the current study was to determine whether a postural stability assessment protocol, reported using the recommendations of previous research, yields reliable and valid results.

The extent of the current literature investigating bilateral asymmetry using a bilateral CMJ is limited, with questionable applicability and comparability of results between studies, due to the dearth of clear information, or in some cases no information, in the studies' methodologies. Therefore, the current study incorporated the criterion method for the determination of peak mechanical power, as a starting point for methodological clarity, in a bilateral CMJ. In terms of asymmetry metrics, both 'sided' and 'un-sided' methods of calculation were used to investigate the impact of reference value selection on asymmetry measurement and interpretation.

Consequent to the limitations in previous literature, the current study sought to provide full details of all methodological processes and acquisition settings as to ensure easy reproducibility. Additionally, the reliability and test-retest agreement of the results obtained from methods used in the current study were investigated in order to form further recommendations for future research.

Chapter 3

Methodology

3.1 Methodological overview

The research comprised of two individual protocols, each with the objective of assessing postural stability and bilateral asymmetry of the lower limbs via a countermovement jump, respectively. To assess the test-retest reliability of all parameters investigated, a repeated measures design was used to carry out identical protocols on two testing sessions a few days apart. The research structure and order of protocols are provided in Appendix B.

3.2 Experimental protocols

3.2.1 Participant population and recruitment

Male (n = 10) and female (n = 9) members of the Swansea University staff and postgraduate offices took part in the study after providing informed and written consent. Participant anthropometric data was recorded prior to testing and can be found in Table 3.1. Prior ethical authorisation was given by the College of Engineering Research and Ethics Committee at Swansea University (Appendix N). To determine fitness and wellbeing and ensure participants were unaffected by any optical, vestibular, musculoskeletal, aural or neurological health condition that may influence their capability to maintain equilibrium, participants were required to complete a PAR-Q (Appendix D); any participant reporting a current health condition or lower back pain in the last 6 months was excluded from the study. Prior to each individual element of data collection, the testing procedure was fully explained and participants were familiarised with the relevant equipment.

To remove any potential consequence of circadian or diurnal variability or 'time-of-day' effects (Atkinson & Reilly, 1996; Faria & Drummond, 1982; Gribble, Tucker & White, 2007; Hill & Smith, 1991; Melhim, 1993; Wyse, Mercer & Gleeson, 1994), each participant completed their trials at the same time of day on both sessions.

3.2.1.1 Determination of anthropometric measurements

Body mass (BM) for each participant was determined from force platform (FP) data collected during the familiarisation stage of the postural stability protocol, as outlined below. During this time, participants were stood with minimal clothing and without shoes as to ensure a valid measure. The FP provided a measure of body weight (BW) across a 100 s period, the average of BW between 10 and 15 s was used and subsequently divided by 9.80665 $m.s^{-2}$ (as the value adopted in the international service of weights and measures (BIPM, 1901)) to give BM in kilograms (kg). Stature data was obtained using a stadiometer (Seca 213, Birmingham, United Kingdom) and reported to the nearest 0.001 m. Age was obtained from the participant's PAR-Q completed prior to testing.

Table 3.1 *Participant anthropometric data.*

	Age (yrs.)		Height (m)		Mass (kg)	
	Mean	± SD	Mean	± SD	Mean	± SD
Female	32.4	8.7	1.662	0.055	70.8	13.5
Male	32.7	9.5	1.797	0.060	88.2	14.4
Entire study population	32.6	9.1	1.733	0.089	80.0	16.4

3.2.2 Postural stability assessment

The following section refers hypotheses 1 through 4 and hypothesis 8, listed below:

1. There is no criterion method for the determination of postural stability performance, however, a methodology devised from the recommendations of previous research will produce valid and reliable results.
2. As sampling duration increases, so will the test-retest reliability of postural stability parameters derived from CoP analysis.
3. As the number of repetitions increases, so will the precision and validity of postural control outcome measures.
4. Visual condition will have a significant impact on postural stability performance. Further, CoP excursion measures will be significantly higher when visual feedback is removed.
8. Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population.

3.2.2.1 Trial protocol

Quiet standing centre of pressure (CoP) analysis was conducted for eight trials under the following conditions, eyes open (EO) and eyes closed (EC) prefaced by a familiarisation trial. The trials were divided equally between two testing sessions with the familiarisation and four trials of each condition on the first day and four trials for each condition on the second day. The order of presentation for EO and EC trials was randomised and counter-balanced across participants to minimise potential order effects; the randomised matrix for order of trials for each individual participant is shown in Appendix E. The sampling duration for all trials was 100 s (Section 2.5.3.2.1) with approximately 120 s period of seated rest in between each standing trial, to minimise any effects due to fatigue.

Each participant was required to stand unshod on a FP (model number 92866A, Kistler Instruments Ltd., Farnborough, United Kingdom) with their feet in a self-selected position (Section 2.5.3.4.3), approximately shoulder width apart, and arms hanging by their sides. Participants were given the instruction to 'stand as still as possible' whilst keeping their head in a normal forward-facing position, with eyes focussed on a stationary target located approximately 1.5 m away at eye-level for the EO condition.

In a rowing boat, although there are shoes, they have only very thin soles and only act as a rigid surface to affix to; they do not provide any additional support to the underside of the foot. With this, boat shoes are a constant variable throughout the crew, as is measuring participants of a postural control study barefooted, as it removes any additional variability in shoes worn.

Environmental conditions (such as lighting and noises) were kept stable between trials, testing sessions and between participants and the FP was set-up on a flat and rigid laboratory floor according to the manufacturer's installation requirements.

For each trial, ground reaction forces and moments in three planes were collected for the duration of 100 s at a sampling frequency of 100 Hz and interfaced to a data recording computer via a 16-bit analog-to-digital converter in a data acquisition system (DAQ system for Bioware®, Type 5691, Kistler Instruments Ltd., Farnborough, United Kingdom). The FP was set-up to record with a shear range of 637 N and vertical range of 5362 N. Displacement of CoP was calculated for each individual trial, at the sampling frequency, following the removal of all data for the first 5 s and last 5 s of the trial. The remaining 90 s epoch was analysed as three consecutive 30 s records, two 60 s records and the entire 90 s of CoP displacement. Figure 3.1 illustrates the section of the sample covered by each epoch.

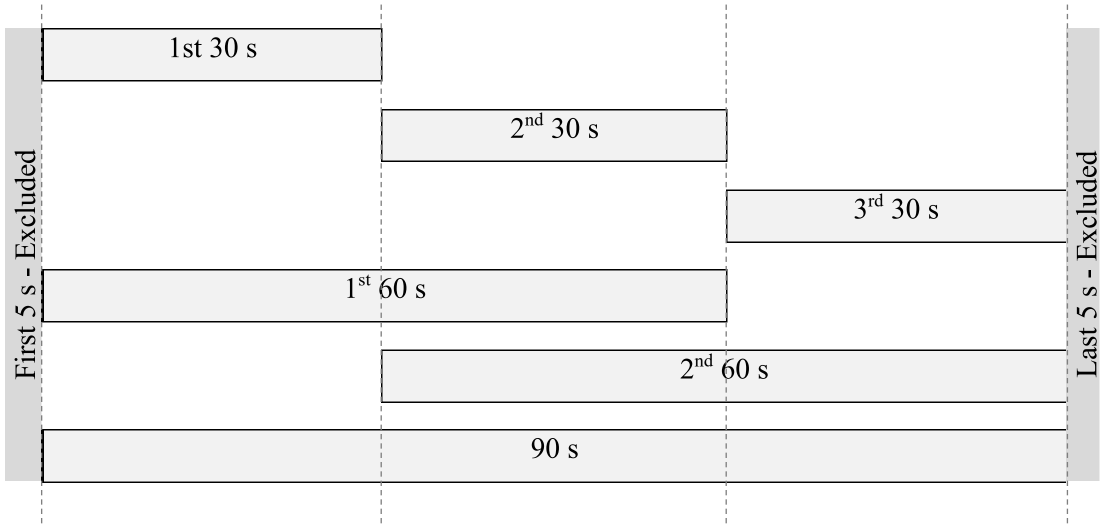

Figure 3.1 *Configuration of each of the six epochs considered in the analysis.*

Prior to time slicing (to avoid any edge effects), force-time histories were filtered with a dual pass Butterworth filter with a low-pass cut-off frequency of 10 Hz to reduce measurement noise (Błaszczyk, 2016). Data signals were recorded onto a computer system using Bioware (Type 2812A, Version 5.3.2.9; Kistler Instruments Ltd., Farnborough, United Kingdom) where, within the global coordinate system, y represents medial-lateral (ML) displacement and x refers to anterior-posterior (AP) displacement. A copy of the ground reaction force-time record was exported from the data collection software to Microsoft Excel (Microsoft Corp. Version 16. Redmond, WA)

Prior to data collection, the validity of static CoP measurement was verified using methods simplified from Blosch *et al* (2019). A known load was placed at five different positions (P1-P5) on the surface of the FP; positions were the geographical centre of the platform and the four corner points. CoP coordinates were recorded over a time interval of 10 s and repeated three times. Calibration checks were performed with calibration weights that were traceable to national standards.

3.2.2.2 Outcome variables

Only the variables previously established as having a high level of validity and reliability were chosen for analysis; these measures were total path length (L_p), mean velocity (V_m) and sway area (A_s), as defined by the area of a 95% confidence interval (CI) ellipse (Blosch *et al*, 2019; Donath *et al*, 2012; Hérbert-losier & Murray, 2020). It was deemed appropriate to use fewer variables that characterise different aspects of postural stability in order to provide a larger, overall, analysis of performance.

The CoP trajectory recorded by a FP across a quiet standing trial can be displayed as a statokinesigram (Directions, 1983), as illustrated in Figure 3.2. The use of a statokinesigram allows for visual interpretation and analysis of data by providing an apparently erratic time-course of the CoP. Appendix F. shows the specialised spreadsheet designed for the calculation of postural stability measures.

3.2.2.2.1 Total path length

Path length (L_p) refers to the quantification of the magnitude of the two-dimensional displacement based on the total distance travelled by the CoP trajectory over the course of the trial. Assessments of performance are based on the premise that the smaller the path length, the better an individual's postural stability (Donath *et al.*, 2012). L_p was calculated following the input of CoP coordinate data into a specialised spreadsheet (Appendix F.) using Equation 3.1; the following equation is representative of the calculation for analysis of a 90 s record.

Equation for calculation of L_p.

$$= \sum_{n=0}^{n=9000} \sqrt{(x_i - x_{i-1})^2 + (y_i - y_{i-1})^2} \quad \cdots\cdots\cdots\cdots \quad 3.1.$$

where: x_i and y_i = coordinates of CoP

n = number of points comprising CoP trajectory

3.2.2.2.2 Mean velocity

For this CoP parameter, the magnitude of the velocity of CoP excursion for each discrete time interval was calculated individually before calculating the average of these values. The following equation is representative of the calculation necessary for analysis of a 90 s record.

Equation for calculation of V_m:

$$= \frac{1}{n}\sum_{n=0}^{n=9000} \frac{\sqrt{(x_{i+1}-x_{i-1})^2+(y_{i+1}-y_{i-1})^2}}{\Delta t} \quad \cdots\cdots\cdots \quad 3.2.$$

where: x_i and y_i = coordinates of CoP

Δt = change in time/ time interval

n = number of points comprising CoP trajectory

3.2.2.2.3 Sway area

Sway area (A_s) refers to the quantification of 95% of the total area covered in the anterior-posterior and medial-lateral direction using the smallest ellipse containing 95% of all CoP coordinate data points. This 95% CI was determined from calculating 1.96 SD in the M/L and A/P directions. The ellipse area is considered to be an index of overall postural performance and it has been suggested by Asseman *et al.*, (2004) that the smaller the ellipse area, the better the "performance".

The standard form of the equation of an ellipse with centre (h, k) and major axis parallel to the x-axis is:

$$= \frac{(x-h)^2}{a^2}+\frac{(y-k)^2}{b^2}=1 \quad \cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots \quad 3.3.$$

where: $a > b$

a and b = the radius of the x- and y- axis, respectively.

the length of the major axis is 2 a

the coordinates of the vertices are $(h \pm a\,, k)$

the length of the minor axis is 2 b

the coordinates of the co-vertices are $(h, k \pm b)$

Following from this, the equation for the area of an ellipse is $A = \pi \cdot a \cdot b$ where, a and b are the radius of the x-axis and y-axis, respectively. Therefore, the area of a 95% confidence interval ellipse can be represented with Equation 3.4.

$$= \quad \pi \cdot 1.96(\sigma \cdot x) \cdot 1.96(\sigma \cdot y) \quad \cdots\cdots\cdots\cdots\cdots\cdots\cdots \quad 3.4.$$

where: x = sway of the CoP coordinate trajectory in the anterior-posterior direction

y = sway of the CoP coordinate trajectory in the medial-lateral direction

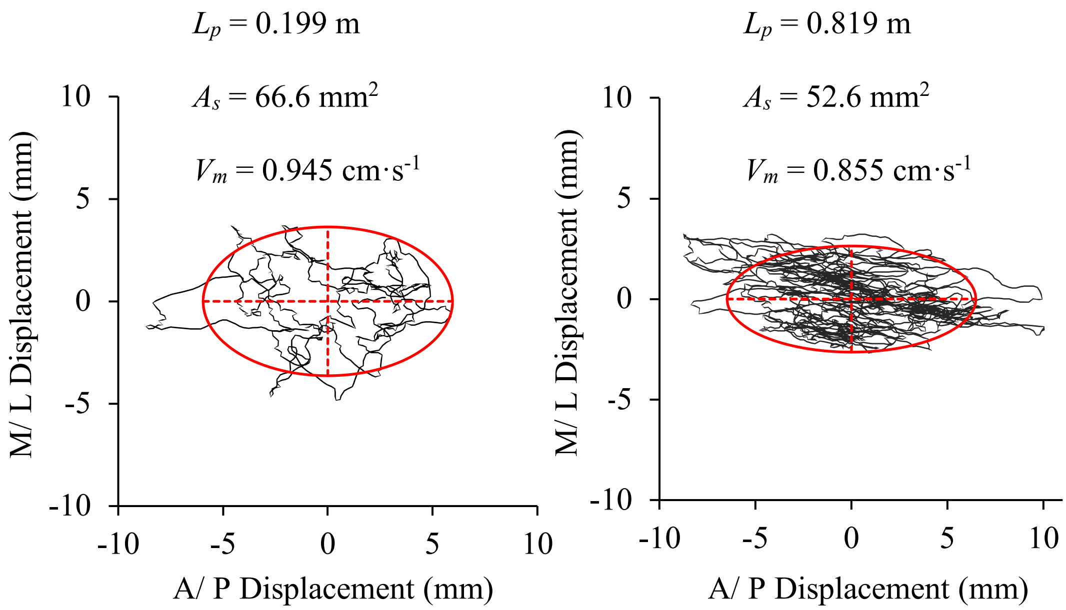

Figure 3.2. *Example statokinesigrams for 20 s (left) and 90 s (right) trials of quiet standing showing the 95% CI ellipse and the corresponding performance variables for each trace.* L_p *= path length,* A_s *= sway area,* V_m *= mean velocity.*

3.2.3 Lower limb bilateral asymmetry assessment

The following section refers to hypotheses 5 through 8, listed below:

5. Bilateral CMJ performance, defined using the criterion method for the determination of neuromuscular variables, will show excellent test-retest reliability and day-to-day agreement.
6. In the execution of a bilateral CMJ, there will be differences in the contribution of each individual limb to overall performance. These variations in the magnitude of vertical component of the ground reaction force are the result of differences in the force generating capacity between limbs.
7. 'Sided' asymmetry calculations will produce statistically significantly different results to 'un-sided'. Additionally, 'sided' methods of asymmetry calculation will have lower reliability and validity than 'un-sided' methods quantifying absolute asymmetry, due to the consequences of nullification when producing group means.

8. Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population.

Assessment of lower limb bilateral asymmetry was assessed via countermovement jumps (CMJs) performed off two FPs, one for each foot. Jumps were carried out over two testing sessions (five trials per testing session) following a standardised, prescribed warm-up, as outlined below. Neuromuscular performance variables were determined from each CMJ by analysis of the vertical component of the ground reaction force-time history (VGRF), in addition to the left and right leg contributions towards each jump, in terms of VGRF, and, consequently, the percentage asymmetry, or symmetry index (SI), between the two lower limbs.

3.2.3.1 Warm-up protocol

Participants completed a standardised warm-up based on the RAMP protocol as outlined by Jeffreys (2006); the warm-up was designed to raise the metabolic indicators, such as heart rate and breathing, and activate, mobilise and potentiate the muscles and joints associated with vertical jumping. After five minutes of sub-maximal rowing on a rowing ergometer, followed by approximately ten minutes of dynamic stretching (Pagaduan, Pojskić, Užičanin & Babajić, 2012), participants carried out ten sub-maximal jumps followed by five maximal effort jumps on the dual-platform set-up to avoid any potentiation effects of repeated jumping. This procedure was developed to improve familiarisation.

3.2.3.2 Trial protocol

Following the warm-up and sufficient rest, to avoid the effects of fatigue, each participant completed five maximal effort countermovement jumps, each with approximately two minutes rest between trials. Participants stood in an upright position with feet positioned approximately shoulder width apart. From this upright position, participants were instructed to squat to a self-selected depth before immediately jumping as high as possible. For the standardisation of jumps, participants kept their arms akimbo (hands on their hips) throughout the entire movement, as this was a measure of lower body power only, and were instructed to jump vertically with as little horizontal displacement as possible.

All CMJs were completed on a dual-platform set-up consisting of two, side-by-side, portable Kistler FPs (model number 92866A, Kistler Instruments Ltd., Farnborough, United Kingdom), as shown in Figure 3.3, so that the vertical component of the ground reaction force of each leg could be assessed independently. Data was collected for a duration of 12 s at a sample rate of 1000 Hz and an absolute vertical force range of ~ 20 kN (5365 N per corner transducer). The force data was interfaced to a data recording computer via a data acquisition system and converted into digital signals using a 16-bit analogue-to-digital converter.

The signal to jump was given following at least 5 s of quiet standing to ensure that the participant remained stationary for the measurement of bodyweight and for the initial conditions for jump analysis to be met i.e., velocity = 0 $m \cdot s^{-1}$. Participants were not tested on their ability to react to the signal, just instructed to jump soon after the signal was given.

***Figure 3.3**. Dual-platform set-up for the assessment of lower limb bilateral asymmetry using two force platforms.*

3.2.3.3 Determination of body weight, mass and jump initiation time for the assessment of CMJ performance

A copy of the VGRF-time record for each trial was exported from Kistler's Bioware® software to Microsoft Excel and, using a custom programme, the body weight was calculated by determining the mean vertical ground reaction force during the 1 s period between 4 and 5 s in the stationary stance phase immediately before the instruction to jump was given. The participant's BM was subsequently determined by dividing this mean BW by acceleration due to gravity (g = 9.80665 m·s^{-2}). The same custom programme determined a jump initiation time (t_i) using the criterion method for the determination of peak mechanical power (Owen *et al.*, 2014), the instant that BW ± 5 SD is exceeded after the signal to jump has been given, minus 30 ms. All data prior to t_i and just after t_{to} was then discarded.

3.2.3.4 Outcome variables

The VGRF-time history for each trial was exported to Microsoft Excel and then processed to generate kinetic and normalised temporal neuromuscular variables, as defined below. As the current study was only interested in the countermovement portion of the record, only data from t_i (determined using the method outlined above) up to and including t_{to} was considered, in order to prevent any landing force from being included in the analysis. BW for each individual trial was determined by applying the method outlined above and the net resultant vertical force-time record was then integrated with respect to time from t_i with the constant of integration set to zero. The process of integration is necessary for the determination of the physical quantities other than force, such as velocity or displacement. In order to determine a participant's vertical velocity, it is necessary to integrate the expression, $(F_z - m \cdot g)/\, m$, numerically (Kibele, 1998). As per the criterion method (Owen *et al.*, 2014), an integration frequency of 1000 Hz, equal to the sampling frequency, was selected.

3.2.3.4.1 Bilateral countermovement assessment- Kinetic variables

3.2.3.4.1.1 Peak force

Peak force (F_{max}) between the instants of jump initiation, t_i, and take-off, t_{to} was determined using the MAX function by defining the range of interest.

3.2.3.4.1.2 Phase specific analysis - Eccentric impulse

Impulse (J), or change in momentum, is calculated as the product of the force and period of time during which this force acts (Hay, 1993, pg. 78) and, as such, is defined with the expression,

$$J = F \times t \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 3.5.$$

The impulse of each sample interval, the area under the net force curve for one sample period, was calculated using the net force, before subsequently generating a cumulative impulse, enabling the total impulse generated to be determined for any time point within the CMJ. Equation 3.6. gives the calculation required for the determination of impulse at each time interval.

$$J = t_s \times 0.5 \times \left(F_{net_{n-1}} + F_{net_n}\right) \quad \cdots\cdots\cdots\cdots \quad 3.6.$$

where: F_{net_n} = net force at time point n (N)

t_s = width of sample time

As the body is stationary at the initiation of the jump, there is no net force acting on the body, and therefore, no impulse; consequently, the determination of cumulative impulse is calculated starting from a value, J = 0, and subsequently calculated at each interval using the expression:

$$J_{cum} = J_{cum_{n-1}} + J_n \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 3.7.$$

where: $J_{cum_{n-1}}$ = Cumulative impulse at the previous interval

J_n = Impulse of the current interval

Benjanuvatra *et al*. (2013), Menzel *et al*. (2013) and Jordan *et al*. (2015) are in agreement that phase-specific kinetic impulse analysis can be beneficial in revealing directional asymmetries and seem a more sensitive criteria for asymmetry identification. Further, because the propulsion and change of velocity of the whole-body CoG are mechanically determined by impulse, its inclusion in bilateral asymmetry assessment is justified.

The eccentric deceleration phase/ the first weighting phase was defined as the time interval from maximum negative (downward) velocity to zero velocity (Jordan *et al* , 2015). In this case, the term velocity refers to the velocity of the CoG. For the determination of instantaneous velocity of the whole-body CoG, the impulse momentum principle was applied, at the sample frequency, to the net vertical force and, as such, instantaneous velocity for each interval was calculated using:

$$v = \frac{J_{cum}}{m} \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 3.8.$$

where: v = instantaneous vertical velocity of CoG ($m \cdot s^{-1}$)

J_{cum} = cumulative impulse (N·s)

m = mass of participant (kg)

The magnitude and time of maximum negative velocity was determined using a custom spreadsheet (Appendix F).

The cumulative impulse at the instant of maximum negative velocity is representative of the first unweighting phase. In order for the CoG to begin upward vertical motion, following the initial negative impulse, the downward velocity must be reduced to zero, i.e., there must be an equal, but opposite, impulse. Therefore, to give the value of zero velocity, eccentric impulse, J_{ecc}, was taken as 0 – [the cumulative impulse at the instant of maximum negative velocity] identified by a custom spreadsheet (Appendix F).

3.2.3.4.1.3 Phase specific analysis- Propulsive impulse

Once the first unweighting and first weighting impulses are balanced, the following positive impulse is referred to as the propulsive impulse, and is responsible for the upward (positive) velocity of the CoG. The concentric propulsion phase was defined from the instant of zero CoG velocity to the instant of t_{to} and, therefore, propulsive concentric impulse, J_{con}, was taken to be the cumulative impulse at the instant of t_{to}.

The instant of t_{to} was defined as the time interval F_z (N) < 3.12 N and was determined by a pilot study (Appendix G).

3.2.3.4.1.4 Peak mechanical power output

Similar to velocity, mechanical power output was calculated for each interval throughout the CMJ. Instantaneous mechanical power output was calculated from the product of instantaneous VGRF and the corresponding instantaneous vertical velocity (Davies & Rennie, 1968; Winter, 2005) as shown in Equation 2.6.

$$P_i = F_i . v_i \quad \cdots\cdots\cdots\cdots\cdots\cdots \quad 2.6.$$

where: P_i = instantaneous power at time t = t_i (W)

F_i = instantaneous vertical ground reaction force (VGRF) at time t = t_i (N)

v_i = instantaneous vertical velocity of CoG at time t = t_i ($m \cdot s^{-1}$)

Peak mechanical power output (PPO) refers to the maximum power output generated by the combined effort of the lower limbs and was identified using a custom spreadsheet. PPO was defined as the maximum value of instantaneous power.

3.2.3.4.1.5 Take-off velocity

Take-off velocity, V_{to}, was defined as the value for the velocity of the whole-body CoG calculated using Equation 3.8. at the instant of t_{to}, determined using the method above from information gathered in a pilot study (Appendix G). Velocity was calculated using the equation:

$$v = \frac{J_{cum}}{m} \qquad \cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots \; 3.8.$$

where: v = instantaneous vertical velocity of CoG ($m \cdot s^{-1}$)

J_{cum} = cumulative impulse ($N \cdot s$)

m = mass of participant (kg)

Once the time point for the instant of t_{to} was determined, the value for instantaneous velocity at this point was identified using a custom spreadsheet (Appendices H and I).

3.2.3.4.2 Bilateral countermovement assessment- Normalised temporal variables

3.2.3.4.2.1 Eccentric to Concentric changeover as percentage of jump

The time, after t_i, at which the changeover of the impulse due to the eccentric or concentric contractions, similarly described as the changeover from braking force to propulsive force, occurs is the point at which the velocity of the whole-body CoG is equal to zero, and was defined as, t_{ec}. However, zero velocity may also occur just after, t_i, due to some participants initiating a jump by first moving slightly upwards, before making a countermovement (Owen *et al*., 2014). Consequently, t_{ec}, was defined as the instant, after the time of a minimum value of VGRF, t_{Fmin}, that the velocity of the CoG was, $v = 0$. In order to determine the point of zero velocity, the magnitude of the instantaneous velocity, after t_{Fmin}, was searched for the first minimum value. The minimum value then being taken as, $v = 0$, and the corresponding time point taken as, t_{ec}. This process was achieved using a custom spreadsheet (Appendices H and I).

To normalise, t_{ec}, it was expressed as a percentage of the total jump duration, Equation 3.9, and defined as, t_{ecN}. Jump duration, t_j, was defined as the difference in time between, $t = t_i$, and, $t = t_{to}$.

$$t_{ecN} = \frac{t_{ec}}{t_j} \times 100 \qquad \cdots\cdots\cdots \; 3.9.$$

3.2.3.4.2.2 Peak Force as percentage of jump

The instant that peak force occurred, t_{Fmax}, after, t_i, was normalised by expressing it as a percentage of, t_j, and defined as t_{FmaxN}.

$$t_{FmaxN} = \frac{t_{Fmax}}{t_j} \times 100 \qquad \cdots\cdots\cdots \; 3.10.$$

For both t_{ecN} and t_{FmaxN}, although the variable titles refer to the percentage of the 'jump', it is only referring to the percentage of the countermovement portion up to the point of t_{to}, not including the flight phase or landing.

3.2.3.4.3 Individual leg assessment

In order to assess the individual contribution of each lower limb to the jump, the VGRF from each FP was examined using the same custom spreadsheet as above (Appendices H and I). Jump initiation time was not calculated independently for each limb but was taken to be the start time obtained from the combined output from the two FPs. For the calculation of net force, the VGRF, derived from individual FPs, of each limb was adjusted to 50% of BW because the left and right limbs are coordinated cooperatively to accelerate the CoG upward. By allowing 50% loading to each limb, it ensured that variations in the bilateral GRF were preserved (Benjanuvatra *et al.*, 2013). Although variables were calculated using 50% of BW, it is reasonable to expect that this distribution of BM was not equal at all times, and, as such, it was deemed inappropriate to consider measurements of PPO for individual limbs due to its reliance on BM. Luk *et al.* (2014) and Heishman *et al.* (2019), two previous studies investigating lower limb asymmetries including the analysis of power variables, have not provided any information regarding how BW was adjusted for in their analysis. Menzel *et al.* (2013) determined instantaneous power using the following equation, but still provide little detail about the influence of BW contribution to individual limbs during a jump.

$$P_{ti} = F_{ti} \times v_{ti} = F_{ti} \times \frac{1}{m}\int_{t_0}^{t_i} F dt \qquad \cdots\cdots\cdots \; 3.11.$$

"where F_{ti} is the instantaneous vertical ground reaction force, and v_{ti} is the instantaneous vertical velocity of the centre of gravity. P_{max} is the maximal value of instantaneous power."

The assumption of exactly 50% contribution from each limb within a jump would likely result in erroneous results due to either overestimation or underestimation as a result of more or less loading on the limb, however, it was beyond the scope of this research to consider the proportion of loading on each limb. Therefore, equal distribution was assumed. Variables chosen for analysis of asymmetry were eccentric and concentric impulse (J_{ecc}, J_{con}) peak force (F_{max}) and the percentage of the jump that peak force occurred (t_{FmaxN}). The calculation of these variables were carried out in the same ways as outlined previously in (Sections 3.2.3.4.1 and 3.2.3.4.2).

3.2.3.4.4 Calculating asymmetry between lower limbs

The symmetry index (SI) is one of the most common methods of quantifying asymmetry between discrete measures. A measurement of the percentage difference between two limbs, the SI ascribes a single value to the level of asymmetry between two sides. A general formula for SI is $SI = \frac{(X_{side1} - X_{side2})}{reference\ value} \times 100\%$

To avoid ambiguity of terms like dominant vs non-dominant, which suggests that you would expect one limb to always be higher, the term high vs low has been chosen as a method to identify differences between the values of each limb, irrespective of whether they originate from the left or right. The study conducted by Menzel *et al*. (2013) also considered the assessment of asymmetry using both left vs right and a method removing the directionality, termed LSI (lateral symmetry index) and absolute LSI, respectively. The inclusion of this method was to account for instances where asymmetry between limbs is changing from favouring the left, to favouring the right leg; if the limb producing the greater value is changing, left vs right analysis may not be as appropriate of a measure as data will be nullified by counteracting values. Despite this, left vs right analysis was included in the current study in order to identify the differences the two methods of assessment have on the outcome. The calculation of SI for each of the two methods was as follows.

3.2.3.4.4.1 Left vs right symmetry index

Using calculation methods similar to Clark (2001), Benjanuvatra *et al*. (2013), Menzel *et al.*(2013) and Jordan *et al*. (2015) the current study calculated left vs right (LvsR) percentage asymmetry as:

$$LvsR\ SI = \frac{(X_{Left} - X_{Right})}{Max\ (X_{Left}, X_{Right})} \times 100 \quad \cdots\cdots\cdots\cdots\cdots\cdots \ 3.12.$$

where: X_{Left} = value produced by the left limb

X_{Right} = value produced by the right limb

A positive value represents a greater magnitude recorded by the left leg.

3.2.3.4.4.2 High vs low symmetry index

Using calculation methods similar to Newton *et al*. (2006) and Impellizzeri *et al.* (2007), the current study calculated high vs low (HvsL) percentage asymmetry as:

$$HvsL\ SI = \frac{\left(Max\ (X_{Left}, X_{Right})\right) - \left(Min\ (X_{Left}, X_{Right})\right)}{Max\ (X_{Left}, X_{Right})} \times 100 \quad \cdots\ 3.13.$$

where: X_{Left} = value produced by the left limb

X_{Right} = value produced by the right limb

3.3 Statistical analysis

All quantitative data from postural stability and CMJ assessments were collated, processed and analysed using Microsoft Excel (Microsoft Corp. Version 16. Redmond, WA) and SPSS 28 (IBM Corp, 2019) for both Windows and Apple macOS. Male and female study populations were assessed independently as well as combining populations to provide a larger, more generalised, study population, in order to form an additional level of analysis. Descriptive statistics for all variables within all three populations were presented using the mean ± standard deviation (SD). Kolmogorov-Smirnov tests were conducted to test for normality; results suggested a small number of cases where distribution showed a significant departure from normality. However, all data was treated as being normally distributed and, therefore, only parametric testing was conducted (le Cessie, Goeman & Dekkers, 2020; Wadgave, 2019). Paired samples student t-tests were conducted to identify statistically significant differences between day 1 and day 2 outcomes; the sample size was deemed large enough for a moderate effect size of 0.5 and statistical power of 95% with an alpha significance level of 0.05.

The reliability of measurements was assessed using intraclass correlation coefficients (ICCs) (Fisher, 1954) and Bland and Altman agreement (B&A) plots (Altman & Bland, 1983; Bland & Altman, 1986, 1999). Bland and Altman introduced the terms bias, defined as the mean of differences between two methods, and limits of agreement (LoA), defined as the range expected to include 95% of the future differences between two methods and is, in essence, a measure of the amount of between-methods agreement (Bartlett & Frost, 2008; de Vet, Terwee, Knol & Bouter, 2006; Haghayegh, Kang, Khoshnevis, Smolensky & Diller, 2020). The results of ICC analysis are influenced by sample heterogeneity (Atkinson & Nevill, 1998) and as such do not provide the most effective assessment; the addition of the B&A 95% LoA (Bland & Altman, 1986) to assess the agreement between repeated measures improves the validity of the conclusions made.

Pearson's product correlation coefficients, r, is an additional method of reliability assessment; however, the correlation coefficient is a reflection of how closely a set of paired observations follow a straight line, regardless of the slope/ gradient of the line. As a result, clinicians may mistake excellent correlation for complete agreement, which is not the case (Vaz, Falkmer, Passmore, Parsons & Andreou, 2013). Unlike Pearson's (r), the ICC accounts for both consistency of performances from test to retest (within-subject change), as well as change in average performance of participants as a group over time, i.e., systematic change in mean (Atkinson & Nevill, 1998; Bland & Altman, 2003; Lexell & Downham, 2005).

Shrout and Fleiss (1979) defined 6 forms of ICC each specific to the raters used and those of interest. They suggest that a two-way mixed-effects model (3,1) is appropriate for testing intra-rater reliability with multiple scores from the same rater, as it is not reasonable to generalise one rater's scores to a larger population of raters. Similarly, a two-way mixed-effects model should also be used in test-retest reliability studies because repeated measurements cannot be regarded as randomised samples. A two-way random-effects model (2,1) is an appropriate choice when we plan to generalise our reliability results to any raters who possess the same characteristics as the selected raters in the reliability study (Koo & Li, 2016). These two ICC models were applied to the current study, with absolute agreement, and point estimates of the ICCs (absolute ICC and 95% confidence intervals (CI)) were interpreted as follows: poor (< 0.5), moderate (0.5 – 0.75), good (0.75 – 0.90) and values greater than 0.9 indicated excellent reliability (Koo & Li, 2016). An absolute ICC value of 0.95 would be indicative of excellent reliability and suggests that 95% of variation in the measurements is due to real differences between subjects and 5% is due to measurement errors (Haghayegh *et al.*, 2020).

With regard to B&A analysis, as multiple plots were generated in the current study, bar charts were used to represent the key information, magnitude of the bias and LoA, such that B&A data could be compressed into a simple chart allowing multiple epochs and variables to be reported simultaneously.

In addition to the SD, for select parameters the standard error of the mean (SEM) was also calculated. The SD is a descriptive parameter that quantifies the variability in a data set and its dispersion in a normal distribution, whereas SEM is an inferential parameter that quantifies the uncertainty in the estimate of the mean and includes statistical inference based on the sampling distribution (Lee, In & Lee, 2015).

Chapter 4

Results

4.1 Postural stability

4.1.1 Assessment of normality

The results of Kolmogorov-Smirnov normality tests, conducted on all variables for each epoch under both EO and EC conditions, established that all data series for both path length (L_p) and mean velocity (V_m) were normally distributed on day 1 and day 2. Sway area (A_s), on the other hand, showed a few cases where the data did not follow a normal distribution. In the ALL study population, A_s showed a significant departure from normality, for the EO condition, within the 3rd 30 s, 1st 60 s and 2nd 60 s epochs on day 2 ($D(19) = 0.211$, $p = 0.026$; $D(19) = 0.235$, $p = 0.007$ and $D(19) = 0.203$, $p = 0.039$, respectively). For EC, A_s was normally distributed for all epochs on both days.

In the FEM population, the 3rd 30 s epoch for EO was not normally distributed on both day 1 ($D(9) = 0.276$, $p = 0.046$) and day 2 ($D(9) = 0.284$, $p = 0.035$) while all remaining epochs in both conditions were considered to demonstrate normal distribution.

In the MALE population, on day 1, only the 2nd 60 s epoch in the EO condition significantly deviated from a normal distribution ($D(10) = 0.274$, $p = 0.032$). However, on day 2, within the EO condition, only the 2nd 30 s epoch was normally distributed ($D(10) = 0.194$, $p = 0.200$) while, for the EC condition, the 1st 30 s ($D(10) = 0.233$, $p = 0.133$), 3rd 30 s ($D(10) = 0.233$, $p = 0.130$) and the 2nd 60s ($D(10) = 0.305$, $p = 0.009$) were the series of data considered to deviate from a normal distribution.

Wadgave (2019) concluded that 'existing evidence from simulation studies suggests that parametric methods are preferred over non-parametric in most situations while analysing non-normally distributed continuous data' and that non-parametric tests should only be considered when the distribution is highly skewed. Additionally, le Cessie *et al*. (2020) report that parametric testing can yield the same results for non-normally distributed data and work well even from moderately skewed distributions of the outcome variable. Therefore, all data identified to be not normally distributed was treated as normally distributed within the current study and parametric tests were used.

4.1.2 Reliability of postural stability measures for individual epochs

The results of a paired samples t-test between day 1 and day 2 for each population within both conditions and all epochs are shown in Tables 4.1, 4.2. and 4.3.

For the variable of L_p, statistical analysis shows that there was only one significant difference ($p \leq 0.05$) between day 1 and day 2 for the 3rd 30 s epoch in the ALL population under the EC condition ($p = 0.05$). All other epochs within the three study populations were not significantly different. A_s demonstrated a significant difference between days for the following conditions and epochs; ALL EO 1st 30 s ($p < 0.001$), ALL EC 3rd 30 s ($p = 0.008$), FEM EO 1st 30 s ($p = 0.032$), FEM EC 3rd 30 s ($p = 0.036$), FEM EC 90 s ($p = 0.034$), and finally, MALE EO 1st 30 s ($p = 0.004$). For V_m, there was a significant difference between day 1 and day 2 for the 3rd 30 s epoch within the EC condition ($p = 0.049$).

Tables 4.4, 4.5 and 4.6 provide the results of intraclass correlation coefficient (ICC) analysis for the assessment of the difference between day 1 and day 2 variable outcomes. As ICC analysis does not allow us to quantify the error/ difference, Figure 4.1 illustrates the results of Bland and Altman (B&A) analysis for the comparison of day 1 and day 2 data. L_p and V_m demonstrated ICC values representative of excellent reliability (> 0.9) for all but one epoch in the FEM (2nd 30s) and MALE (1st 30s) populations, although still indicative of good reliability (0.796 & 0.886, respectively). 95% confidence intervals (CIs), however, showed a wider range for a number of epochs in both EO and EC conditions for all populations. The majority of A_s ICC values were indicative of either good or excellent reliability, however, when considering the 95% CI, many epochs in both conditions gave values at the lower end of 'poor'.

B&A analysis (Figure 4.1) shows a much lower magnitude of limits of agreement (LOA) for L_p and V_m. Analysis of A_s generated LOA with magnitudes up to 80% of the mean and biases close to 40% of the mean in individual populations. From the B&A analysis it is not clear which epoch generated results with the best, or worst agreement, due to similarities in the magnitude of the bias and LOA across the epochs; L_p and V_m appear to favour the 3rd 30 s, however, this could be considered one of the epochs with the least agreement for A_s. The 90 s epoch demonstrated some of the smallest variation within the FEM population under the EC condition, and EO condition for MALE. Although not the case for all populations and conditions, the 1st

30 s appeared to be one of the lowest performing epochs with regard to day 1 to day 2 agreement. A table presenting the results of B&A analysis, both raw, and as a percentage of the mean, is given in Appendix J.

4.1.3 Effect of visual conditions on postural stability measures

Due to the level of agreement between day 1 and day 2 data sets for both EO and EC conditions, the four trials of day 1 and four of day 2 were combined to produce a mean of all eight trials within each condition. These combined mean values were used to assess whether the EC condition produced significantly different performance outcomes. As shown in Tables 4.1 and 4.3, for both L_p and V_m, there was a significant ($p \leq 0.05$) difference between EO and EC conditions within all epochs; both L_p and V_m were significantly higher in the EC condition. For A_s (Table 4.2), within the ALL population, there was only a significant difference between EO and EC within the 2nd 30 s, 3rd 30 s and 2nd 60 s epochs. When analysing only the FEM participants, the only epoch in which there was a significant difference was the 3rd 30 s. All other epochs demonstrated a difference between the two conditions; however, these were not considered significant. In the MALE study population, similar to the ALL population, only the 2nd 30 s, 3rd 30s and 2nd 60 s epochs showed statistical significance. In all seven cases of significance, A_s was greater in the EC condition.

For L_p, with the exception of the 3rd 30 s in the FEM population, all ICC values and their 95% CIs were higher in the EC condition compared to EO (Table 4.4). In the B&A analysis, the LOA were, in the majority of cases, smaller in the EC condition, although this difference was negligible for the variables L_p and V_m. The bias, however, tended to be smaller in the EO condition for the ALL and FEM population. The MALE population showed biases of smaller magnitude in the EC condition for L_p and V_m and EO for A_s. In the FEM population, with the exception of the 1st 30 s, the LOA were smaller in the EC condition, while LOA were only smaller in the EC condition for the 1st 30 s and 3rd 30 s.

4.1.4 Population differences

A paired samples t-test comparing FEM to MALE data series for all epochs under both EO and EC conditions identified a significant difference between sexes for a number of cases, given below.

In the EO condition, for the variables L_p and V_m, there was a significant difference ($p < 0.05$) between populations in the 1st 30 s and 1st 60 s epochs on day 1 (L_p: $p = 0.021$ & $p = 0.046$; V_m: $p = 0.02$ & 0.045). On day 2 and days combined there was no significant difference ($p > 0.05$) between FEM and MALE. For A_s, there was no significant difference between populations for either day, and thus, days combined.

In the EC condition, for the variables L_p and V_m, there was a significant difference between populations on day 2 in the 2nd 30 s ($p = 0.043$ & $p = 0.042$, respectively), 3rd 30 s ($p = 0.039$), 2nd 60 s ($p = 0.04$) and 90 s ($p = 0.049$) epochs (Table 4.1 and 4.3). Day 1, however, showed no significant differences. The 1st 60 s epoch for V_m on day 1 was close to significant ($p = 0.054$). When days were combined, a significant difference was identified in the 3rd 30 s for L_p ($p = 0.044$), while the 2nd 60 s epoch was close to significant ($p = 0.051$). For V_m, the 3rd 30 s and 2nd 60 s were significantly different ($p = 0.043$ and $p = 0.05$, respectively). There was no significant difference between populations for the variable A_s.

With the exception of the 1st 30 s and 3rd 30 s for L_p and V_m in the EO condition, the MALE population produced slightly higher ICC values for all epochs in both the EO and EC conditions (Table 4.4 and 4.6). In most cases, 95% CIs remained in the good to excellent range for MALE, while the FEM population, although still generating ICC values indicative of good and excellent reliability, had wider ranging 95% with more 'poor' inferences. The biggest differences between sexes were in A_s; although MALE still generated ICC 95% CIs that ranged from poor in the EC condition, the FEM population was a lot more varied with all 95% CIs ranging from poor to excellent in the EO condition and the 1st 30 s and 3rd 30 s in the EC condition (Table 4.5).

The B&A plots (Figure 4.1) show that for some variables within each condition, the MALE population displayed LOA and biases smaller in magnitude, while the FEM population showed favourable variability in others. To generalise the data, the MALE population tended to generate LOA and biases of smaller magnitude in both EO and EC conditions for A_s. L_p and V_m didn't tend to favour a particular population and often differences appeared to be negligible.

4.1.5 Results of cumulative averaging analysis

Cumulative moving average analysis was conducted in order to report the impact of increasing the number of trials included in analysis on the mean and, subsequently, the precision of the mean value. Figures 4.2, 4.3 and 4.4 illustrate the standard error of the mean (SEM), reported as a percentage of the mean, as the number of trials averaged increases for L_p, A_s and V_m for all three populations. For all epochs, the greater the number of trials included in analysis, the smaller the SEM. Tables presenting the mean, SEM and SEM as a percentage of the mean for each variable, condition and epoch within each population are provided in Appendix K.

Table 4.1. *Descriptive statistics (Mean ± SD) for the variable Path Length under eyes open (EO) and eyes closed (EC) conditions for all epochs and populations on day 1, day 2 and days combined.*

		Path Length (m)												
		Eyes Open						Eyes Closed						
Pop.	Epoch	Day 1		Day 2		Combined		Day 1		Day 2		Combined		
		Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	
All Participants	1st 30 s	0.245	0.060	0.244	0.057	0.245	0.056	0.334	0.100	0.328	0.109	0.331	0.103	†
	2nd 30 s	0.220	0.052	0.218	0.056	0.219	0.052	0.292	0.086	0.290	0.092	0.291	0.087	†
	3rd 30 s	0.222	0.051	0.219	0.050	0.221	0.049	0.299	0.093	0.285	0.092 ⁎	0.292	0.092	†
	1st 60 s	0.465	0.109	0.462	0.111	0.464	0.107	0.626	0.182	0.619	0.198	0.622	0.188	†
	2nd 60 s	0.442	0.101	0.438	0.104	0.440	0.100	0.591	0.177	0.575	0.183	0.583	0.178	†
	90 s	0.687	0.158	0.681	0.159	0.684	0.154	0.921	0.273	0.903	0.288	0.912	0.278	†
Female	1st 30 s	0.209	0.040	0.222	0.048	0.216	0.043	0.286	0.059	0.273	0.054	0.280	0.056	†
	2nd 30 s	0.197	0.033	0.193	0.032	0.195	0.029	0.255	0.061	0.246	0.046	0.251	0.052	†
	3rd 30 s	0.200	0.035	0.201	0.035	0.200	0.034	0.259	0.063	0.237	0.047	0.248	0.054	∇
	1st 60 s	0.406	0.070	0.415	0.078	0.411	0.071	0.541	0.118	0.519	0.095	0.530	0.105	†
	2nd 60 s	0.397	0.066	0.394	0.067	0.395	0.063	0.514	0.123	0.483	0.091	0.499	0.105	∇
	90 s	0.606	0.101	0.616	0.112	0.611	0.103	0.797	0.174	0.757	0.140	0.777	0.155	†
Male	1st 30 s	0.277	0.057	0.263	0.058	0.270	0.055	0.377	0.109	0.378	0.122	0.377	0.114	∇
	2nd 30 s	0.241	0.056	0.241	0.063	0.241	0.058	0.325	0.091	0.330	0.104	0.328	0.096	†
	3rd 30 s	0.242	0.055	0.236	0.055	0.239	0.053	0.334	0.101	0.327	0.102	0.331	0.101	†
	1st 60 s	0.518	0.111	0.504	0.118	0.511	0.111	0.702	0.195	0.708	0.222	0.705	0.206	†
	2nd 60 s	0.483	0.109	0.477	0.116	0.480	0.109	0.660	0.190	0.657	0.205	0.659	0.196	†
	90 s	0.760	0.164	0.740	0.172	0.750	0.163	1.032	0.297	1.035	0.321	1.034	0.306	†

⁎ *Indicates a statistically significant difference between day 1 and day 2 ($p \leq 0.05$)*

† & ∇ *Indicate a statistically significant difference between EO and EC for days combined values († = $p \leq 0.001$ and ∇ = $p \leq 0.05$)*

***Table 4.2** Descriptive statistics (Mean ± SD) for the variable Sway Area under eyes open (EO) and eyes closed (EC) conditions for all epochs and populations on day 1, day 2 and days combined.*

		Sway Area (mm^2)												
		Eyes Open						Eyes Closed						
		Day 1		Day 2		Combined		Day 1		Day 2		Combined		
Pop.	Epoch	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	
All Participants	1st 30 s	96.89	53.51	72.23	42.65 ⁎	71.36	40.49	95.13	43.96	81.86	29.72	88.49	34.46	
	2nd 30 s	47.37	21.38	51.74	32.28	49.56	25.71	61.24	31.73	63.78	37.25	62.51	32.56	∇
	3rd 30 s	46.63	25.78	46.31	29.66	46.47	25.85	71.70	46.15	52.68	30.11⁎	62.19	36.52	†
	1st 60 s	89.28	49.65	96.35	64.88	92.81	55.16	113.09	54.64	109.77	45.83	111.43	47.76	
	2nd 60 s	65.17	34.60	71.77	52.31	68.47	42.22	83.54	46.36	77.78	46.08	80.66	44.78	∇
	90 s	96.89	53.51	108.66	69.42	102.78	60.10	121.59	58.32	117.32	47.96	119.45	51.86	
Female	1st 30 s	86.73	39.92	68.90	32.57 ⁎	65.32	27.63	97.72	49.37	78.83	33.30	88.28	39.04	
	2nd 30 s	42.52	18.11	47.82	29.33	45.17	22.39	52.84	27.24	62.56	36.85	57.70	30.96	
	3rd 30 s	43.96	22.79	44.87	29.30	44.41	23.44	71.71	45.22	44.38	26.28⁎	58.05	33.64	∇
	1st 60 s	80.78	37.66	91.89	58.64	86.33	45.00	109.35	52.16	100.74	45.75	105.05	47.65	
	2nd 60 s	59.63	25.67	64.41	40.54	62.02	30.53	78.72	44.58	69.29	36.12	74.01	38.79	
	90 s	86.73	39.92	104.71	58.69	95.72	47.40	117.30	54.88	106.36	47.16 ⁎	111.83	50.80	
Male	1st 30 s	106.04	61.88	75.22	49.82 ⁎	76.80	48.63	92.80	38.30	84.58	25.78	88.69	29.73	
	2nd 30 s	51.74	23.09	55.27	34.33	53.51	27.78	68.80	33.54	64.87	37.57	66.84	33.35	∇
	3rd 30 s	49.03	27.99	47.61	29.93	48.32	27.70	71.69	46.97	60.15	31.36	65.92	38.55	∇
	1st 60 s	96.92	57.31	100.37	69.78	98.65	62.34	116.45	56.57	117.90	44.37	117.18	47.12	
	2nd 60 s	70.16	40.37	78.39	60.23	74.28	49.76	87.89	47.49	85.43	52.31	86.66	48.79	∇
	90 s	106.04	61.88	112.22	77.65	109.13	68.97	125.45	61.00	127.19	46.51	126.32	51.83	

⁎ *Indicates a statistically significant difference between day 1 and day 2 ($p \leq 0.05$)*

† & ∇ *Indicate a statistically significant difference between EO and EC for days combined values († = $p \leq 0.001$ and ∇ = $p \leq 0.05$)*

Table 4.3 *Descriptive statistics (Mean ± SD) for the variable Mean Velocity under eyes open (EO) and eyes closed (EC) conditions for all epochs and populations on day 1, day 2 and days combined.*

		Mean Velocity ($cm \cdot s^{-1}$)												
		Eyes Open						Eyes Closed						
Pop.	Epoch	Day 1		Day 2		Combined		Day 1		Day 2		Combined		
		Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	
All Participants	1st 30 s	0.812	0.200	0.807	0.191	0.810	0.187	1.108	0.332	1.088	0.363	1.098	0.344	†
	2nd 30 s	0.728	0.172	0.722	0.186	0.725	0.173	0.968	0.285	0.962	0.307	0.965	0.290	†
	3rd 30 s	0.735	0.169	0.725	0.166	0.730	0.163	0.989	0.311	0.943	0.308 ⁎	0.966	0.306	†
	1st 60 s	0.770	0.182	0.765	0.184	0.767	0.178	1.025	0.329	1.038	0.303	1.032	0.312	†
	2nd 60 s	0.732	0.168	0.724	0.173	0.728	0.166	0.979	0.295	0.953	0.306	0.966	0.297	†
	90 s	0.758	0.175	0.751	0.177	0.755	0.171	1.018	0.303	0.998	0.320	1.008	0.308	†
Female	1st 30 s	0.693	0.133	0.734	0.160	0.713	0.142	0.948	0.198	0.905	0.181	0.926	0.187	†
	2nd 30 s	0.651	0.109	0.638	0.107	0.644	0.098	0.846	0.204	0.814	0.153	0.830	0.173	†
	3rd 30 s	0.660	0.115	0.662	0.116	0.661	0.113	0.856	0.212	0.786	0.157	0.821	0.181	∇
	1st 60 s	0.672	0.117	0.686	0.130	0.679	0.117	0.860	0.159	0.897	0.197	0.878	0.175	†
	2nd 60 s	0.656	0.110	0.650	0.110	0.653	0.105	0.851	0.205	0.800	0.152	0.826	0.176	∇
	90 s	0.667	0.112	0.678	0.124	0.673	0.114	0.881	0.193	0.835	0.156	0.858	0.172	†
Male	1st 30 s	0.919	0.189	0.873	0.193	0.896	0.181	1.251	0.361	1.253	0.405	1.252	0.378	†
	2nd 30 s	0.798	0.187	0.798	0.209	0.798	0.192	1.079	0.302	1.095	0.347	1.087	0.319	†
	3rd 30 s	0.802	0.182	0.781	0.183	0.791	0.176	1.109	0.337	1.084	0.341	1.097	0.336	†
	1st 60 s	0.859	0.184	0.835	0.197	0.847	0.185	1.174	0.369	1.165	0.325	1.170	0.343	†
	2nd 60 s	0.800	0.181	0.790	0.192	0.795	0.181	1.094	0.316	1.090	0.342	1.092	0.326	†
	90 s	0.840	0.181	0.817	0.191	0.828	0.180	1.142	0.329	1.144	0.356	1.143	0.339	†

⁎ *Indicates a statistically significant difference between day 1 and day 2 ($p \leq 0.05$)*

† & ∇ *Indicate a statistically significant difference between EO and EC for days combined values († = $p \leq 0.001$ and ∇ = $p \leq 0.05$)*

Table 4.4 *Intraclass Correlation Coefficient (ICC) (3,1) absolute agreement, average measures: Day 1 vs Day 2. Variable: Path Length*

		Eyes Open			Eyes Closed		
			95% CI			95% CI	
Pop.	Epoch	ICC	Lower bound	Upper bound	ICC	Lower bound	Upper bound
All Participants	1st 30 s	0.913	0.772	0.966	0.978	0.944	0.992
	2nd 30 s	0.928	0.812	0.972	0.961	0.898	0.985
	3rd 30 s	0.944	0.854	0.978	0.973	0.921	0.990
	1st 60 s	0.941	0.845	0.977	0.975	0.936	0.990
	2nd 60 s	0.943	0.852	0.978	0.976	0.939	0.991
	90 s	0.948	0.864	0.980	0.979	0.947	0.992
Female	1st 30 s	0.906	0.614	0.979	0.968	0.816	0.993
	2nd 30 s	0.796	0.400	0.955	0.918	0.661	0.981
	3rd 30 s	0.957	0.806	0.990	0.912	0.537	0.981
	1st 60 s	0.902	0.570	0.978	0.954	0.804	0.990
	2nd 60 s	0.908	0.579	0.979	0.934	0.697	0.985
	90 s	0.933	0.706	0.985	0.950	0.755	0.989
Male	1st 30 s	0.886	0.572	0.971	0.974	0.896	0.994
	2nd 30 s	0.942	0.761	0.986	0.963	0.853	0.991
	3rd 30 s	0.929	0.720	0.982	0.982	0.932	0.996
	1st 60 s	0.937	0.757	0.984	0.974	0.893	0.993
	2nd 60 s	0.941	0.763	0.985	0.982	0.927	0.995
	90 s	0.940	0.767	0.985	0.981	0.922	0.995

Table 4.5 *Intraclass Correlation Coefficient (ICC) (3,1) absolute agreement, average measures: Day 1 vs Day 2. Variable: Sway Area*

		Eyes Open			Eyes Closed		
			95% CI			95% CI	
Pop.	Epoch	ICC	Lower bound	Upper bound	ICC	Lower bound	Upper bound
All Participants	1st 30 s	0.885	0.275	0.967	0.794	0.472	0.920
	2nd 30 s	0.866	0.659	0.948	0.875	0.676	0.952
	3rd 30 s	0.851	0.608	0.943	0.815	0.410	0.934
	1st 60 s	0.904	0.755	0.963	0.889	0.712	0.957
	2nd 60 s	0.896	0.735	0.960	0.934	0.833	0.974
	90 s	0.931	0.821	0.974	0.941	0.849	0.977
Female	1st 30 s	0.880	0.307	0.975	0.809	0.241	0.956
	2nd 30 s	0.821	0.233	0.959	0.895	0.569	0.976
	3rd 30 s	0.768	-0.138	0.949	0.705	-0.130	0.931
	1st 60 s	0.807	0.173	0.956	0.939	0.752	0.986
	2nd 60 s	0.781	-0.018	0.951	0.903	0.607	0.978
	90 s	0.863	0.441	0.968	0.977	0.824	0.995
Male	1st 30 s	0.892	0.074	0.978	0.797	0.221	0.949
	2nd 30 s	0.897	0.592	0.974	0.869	0.467	0.968
	3rd 30 s	0.914	0.647	0.979	0.915	0.663	0.979
	1st 60 s	0.955	0.820	0.989	0.850	0.365	0.963
	2nd 60 s	0.938	0.767	0.984	0.956	0.821	0.989
	90 s	0.966	0.867	0.991	0.913	0.643	0.979

Table 4.6 *Intraclass Correlation Coefficient (ICC) (3,1) absolute agreement, average measures: Day 1 vs Day 2. Variable: Mean Velocity*

		Eyes Open			Eyes Closed		
			95% CI			95% CI	
Pop.	Epoch	ICC	Lower bound	Upper bound	ICC	Lower bound	Upper bound
All Participants	1st 30 s	0.913	0.774	0.967	0.978	0.944	0.992
	2nd 30 s	0.929	0.814	0.973	0.961	0.898	0.985
	3rd 30 s	0.943	0.853	0.978	0.972	0.920	0.990
	1st 60 s	0.942	0.849	0.978	0.975	0.936	0.990
	2nd 60 s	0.944	0.854	0.978	0.976	0.939	0.991
	90 s	0.948	0.866	0.980	0.979	0.947	0.992
Female	1st 30 s	0.907	0.616	0.979	0.969	0.814	0.993
	2nd 30 s	0.802	0.077	0.956	0.918	0.660	0.981
	3rd 30 s	0.957	0.806	0.99	0.913	0.538	0.981
	1st 60 s	0.904	0.580	0.978	0.955	0.805	0.990
	2nd 60 s	0.911	0.595	0.980	0.934	0.702	0.985
	90 s	0.935	0.716	0.985	0.951	0.758	0.989
Male	1st 30 s	0.886	0.575	0.971	0.974	0.896	0.994
	2nd 30 s	0.942	0.761	0.986	0.963	0.853	0.991
	3rd 30 s	0.928	0.716	0.982	0.982	0.931	0.995
	1st 60 s	0.938	0.760	0.984	0.974	0.893	0.993
	2nd 60 s	0.941	0.764	0.985	0.982	0.926	0.995
	90 s	0.940	0.768	0.985	0.981	0.922	0.995

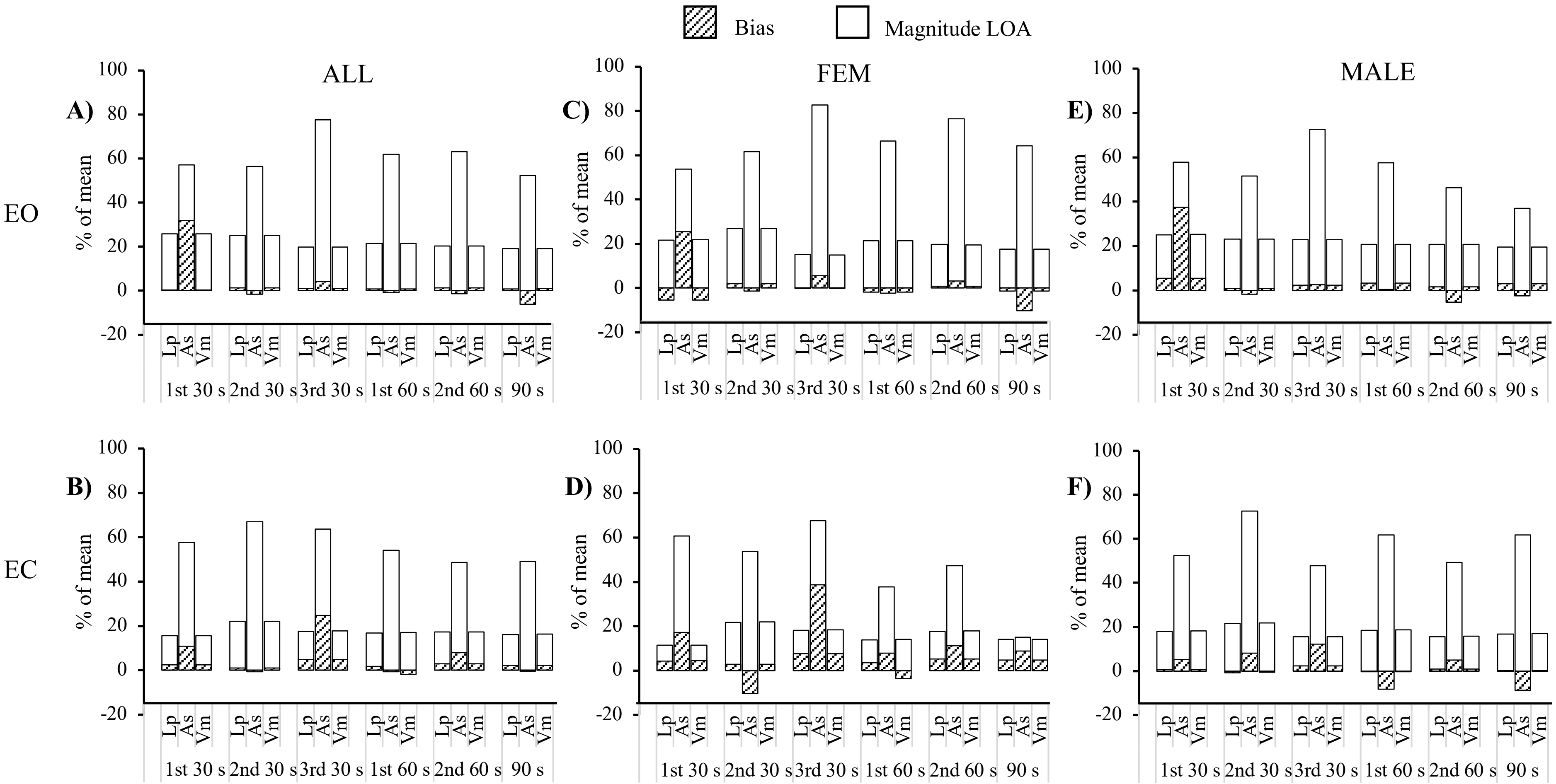

***Figure 4.1**. Summary of Bland and Altman analysis for postural stability measures.* A = ALL EO, B = ALL EC, C = FEM EO, D = FEM EC, E = MALE EO, F = MALE EC. Note: L_p, path length; A_s, sway area; V_m, mean velocity; EO, eyes open; EC, eyes closed

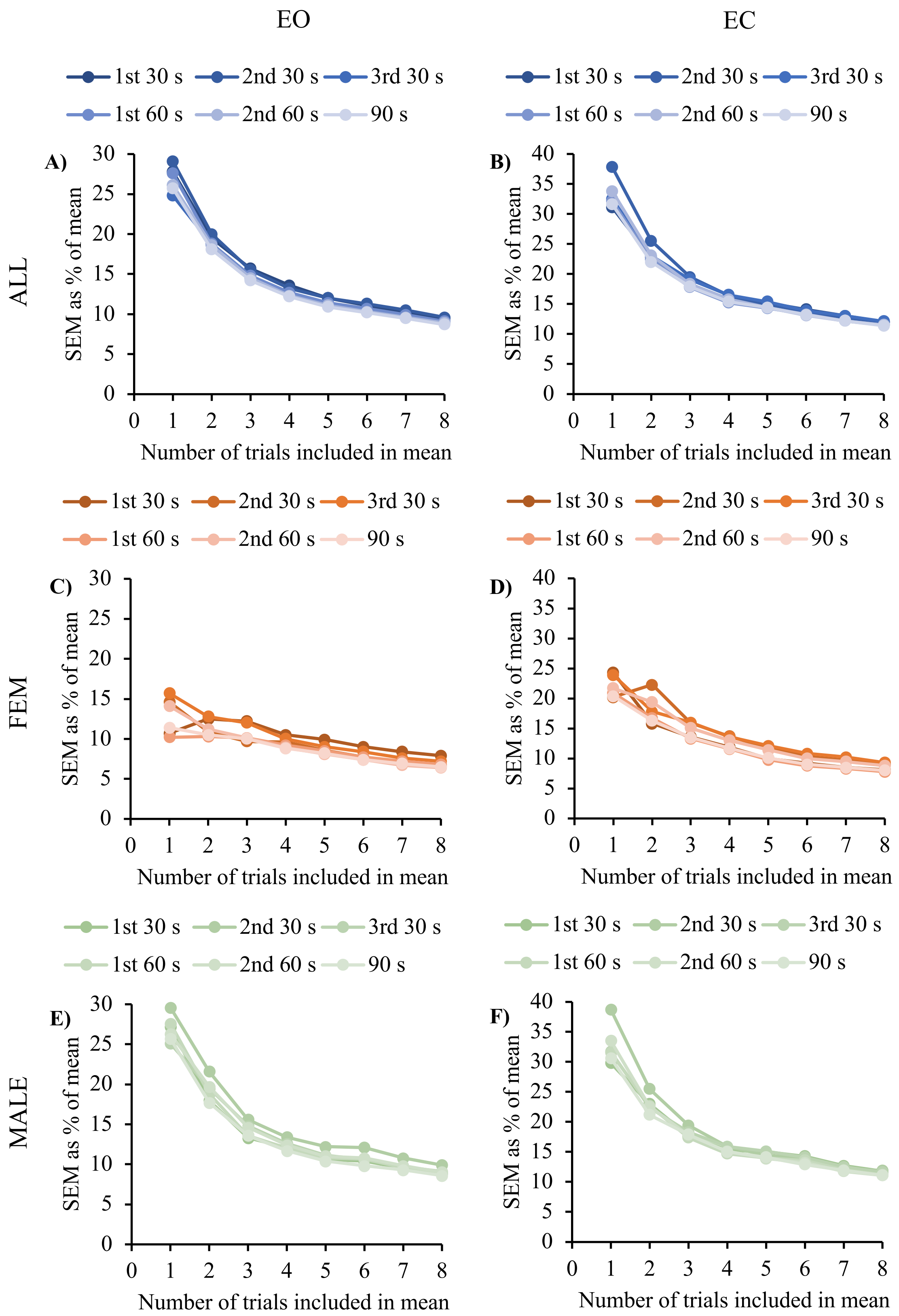

Figure 4.2. *Cumulative averaging analysis for the variable Path Length with the standard error of the mean (SEM) expressed as a percentage of the mean.*

A = ALL EO, B = ALL EC, C = FEM EO, D = FEM EC, E = MALE EO, F = MALE EC

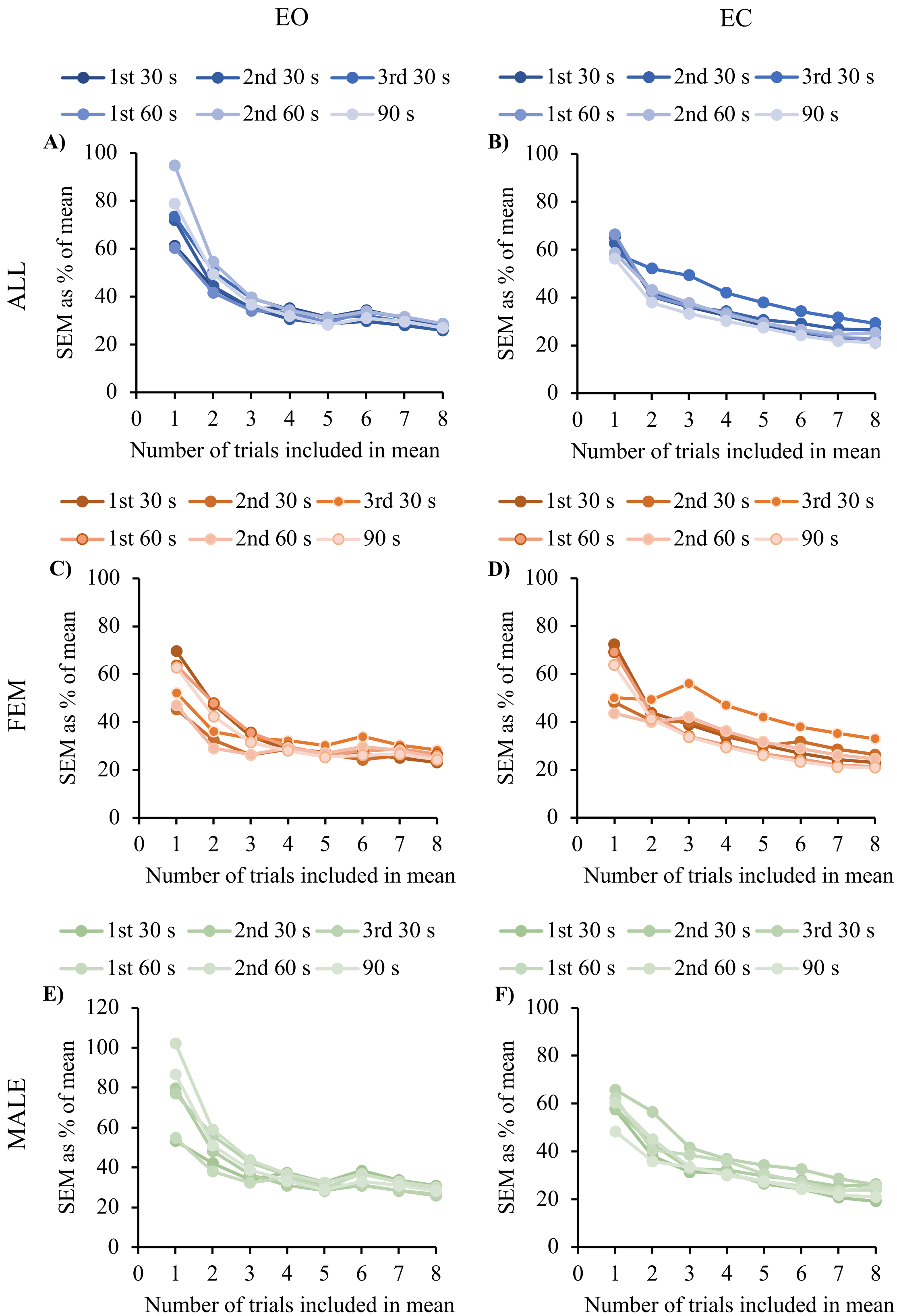

Figure 4.3. *Cumulative averaging analysis for the variable Sway Area with the standard error of the mean (SEM) expressed as a percentage of the mean.*

A = ALL EO, B = ALL EC, C = FEM EO, D = FEM EC, E = MALE EO, F = MALE EC

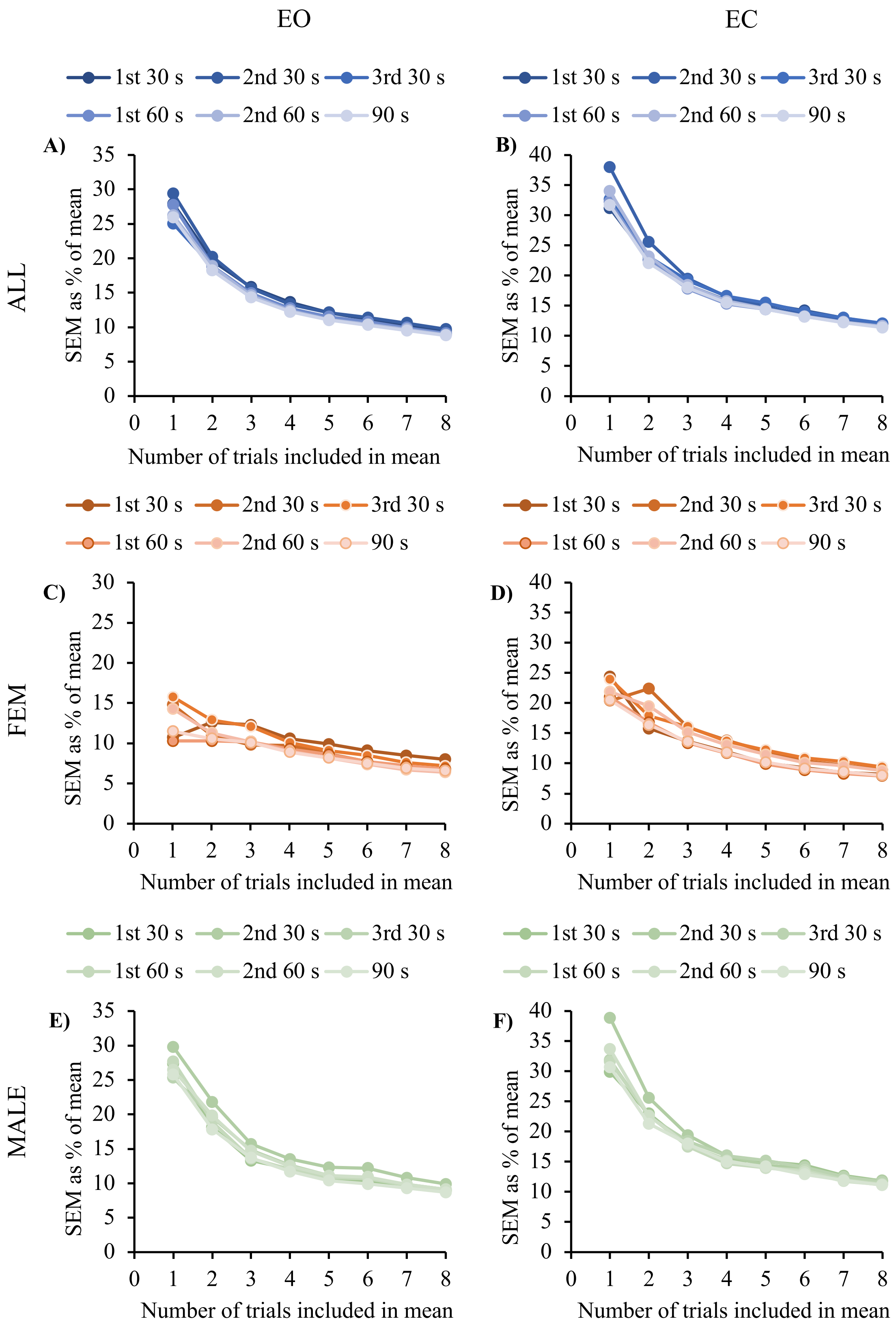

Figure 4.4. *Cumulative averaging analysis for the variable Mean Velocity with the standard error of the mean (SEM) expressed as a percentage of the mean.*

A = ALL EO, B = ALL EC, C = FEM EO, D = FEM EC, E = MALE EO, F = MALE EC

4.2 Lower limb bilateral asymmetry

4.2.1 Results of bilateral countermovement jump analysis

4.2.1.1 Assessment of normality

The results of Kolmogorov-Smirnova normality tests, conducted on all variables for each population on both day 1 and day 2, established that all data series were normally distributed, with the exception of a few select data series. Within the ALL-participants population, on day 1, only J_{ecc} showed a significant departure from normality (D(19) = 0.204, p = 0.036). All data sets within the FEM population were normally distributed. The MALE population J_{ecc} on day 1 and t_{FmaxN} on day 2 data series significantly departed from normality (D(10) = 0.297, p = 0.013, and D(10) = 0.271, p = 0.037, respectively).

As outlined in 4.1, despite the identification of non-normally distributed data, parametric testing was conducted for all data series. This is due to reports from Wadgave (2019) and le Cessie *et al*. (2020) concluding that parametric testing was appropriate for use on both normal and non-normally distributed data.

4.2.1.2 Assessment of neuromuscular performance test-retest reliability

Descriptive statistics (mean $\pm$ SD) for the average of all five trials on day 1 and five trials on day 2, in addition to the mean of all ten trials when days are combined are reported in Table 4.7. The results of each of the five individual trials on both day 1 and day 2 are also reported in Appendix L. to provide insight into the consistency of results across trials.

For all three study populations, paired sample t-tests showed no significant difference between day 1 and day 2 for all seven assessed variables (p > 0.05). Figure 4.5 illustrates the B&A analysis of day 1 vs day 2 for each variable. For all 3 populations, J_{ecc} and t_{FmaxN} were the variables with the greatest magnitude of bias and LOA, and therefore, the most variability across days.

4.2.1.3 Population differences

A paired samples t-test comparing FEM to MALE data series for all variables identified a significant difference between sexes in the following cases. For both day 1 and day 2 there was a significant difference between FEM and MALE for the variables F_{max} (D1 p = 0.015; D2 p = < 0.001), J_{con} (D1 & D2 p = 0.002), *PPO* (D1 p = 0.001; D2 p = 0.002) and V_{to} (D1 p = 0.01; D2 p = 0.007). J_{ecc} was only significantly different on day 1 (p = 0.007). Normalised temporal variables were not significant between populations (p > 0.05) on both day 1 and day 2.

For the variables F_{max}, J_{con} and *PPO*, ICC and their 95% CIs were greater than 0.90 for both FEM and MALE populations, indicating excellent reliability. For these three variables, ICC values were slightly higher in the MALE population. Considering the B&A analysis (Figure 4.5), for J_{con} and *PPO*, the bias and LOA were smaller in the MALE population. F_{max}, however, showed the magnitude of the bias and LOA (as a percentage of the mean) were smaller in the FEM population, suggesting when the measured error was quantified, the FEM population showed a smaller magnitude of variability in the data. For the variable J_{ecc}, ICC values for MALE were excellent, while FEM indicated moderate to excellent reliability; although the obtained ICC value is 0.883 (suggesting good reliability), its 95% CI ranges between 0.519 and 0.973, meaning that there is a 95% chance that the true ICC value lands on any point between these two values. B&A analysis demonstrated a LOA almost twice as large in the FEM population than MALE. Both V_{to} and t_{ecN} produced greater ICC values in FEM than MALE, however, still greater than 0.90. The 95% CI for MALE V_{to} ranged from 0.838 to 0.990 (good to excellent reliability) compared to 0.947 to 0.997 in FEM. The 95% CI in t_{ecN} for FEM indicated good to excellent reliability (0.784 – 0.988) and moderate to excellent (0.673 – 0.980) for MALE. B&A analysis, however, showed LOA and a bias that was smaller in the MALE population than FEM for both V_{to} and t_{ecN}. Finally, for t_{FmaxN}, the ICC value was much greater in MALE (0.928, 95% CI 0.713 – 0.982) than FEM, which displayed an ICC of 0.622 indicating moderate reliability. The 95% CI, however, ranged from -0.325 to 0.909, indicating that there is a 95% chance the true value lies between a very large range that could indicate reliability anywhere from very poor, to excellent.

Table 4.7 *Bilateral Countermovement Jump Performance Variables, Kinetic and Normalised Temporal Data. Descriptive statistics (Mean ± SD) for the average of the five trials conducted on day 1, five trials on day 2 and all ten trials combined.*

		All Participants		Female		Male	
Variable	Day	Mean	±SD	Mean	±SD	Mean	±SD
Peak Force (N)	Day 1	1704	465	1433	242	1948	482
	Day 2	1695	428	1431	232	1933	424
	Combined	1700	445	1432	235	1940	452
Eccentric Impulse (N·s)	Day 1	88.9	26.7	77.2	17.9	99.5	28.8
	Day 2	92.7	25.6	81.6	20.1	102.8	25.9
	Combined	90.8	25.6	79.4	18.0	101.1	27.1
Concentric Impulse (N·s)	Day 1	180.6	58.5	135.8	29.4	220.8	48.0
	Day 2	180.8	57.9	136.3	28.5	220.8	47.6
	Combined	180.7	58.2	136.0	28.8	220.8	47.7
Peak Mechanical Power (W)	Day 1	3360	1126	2476	555	4156	892
	Day 2	3372	1097	2522	554	4138	879
	Combined	3366	1110	2499	552	4147	884
Take Off Velocity ($m \cdot s^{-1}$)	Day 1	2.22	0.38	1.93	0.33	2.48	0.18
	Day 2	2.22	0.37	1.94	0.32	2.47	0.18
	Combined	2.22	0.37	1.93	0.33	2.47	0.18
Ecc/Con Changeover % of jump	Day 1	68.3	2.9	68.5	3.5	68.2	2.3
	Day 2	68.1	3.2	68.0	4.1	68.1	2.1
	Combined	68.2	3.0	68.2	3.7	68.1	2.1
Peak Force % of jump	Day 1	78.5	10.1	76.7	10.5	80.0	9.6
	Day 2	80.4	8.8	81.6	5.4	79.2	10.8
	Combined	79.4	8.7	79.2	7.2	79.6	9.8

Table 4.8 *Intraclass Correlation Coefficient (ICC) (2,1) absolute agreement, average measures: Day 1 vs Day 2*

Variable	All Participants			Female			Male		
	ICC	95% CI Lower bound	95% CI Upper bound	ICC	95% CI Lower bound	95% CI Upper bound	ICC	95% CI Lower bound	95% CI Upper bound
Peak Force	0.993	0.982	0.997	0.988	0.945	0.997	0.991	0.963	0.998
Eccentric Impulse	0.957	0.888	0.984	0.883	0.519	0.973	0.978	0.915	0.994
Concentric Impulse	0.997	0.993	0.999	0.990	0.958	0.998	0.996	0.984	0.999
Peak Mechanical Power	0.997	0.993	0.999	0.990	0.960	0.998	0.996	0.984	0.999
Take Off Velocity	0.990	0.975	0.996	0.988	0.947	0.997	0.960	0.838	0.990
Ecc/Con Changeover % of jump	0.938	0.841	0.976	0.949	0.784	0.988	0.920	0.673	0.980
Peak Force % of jump	0.809	0.516	0.926	0.622	-0.325	0.909	0.928	0.713	0.982

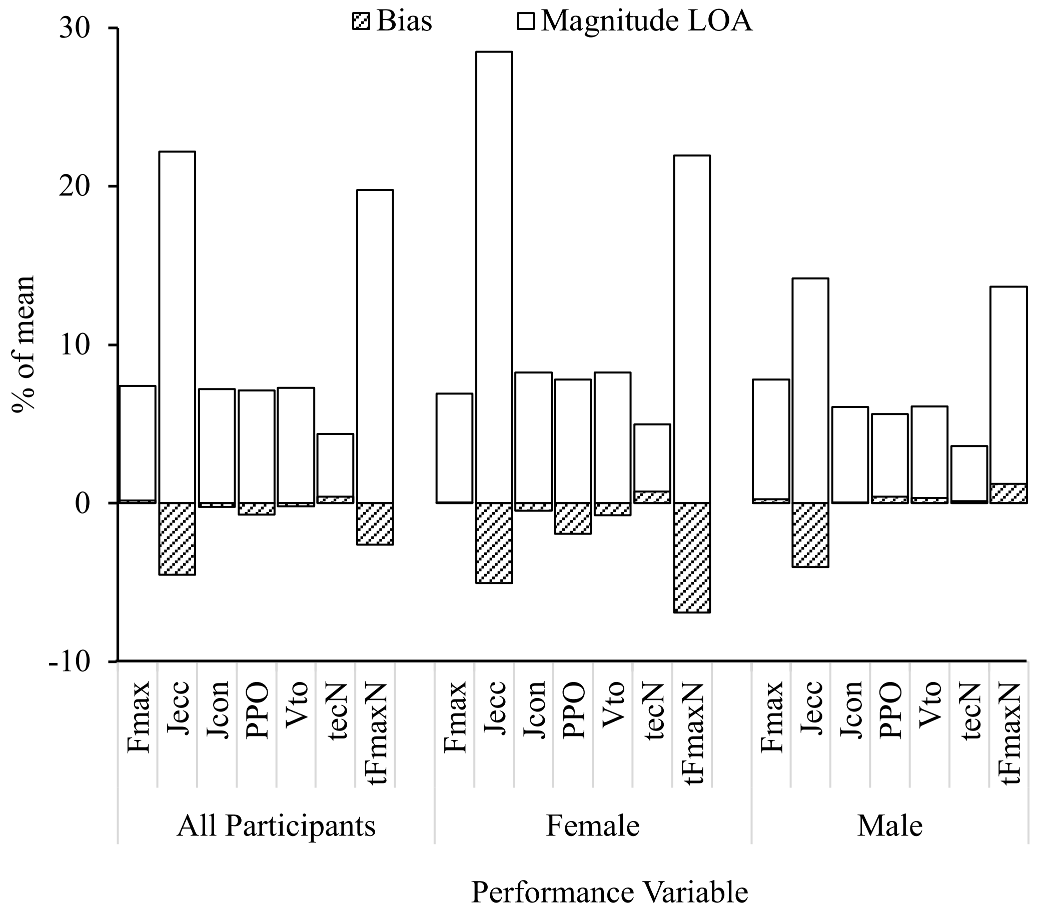

Figure 4.5. *Summary of Bland and Altman analysis comparing day 1 vs day 2 for all performance variables within all populations. Both bias and magnitude of LOA are reported as a percentage of the mean.*

4.2.2 Analysis of left and right leg contribution to bilateral countermovement jump performance

This area of analysis investigated how individual limbs contributed to each jump. This was achieved by measuring the force contribution of the left and right leg independently during a bilateral jump. Percentage asymmetry between limbs was then calculated using two forms of symmetry index; left leg vs right leg (LvsR) and higher value vs lower value (HvsL). HvsL was the method used to determine 'absolute' asymmetry by removing the directionality of the asymmetry, i.e., whether it was the right or left leg that produced the greatest value was not considered.

4.2.2.1 Assessment of normality of left and right leg data

For the variable F_{max}, a Kolmogorov-Smirnova normality test showed a departure from normality within the left leg data for day 1 in the ALL population (D(19) = 0.212, p = 0.025). All remaining left and right leg data for both day 1 and day 2 were normally distributed. Left and right leg data for J_{ecc} and J_{con} were shown to be normally distributed across all participant populations. Within the variable t_{FmaxN} only the MALE population left leg data, on day 2, deviated from normality (D(10) = 0.266, p = 0.043).

4.2.2.2 Assessment of left and right leg individual performance across days

For each variable the mean result of the 5 trials of day 1 were compared to the mean of those for day 2; both the left and right leg, there was no statistically significant difference ($p > 0.05$) between day 1 and day 2 for any of the four variables within all three populations.

Table 4.9 provides the mean values for left and right contributions on day 1, day 2 and days combined. Statistically, there was no significant difference between left and right legs for all variables on both day 1 and day 2 within any of the three populations. Within the FEM study population, the difference in J_{ecc} between left and right legs on day 2 was close to significant ($p = 0.052$).

4.2.2.3 Assessment of symmetry index validity and reliability

There was no statistically significant difference between day 1 and day 2 for both LvsR and HvsL assessment methods, however, in many cases, there was a statistically significant difference between the two methods of calculating symmetry index (SI); Table 4.10 shows the cases that had a significant difference.

Table 4.11 provides the ICCs for the comparison of SI scores on day 1 and day 2 for both LvsR leg and HvsL analysis. For all variables, ICC values were higher for LvsR than HvsL, with the exception of J_{con} in the MALE population, although both values were similar in indicating good reliability (LvsR = 0.788; HvsL = 0.794). B&A analysis (Figure 4.10), however, identified that when the measured errors were quantified, all variables, in all three populations, showed lower variability within the data in the HvsL SI analysis than LvsR.

4.2.2.4 Population differences

A paired samples t-test comparing FEM to MALE left and right leg data series, for all variables, identified a significant difference ($p \leq 0.05$) between sexes in the following cases.

On day 1 and day 2 there was a significant difference ($p > 0.05$) between populations for the variables F_{max} and J_{con} for both left and right legs. J_{ecc} was significantly different for the right leg on day 2 only ($p = 0.017$). t_{FmaxN} was not significant between populations for both left and right legs on both day 1 and day 2.

For the LvsR SI method there was no significant difference between FEM and MALE for either day 1, day 2 or days combined. For HvsL SI, there was only a significant difference between populations for F_{max} on day 1 ($p = 0.031$) and days combined, although close to insignificance ($p = 0.048$). All other variables on day 1, day 2 and days combined showed no statistically significant difference between populations.

For all 4 variables, in both the LvsR and HvsL SI methods, ICC values were higher in the FEM population. For F_{max}, ICC values suggest FEM data showed excellent reliability for LvsR (0.988, 95% CI 0.947 – 0.997) and good to excellent for HvsL (0.960, 95% CI 0.822 – 0.991), while reliability of MALE data was poor to excellent for both LvsR and HvsL (LvsR = 0.869, 95% CI 0.456 – 0.948; HvsL = 0.859, 95% 0.420 – 0.965). In fact, for all 4 variables in MALE, the range of the 95% CI extends from very poor to either good (0.75 - 0.90) or excellent (> 0.90). For example, in the LvsR analysis, for both J_{ecc} and J_{con}, ICC values indicate good reliability (0.781 & 0.788, respectively), however, the 95% CI for J_{ecc} was 0.058 – 0.947 and 0.087 – 0.948 for J_{con}. FEM data displayed a little more consistency in regards to tighter CIs for all variables. Despite this, both J_{ecc} and t_{FmaxN} had poor to excellent reliability in the HvsL SI method of analysis (J_{ecc} = 0.816, 95% CI 0.229 – 0.958; t_{FmaxN} = 0.844, 95% CI 0.315 – 0.965).

B&A analysis (Figure 4.10) shows that there were some differences in the reliability exhibited by the two populations. For F_{max}, the LOA and bias were smaller in the FEM population, especially in the LvsR SI analysis. The magnitude of the LOA and bias in J_{ecc} (LvsR) was considerably smaller in the MALE population, despite ICC suggesting higher reliability in FEM population (0.879, 95% CI 0.509 – 0.972 compared to 0.781, 95% CI 0.058 – 0.947). J_{con} showed substantially more variation in the MALE population for LvsR than FEM (bias = - 49%, LOA = 587% for MALE compared to bias = - 3%, LOA = 62% for FEM). The magnitude of LOA and bias for t_{fmaxN}, however, was far more comparable. In the HvsL method of analysis, F_{max} was smaller in the FEM population for both the magnitude of LOA and bias. The other three variables, however, were not as decisive. For the variables J_{ecc} and J_{con}, the magnitude of the LOA were marginally smaller in the FEM population, while the magnitude of the bias was slightly greater. For t_{FmaxN}, the magnitude of LOA were also twice as high for MALE, while the bias was just over 6 times smaller than FEM.

Table 4.9 *Independent contribution of left and right legs during the countermovement jump.*

		All Participants				Female				Male			
		Left		Right		Left		Right		Left		Right	
Variable	Day	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD
Peak Force (N)	Day 1	854	242	858	224	705	104	738	141	988	253	966	229
	Day 2	848	221	854	209	703	99	737	138	977	220	959	206
	Combined	851	231	856	216	704	101	738	139	982	235	962	217
Eccentric Impulse (N·s)	Day 1	46.6	13.7	45.2	15.5	44.4	10.6	37.5	10.5	48.5	15.8	52.2	15.9
	Day 2	48.6	13.7	44.2	16.4	46.4	13.6	33.9	12.3	50.5	13.4	53.4	13.9
	Combined	47.6	13.3	44.7	15.5	45.4	11.9	35.7	11.0	49.5	14.2	52.8	14.5
Concentric Impulse (N·s)	Day 1	88.5	37.9	94.0	27.5	59.7	20.6	77.7	19.9	114.5	30.4	108.7	25.0
	Day 2	89.1	38.4	93.9	25.2	59.9	19.9	78.6	20.3	115.4	31.4	107.6	20.8
	Combined	88.8	37.7	93.9	26.0	59.8	19.8	78.1	20.0	114.9	30.0	108.2	22.4
Peak Force % of jump	Day 1	77.7	10.5	80.0	9.7	75.7	10.9	78.4	9.6	79.4	9.8	81.5	9.6
	Day 2	79.2	9.7	80.8	9.1	78.5	8.8	82.5	7.6	79.9	10.4	79.2	10.0
	Combined	78.5	9.7	80.4	8.6	77.1	9.3	80.4	7.8	79.7	9.9	80.4	9.3

Note: Each day is representative of the mean of the five trials of that day, while combined is the mean of all ten trials across both days.

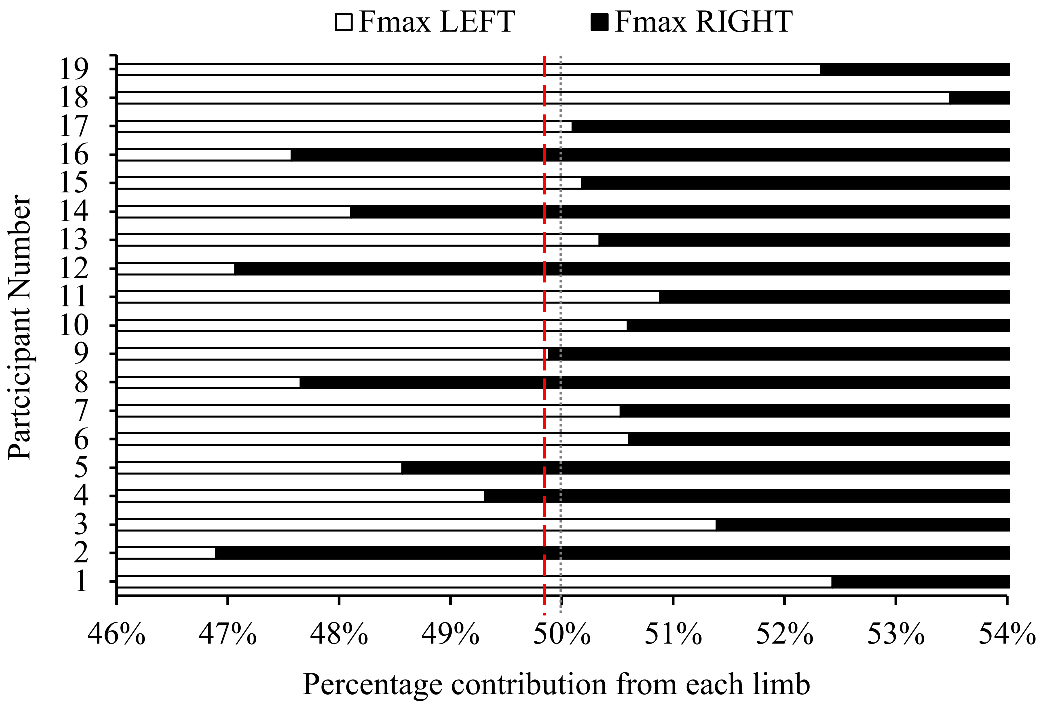

Figure 4.6. *Example illustration of the variability between individual participants within a single jump trial (J1) and the subsequent consequences of averaging; Mean contribution= Left 49.89% : 50.11% Right (as illustrated by the red line).*

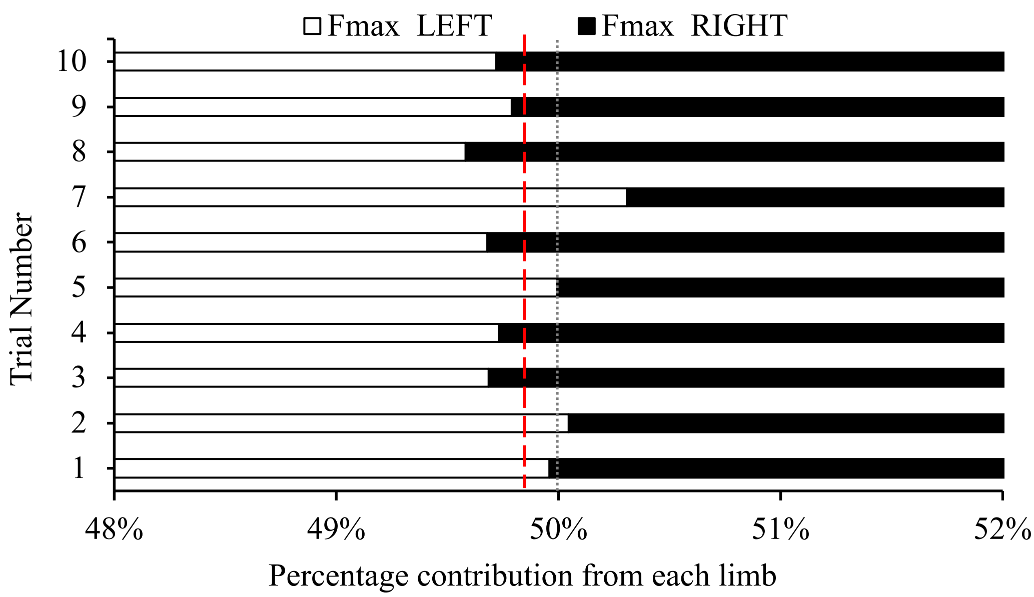

Figure 4.7. *Example illustration of the variability between the 10 trials and the subsequent consequence of averaging; Mean contribution= Left 49.85% : 50.15% Right of the total peak force (as illustrated by the red line). Each trial is the mean of all 19 participants.*

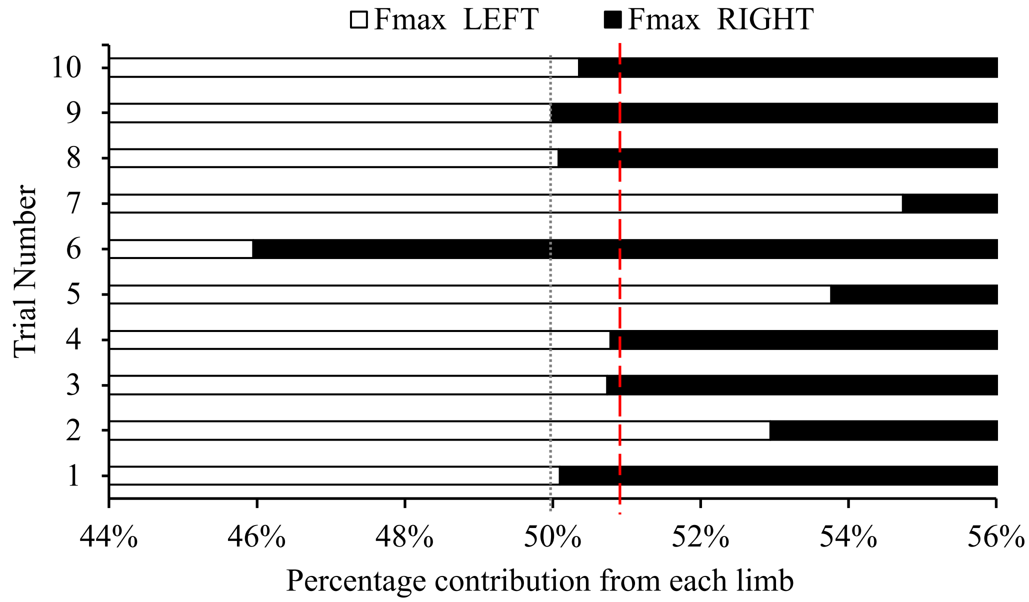

Figure 4.8. *Example illustration of the variability in contribution to total peak force each limb makes between the 10 jumps of a single participant (P17). The subsequent consequence of averaging is demonstrated by the red line; Mean contribution= Left 50.95% : 49.05% Right.*

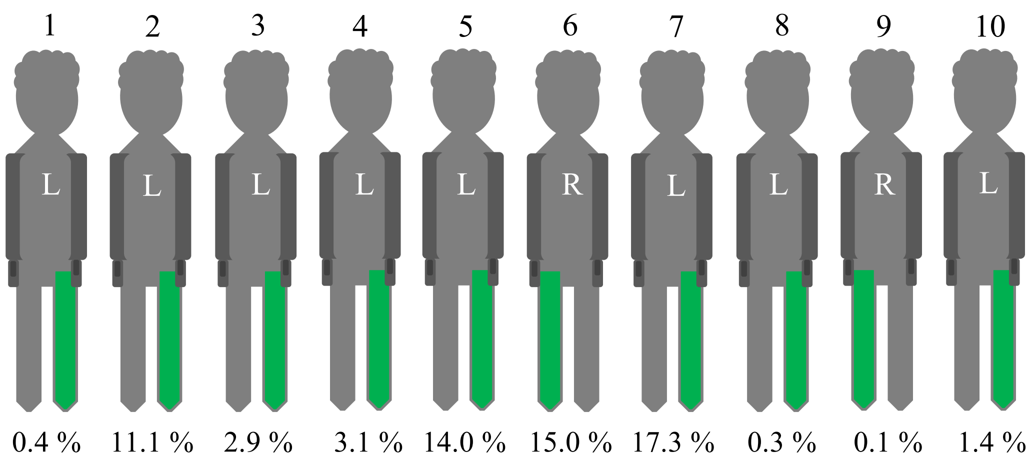

Figure 4.9. *Example illustration of the variability in percentage asymmetry between the 10 jumps for a single participant (P17). The leg that generated a higher peak force is colour coded green. Mean symmetry index = LvsR (3.6%) HvsL (6.6%).*

Table 4.10 *Symmetry Index (SI) Scores for each variable assessed as left leg vs right leg and higher vs lower value, irrespective of direction.*

Method of Symmetry Index Calc.		All Participants					Female					Male				
		Left vs Right (%)		High vs Low (%)			Left vs Right (%)		High vs Low (%)			Left vs Right (%)		High vs Low (%)		
Variable	Day	Mean	± SD	Mean	± SD		Mean	± SD	Mean	± SD		Mean	± SD	Mean	± SD	
Peak Force	Day 1	-0.61	7.14	6.20	3.97	⁎	-3.39	8.63	8.11	4.56	⁎	1.88	4.08	4.48	2.24	⁎
	Day 2	-0.82	6.98	6.23	4.01	⁎	-3.56	8.65	7.94	4.94	⁎	1.65	3.52	4.69	1.90	⁎
	Combined	-0.72	6.96	6.22	3.89	⁎	-3.47	8.58	8.03	4.65	⁎	1.76	3.56	4.59	1.94	⁎
Eccentric Impulse	Day 1	3.01	25.35	24.90	12.83	⁎	13.41	18.67	24.75	10.29	⁎	-6.35	26.88	25.05	14.74	⁎
	Day 2	6.97	24.74	25.68	14.99	⁎	20.21	24.50	32.15	18.23	⁎	-4.95	17.98	19.86	7.49	⁎
	Combined	4.99	23.47	25.29	12.33	⁎	16.81	20.67	28.45	13.83	⁎	-5.65	20.56	22.45	9.97	⁎
Concentric Impulse	Day 1	-6.92	28.16	25.39	14.63	⁎	-18.43	29.97	30.43	18.44	⁎	3.45	21.72	20.86	7.58	⁎
	Day 2	-6.94	27.51	25.55	14.30	⁎	-19.42	30.55	32.50	17.27	⁎	4.30	18.22	19.30	6.13	⁎
	Combined	-6.93	26.87	25.47	13.80	⁎	-18.93	29.79	31.46	17.05	⁎	3.87	18.08	20.08	6.26	⁎
Peak Force % of jump	Day 1	-2.59	7.13	4.87	6.21	⁎	-2.90	8.94	6.15	7.40		-2.31	4.96	3.72	4.60	
	Day 2	-1.62	9.27	6.47	8.64		-4.33	12.49	10.52	10.89		0.81	3.19	2.82	2.65	
	Combined	-2.10	7.87	5.67	6.83	⁎	-3.61	10.59	8.34	8.79		-0.75	3.60	3.27	2.65	

Note: In the Left vs Right method of assessment, (-) is indicative of a greater value for the right leg.

⁎ *Is indicative of a statistically significant difference ($p \leq 0.05$) between the two methods of % asymmetry/ SI assessment.*

Table 4.11 *Intraclass Correlation Coefficient (ICC) (2,1) absolute agreement, average measures: Day 1 vs Day 2 for both methods of symmetry index (SI) calculation.*

		Left vs Right SI			High vs Low SI		
			95% CI			95% CI	
	Variable	ICC	Lower bound	Upper Bound	ICC	Lower bound	Upper Bound
All Participants	Peak Force	0.972	0.927	0.989	0.952	0.875	0.982
	Eccentric Impulse	0.861	0.646	0.946	0.729	0.282	0.896
	Concentric Impulse	0.930	0.817	0.973	0.906	0.755	0.964
	Peak Force % of Jump	0.899	0.741	0.961	0.786	0.458	0.917
Female	Peak Force	0.988	0.947	0.997	0.960	0.822	0.991
	Eccentric Impulse	0.879	0.509	0.972	0.816	0.229	0.958
	Concentric Impulse	0.971	0.872	0.993	0.909	0.600	0.979
	Peak Force % of Jump	0.950	0.791	0.988	0.844	0.315	0.965
Male	Peak Force	0.869	0.456	0.968	0.859	0.420	0.965
	Eccentric Impulse	0.781	0.058	0.947	0.611	-0.384	0.900
	Concentric Impulse	0.788	0.087	0.948	0.794	0.194	0.949
	Peak Force % of Jump	0.584	-0.277	0.888	-0.012	-4.409	0.763

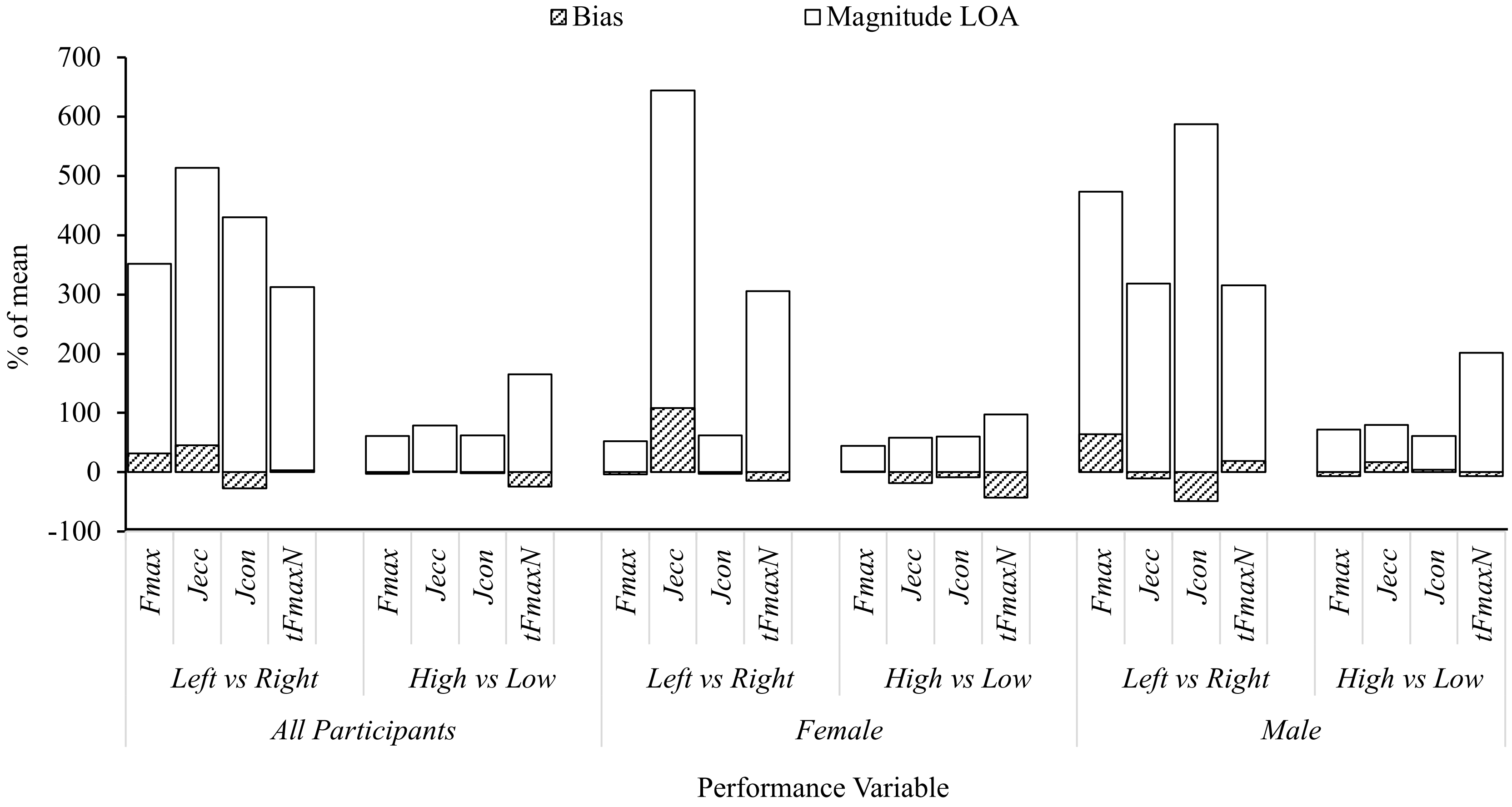

Figure 4.10. *Summary of Bland and Altman analysis for the comparison of day 1 to day 2 for the two methods used to calculate symmetry index when we consider the percentage asymmetry between left and right leg in each of the trials before then forming a mean of the five trials from each respective day.*

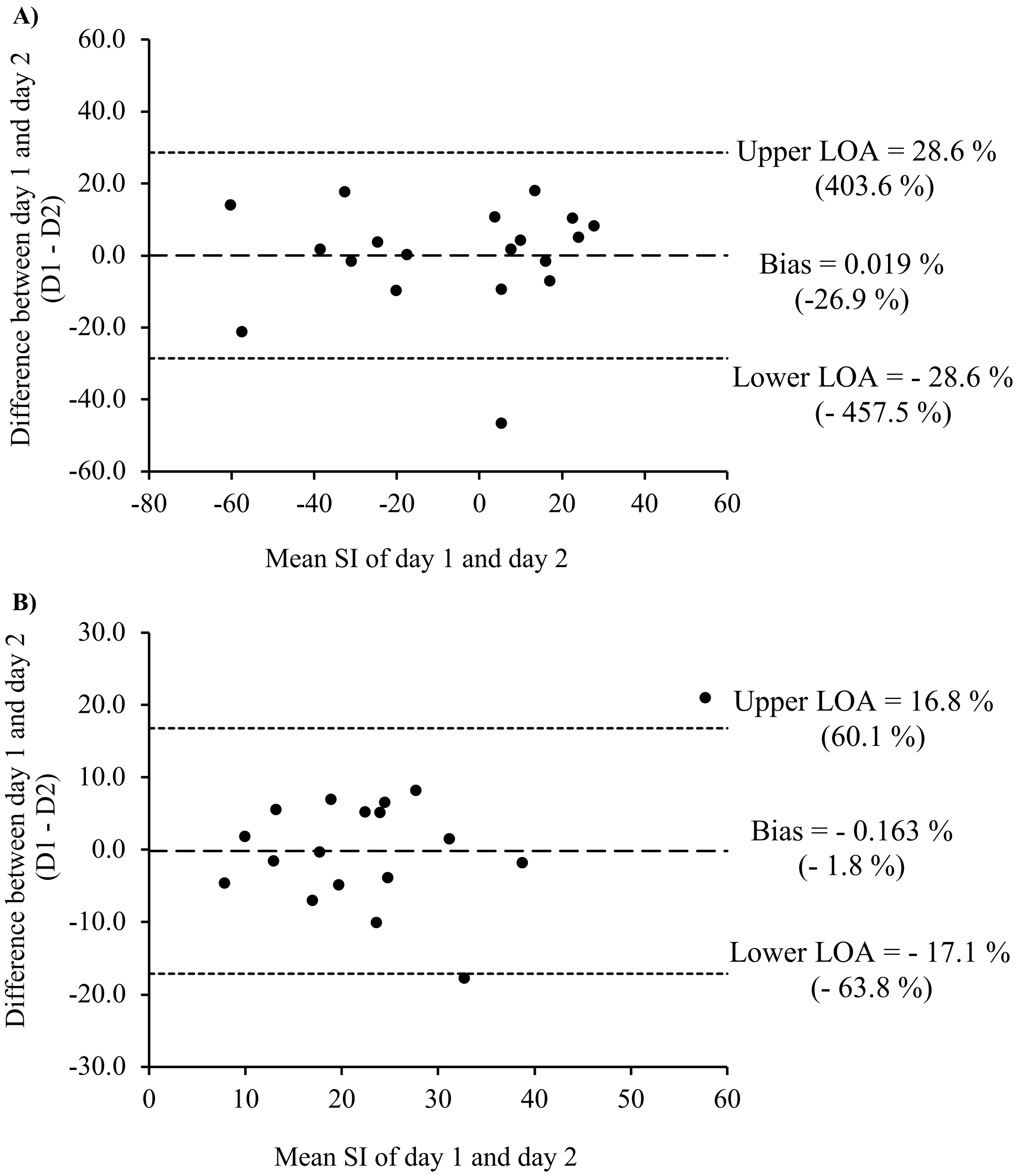

Figure 4.11. *Bland and Altman Analysis for the comparison of day 1 to day 2 for Concentric impulse in the ALL population using the Left vs Right (A) and High vs Low (B) assessment of percentage asymmetry/ symmetry index. Values given in brackets report the bias and LOA as percentages of the mean.*

Chapter 5

Discussion

5.1 Introduction

The purpose of the study was to investigate the test-retest reliability and agreement of postural stability, as measured by a statokinesigram, and lower limb bilateral asymmetry of neuromuscular variables, using two different methods of symmetry index (SI). Although protocols were conducted during the same testing sessions, postural stability and lower limb bilateral asymmetry assessments were treated independently. Overall, the findings identify the importance of key methodological considerations in both the assessment of postural stability and neuromuscular performance asymmetry.

5.2 Postural Stability

The following section refers hypotheses 1 through 4 and hypothesis 8, listed below:

1. There is no criterion method for the determination of postural stability performance, however, a methodology devised from the recommendations of previous research will produce valid and reliable results.
2. As sampling duration increases, so will the test-retest reliability of postural stability parameters derived from CoP analysis.
3. As the number of repetitions increases, so will the precision and validity of postural control outcome measures.
4. Visual condition will have a significant impact on postural stability performance. Further, CoP excursion measures will be significantly higher when visual feedback is removed.
8. Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population.

The current study conducted an assessment of postural stability using the most common variables in centre of pressure (CoP) analysis, total path length (L_p), area of 95% confidence interval (CI) ellipse (A_s) and CoP velocity (V_m) (Blosch *et al*., 2019). The results of the study have shown that with respect to test-retest reliability and magnitude of variability within the data, L_p and V_m are more reliable and valid measures of postural stability than A_s in both eyes open (EO) and eyes closed (EC) visual conditions.

5.2.1 Path length and Mean Velocity

Paired sample t-tests have shown only one instance where the mean L_p of the ALL population was significantly different between testing sessions. This difference was seen in the EC condition, however, was close to insignificance ($p = 0.05$). Similarly, V_m in the same epoch and condition showed, between day 1 and day 2, a difference on the borderline of statistical significance ($p = 0.049$). Despite this, the corresponding ICC values for both L_p and V_m demonstrated an excellent level of reliability (ICC = 0.973) and when the error was quantified using Bland and Altman (B&A) plot analysis, for both variables the 3rd 30 s epoch reported a magnitude of bias of 5% and limits of agreement (LOA) of 18% of the mean, which was by no means the largest and were, in fact, very comparable to the other epochs. For all epochs in both EO and EC of the ALL population, absolute ICC values were indicative of excellent reliability (> 0.90) with close 95% CI ranging from good to excellent. FEM and MALE populations individually were slightly lower, however, still reported mostly excellent ICCs despite 95% CIs that were slightly less definitive, ranging from moderate to excellent in the majority of cases. Variables, L_p and V_m showed higher ICCs in the EC condition. When B&A analysis is considered (Figure 4.1), for the majority of cases in all three populations, the magnitude of the LOA were smaller in the EC condition, but biases were larger.

The B&A plot method only defines the intervals of agreements, it does not say whether those limits are acceptable or not. Acceptable limits must be defined a priori, based on clinical necessity, biological considerations or other goals (Bland & Altman, 2003; Giavarina, 2015). Due to the lack of existing postural stability research and even more limited research including B&A, or quantified error, there is no current agreed level of error. As such, despite the apparent strong agreement between testing sessions, it is unclear whether the level of variability within the data is representative of the inherent variability we should expect to see. Consequently, it is not possible to state that the findings of the current study are acceptable, as this will be dependent on many other factors. However, a recommendation for the use of L_p and V_m for future research has been made based on the current findings of agreement and reliability.

Due to the heterogeneity of the existing literature, the comparison of outcomes and recommendations becomes increasingly difficult. No direct comparisons can be made between results as it is likely that any differences are the result of variability in the methodology. Nevertheless, across the literature there is a general consensus towards good reliability in both L_p and V_m. Systematic reviews by both Ruhe *et al.,* (2010) and Hébert-Losier and Murray (2020) have identified that L_p and V_m are two of the most reliable CoP parameters. Clark *et al.* (2010) reported high test-retest reliability of L_p for both EO and EC conditions (EO: ICC 0.86, 95% CI 0.71 – 0.93; EC: ICC 0.94, 95% CI 0.87 – 0.97) using a sampling duration of 30 s. Similar to the current study, reliability was reported to be higher in the EC condition. However, the results should be interpreted with caution as although the study has provided sufficient methodological details, there are some distinct limitations. Firstly, for the EC condition, foot placement was a narrow stance/ feet together, which has previously been reported to lead to lower overall reliability (Hill *et al.,* 1995). Additionally, the body orientation was reported to be 'hands on hips', which, as discussed in Section 2.5.3.4.2, raises the body's CoG, reducing the stability of the individual. Further limitations of this study include the number of trial repetitions, as only the average of three trials on each testing session were reported. If the current study had used only three repetitions, there would have been a standard error of the mean of around 15% (Figure 4.2). The study also reported a lower sampling frequency (40 Hz) than considered appropriate (Schmid *et al*., 2002; Scoppa *et al*., 2013) as well as a cut-off filtering frequency (12 Hz) slightly higher than the recommended 10 Hz (Salavati *et al.,* 2009; Schmid *et al*., 2002). The consequences of these differences is unclear.

V_m is a commonly reported variable in the literature, often reported to have high reliability in both EO and EC conditions (e.g., Chiari *et al*., 2000; Schmid *et al*., 2002; Takala, Korhonen & Viikari-Juntura, 1997). Kitabayashi *et al*. (2003) reported V_m as one of the most reliable from 34 parameters (ICC 0.96). The study had a large population (220 M & F), however, only a sampling frequency of 20 Hz was used and only 3 repetitions were conducted, reducing the ability to confidently interpret their results and compare to those of the current study. There are studies, however, that do not support the confidence in V_m conclusions; Doyle *et al*. (2005) noted low reliability coefficients ($ICC_{2,1}$ 0.05-0.71), however, it is important to consider methodological differences. Doyle *et al*. (2005) recorded data for only 10 s, compared to up to 100 s in the current study, which is contrary to previous research quoted in their own study

(Lafond *et al*., 2004) which indicated that this would be an insufficient sampling duration to gain reliable data.

The current study reported, overall, ICC values much higher than those of Doyle *et al.* (2005), although a smaller duration epoch, the 2nd 30 s in the FEM population, reported 95% CI of 0.077 – 0.956 which, if used alone, could potentially form comparable conclusions. However, due to the extended trial duration and multiple epochs evaluated in the current study, it is clear that the low reliability is not the case and would not be truly representative. With regard to L_p, it is interpreted that the smaller the L_p, the better an individual's postural stability (Donath *et al*., 2012). The current study didn't report the level of performance of athletes, but this would certainly be applicable to studies comparing populations. From an inspection of the results of the current study, on this basis, females demonstrated better performance than males for all epochs, however it is not known if this difference was significant.

5.2.2 Sway area

Both ICC and B&A plot analysis (Figure 4.1) have shown that A_s was the least reliable variable, with B&A plots exhibiting considerably larger LOA compared to L_p and V_m. In addition to this, both ICC and B&A analysis have demonstrated that there is minimal variation between any of the epochs for this variable. Whilst 90 s does generally produce higher correlations, there was minimal difference in terms of both the absolute ICC and the CIs, and magnitude of biases and LOAs. Therefore, these results would suggest that studies that have used shorter trial durations can still be considered valid.

The recommendation of the current study is that A_s is not used as a variable to define postural stability, as it shows limited reliability in all of the epochs. In addition to this, the variable itself is representative of the area covered by the L_p over the duration of the trial/epoch. L_p has been identified in the current study to be very reliable and so the combination of these results have identified that the fact that an individual has covered a particular area drawn out from their L_p, doesn't mean to say it is a reliable variable, as the same L_p can result in wildly different A_s. For example, in the current study, for a single participant, P1, the reported L_p of two separate 30 s epochs was 0.230 m, but the associated A_s were different, 34.8 mm^2 and 58.1 mm^2. Similarly, when comparing between participants, participants who recorded comparable L_p's of 0.875 m (P18) and 0.872 m (P2), each reported a A_s of 37.7 mm^2 and 358.0 mm^2, respectively. This demonstrates the unpredictability and large variation of A_s results, which could explain the low reliability.

Previous literature has reported A_s as an index of overall postural performance. Asseman *et al.* (2004) stated that a smaller area was representative of better performance. Given the poor reliability the variable has, defining or classifying an individual's performance based on A_s would not be appropriate. Clearly, individuals who produced a small ellipse area in one trial may be considerably different in another, or within the same trial the A_s may fluctuate. For example, the current study found that for one of the participants, P14, within one 30 s epoch of the 100 s trial, they recorded a mean area of 140 mm^2, which, depending on a study's classifications, could be considered as poor performance, however in the proceeding 30 s epoch, their A_s was, on average, more than 3 times smaller at 42 mm^2, which would be indicative of a far better 'performance'. Not only does this bring into question the suitability and validity of A_s as a variable, it also questions the effectiveness of shorter trial durations in gathering a sufficient volume of data to make valid conclusions.

A_s might become important in populations with clinical problems that may impact balance, such as injuries, like lower back pain (Harringe *et al.,* 2008; Lafond *et al.,* 2009; Ruhe, Fejer & Walker, 2011) or in elderly populations (Yennan, Suputtutada & Yuktanandana, 2009) in which a greater magnitude or volume of postural corrections may be required. However, in the current study with a healthy population, its limited reliability compromises any justification for future application.

5.2.3 Sampling duration

In the current study, the 90 s epoch generally had the highest test-retest reliability, particularly in the L_p and V_m variables, supporting its application in future research. However, the smaller epochs have also demonstrated very similar results (Tables 4.4 and 4.6) which could suggest that their use would not be as much of a limitation to a study. Despite this, if provisions allow, even though shorter epochs may provide equally reliable results, if given the opportunity to use a larger trial duration and, therefore, characterise longer periods of postural stability measures, longer durations would be recommended, as longer epochs would contain more information. The ability to characterise longer durations may also increase the validity of the outcomes. For A_s, the 3rd 30 s epoch showed lowest agreement in ALL and FEM populations (ICC ALL, 0.608-0.943; FEM, -0.138-0.949) which has consequently impacted the 2nd 60 s epoch. Despite results showing some favourability towards a 90 s trial duration, these findings would suggest that the inclusion of the 3rd 30 s may not be the most appropriate.

Alternatively, these results could be interpreted as showing that the 30 s epoch is, in fact, important to the validity of the study as it demonstrates the amount of variability exhibited by longer durations. This may be important to characterise, as without it, postural stability performance may be misrepresented by smaller trial durations and, consequently, misinterpreted.

As the current study is unable to definitively determine an optimal trial/ sampling duration for the assessment of postural stability, future research should consider longer trial durations. This should include the analysis of a wider range of varying epochs to identify whether an optimal epoch exists that enables the characterisation of robust traits of human postural control and strong test-retest reliability. Using a longer trial duration and subsequently dividing each trial into smaller epochs allows the comparison of different length samples partitioned from the same original CoP signal. Thus, controlling for any confounding effects of the initial transient component of CoP signal, by keeping this factor constant within each sample duration (Blosch *et al.,* 2019).

5.2.4 Effect of Visual Condition

The current study identified a significant difference between EO and EC for L_p and V_m. Higher ICC values for EC indicate a higher reliability for this condition. A_s, however, was more variable; in some epochs ICC values were higher in the EO condition while others favoured EC (Table 4.5). When considering the B&A plot, the magnitude of the LOA, although very similar, appear to be better in the EC condition, more so in the FEM population. For example, in the 90 s epoch, the magnitude of LOA for A_s were 64% of the mean in the EO condition, compared to 16% of the mean in EC. These results, however, do not enable the recommendation of one condition over another as the two methods are representative of, and testing, different control mechanisms. That is, the two conditions reflect visual acuity of a participant and might be a factor in predicting physical performance, potentially of use to both rowing, and other sports.

Van der Kooij *et al.* (2011) identified that significantly greater sampling durations were required to achieve stable standard deviations in the EC condition. This would indicate that removal of vision introduces larger amplitude displacements in the CoP signal that may only emerge after longer periods of stance. With the exception of A_s, the current study reported significant differences between EO and EC conditions, even in the shortest trial durations. Thus, indicating that the effect of lack of vision is identifiable from the offset of postural stability measurement. However, it should be considered if the magnitude of this impact changes as trial duration increases. For example, in the current study, for L_p, closer inspection of the data revealed that the percentage difference between visual conditions decreased as the duration of the epoch increased; the percentage difference between EO and EC, calculated as $\left(\frac{EO-EC}{EO}\right) \times 100$, was, on average, 24% in the 1st 30 s epoch, which decreased to 23.5% for the 1st 60 s epoch and finally reduced further to 22.8% in the 90 s epoch. Similarly, a closer inspection of V_m data revealed that as the trial duration increased, the magnitude of differences between visual conditions decreased; for the 1st 30 s the arithmetic difference between conditions was, on average, 0.288 $cm \cdot s^{-1}$, which reduced to 0.264 $cm \cdot s^{-1}$ for the 1st 60 s epoch and finally reduced again to 0.253 $cm \cdot s^{-1}$ for the 90 s epoch. Although only very small changes, these results suggest that despite a difference between EO and EC conditions being observed within a 30 s trial duration, a longer trial duration is needed to uncover the true magnitude of these differences and quantify the effect of lack of vision on stability measures.

5.2.5 Number of Trials

The number of repetitions required to gain acceptable reliability varied with the parameter being investigated. Velocity measures have previously demonstrated good to excellent test-retest reliability when two trials or more were performed (Golriz *et al.,* 2012; Lafond *et al.,* 2004), however the average of five measures was required for A_s. The current study has not directly investigated the test-retest reliability of multiple numbers of trials, however, from the cumulative moving average analysis (Figures 4.2, 4.3 and 4.4) it can be seen that a trend exists towards reduced standard error of the mean (SEM) as number of trials averaged increases. The determination of an appropriate number of repetitions is dependent on the definition of an acceptable SEM, so it would not be appropriate for the current study to recommend a number of repetitions, however, if, for example, a SEM (as a percentage of the mean) of 10% for the variable L_p was deemed acceptable, then, for the ALL study population in the current study, a minimum of seven trials would be necessary in the EO condition. However, for EC this level of precision was not met, indicating the need for more repetitions to achieve the required SEM. Within the current study, the FEM population reported lower SEMs than MALE.

5.3 Lower Limb Bilateral Asymmetry

The following section refers to hypotheses 5 through 8, listed below:

5. Bilateral CMJ performance, defined using the criterion method for the determination of neuromuscular variables, will show excellent test-retest reliability and day-to-day agreement.
6. In the execution of a bilateral CMJ, there will be differences in the contribution of each individual limb to overall performance. These variations in the magnitude of vertical component of the ground reaction force are the result of differences in the force generating capacity between limbs.
7. 'Sided' asymmetry calculations will produce statistically significantly different results to 'un-sided'. Additionally, 'sided' methods of asymmetry calculation will have lower reliability and validity than 'un-sided' methods quantifying absolute asymmetry, due to the consequences of nullification when producing group means.
8. Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population.

This part of the study analysed the test-retest reliability of seven bilateral countermovement jump (CMJ) neuromuscular variables as well as the contribution of each limb to the jump and the percentage 'difference' between limbs for the variables of interest. Overall, the study found that bilateral CMJ performance had high test-retest reliability (ICC >0.90). However, when assessing lower limbs individually, significant levels of asymmetry were identified. The current study also determined that the method of symmetry index (SI) calculation was a highly important factor in asymmetry determination and interpretation due to significant differences between calculation methods and the influence of 'dominant' limb interchangeability between trials and the nullification of values as a result of equal magnitude, but opposite direction, asymmetries.

5.3.1 Bilateral Countermovement Jump Performance

The seven variables derived from the force-time history of the VGRF were peak force (F_{max}), impulse due to eccentric (J_{ecc}) and concentric (J_{con}) phases, peak instantaneous mechanical power (*PPO*), take-off velocity (V_{to}) and normalised temporal variables, percentage of jump the changeover from eccentric to concentric phases occurs (t_{ecN}) and percentage of the jump that peak force occurred (t_{FmaxN}). There was no significant difference between the five trials conducted on each of the two testing sessions. The ICC for the first six variables were deemed excellent (ICC > 0.90 is excellent, 0.75 – 0.90 is good, 0.50 – 0.75 is moderate and < 0.5 is poor), as defined by Koo and Li (2016), with the exception of one variable in the FEM population (J_{ecc}) which was still good (0.883). The 95% CI of each variable within each population is given in Table 4.8, however, in the majority of cases, the CIs still fell within the excellent definition. For three of the variables, J_{ecc}, t_{ecN} and t_{FmaxN}, it was evident that there were individual participants within each of the populations where performance was not as consistent as others. *PPO* and J_{con} reported ICC values, and 95% CI that were very high with very small confidence intervals (e.g., 0.997, 95% CI 0.993 – 0.999 in the ALL population) suggesting they are almost 'perfect'. Yet, when the corresponding Bland and Altman (B&A) plots are considered, whilst there is still a very low bias e.g., J_{con} = 0.17% and PPO = - 0.7%), the LOA show an error of ± 7%. The result is in contrast to the ICC, which would indicate an almost perfect test-retest reliability; therefore, it is important when researchers use these variables that they are aware of the error in terms of quantitative values as well as the generalised scores that ICCs produce. These results suggest that if testing an individual athlete, as opposed to a group, it is going to be of great benefit producing means of at least 5 or 6 trials to minimise uncertainty. Minimising the uncertainty in an individual's measures would subsequently improve the confidence in changes detected by test-retest scenarios.

The ICCs reported in Table 4.8 are generally all high, with the exception of t_{FmaxN}, but only in the FEM population. This has consequently carried over to the ALL population statistics. As illustrated in Figure 5.1, the trace of a CMJ force-time history is bimodal with two local maxima, or 'peaks', corresponding to the breaking force peak and propulsive force peak. Although within two separate and distinct phases of a CMJ, depending of the movement strategy adopted by an individual, both have the potential to exhibit the peak force within a jump.

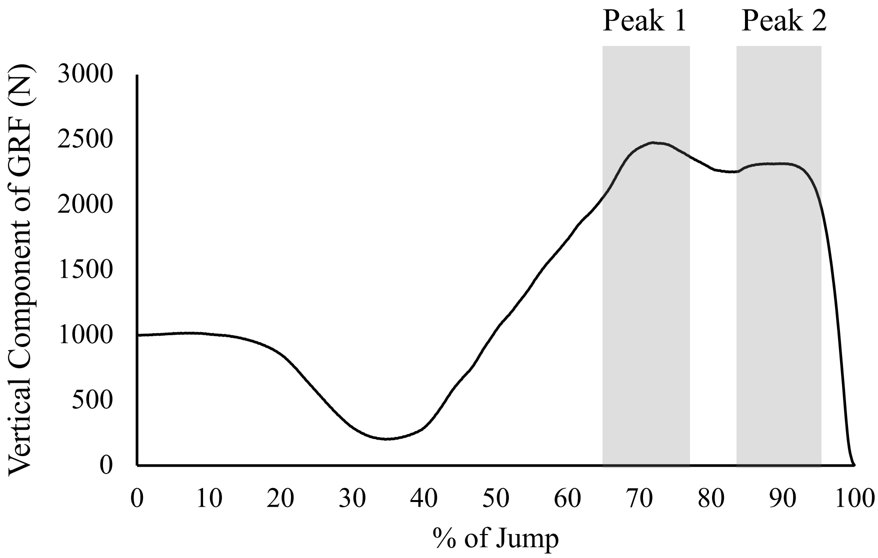

Figure 5.1. *Illustration of the two peaks in a countermovement jump force-time history and, as a result, the two potential locations for peak force to occur.*

The primary findings of Kennedy and Drake (2018) indicated that "a bimodal force-time curve does not represent an optimal pattern of CMJ performance and simply reflects the adoption of a movement strategy that can be characterised as an inefficient stretch-shortening cycle". As t_{FmaxN} is a normalised temporal variable, and these two peaks occur at different phases of the jump, the consequent percentage location either sits within the ~55-75% or ~85-95% range (data from current study). Considering the effect adopted movement strategy has on whether an individual achieves F_{max} in either peak 1 or peak 2 (as illustrated in Figure 5.1), the resultant percentage location of some participants may be within the first peak, while many may be in the second, or further still, within a single participant, across jumps this percentage outcome may vary. As a consequence, when averaged, the results have the potential to differ considerably based on how many jumps from within peak 1 or peak 2 are contained in that mean value. This provides reason for the considerably large magnitude of standard deviation within the data of the current study (Table 4.7) and also why the 95% CI for ICC analysis are so poor (Table 4.8); a 95% CI range from -0.325 to 0.909 in the FEM population suggests that there were cases within the population where a single participant produced their F_{max} in the braking phase for some jumps and the propulsive phase for others, meaning that the resultant percentage location of F_{max} had a wide range that will have appeared very inconsistent. Lake and McMahon (2018) identified in their population that over ten trials, 40% demonstrated inconsistencies in the shape of their CMJ force-time curve and, as a result, suggest that researchers look to assess individual consistency across trials before attempting to produce group averages. This is supported by the results of the current study, as the lack of consistency within participants cannot be discerned in the group means reported in Table 4.7. This can be attributed to lower t_{FmaxN} values for one participant being nullified by higher values of another and vice versa.

Due to the uncertainty within this variable, it may not be an appropriate metric to use; as expressed in Kennedy and Drake (2018). Caution should be exercised when using temporal phase analysis as pooling bimodal shaped curves may be a confounding factor, due to the force-time profiles of sub-groups that utilise small braking displacements during the execution of a CMJ. In support of this, results from Lake and McMahon (2018) showed that grouping individual time-normalised curves together could lead to the misrepresentation of individual movement strategies. Consequently, it was recommended that researchers and practitioners consider data on an individual basis.

5.3.2 Results of Asymmetry Assessment

Four neuromuscular variables were used to characterise the asymmetry between the lower limbs, they were F_{max}, J_{ecc}, J_{con} and t_{FmaxN}. Asymmetry was assessed as left leg versus right leg (LvsR) and using asymmetry magnitude, which only considered the size of the asymmetry, but not which leg produced the larger value i.e., un-sided (HvsL), otherwise termed absolute. The two methods of symmetry index (SI) calculation were then compared to establish the differences between them and the suitability of their use. Overall, there was little difference in the percentage asymmetry reported on day 1 and day 2 for all variables, and for three out of the four variables there was a significant difference between SI calculation methods in all three study populations, indicating the impact of directionality and reference value in the calculation of asymmetry. Of the variables assessed, J_{ecc} and J_{con} demonstrated the largest magnitude of asymmetries.

5.3.2.1 Peak Force

The assessment of the maximum force exerted by the left and right legs independently, showed, on average, an absolute inter-limb asymmetry, i.e., asymmetry irrespective of direction, of magnitude 6.22 ± 3.89% for the ALL study population. This result is in line with existing research, such as Newton *et al*. (2006) and Menzel *et al*. (2013), who reported asymmetry values of 5.68 ± 3.95% and 5.58 ± 4.55%, respectively. Similarly, the results of Impellizzeri *et al*. (2007) suggested mean F_{max} asymmetries of 6.18% in their formulation of normative data/ reference values for bilateral strength asymmetry. Additionally, the same study showed asymmetry indexes of 6.7% and 6.5% between 'strong' and 'weak' legs for testing sessions one and two, respectively, when assessing the test-retest reliability of their methodology. These results, specifically, are not reported by the authors, so SIs were calculated from the reported values of 'strong' and 'weak' limbs in order to allow the formation of comparisons with the current study.

Luk *et al*., (2014) also reported a SI of 6.73% for 'force' within one of their populations (jumpers), however, the study provided no definition of the variable beyond referring to their protocol as 'similar' to Newton *et al.*, (2006); consequently, it is unclear whether the variable 'force' in Luk *et al*., (2014) was referring to peak or average force, as both of these variables were assessed in Newton *et al*., (2006). Consequently, the outcomes of this study need to be interpreted with caution.

When assessing SI as LvsR, and, therefore, considering the directionality of the differences, the current study identified, on average, an asymmetry of -0.72% ± 6.96% for the whole population, -3.47% ± 8.58% in the FEM population and 1.76 ± 3.56% for the MALE population, with the magnitude of asymmetry recorded ranging from -30.7% to 17.3% ('–' indicative of a greater value recorded from the right leg). Newton *et al*. (2006), Impellizzeri *et al*. (2007), Benjanuvatra *et al*. (2013) and Menzel *et al*. (2013) were the only studies to consider F_{max} using LvsR analysis. Newton *et al.* (2006) reported a mean asymmetry of 2.97 ± 6.49% (indicative of a higher value from the left leg) while Menzel *et al* (2013) reported -0.69 ± 7.21% ('-' indicative of higher value from left leg). Impellizzeri *et al.* (2007) reported that the 'normal range (95% ref. interval) was -15.1% and 15.0%' with an 'average value' of 0.8%. These results show an outcome similar from the current study. The reasoning behind the small mean values and large standard deviations, or ranges, for these studies are proposed in

Sections 5.3.3 and 5.3.4 of this discussion and relate to the averaging of LvsR analysis resulting in the nullification of higher magnitude asymmetries both within- and between-subjects.

Benjanuvatra *et al*. (2013) provided results for the asymmetry assessment of LvsR legs, however, they also split their study populations into left and right 'dominant' and 'symmetrical' groups. In their study, 'the dominant limb was defined as the one that consistently recorded higher jumps than the other. To be in an asymmetric group, the best 4 performances had to be from the same side' and, with that, 'left and right side differences were required to be statistically significant as determined by an independent sample t-statistics (a = 0.05)'. Using the division of groups, the study reported SIs for F_{max} of -10.0 ± 7.8% and -7.9 ± 4.4% for 'left dominant' men (n = 5) and women (n = 5), respectively, -0.5 ± 3.4% and 0.0 ± 2.1% for 'symmetrical' men (n = 13) and women (n = 9), respectively, and finally 10.0 ± 2.5% and 12.7 ± 6.4% for 'right dominant' men (n = 10) and women (n = 16), respectively. The division of participants into side-dominant groups, in theory, would remove the risk of nullification due to one participant favouring one leg while another favours the other, however, as will be discussed, the current study has demonstrated the interchangeability of 'dominant' limb from one trial to another which could, consequently, question the rationale and confidence in grouping individuals due to their 'dominant limb'. Despite this, it is evident from the results of Benjanuvatra *et al.* (2013) that those grouped in the left or right 'dominant' groups displayed larger magnitudes of asymmetry. Additionally, the 'symmetrical' population displayed far smaller magnitudes, as would be expected from the grouping title. However, as given in the methodology of the study, in order to deem a limb 'dominant', there had to have been a significant difference between limbs; it is possible that asymmetries were present within the 'symmetrical' group that were not quite statistically significant and, further, the small group mean may be the result of the aforementioned nullification of values.

Although the existing results of asymmetry when assessing a 'stronger', or 'dominant', limb compared to a 'weaker', or 'non-dominant', limb or as LvsR, are very similar to the outcome of the current study, the results of F_{max} asymmetry in each of the previous studies identified should be interpreted with caution. In the studies conducted by Newton *et al*. (2006), Menzel *et al*. (2013) and Luk *et al*. (2014), results of SIs are the average of only a small number of trials (2, 3 and 3, respectively) which are unlikely to present the greatest reliability or be truly representative of the variables being studied, for reasons that will be discussed in Section 5.3.3 and 5.3.4. Secondly, Newton *et al*. (2006) included the use of an arm swing, which has been shown to influence vertical jump performance (Lees *et al*., 2004) and the variability of lower extremity inter-limb asymmetries in that it can influence CMJ symmetry during both the loading and propulsion phases (Heishman *et al*., 2019; 2020). In addition, the methodology given by Newton *et al*. (2006) and Luk *et al*. (2014), presented no information regarding the acquisition settings of the force-platform set-ups, and, as such, the accuracy, reliability or validity of the subsequent data need to be interpreted with caution.

It is also important to note that the study conducted by Impellizzeri *et al*. (2007) was using the single force-platform methodology, in which only one limb can be assessed at a time for each jump; therefore, these results must be interoperated with extreme caution as this method requires two separate jumping attempts in order to form just one inter-limb comparison. Consequently, although the methodology may allow the assessment of the force generating capacity of each limb, it is not truly representative of the asymmetrical generation of force demonstrated within a single trial, or able to quantify individual limb contributions in response to changing jump strategies and is, therefore, not reporting the same information the current study has investigated.

It may not be possible to look to existing studies to confirm the results of the current study, due to methodological differences etc, however, the test-retest reliability of F_{max} SI for both LvsR and HvsL methods of analysis appear to be very good, in terms of reliability expressed as ICC. For both methods in the ALL and FEM populations, ICCs were indicative of excellent reliability (> 0.9) while the MALE population was also indicative of good reliability (LvsR = 0.869; HvsL = 0.859). However, the 95% CI would suggest that there were cases within the population where reliability may have been poor (< 0.5). B&A plot analysis on the other hand identified that, even in the best cases, the LOA had a magnitude of 44% of the mean (HvsL, FEM) with some substantial values reported for LvsR analyses, such as 473% of the mean for F_{max} in the MALE population. There are few reports of the test-retest reliability of F_{max} within the existing literature, however of those provided, Impellizzeri *et al.* (2007) reported an ICC value of 0.91 when including 5 jumps in their analysis, Menzel *et al*. (2013) reported an ICC value of 0.74 and Luk *et al*. (2014) reported that the 'test-retest reliability of the dependent variables showed intraclass correlation coefficients of R ≥0.947'. Even though the author has used a 'R' it is clear from the context that they were referring to ICC. Additionally, Heishman *et al*. (2019) reported, for the NAS condition, inter-session ICC values of 0.815 and 0.911 and intra-session reliability of 0.842 and 0.918 for F_{max} in the eccentric and concentric phases, respectively. Heishman *et al.* (2019), however, did not use an SI in order to quantify asymmetries; instead, they reported the lower limb difference score as only the arithmetic difference with respect to each variable, not the percentage difference. Hence, the reported asymmetry results of this study have not been presented for comparison with the current study as they are raw, absolute, differences that bear no relation to the SI reported in the current study.

Impellizzeri *et al.* (2007) report the use of B&A analysis for the assessment of test-retest reliability, however, only provide the interpretation that all residuals in the plots showed no evidence of significant heteroscedasticity, and changes in the means between test and retest were not substantial (lower than ± 1.52%). The limited information reduces the ability to form comparisons with the current study considerably.

Benjanuvatra *et al.* (2013) provided no information regarding the reliability of F_{max}, but do, however, report that the ten participants, who were tested on two separate occasions, recorded asymmetry indexes that were 'consistent'. They also reported that 'this suggested that asymmetry was not because of random variability of the motor system but is a robust trait of CMJ performances'.

The reported test-retest reliability of all of the above studies report good to excellent reliability, however, they should be interpreted with extreme caution as a consequence of the significant methodological and analytical differences they have with the current study and the absence of 95% CI reports in all studies. If 95% CIs had been excluded from the current study, and only the ICC value alone used, the reliability of measures, in particular the MALE population, would have been overestimated.

5.3.2.2 Eccentric Impulse

The results of the current study found, on average, an asymmetry between the lower limbs of 25.29 ± 12.33% when the directionality of the asymmetry is not considered.

When considering the SI using LvsR, the current study reported a significantly lower mean value of 4.99% but much higher SD (23.47). As will be discussed in more detail, the dramatic reduction in mean value with LvsR SI is very likely the result of nullification due to asymmetries of equal magnitude, but opposite signs. For example, the current study reports that for the FEM population J_{ecc} displayed SI values of 16.81 ± 20.67% while the MALE population displayed a mean SI of -5.65 ± 20.56% ('-' indicative of greater values from the right leg). These results demonstrate the vast range of asymmetry results obtained within the population. For the ALL group, SI values for individual trials ranged from as little as 0.02% up to 92.23%, so it is likely that a group mean would never be truly representative of the true magnitude of observed asymmetries.

There are limited reports of J_{ecc} asymmetry assessments in the literature, phase-specific analysis has tended to report propulsive impulses (J_{con}) with no consideration of the asymmetries that can be observed during the braking phase of a CMJ (J_{ecc}). One study that did include J_{ecc} in their assessment was Jordan *et al.* (2015). The results of their study differ greatly to those of the current study in that they reported, in their uninjured/ healthy population, asymmetries of magnitude 1.0%, 95% CI: -1.5 to 3.5. The results of the current study would be more comparable to their ACL-R population, of whom are individuals with previous ACL injury but have fully returned to activity. The results of that population were 5.2%, 95% CI: -4.5 to 14.9. Results of this study should, however, be interpreted with caution as although the study defines the two phases of J_{ecc} and J_{con} as 'the time interval from the maximum negative velocity to zero velocity (deepest BCM position)' and 'the instant of zero BCM velocity to the instant of jump take-off', respectively (BCM representative of body centre of mass), they do not provide information about the determination of instant of t_{to} or t_i. As the integration necessary for impulse determination is reliant on the correct identification of t_i (Owen *et al*., 2014), we cannot be confident in the validity of the subsequent impulse measurements. Additionally, the sample size of nine in each population of Jordan *et al.* (2015), (ACL-R = 5 female, 4 male; uninjured = 4 female, 5 male), is lower than could be considered ideal for statistical power (Houser, 2007; Prajapati, Dunne & Armstrong, 2010), however, their statistical power calculation was based on pilot data in which a minimum sample size of eight subjects per group was deemed necessary.

Despite the errors these limitations may impose on the study, it is still clear that the magnitudes of asymmetry reported in the current study far exceed those of

Jordan *et al*. (2015), or what would be expected considering the consistency of bilateral performance across trials.

As far as can be ascertained, the test-retest reliability of J_{ecc} asymmetry has not previously been reported in the literature. The current study reports that, using absolute ICC values, the reliability of J_{ecc} when assessed using LvsR was good for the ALL and FEM populations and moderate for MALE. The HvsL methodology ICCs were lower in each population, although still moderate and good. 95% CI were, however, in most cases (very) poor to excellent, suggesting that within the population some outcomes may be the result of random variability instead of a robust trait of human performance. In general, ICC analysis would suggest that LvsR was a more reliable method of SI assessment, however, when the errors in the data were quantified, using B&A plot analysis, HvsL SI had substantially less variation compared to LvsR. For example, in the ALL population, LvsR reported a 46% of the mean bias and 513% LOA, while HvsL showed a bias of only 0.3% of the mean and 70% LOA. Despite being considerably smaller than LvsR, the magnitude of the LOAs for HvsL are still very large and suggest that there is a considerable amount of variation in their data.

5.3.2.3 Concentric Impulse

The current study reported the largest magnitude of asymmetries in the J_{con} variable. Based on these results, an absolute SI of, on average, 25.47 ± 13.80% would be expected across a population of male and female individuals. Between populations, a larger magnitude of asymmetry was observed in the FEM population (31.46 ± 17.05% with range from 1.13% to 89.04%) compared to MALE (20.08 ± 6.26% with range 1.23% to 49.67%). LvsR assessment reported smaller mean SI magnitudes than HvsL but, like J_{ecc}, larger SDs for all three populations (ALL: -6.93 ± 26.87, FEM: -18.93 ± 29.79 and MALE: 3.87 ± 18.08). Similar to the other neuromuscular variables, the appearance of smaller SI magnitudes reported for the LvsR SI was likely the result of opposing values favouring the left or right limb; in the current study, although values of absolute (HvsL) SI across the ALL population ranged from 1.13% to 89.04%, values for LvsR SI, in the same population, ranged from -89.04% to 49.67% ('-' indicative of a greater value from the right leg), which would account for the reduction in LvsR SI magnitude through nullification of values.

Studies reporting J_{con} asymmetry have provided similar findings; Benjanuvatra *et al*. (2013) reported impulse during the 'extension phase', which appears to mean to mean the concentric phase, SIs of magnitude -16.0 ± 9.8% and -13.4 ± 6.0% for 'left dominant' men (n = 5) and women (n = 5), respectively, -0.5 ± 5.2% and 1.6 ± 4.1% for 'symmetrical' men (n = 13) and women (n = 9), respectively, and finally 16.1 ± 8.0% and 19.1 ± 8.7% for 'right dominant' men (n = 10) and women (n = 16), respectively. It is evident that this study identified mean SIs of similar magnitudes to the current study, however, the range reported in the population appears to be smaller, i.e., the SIs reported in the current study have extended up to 89%, but based on the SDs reported in Benjanuvatra *et al*. (2013) this may not have been the case. More comparable to the current study were the results of Menzel *et al*. (2013) whom reported an absolute SI of 20.66 ± 14.60% for 'net impulse' and, when assessed using LvsR, reported an SI of 6.43 ± 24.64%. Although the results of Menzel *et al.* (2013) are very similar to those of the current study, their results should be interpreted with caution. The calculation of impulse is dependent on the determination of t_i and t_{to} (Owen *et al*., 2014), however, this study defined t_i as the 'moment when the GRF dropped below body weight', a method that is not validated or acknowledged as being correct. Additionally, the end of the CMJ was defined as 'the moment when total vertical GRF was zero for the first time', which will not have accounted for systematic noise of the force platform, as identified in the pilot study within the current study (Appendix G) and may, therefore, overestimate the duration of the CMJ. Furthermore, Menzel *et al.* (2013) provided no details of the integration method or origin of instantaneous impulse. As with J_{ecc}, Jordan *et al.* (2015) reported smaller magnitudes of SI for both their uninjured and ACL-R populations compared to the current study. Although their results may be the product of nullification due to LvsR analysis, they reported asymmetries of magnitudes 0.5%, 95% CI: -1.3 to 2.4 (uninjured) and 6.8%, 95% CI: 1.5 to 12.0 (ACL-R). However, the results of this study should be interpreted with caution due to its methodological and analytical limitations.

The current study reported absolute ICC values that would suggest excellent test-retest reliability; however, the 95% CI for the entire study population indicated this reliability was good to excellent for LvsR and moderate to excellent for HvsL. Within the individual FEM and MALE populations, FEM participants appeared to show better reliability (LvsR ICC 0.971, 95% CI: 0.872 - 0.993; HvsL ICC 0.909, 95% CI: 0.600 - 0.979) compared to MALE (LvsR ICC 0.788, 95% CI: 0.087 - 0.948; HvsL

ICC 0.794, 95% CI: 0.194 - 0.949). These results would suggest that, particularly in the MALE population, results were the consequence of random variability as opposed to representing a robust trait of human performance. B&A plot analysis (Figures 4.10 and 4.11) support these findings by demonstrating that when the error within the data is quantified, for the ALL population, a bias of -27% of the mean and LOA of magnitude 431% of the mean are reported for the LvsR analysis. HvsL SI assessment methods appear far more favourable, with a bias of -1.8% and LOA of 62%. Although considerably smaller, these results are still indicative of a large magnitude of variability within the data. With regard to population differences, the B&A analysis clearly illustrates the much smaller magnitude of variation for J_{con} in the FEM population compared to MALE for LvsR SI analysis.

Previous reports of J_{con} reliability have been varied; Benjanuvatra *et al*. (2013) reported that analysis of J_{con} revealed a strong relationship using Pearson's product correlation coefficient ($r = 0.99$, 95% CI: 0.96 - 0.99) between testing sessions. Menzel *et al*. (2013), however, reported an ICC of bilateral differences of impulse of 0.71 ($p < 0.05$), indicative of only moderate reliability (Koo & Lee, 2016). Both of these studies did, however, use very different methodologies which makes the comparability of results to each other, and the current study, challenging. The sample size for the assessment of test-retest reliability was a limitation in Benjanuvatra *et al*. (2013); despite having an initial study population of 58 (men, n = 28; female, n = 30), only five men and five women were selected to return for a second assessment. Menzel *et al.* (2013) had a larger population of sixteen participants, however, reliability assessments were only conducted as part of a pilot study in which only three 'valid attempts' were used for analysis. Jordan *et al*. (2015) did not include any assessment of test-retest reliability in their research. Heishman *et al.* (2019) reported an additional assessment of J_{con} test-retest reliability. The study did not utilise SI for the assessment of asymmetry, only arithmetic differences, however the test-retest reliability of those measures still relates to the reliability of J_{con}. They reported ICC values of 0.868 and 0.886 for inter- and intra-session reliability, respectively. The results would suggest a good level of reliability, however without the inclusion of 95% CI, the true level of variability in the data is unclear. In addition to this, the methodology of impulse calculation within the study compounds the limitations to confidently interpret their results. A set value of 20 N offset from BW, was selected to define jump initiation. As established by Owen *et al*. (2014), despite there being no agreed method for the determination of t_i, methods that define the start of a jump as the point when VGRF exceeds a set value are limited, as they are likely to produce an erroneous initiation point, especially in participants of higher bodyweight.

5.3.2.4 Temporal Analysis of Peak Force Production

The results of the current study reported a mean absolute SI of 5.67 ± 6.83% for the variable t_{FmaxN}, with a greater magnitude of asymmetry reported for FEM compared to MALE (8.34 ± 8.79% and 3.27 ± 2.65%, respectively). As established in 5.2.1, this normalised temporal variable has some inherent error due to the nature of a CMJ force-time history. That is, the bimodal nature of local maxima in the concentric phase of the jump, either of which can be a global maximum. Consequently, using this variable could result in a substantial increase of variation within the data. Therefore, this variable may not be the most appropriate when represented by a group mean and is clearly a limitation. On an individual basis however, this variable may be beneficial in identifying the timing, in conjunction with, utilisation of each limb. For example, upon closer inspection of data from the current study, it was noted that for a jump (J8) of P3, the reported t_{FmaxN} had an SI of 0.00% due to the timing of F_{max} occurring at exactly 91.26% of the jump for both limbs; within this, however, that same jump reported a 6.58% SI between the F_{max} of left and right leg. This meant that the left and right limbs produced their F_{max} at the exact same instant, but the magnitude of the force produced was considerably different. An equally notable finding was P10 (J1) whom reported a 26.83% SI between left and right limbs for t_{FmaxN}, but the magnitude of F_{max} was only 2.33% different. This meant that they produced a similar magnitude of force, but at completely different times within the jump. The implications of both of these findings might be important, however, further research into individual variation in temporal variables is needed to determine the usefulness of the variable.

With regard to the test-retest reliability of this variable, absolute ICC values suggest that for the FEM population, reliability was excellent (0.950) and good (0.844) for the LvsR and HvsL methods, respectively, while the MALE population reported moderate and poor ICCs for both methods (0.584 and -0.012). B&A analysis demonstrated the variable to be the worst performing, with respect to variability and error, reporting the greatest magnitudes of bias and LOA for all three populations in the HvsL. Following the findings of bilateral t_{FmaxN} performance, these findings were not unexpected and confirm that, at least when pooling results, it is not a suitable outcome variable. As far as can be ascertained, there is no existing research including the assessment of t_{FmaxN} in the evaluation of lower limb asymmetry.

5.3.3 'Dominant' Limb Interchangeability

HvsL results suggest that we would expect to see, on average, an asymmetry of 6% between legs in the generation of F_{max} within a single CMJ, however, this result, when used in conjunction with the LvsR data, allows us to identify that the asymmetry is only the difference present between the two limbs, which may be interchangeable between left 'dominant' or right 'dominant', not a single limb producing the greater performance each time.

The results of the current study have identified that whether it is the left or right leg that contributes more to the execution and performance of a jump can change between trials (for example Figure 4.7, 4.8 and 4.9). As a result, it is necessary to conduct multiple trials in order to assess and/ or monitor this. Consequently, when considering left and right legs as only LvsR, it would not be appropriate to produce a group mean, as the absence of imbalance when comparing right and left legs could be attributed to the fact that some people are right leg- 'dominant' whereas others are left leg-'dominant', therefore nullifying strength differences when these are averaged across the group (Newton *et al.*, 2006). For example, a 3% difference in contribution favouring the left leg in one participant, may be counteracted by a 3% difference in contribution favouring the right leg in a different participant (Figure 4.6). Furthermore, within the same participant any percentage difference between legs that may vary from left to right from one jump to the next may also be muted as a consequence of averaging. Similarly, in a study where asymmetry is measured/ reported using only one trial per participant, this data would show an erroneous result, in terms of asymmetry analysis, as a single jump cannot describe the asymmetry of an individual reliably; as demonstrated in the current study, one jump is not representative due to the sizeable level of between-limb variation within participants.

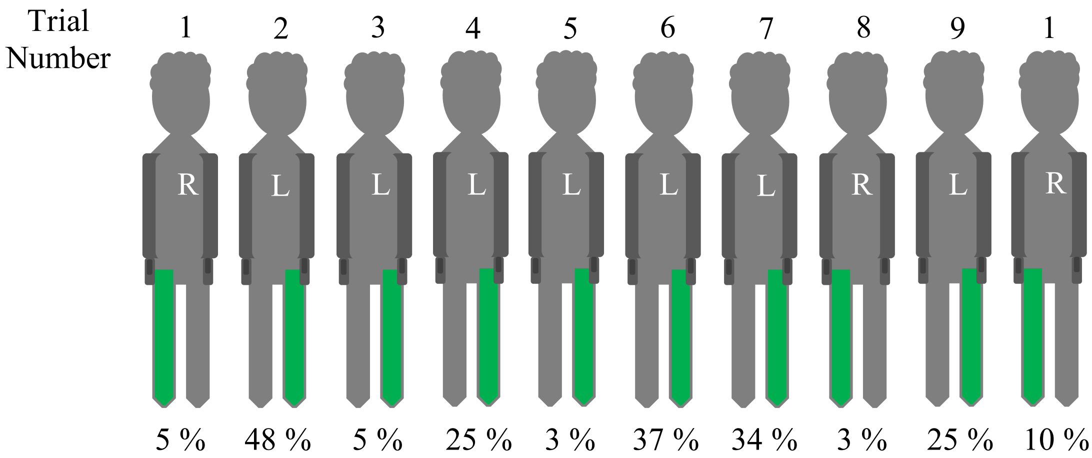

***Figure 5.2**. Concentric impulse percentage asymmetry between lower limbs for all ten trials of a single participant (P15), demonstrating the variability in symmetry index scores and interchangeability of the limb producing a greater value from one trial to the next. The limb generating the greatest impulse due to concentric contraction is illustrated in green and labelled L, left or R, right.*

A variation of left to right contribution from one jump to the next may be indicative of the absence of a preferred or dominant leg and could suggest that asymmetries are simply the consequence of changing movement strategies. Bilateral data analysis clearly shows that there was excellent consistency across trials for the combined effect of left and right limbs. Consequently, any difference between limbs that alternates from left to right within these jumps is potentially the result of the jump strategy i.e., varying limb contribution, as combined they are able to produce the same level of performance outcome, but evidently in a different way. Despite most studies referring to the difference between limbs, as demonstrated in the current study, it is important to note that this is not always the case and, in fact, intra-limb variations (differences within the same limb) will be evident when performing repeated athletic tasks and most likely magnified during maximal efforts (Bishop *et al*., 2016). Consequently, Exell, Irwin, Gittoes and Kerwin (2012) suggest that asymmetry for each neuromuscular variable should only be deemed significant if the inter-limb variability is larger than intra-limb variation.

5.3.4 Method of Calculating Inter-Limb Asymmetries

The ICC results indicate that LvsR SI was a more reliable method of assessing asymmetry. However, B&A plot analysis shows that when the error in the measures is quantified, there is a considerable difference between the two methods that favours the use of HvsL SI, due to a much lower level of variability within the data. Table 4.10 shows that the mean SI for LvsR is, for all variables, smaller in magnitude than HvsL; this difference is, in the majority of cases, statistically significant. However, the interchangeability of asymmetry between limbs resulted in the nullification of results, which would explain the significant difference between asymmetry calculation methods.

With this, the LvsR SI results show, in particular for F_{max}, that there is little evidence of a 'dominant' limb. However, when assessed using the HvsL SI method, there are in fact a greater magnitude of asymmetries, which demonstrates the proposed variability between legs. The HvsL method cannot determine which of the limbs has produced a greater value, but does allow the quantification of all of the asymmetries present, without risk of one result being nullified, or muted, by another.

This identification of varying limb asymmetries in CMJs potentially calls into question the use of this method of asymmetry assessment, for example, in the evaluation of rehabilitation and recommendation of 'return to play' for individuals post injury or procedure. The results of the current study emphasise the presence of inherent asymmetries in CMJs, however, it also makes it clear that the interpretation of the results of asymmetry calculation must be considered with extreme caution. To fully quantify asymmetry, multiple trials need to be conducted. Then, conclusions or assumptions about the outcome of CMJ asymmetry assessment should only be made when the direction of the asymmetries have been identified. This should be conducted on an individual basis and confirmed to follow the expected outcome, or are, at least, consistent across trials.

The results of the current study, similar to Exell *et al.* (2012), have also identified that the variables displaying considerable asymmetry were not consistent between participants; for example, P5 demonstrated a mean absolute SI of only 2.5 ± 1.3% for F_{max}, while another, P16, generated asymmetry of 14.7 ± 3.8%. Additionally, for J_{con}, P8 showed mean asymmetries of 60.4 ± 17.0% while P7 was much smaller in comparison with 7.8 ± 6.0%. The individual nature of asymmetry has been highlighted by the measurement of substantial asymmetries by one participant but not another, reinforcing the recommendation of assessment on an individual basis.

Although one of the most common methods of quantifying inter-limb asymmetries, the calculation of a SI requires the normalisation of differences to a reference value; the choice of value is typically based upon the study objective and, as outlined by Bishop *et al*. (2016), there are a number of potential approaches. In a clinical setting, when generating a SI between injured and uninjured limbs, it would be logical to use the uninjured limb as the reference. However, in a healthy population, there is no obvious choice and, as such, there are differing recommendations within the literature. The limitation of this in research is that when faced with the situation of no clear reason for choosing a particular reference value, an arbitrarily chosen side can lead to inconsistent results. Alternatively, using the average of the two limbs produces significantly lower SI values (Zifchock *et al.,* 2008). Consequently, it is possible that studies that employ this method could be underestimating asymmetry. Another limitation of the SI is its potential for 'artificial inflation' (Zifchock *et al.,* 2008) as a result of an inappropriate reference value being implemented into the equation. As far as can be ascertained, distinguishing between equations and their suitability has not yet been established and, therefore, it is difficult to completely justify which method should be used over another (Bishop *et al*., 2016).

In addition to the use of a SI, as used in the current study, additional methods of asymmetry quantification have been considered. Proposed by Zifchock *et al.* (2008), the symmetry angle (SA), like a SI, is a measure of the relationship between discrete values obtained from the left and right sides. Zifchock *et al*. (2008) reported that although values of SA tended to be much lower than SI values, they were highly correlated to the widely accepted SI. Consequently, it may be beneficial for future studies of inter-limb symmetry to use SA as a method of asymmetry assessment as it removes the requirement of a reference value and is, therefore, not prone to the problems of overestimation. Additionally, the SA method provides a standard scale (± 100%) to interpret results and provide comparisons.

Despite the very limited research regarding the 'ideal' asymmetry calculation method, existing comments in the literature may indicate that reporting asymmetries using the SA method holds some advantages over other options, primarily its immunity to reference values and inflated scores, which may suggest it is a more robust method for asymmetry detection and has the potential to offer a more consistent and universal approach to asymmetry detection (Bishop *et al.,* 2016). However, although SA allows quantification of asymmetry for discrete kinematic and kinetic values, it does not inherently consider intra-limb variability in its calculation. In addition to this, the study conducted by Zifchock *et al*. (2008) was based on investigations of asymmetry during gait analysis and, therefore, the applicability of this method to vertical jumping methods of inter- and intra-limb neuromuscular asymmetry assessment should first be investigated, before its practicality and suitability is assumed.

Often, cut-off values of 15%, or even 10% have been presented as the point at which asymmetries become associated with injury risk or the criteria for return-to-sport (Ardern, Webster, Taylor & Feller, 2011; Knapik, Bauman, Jones, Harris & Vaughan, 1991; McCormick *et al.,* 2014; Myer, Paterno *et al.,* 2006; Wellsandt, Failla & Snyder-Mackler, 2017). The magnitude of asymmetries reported from the healthy population of the current study greatly exceeded this value, Table 4.10. Therefore, future studies are necessary to verify, firstly, the most appropriate method of asymmetry assessment, and secondly, whether the reported cut-off value is adequate for all neuromuscular variables. Menzel *et al.,* (2013) consider that impulse has a higher dispersion compared with, say, maximal force, which has led to a higher number of identified bilateral differences as a result of the same cut-off value applied to different variables. They suggest that if the dispersion of a variable was considered, this would lead to different cut-off values for distinct variables and assessment methods and may enhance the concordance of the diagnostic information from different methods.

5.4 Differences Between Study Populations

The following section refers to hypothesis 8, "Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population."

The current study consisted of three study populations, female-only (FEM), male-only (MALE) and female and male combined (ALL). In both postural stability and bilateral asymmetry assessments, although differences were present between sexes for individual parameters, all three populations resulted in comparable levels of reliability.

In the systematic review of Ruhe *et al.* (2010), of the studies included in their search, 83% enrolled mixed sex groups, and from these they determined that it was difficult to reach a conclusion regarding the effect of gender on the reliability of CoP measures. Hageman, Leibowitz and Blanke (1995), however, reported that sex was not significant for any outcome measure within their study. Differences between age groups has been shown in CoP analysis (e.g., Hageman *et al.*, 1995; Choy, Brauer & Nitz, 2003) however, these differences are often associated with differences between adolescent and elderly populations. The sex groups in the current study were of similar ages (FEM: 32.4 ± 8.7 yrs., MALE: 32.7 ± 9.5 yrs.) and so the influence of age differences is unlikely to have been a confounding factor.

For the assessment of bilateral asymmetry, the influence of sex differences may be more prominent. The current study demonstrated that the reliability of CMJ variables varied only slightly between sexes, with some variables favouring FEM while others favoured MALE. However, in terms of the performance of neuromuscular variables, significant differences between populations were identified for force generating capacity variables, F_{max}, J_{ecc}, J_{con}, and TO_{vel}. Differences in normalised temporal variables, however, were not significant between populations ($p > 0.05$) on both day 1 and day 2. This was not unexpected as females typically display lesser relative force application abilities during CMJs than males (Heishman *et al.,* 2019). It has also been previously reported that females tend to exhibit greater asymmetries than males during dynamic movements with rapid eccentric decelerations (Pappas & Carpes, 2012). The current study reported higher magnitudes of absolute asymmetry for F_{max}, J_{ecc} and J_{con} in the FEM population compared to MALE (Table 4.10.), however, the reason of this is unclear. Harry *et al.* (2021) propose that inter-limb asymmetries during bilateral CMJs may be 'more severe for females as less relative strength and increased eccentric force application asymmetry might alter intended CMJ techniques and lead to requisite compromises in performance'.

Despite the identified differences between sexes, the results of the pooled group suggest a good enough level of reliability that wouldn't deter the recommendation for pooled groups in future research. The combining of population also allows for a larger sample size in order to generate greater statistical power.

5.5 Limitations of Current Study

5.5.1 Limitations Due to Participants

Due to the difficulty of access to participants, the lower sample size of the current study may have limited the statistical power to show differences between conditions, or epochs, in postural stability assessments or to give the most reliable interpretations of SI calculations in the assessment of bilateral asymmetry. Loken and Gelman (2017) note that greater measurement error leads to greater variation in the measured effect sizes which, particularly for smaller sample sizes, resulting in some measured effect sizes overestimating the true effect size by chance. Additionally, although all participants of the study were an active population who regularly participate in sport, this was mostly only on a recreational basis. In order to assess the suitability of methodology or draw conclusions as to its validity and reliability when applied to other populations, it would be necessary to repeat the research in a more demographically consistent group such as athletes at a specific level of one particular sport.

With respect to the controlling of participants, besides ensuring the trials were conducted at the same time of day to account for circadian or diurnal variability, or 'time-of-day' (Atkinson & Reilly, 1996; Faria & Drummond, 1982; Gribble *et al*., 2007; Hill & Smith, 1991; Melhim, 1993; Wyse *et al.,* 1994), the current study did not control the movements and/ or nutrition of the participants prior to each visit to the lab. As a result, the influence of fatigue or, inversely, increased energy may have impacted each participant's ability to perform on each of the two testing sessions, for example the influence of caffeine ingestion (Grgic, Venier & Mikulic, 2020; Venier, Grgic & Mikulic, 2019; Zbinden-Foncea *et al.,* 2018).

The leg length of participants was not a variable measured and was, therefore, not considered during the selection, or rejection, of participants. In studies assessing bilateral asymmetry, for example Benjanuvatra *et al*. (2013), only participants with inter-limb leg length differences of 2% or less, as determined by the trochanterion-to-ground measure, were included to ensure that asymmetries were not associated with leg length discrepancies. The studies then referenced in this paper, (Perttunen, Anttila, Södergård, Merikanto & Komi, 2004; Song, Halliday & Little, 1997) in addition to many others (e.g., Saaid, 2017) relate to the impact of leg length differences on asymmetrical gait and so may not be as easily applicable to jumping forms of asymmetry analysis. As far as can be ascertained, there is no existing research reporting the impact of limb length discrepancies on bilateral asymmetry within jumping or their correlations.

5.5.2 Limitations Due to Methodology

The current study provided a thorough warm-up in order to prevent any consequences of potentiation, however, it is unclear if this protocol was the most effective as there appear to be no studies reporting on the effect of warm-up on the reliability of jump performance. The consistency of outcome variables across trials in the current study (Appendix L), as opposed to an increase over trials, would suggest the absence of potentiation effect, however, it Is not clear this is necessarily the case without investigating the consistency and time-order effects of each individual participant.

A potential limitation of the study was its use of a self-selected foot placement in the assessment of postural stability; using a pre-determined and set position may have ensured repeatability and consequently improved reliability, but this would have been at the potential cost of validity. Pre-determined positions with specified heel distances and angles are unlikely to represent the habitual foot placement of different individuals (Hébert-losier & Murray, 2020). Investigations into varied foot placements should be made in order to establish its effect on postural stability measures and their reliability. However, for such an investigation to be worthwhile, it would first be necessary to develop a criterion method for postural stability assessment. Due to the uncertainty surrounding foot placement, it was reasonable to use self-selection for foot placement. However, it may have been of benefit to enable a self-selected position that was then traced or measured and recreated to allow for individuality and representation of habitual placement, but also guarantee consistency across trials. Conversely, it has been identified that when habitual placement is measured, individuals naturally place their feet at close to the same distance and angle (N. Owen, personal communication, October 13, 2020).

Data collected for the assessment of postural stability in the current study was measured using a piezoelectric FP. Because the platform uses charge as opposed to voltage, it is not recommended for extended periods of time, due to charge dissipation resulting in consequential drift. Quagliarella et al. (2008) identified drift in a piezoelectric FP to be a linear trend with time predominantly affecting the F_z

component of GRF and F_x component of CoP coordinates. Quagliarella et al. (2008) also reported that drift in a paediatric population should be compensated if the trials were over 60 s. However, in the current study, the population was adult and reliability in the epochs under 60 s and over 60 s were both measured to be reliable.

Statistical analysis techniques may also have limited the validity and reliability of the conclusions made. Firstly, in relation to B&A analysis , without large numbers of results, i.e., a large sample size, there is a very real potential for incorrectly finding a new method, or methodology, to be acceptable as a result of the reported bias and LOA not being truly representative of a population, and for such methods to be recommended for widespread use without justification (Bunce, 2009). Additionally, the results of t-test statistics in the current study should be interpreted with some caution. Generally, the t-tests have shown there to be no significant difference between days, although there were some exceptions. However, one of the conditions that was flagged up by a t-test as significant, when using more sensitive tests and quantifying the differences, these differences are amongst the lowest reported. Therefore, it is important to interpret scores with caution and further investigate measures of agreement. Further to this, the t-tests identified to be significant, or vice versa, could be explained by the probability levels selected. The value used for assessment was $p \leq 0.05$, which suggests that the test would return the 'correct' answer, on average, 95% of the time. The current study conducted over 100 t-tests in the postural stability analysis alone, so it would be reasonable to assume there were some inherent errors; considering the results of B&A analysis, it could be argued that some t-tests may be incorrect. Ultimately, B&A provides a better assessment of agreement as it is providing a quantification of any error as opposed to the use of generalised scores with associated probabilities.

5.5.3 Alternative Method of Postural Stability Assessment

Alongside quiet standing, examples of seated posture and stability assessment are also reported in the literature (Appendix M). The initial population of interest for this research was rowers, consequently, it would have been reasonable to consider assessing balance/ postural stability using body positioning similar to the way in a rowing boat, seated.

With seated posture it is typically far easier to achieve balance due to only half of the body in need of stabilisation and typically a larger base of support. The use of seated posture would likely be a favourable choice in some clinical populations, such as those with lower back pain, or if the lower limbs are compromised in a way that would negatively impact the reliability and validity of standing trials. Or, if standing trials are simply not possible, for example in lower limb amputation or spinal cord injury. It may be favourable in such populations as postural control of the lumbar spine, while seated, is isolated from the control of lower extremity joints (Cholewicki, Polzhofer & Radebold, 2000; Radebold, Cholewicki, Polzhofer & Greene, 2001).

The validity of seated assessment, however, in quantifying postural control in healthy populations is limited due to the numerous additional factors to consider. The validity of the results can be altered depending on the absence, or presence, of a back rest, in addition to, the seated position adopted (anterior, middle, or posterior) and the positioning of the legs. For example, a middle position would cause the CoG to be directly over the ischial tuberosities (IT) and, therefore, the IT would be supporting ~45% of BW (Boggs & Ahmadian, 2007). An anterior, posterior or lateral position, however, would result in the CoG being positioned differently over the IT resulting in other structures, such as the coccyx, lower sacrum and trochanters, supporting more or less body weight, respectively (Boggs & Ahmadian, 2007, Figure 2-2). Researchers can be less certain of the reliability and validity of CoP measures when there are segment masses outside the field of measurement and when the positioning of this mass is going to impact how the CoP responds. Although studies often use foot and leg rests to reduce lower limb movement, when the lower limbs, whether legs or feet, are in contact with the ground or any supporting surface, they are likely to be used to alter weight distribution and are, therefore, going to be exerting a reaction force- introducing another factor that would likely be necessary to measure and quantify, also.

Due to the considerable variation in methodologies and absence of any criteria for valid and reliable measures, seated measures of postural stability would need to be thoroughly piloted before its suitability was assumed.

Chapter 6

Conclusions and Future Directions

6.1 Conclusion

There have been a number of notable findings identified in the results and discussion of this research, however, as outlined in Section 1.4, eight main hypotheses were formed to provide direction for the analysis. The conclusions of each hypothesis are listed below.

1. For hypothesis 1, "There is no criterion method for the determination of postural stability performance, however, a methodology devised from the recommendations of previous research will produce valid and reliable results", results of the current study have identified that for the assessment of postural stability, two out of three of the variables used, L_p and V_m, were better than the other, A_s, in terms of reliability and variability. Consequently, future studies that replicate the current methodology should consider using the variables, L_p and V_m in preference to A_s. Although A_s has been extensively reported in postural stability studies, and, in theory quantifies a variable of interest, consideration should be given to the future use of A_s, due to its poor reliability. Although the current study did not compare different acquisition settings and their consequent reliability, the acquisition settings and protocol used generated reliable results and, in conjunction with the existing literature, would be recommended for future application.

2. The results of this study suggest that hypothesis 2, "As sampling duration increases, so will the test-retest reliability of postural stability parameters derived from CoP analysis" was incorrect. It has been identified that although the longest trial duration, 90 s, demonstrated one of the highest levels of reliability, there was little to no real difference between the reliability of shorter durations in comparison. This would suggest that the current study has not identified an 'optimum' trial duration and development of this hypothesis would require further research.

3. Similarly, the current study cannot provide a recommendation for the optimal number of trial repetitions (as this would depend on study context), but based on cumulative average analysis, a trend towards improved precision and validity is

present with increasing repetitions. As such, hypothesis 3 can be considered to be correct.

4. In regard to hypothesis 4, "Visual condition will have a significant impact on postural stability performance. Further, CoP excursion measures will be significantly higher when visual feedback is removed", with the exception of A_s, the current study reported significant differences between EO and EC conditions, even in the shortest trial durations. Thus, indicating that the effect of lack of vision is identifiable from the offset of postural stability measurement. In all cases of significance, postural stability parameters were higher (suggesting poorer 'performance') in the EC condition. Closer inspection of data has also shown that that despite a difference between EO and EC conditions being observed within the shorter epochs, a longer trial duration is needed to uncover the true magnitude of these differences and quantify the effect of lack of vision on stability measures. Both EO and EC conditions have shown sufficient reliability coefficients and agreement and, as such, the results of this study would support their use in future research.

5. Hypothesis 5 states that "Bilateral CMJ performance, defined using the criterion method for the determination of neuromuscular variables, will show excellent test-retest reliability and day-to-day agreement". Through the assessment of bilateral asymmetry, the current study has identified that the replication of the criterion method for the determination of PPO (Owen *et al.* 2014) has resulted in high test-retest reliability and day to day agreement for F_{max}, J_{ecc}, J_{con}, PPO, V_{to} and t_{ecN} calculated from a bilateral CMJ. However, t_{FmaxN}, was a far less reliable variable as a consequence of the characteristic bimodal feature of a CMJ VGRF-time history, individual jump strategies and pooling of group data. Future studies should investigate a viable method to account for the bimodal feature, in terms of peak force temporal variables.

6. In a bilateral CMJ, the differences in the force generating capacity between limbs does not necessarily determine the variation in the magnitude of VGRF generated during the jump. Instead, variations in VGRF symmetry should be considered to represent bilateral variations in limb loading that stem from the jumping and

compensatory strategies adopted by individuals, i.e., the coordinated action of the lower limbs, pelvis and trunk, and not only the strength and power differences. With that, the first half of hypothesis 6 “In the execution of a bilateral CMJ, there will be differences in the contribution of each individual limb to overall performance” can be considered to be proven true, however, the latter “These variations in the magnitude of vertical component of the ground reaction force are the result of differences in the force generating capacity between limbs” is shown to be incorrect.

7. An important finding of the current study, in the assessment of lower limb bilateral asymmetry, was that LvsR and HvsL methods of SI calculation are significantly different and have the ability to quantify very different inherent characteristics of bilateral CMJ performance. In the calculation of SI between lower limbs, the current study has identified the importance of determining a suitable set of reference values and the importance of considering the directionality of asymmetries on an individual basis. Reliability assessments of the two methods reported that LvsR was not less reliable than HvsL using ICC analysis, however, B&A plots clearly demonstrate a lower variability and smaller dispersion of data in the HvsL calculation method. With respect to the validity of measures, each calculation method has its limitations, as reviewed in Section 5.3, however, when creating conclusions based off of group means, LvsR is not a valid measure of asymmetry magnitude, due to the nullification of opposing asymmetry values from one trial to the next, or one individual to another.

8. Hypothesis 8, “Individual male and female populations will have comparable reliability but, when combined to form a mixed population, there will be a lower level of reliability due to increased heterogeneity of the population”, has shown to be partly disproven by the current study. For both postural stability and bilateral asymmetry assessments, only slight differences between the sexes were found, as first hypothesised. However, the results of the combined population (ALL) were also indicative of similar levels of reliability, suggesting that the grouping of both female and male individuals did not substantially impact the reliability and validity of outcome measures, and, therefore, the latter half of this hypothesis is incorrect, based on the data of this study.

6.2 Recommendations For Future Research

There are a number of potential avenues future research could benefit from investigating in order to consolidate or expand the current understanding of measurement and interpretation of both postural stability quantification and bilateral asymmetry. The main objective of future research should be the determination of reliability and validity of chosen methods before they are applied to different populations and used to form comparisons as a method of reducing the heterogeneity reported in the literature.

The stability of inanimate objects is determined and altered by the height of the CoG. It would be reasonable to propose that the same principle could be applied to postural stability of the human body. Therefore, a recommendation for future research would be to look at the assessment of postural stability parameters with normalisation to, or at least consideration of, the relative height of CoG. It would also be of benefit to future research to quantify the differences observed between variables at different epochs. This approach would help identify an optimal trial duration for postural stability assessment without the need for multiple trials. Deriving smaller epochs from a larger sample will, as best as possible, control for the influence of changing strategies and CoP control mechanisms.

In the research of bilateral asymmetry, it was of primary importance to choose an appropriate SI calculation method for the identification of directional asymmetries i.e., sidedness. Future research should consider the significance of variability, in terms of sidedness, of the 'dominant' leg with regard to movement strategies in uncertain or boundary conditions. It has become clear from the results of the current study that although the performance of an individual in a CMJ, with respect to neuromuscular variables e.g., mechanical power, is highly reliable, the coordinated movement and utilisation of muscles during the jumps can be highly variable. An example of this is illustrated in Figure 6.1, where the force-time history of Jump A shows that the contribution of the left and right leg is highly asymmetric, however Jump 2, from the same participant, is much more symmetric. The previous work of A N Bernstein, as reported by Latash and Latash (1996), has proposed an alternative to the accepted understanding of motor control. Latash & Latash (1996) provides an overview of some of his work and, of particular interest to the current study, the idea of flexible motor strategies and the concept that signals from the central nervous system to muscles will be different for repetitions of the same task. Also of relevance are the principles of dynamic degrees of freedom which allow someone to perform movements at different speeds and against different loads by changing the time pattern of equilibrium. This would characteristically alter the execution of a CMJ in response to altered limb loading, while allowing optimal performance outcomes. The presence of trial-to-trial movement variation is also supported by Davids, Glazier, Araújo and Bartlett (2003). In light of this, future research may benefit from study the of asymmetry and consistency of performance, in terms of neuromuscular variables, to create a picture of the entire coordinated movement.

The current study has shown the inherent variability of asymmetry in a healthy population, so this magnitude of variability needs to be established before conclusions are made about injury risk or consequence due to injury in clinical populations due to the gross misinterpretation of results this would cause. Additionally, research looking at the influence of leg length discrepancies on the direction of inter-limb asymmetries, or at least the correlation between the two measures, would help to determine whether this is an important factor in the acceptance of participants into a study population.

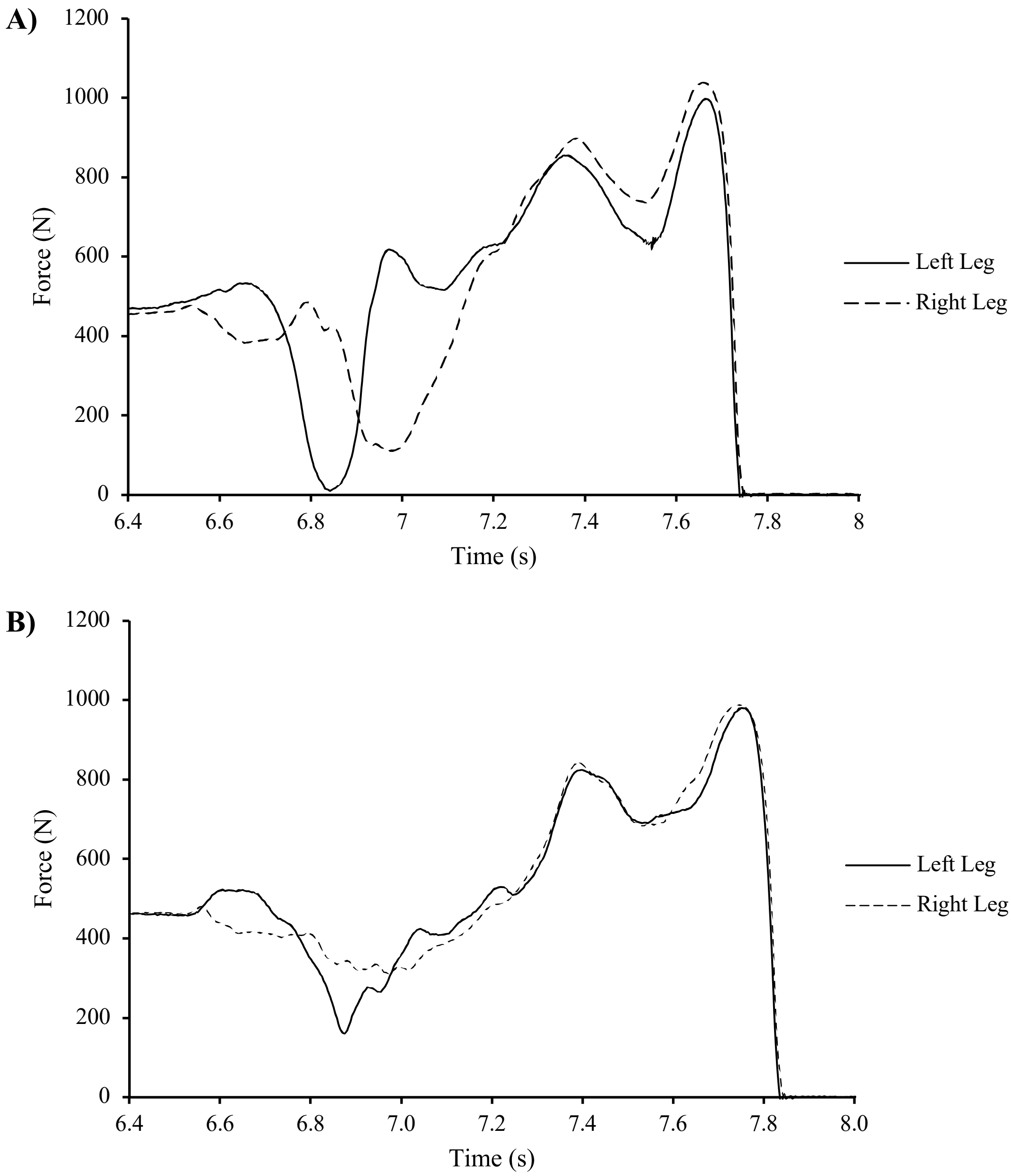

Figure 6.1 *Force-time histories for two countermovement jumps of the same participant that each produced very similar bilateral peak mechanical power. Jump A shows considerably more bilateral variation than Jump B.*

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Appendices

Appendix A: Participant Information Sheet

Swansea University
Prifysgol Abertawe
School of Sport and Exercise Sciences
Ysgol Chwaraeon a Gwyddorau Ymarfer Corff

Participant information sheet

(Version 3, Date: 02 / 06 / 2021)

Jordan Groom
[redacted]

Nick Owen
[redacted]

Investigation of the reliability of postural stability and lower-limb bilateral asymmetry parameters.

This document describes the purpose of the study, and explains what is involved. Please read the information below before deciding whether or not you would like to take part.

Invitation Paragraph
I would like to invite you to take part in a research study that I am conducting for my Postgraduate research project. I am interested in investigating and understanding the relationship between postural stability and bilateral asymmetry parameters. In addition to this, I would like to investigate the reliability of the results produced by the protocol design.
I hope that by developing this insight, we can look to advise in the development of an appropriate and reliable protocol for postural stability assessment and comment on the reliability of the criterion method for countermovement jump assessment when applied to the principle of bilateral asymmetry.

Purpose of Research
The purpose of this study is to determine whether a postural stability protocol designed using the recommendations of previous research is appropriate and yields reliable and valid results. In addition to this, the criterion method for determination of countermovement jump variables is being applied to the assessment of bilateral asymmetry; the reliability of this form of assessment should be assessed in order to advise on its future application.

Why have I been chosen?
You have been chosen to be a part of this study as you are either a staff member/ postgraduate student, or a member of Swansea University Rowing Club. Your participation in this study is voluntary and you have the right to withdraw at any time if you wish.

What will happen to me if I take part?
To take part in this study you will be required to provide your written informed consent (if you choose to do so). You will then first be asked to have your anthropometric measurements taken (such as height and mass) and the height of your centre of gravity will be measured/ calculated. The practical element of the study requires you to complete a series postural stability test- which consists of standing on a force platform

for 100 seconds with both your eyes open and then closed. You will then be asked to complete a minimum of five countermovement jumps on two side-by-side force plates which will be used to assess bilateral asymmetry.
These two protocols will be conducted on two separate occasions in order to comment on inter-session reliability. The completion of these tests should require no more than an hour of your time on each occasion.

What are the possible disadvantages of taking part?
This study requires physical activity (jumping with a suitable warm-up), the risks associated with the study include the resultant fatigue, pain or discomfort. The study involves procedures that you may not be familiar with, however there will be familiarisation periods for both protocols.
In order to participate, the study is going to require around 60 mins of your time on two occasions; this could be a potential disadvantage as it may take time away from your own study time and with the study being held on bay campus it will require travel for those who are not based there.

What are the possible benefits of taking part?
The benefits of taking part in the study are that the greater amount of data we are able to collect regarding this study, the greater understanding we will be able to gain about appropriate methods of postural stability and bilateral asymmetry assessment. Knowledge on this topic could help to further develop assessment techniques.

Will my taking part in the study be kept confidential?
Any information you provide will be kept strictly confidential. Participants' names will not be given in any reports of the findings, and personal information will not be linked in any way to your data. The results of the study will be used to form a conclusion and develop a discussion within my thesis. We will hold any personal data and special categories for a maximum period of 5 years (following Swansea University requirements) after the completion of the research project, as required by Research Councils. Upon completion of the 5-year period or publication of the data, anonymous electronic data files will be deleted and destroyed by the principal researcher (Jordan). Hard copies of informed consent forms will be destroyed using a confidential waste system.

Data Protection and Confidentiality
Your data will be processed in accordance with the Data Protection Act 2018 and the General Data Protection Regulation 2016 (GDPR). All information collected about you will be kept strictly confidential. Your data will only be viewed by the researcher/research team.
All electronic data will be stored on a password-protected computer file kept by the research team. All paper records will be stored in a locked filing cabinet in engineering east of Bay Campus. Your consent information will be kept separately from your responses to minimise risk in the event of a data breach.

Please note that the data we will collect for our study will be made anonymous-anonymization will take place upon your entry into the study, thus it will not be possible to identify and remove your data at a later date, should you decide to withdraw from the study. Therefore, if at the end of this research you decide to have your data withdrawn, please let us know before you leave.

Data Protection Privacy Notice
The data controller for this project will be Swansea University. The University Data Protection Officer provides oversight of university activities involving the processing of personal data, and can be contacted at the Vice Chancellors Office.

Your personal data will be processed for the purposes outlined in this information sheet.
Standard ethical procedures will involve you providing your consent to participate in this study by completing the consent form that has been provided to you.

The legal basis that we will rely on to process your personal data will be processing is necessary for the performance of a task carried out in the public interest. This public interest justification is approved by the College of Engineering Research Ethics Committee, Swansea University.

The legal basis that we will rely on to process special categories of data will be processing is necessary for archiving purposes in the public interest, scientific or historical research purposes or statistical purposes.

What are your rights?
You have a right to access your personal information, to object to the processing of your personal information, to rectify, to erase, to restrict and to port your personal information. Please visit the University Data Protection webpages for further information in relation to your rights. Any requests or objections should be made in writing to the University Data Protection Officer:-

University Compliance Officer (FOI/DP)
Vice-Chancellor's Office
Swansea University
Singleton Park
Swansea
SA2 8PP
Email: dataprotection@swansea.ac.uk

How to make a complaint
If you are unhappy with the way in which your personal data has been processed you may in the first instance contact the University Data Protection Officer using the contact details above.
If you remain dissatisfied then you have the right to apply directly to the Information Commissioner for a decision. The Information Commissioner can be contacted at: -

Information Commissioner's Office,
Wycliffe House,
Water Lane,
Wilmslow,
Cheshire,
SK9 5AF
www.ico.org.uk

What if I have any questions?
If you would like to know any more about the research project, then you can contact the main researcher via email [redacted]

If you have further questions or have concerns/complaints please contact:
Dr Andrew Bloodworth, Chair of the College of Engineering Research Ethics Committee, Swansea University: [redacted]

The institutional contact for reporting cases of research conduct is Registrar & Chief Operating Officer Mr. Andrew Rhodes. Email: researchmisconduct@swansea.ac.uk.

Further details are available at the Swansea University webpages for Research Integrity. http://www.swansea.ac.uk/research/researchintegrity/.

Appendix B: Research protocol

Swansea University
Prifysgol Abertawe
School of Sport and Exercise Sciences
Ysgol Chwaraeon a Gwyddorau Ymarfer Corff

Experimental Protocol

Jordan Groom
[redacted]

Nick Owen
[redacted]

Session 1

- Baseline anthropometrics including height, mass and location of body centre of gravity will be collected.

- Postural Stability Assessment
  - 1 familiarisation trial
  - 4 eyes open and 4 eyes closed 100 s trials (2 mins rest between)

- Bilateral Asymmetry Assessment
  - Warm-up
  - 5 Countermovement jumps on side-by-side force plates

Session 2

- Location of Centre of gravity

- Postural Stability Assessment
  - 1 familiarisation trial
  - 4 eyes open and 4 eyes closed 100 s trials (2 mins rest between)

- Bilateral Asymmetry Assessment
  - Warm-up
  - 5 Countermovement jumps on side-by-side force plates

Appendix C: Participant Consent Form

Swansea University
Prifysgol Abertawe
School of Sport and Exercise Sciences
Ysgol Chwaraeon a Gwyddorau Ymarfer Corff

Participant Consent Form

(Version 2, Date: 12 / 04 / 2021)

Jordan Groom
[redacted]

Nick Owen
[redacted]

Project Title:

Investigation of the reliability of postural stability and lower-limb bilateral asymmetry parameters.

Please initial box

1. I confirm that I have read and understood the information sheet dated 02/06/21 (version number 3) for the above study and have had the opportunity to ask questions. ☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my medical care or legal rights being affected. ☐
3. I understand that sections of any of data obtained may be looked at by responsible individuals from the Swansea University or from regulatory authorities where it is relevant to my taking part in research. I give permission for these individuals to have access to these records. ☐

1. I understand that data I provide may be used in reports and academic publications in anonymous fashion ☐
2. I agree to take part in the above study. ☐

Name of Participant	Date	Signature
Name of Person taking consent.	Date	Signature
Researcher	Date	Signature

Appendix D. Participant PAR-Q

Swansea University
Prifysgol Abertawe
School of Sport and Exercise Sciences
Ysgol Chwaraeon a Gwyddorau Ymarfer Corff

Participant PAR-Q

(Version 2, Date: 12 / 04 / 2021)

Jordan Groom
[redacted]

Nick Owen
[redacted]

Years Rowing experience:

………………

Name .... Ref. No. ....

Date of Birth .... Age: ....

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

Please tick appropriate box	YES	NO
Has the test procedure been fully explained to you?	☐	☐

Any information contained herein will be treated as confidential

Participant Health and Wellbeing

	YES	NO
Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?	☐	☐
Do you feel pain in your chest when you do physical activity?	☐	☐
In the past month, have you had chest pain when you were not doing physical activity?	☐	☐
Do you lose your balance because of dizziness or do you ever lose consciousness?	☐	☐
Do you have a bone or joint problem that could be made worse by a change in your physical activity?	☐	☐
Is your doctor currently prescribing drugs for your blood pressure or heart condition?	☐	☐
Do you know of any other reasons why you should not undergo physical activity? This might include severe asthma, diabetes, a recent sports injury, or serious illness.	☐	☐

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

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

Date ..../….........../……….....

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

• If you have answered YES to one or more questions:
Talk with your doctor in person discussing with him/her those questions you answered yes. Ask your doctor if you are able to conduct the physical activity requirements.

Doctor’s signature .... Date ....../....../.......

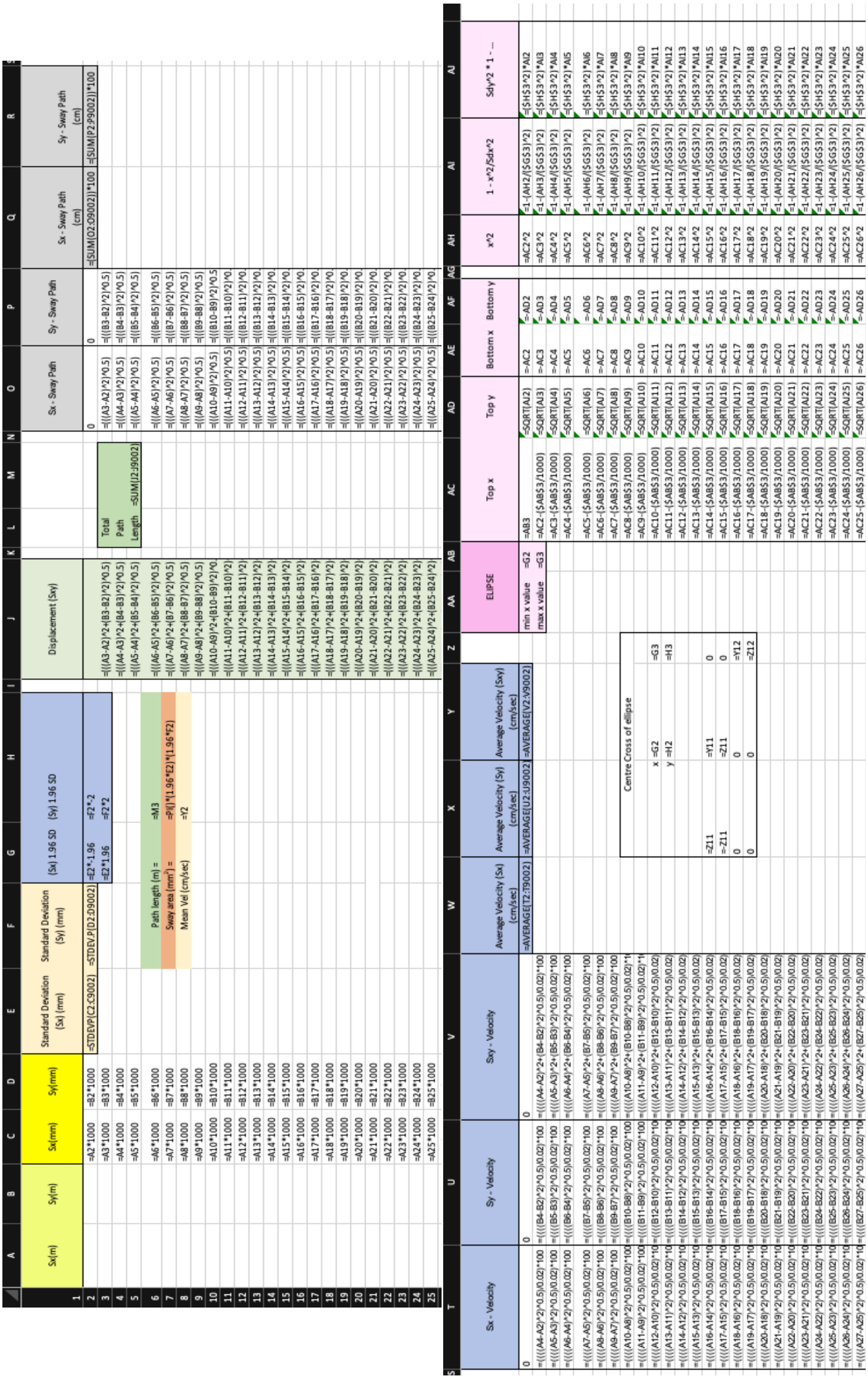
Signature of Experimenter.. Date ....../....../....

Prior to each individual element of the study/ each visit to the laboratory you will be asked to complete a COVID-19 questionnaire to ensure the safety of yourself and others.

Appendix E: Randomised and counter-balanced trial order for individual participants for postural control data collection. C = eyes closed; O = eyes open.

Participant	Day 1									Day 2							
1	C	C	O	O	O	C	O	C		O	O	O	C	O	C	C	C
2	O	O	O	C	C	O	C	C		O	O	O	C	C	O	C	C
3	C	C	O	O	O	C	C	O		C	O	O	O	C	O	C	C
4	C	O	O	C	O	C	C	O		C	O	C	C	O	O	C	O
5	C	C	C	O	C	O	O	O		C	C	O	C	O	O	C	O
6	C	O	O	C	O	O	C	C		C	O	C	O	O	C	O	C
7	C	C	C	O	O	C	O	O		C	O	C	C	O	O	O	C
8	O	C	O	C	O	C	C	O		O	O	O	C	O	C	C	C
9	O	O	C	O	C	O	C	C		O	C	O	C	O	C	C	O
10	O	C	O	O	C	O	C	C		O	C	C	O	O	C	C	O
11	C	O	O	C	O	C	O	C		O	O	C	C	O	O	C	C
12	O	C	O	C	O	C	C	O		C	O	O	C	O	O	C	C
13	C	C	C	O	O	O	C	O		O	C	O	O	O	C	C	C
14	C	O	O	C	O	C	C	O		C	O	C	O	C	O	C	O
15	O	C	O	C	O	C	O	C		O	O	C	C	C	C	O	O
16	C	O	C	C	O	O	C	O		O	C	O	C	O	O	C	C
17	C	O	C	O	C	O	C	O		O	C	O	C	O	C	O	C
18	O	C	C	O	C	O	O	C		O	C	O	O	O	C	C	C
19	O	O	C	C	C	O	C	O		O	C	C	O	O	O	C	C
20	C	O	C	C	C	O	O	O		O	C	O	O	C	C	C	O

Appendix F: Custom Spreadsheet for the calculation of Postural Stability measures. This spreadsheet is applicable to the 90 s epoch.



	A	B	C	D	E	F	G	H	J	L	M	O	P	Q	R
1	Sx(m)	Sy(m)	Sx(mm)	Sy(mm)	Standard Deviation (Sx) (mm)	Standard Deviation (Sy) (mm)	(Sx) 1.96 SD	(Sy) 1.96 SD	Displacement (Sxy)			Sx - Sway Path	Sy - Sway Path	Sx - Sway Path (cm)	Sy - Sway Path (cm)
2			=A2*1000	=B2*1000	=STDEVP(C2:C9002)	=STDEV.P(D2:D9002)	=E2*-1.96	=F2*-2				0	0	=(SUM(O2:O9002))*100	=(SUM(P2:P9002))*100
3			=A3*1000	=B3*1000			=E2*1.96	=F2*2	=(((A3-A2)^2+(B3-B2)^2)^0.5)	Total		=(((A3-A2)^2)^0.5)	=(((B3-B2)^2)^0.5)		
4			=A4*1000	=B4*1000					=(((A4-A3)^2+(B4-B3)^2)^0.5)	Path		=(((A4-A3)^2)^0.5)	=(((B4-B3)^2)^0.5)		
5			=A5*1000	=B5*1000					=(((A5-A4)^2+(B5-B4)^2)^0.5)	Length	=SUM(J2:J9002)	=(((A5-A4)^2)^0.5)	=(((B5-B4)^2)^0.5)		
6			=A6*1000	=B6*1000		Path length (m) =		=M3	=(((A6-A5)^2+(B6-B5)^2)^0.5)			=(((A6-A5)^2)^0.5)	=(((B6-B5)^2)^0.5)		
7			=A7*1000	=B7*1000		Sway area (mm²) =		=PI()*(1.96*E2)*(1.96*F2)	=(((A7-A6)^2+(B7-B6)^2)^0.5)			=(((A7-A6)^2)^0.5)	=(((B7-B6)^2)^0.5)		
8			=A8*1000	=B8*1000		Mean Vel (cm/sec)		=Y2	=(((A8-A7)^2+(B8-B7)^2)^0.5)			=(((A8-A7)^2)^0.5)	=(((B8-B7)^2)^0.5)		
9			=A9*1000	=B9*1000					=(((A9-A8)^2+(B9-B8)^2)^0.5)			=(((A9-A8)^2)^0.5)	=(((B9-B8)^2)^0.5)		
10			=A10*1000	=B10*1000					=(((A10-A9)^2+(B10-B9)^2)^0.			=(((A10-A9)^2)^0.5)	=(((B10-B9)^2)^0.5		
11			=A11*1000	=B11*1000					=(((A11-A10)^2+(B11-B10)^2)			=(((A11-A10)^2)^0.5)	=(((B11-B10)^2)^0.		
12			=A12*1000	=B12*1000					=(((A12-A11)^2+(B12-B11)^2)			=(((A12-A11)^2)^0.5)	=(((B12-B11)^2)^0.		
13			=A13*1000	=B13*1000					=(((A13-A12)^2+(B13-B12)^2)			=(((A13-A12)^2)^0.5)	=(((B13-B12)^2)^0.		
14			=A14*1000	=B14*1000					=(((A14-A13)^2+(B14-B13)^2)			=(((A14-A13)^2)^0.5)	=(((B14-B13)^2)^0.		
15			=A15*1000	=B15*1000					=(((A15-A14)^2+(B15-B14)^2)			=(((A15-A14)^2)^0.5)	=(((B15-B14)^2)^0.		
16			=A16*1000	=B16*1000					=(((A16-A15)^2+(B16-B15)^2)			=(((A16-A15)^2)^0.5)	=(((B16-B15)^2)^0.		
17			=A17*1000	=B17*1000					=(((A17-A16)^2+(B17-B16)^2)			=(((A17-A16)^2)^0.5)	=(((B17-B16)^2)^0.		
18			=A18*1000	=B18*1000					=(((A18-A17)^2+(B18-B17)^2)			=(((A18-A17)^2)^0.5)	=(((B18-B17)^2)^0.		
19			=A19*1000	=B19*1000					=(((A19-A18)^2+(B19-B18)^2)			=(((A19-A18)^2)^0.5)	=(((B19-B18)^2)^0.		
20			=A20*1000	=B20*1000					=(((A20-A19)^2+(B20-B19)^2)			=(((A20-A19)^2)^0.5)	=(((B20-B19)^2)^0.		
21			=A21*1000	=B21*1000					=(((A21-A20)^2+(B21-B20)^2)			=(((A21-A20)^2)^0.5)	=(((B21-B20)^2)^0.		
22			=A22*1000	=B22*1000					=(((A22-A21)^2+(B22-B21)^2)			=(((A22-A21)^2)^0.5)	=(((B22-B21)^2)^0.		
23			=A23*1000	=B23*1000					=(((A23-A22)^2+(B23-B22)^2)			=(((A23-A22)^2)^0.5)	=(((B23-B22)^2)^0.		
24			=A24*1000	=B24*1000					=(((A24-A23)^2+(B24-B23)^2)			=(((A24-A23)^2)^0.5)	=(((B24-B23)^2)^0.		
25			=A25*1000	=B25*1000					=(((A25-A24)^2+(B25-B24)^2)			=(((A25-A24)^2)^0.5)	=(((B25-B24)^2)^0.		

T	U	V	W	X	Y	Z	AA	AB	AC	AD	AE	AF	AH	AI	AJ
Sx - Velocity	Sy - Velocity	Sxy - Velocity	Average Velocity (Sx) (cm/sec)	Average Velocity (Sy) (cm/sec)	Average Velocity (Sxy) (cm/sec)		ELIPSE		Top x	Top y	Bottom x	Bottom y	x^2	1 - x^2/Sdx^2	Sdy^2 * 1 - ...
0	0	0	=AVERAGE(T2:T9002)	=AVERAGE(U2:U9002)	=AVERAGE(V2:V9002)		min x value	=G2	=AB3	=SQRT(AJ2)	=-AC2	=-AD2	=AC2^2	=1-(AH2/($G$3)^2)	=($H$3^2)*AI2
=((((A4-A2)^2)^0.5)/0.02)*100	=((((B4-B2)^2)^0.5)/0.02)*100	=((((A4-A2)^2+(B4-B2)^2)^0.5)/0.02)*100					max x value	=G3	=AC2-($AB$3/1000)	=SQRT(AJ3)	=-AC3	=-AD3	=AC3^2	=1-(AH3/($G$3)^2)	=($H$3^2)*AI3
=((((A5-A3)^2)^0.5)/0.02)*100	=((((B5-B3)^2)^0.5)/0.02)*100	=((((A5-A3)^2+(B5-B3)^2)^0.5)/0.02)*100							=AC3-($AB$3/1000)	=SQRT(AJ4)	=-AC4	=-AD4	=AC4^2	=1-(AH4/($G$3)^2)	=($H$3^2)*AI4
=((((A6-A4)^2)^0.5)/0.02)*100	=((((B6-B4)^2)^0.5)/0.02)*100	=((((A6-A4)^2+(B6-B4)^2)^0.5)/0.02)*100							=AC4-($AB$3/1000)	=SQRT(AJ5)	=-AC5	=-AD5	=AC5^2	=1-(AH5/($G$3)^2)	=($H$3^2)*AI5
=((((A7-A5)^2)^0.5)/0.02)*100	=((((B7-B5)^2)^0.5)/0.02)*100	=((((A7-A5)^2+(B7-B5)^2)^0.5)/0.02)*100							=AC5-($AB$3/1000)	=SQRT(AJ6)	=-AC6	=-AD6	=AC6^2	=1-(AH6/($G$3)^2)	=($H$3^2)*AI6
=((((A8-A6)^2)^0.5)/0.02)*100	=((((B8-B6)^2)^0.5)/0.02)*100	=((((A8-A6)^2+(B8-B6)^2)^0.5)/0.02)*100							=AC6-($AB$3/1000)	=SQRT(AJ7)	=-AC7	=-AD7	=AC7^2	=1-(AH7/($G$3)^2)	=($H$3^2)*AI7
=((((A9-A7)^2)^0.5)/0.02)*100	=((((B9-B7)^2)^0.5)/0.02)*100	=((((A9-A7)^2+(B9-B7)^2)^0.5)/0.02)*100							=AC7-($AB$3/1000)	=SQRT(AJ8)	=-AC8	=-AD8	=AC8^2	=1-(AH8/($G$3)^2)	=($H$3^2)*AI8
=((((A10-A8)^2)^0.5)/0.02)*100	=((((B10-B8)^2)^0.5)/0.02)*100	=((((A10-A8)^2+(B10-B8)^2)^0.5)/0.02)*1		Centre Cross of ellipse					=AC8-($AB$3/1000)	=SQRT(AJ9)	=-AC9	=-AD9	=AC9^2	=1-(AH9/($G$3)^2)	=($H$3^2)*AI9
=((((A11-A9)^2)^0.5)/0.02)*100	=((((B11-B9)^2)^0.5)/0.02)*100	=((((A11-A9)^2+(B11-B9)^2)^0.5)/0.02)*1							=AC9-($AB$3/1000)	=SQRT(AJ10)	=-AC10	=-AD10	=AC10^2	=1-(AH10/($G$3)^2)	=($H$3^2)*AI10
=((((A12-A10)^2)^0.5)/0.02)*10	=((((B12-B10)^2)^0.5)/0.02)*10	=((((A12-A10)^2+(B12-B10)^2)^0.5)/0.02)		x =G2		=G3			=AC10-($AB$3/1000)	=SQRT(AJ11)	=-AC11	=-AD11	=AC11^2	=1-(AH11/($G$3)^2)	=($H$3^2)*AI11
=((((A13-A11)^2)^0.5)/0.02)*10	=((((B13-B11)^2)^0.5)/0.02)*10	=((((A13-A11)^2+(B13-B11)^2)^0.5)/0.02)		y =H2		=H3			=AC11-($AB$3/1000)	=SQRT(AJ12)	=-AC12	=-AD12	=AC12^2	=1-(AH12/($G$3)^2)	=($H$3^2)*AI12
=((((A14-A12)^2)^0.5)/0.02)*10	=((((B14-B12)^2)^0.5)/0.02)*10	=((((A14-A12)^2+(B14-B12)^2)^0.5)/0.02)							=AC12-($AB$3/1000)	=SQRT(AJ13)	=-AC13	=-AD13	=AC13^2	=1-(AH13/($G$3)^2)	=($H$3^2)*AI13
=((((A15-A13)^2)^0.5)/0.02)*10	=((((B15-B13)^2)^0.5)/0.02)*10	=((((A15-A13)^2+(B15-B13)^2)^0.5)/0.02)							=AC13-($AB$3/1000)	=SQRT(AJ14)	=-AC14	=-AD14	=AC14^2	=1-(AH14/($G$3)^2)	=($H$3^2)*AI14
=((((A16-A14)^2)^0.5)/0.02)*10	=((((B16-B14)^2)^0.5)/0.02)*10	=((((A16-A14)^2+(B16-B14)^2)^0.5)/0.02)		=Z11	=Y11	0			=AC14-($AB$3/1000)	=SQRT(AJ15)	=-AC15	=-AD15	=AC15^2	=1-(AH15/($G$3)^2)	=($H$3^2)*AI15
=((((A17-A15)^2)^0.5)/0.02)*10	=((((B17-B15)^2)^0.5)/0.02)*10	=((((A17-A15)^2+(B17-B15)^2)^0.5)/0.02)		=-Z11	=Z11	0			=AC15-($AB$3/1000)	=SQRT(AJ16)	=-AC16	=-AD16	=AC16^2	=1-(AH16/($G$3)^2)	=($H$3^2)*AI16
=((((A18-A16)^2)^0.5)/0.02)*10	=((((B18-B16)^2)^0.5)/0.02)*10	=((((A18-A16)^2+(B18-B16)^2)^0.5)/0.02)		0	0	=Y12			=AC16-($AB$3/1000)	=SQRT(AJ17)	=-AC17	=-AD17	=AC17^2	=1-(AH17/($G$3)^2)	=($H$3^2)*AI17
=((((A19-A17)^2)^0.5)/0.02)*10	=((((B19-B17)^2)^0.5)/0.02)*10	=((((A19-A17)^2+(B19-B17)^2)^0.5)/0.02)		0	0	=Z12			=AC17-($AB$3/1000)	=SQRT(AJ18)	=-AC18	=-AD18	=AC18^2	=1-(AH18/($G$3)^2)	=($H$3^2)*AI18
=((((A20-A18)^2)^0.5)/0.02)*10	=((((B20-B18)^2)^0.5)/0.02)*10	=((((A20-A18)^2+(B20-B18)^2)^0.5)/0.02)							=AC18-($AB$3/1000)	=SQRT(AJ19)	=-AC19	=-AD19	=AC19^2	=1-(AH19/($G$3)^2)	=($H$3^2)*AI19
=((((A21-A19)^2)^0.5)/0.02)*10	=((((B21-B19)^2)^0.5)/0.02)*10	=((((A21-A19)^2+(B21-B19)^2)^0.5)/0.02)							=AC19-($AB$3/1000)	=SQRT(AJ20)	=-AC20	=-AD20	=AC20^2	=1-(AH20/($G$3)^2)	=($H$3^2)*AI20
=((((A22-A20)^2)^0.5)/0.02)*10	=((((B22-B20)^2)^0.5)/0.02)*10	=((((A22-A20)^2+(B22-B20)^2)^0.5)/0.02)							=AC20-($AB$3/1000)	=SQRT(AJ21)	=-AC21	=-AD21	=AC21^2	=1-(AH21/($G$3)^2)	=($H$3^2)*AI21
=((((A23-A21)^2)^0.5)/0.02)*10	=((((B23-B21)^2)^0.5)/0.02)*10	=((((A23-A21)^2+(B23-B21)^2)^0.5)/0.02)							=AC21-($AB$3/1000)	=SQRT(AJ22)	=-AC22	=-AD22	=AC22^2	=1-(AH22/($G$3)^2)	=($H$3^2)*AI22
=((((A24-A22)^2)^0.5)/0.02)*10	=((((B24-B22)^2)^0.5)/0.02)*10	=((((A24-A22)^2+(B24-B22)^2)^0.5)/0.02)							=AC22-($AB$3/1000)	=SQRT(AJ23)	=-AC23	=-AD23	=AC23^2	=1-(AH23/($G$3)^2)	=($H$3^2)*AI23
=((((A25-A23)^2)^0.5)/0.02)*10	=((((B25-B23)^2)^0.5)/0.02)*10	=((((A25-A23)^2+(B25-B23)^2)^0.5)/0.02)							=AC23-($AB$3/1000)	=SQRT(AJ24)	=-AC24	=-AD24	=AC24^2	=1-(AH24/($G$3)^2)	=($H$3^2)*AI24
=((((A26-A24)^2)^0.5)/0.02)*10	=((((B26-B24)^2)^0.5)/0.02)*10	=((((A26-A24)^2+(B26-B24)^2)^0.5)/0.02)							=AC24-($AB$3/1000)	=SQRT(AJ25)	=-AC25	=-AD25	=AC25^2	=1-(AH25/($G$3)^2)	=($H$3^2)*AI25
=((((A27-A25)^2)^0.5)/0.02)*10	=((((B27-B25)^2)^0.5)/0.02)*10	=((((A27-A25)^2+(B27-B25)^2)^0.5)/0.02)							=AC25-($AB$3/1000)	=SQRT(AJ26)	=-AC26	=-AD26	=AC26^2	=1-(AH26/($G$3)^2)	=($H$3^2)*AI26

Appendix G: Pilot study for the determination of instant of take-off.

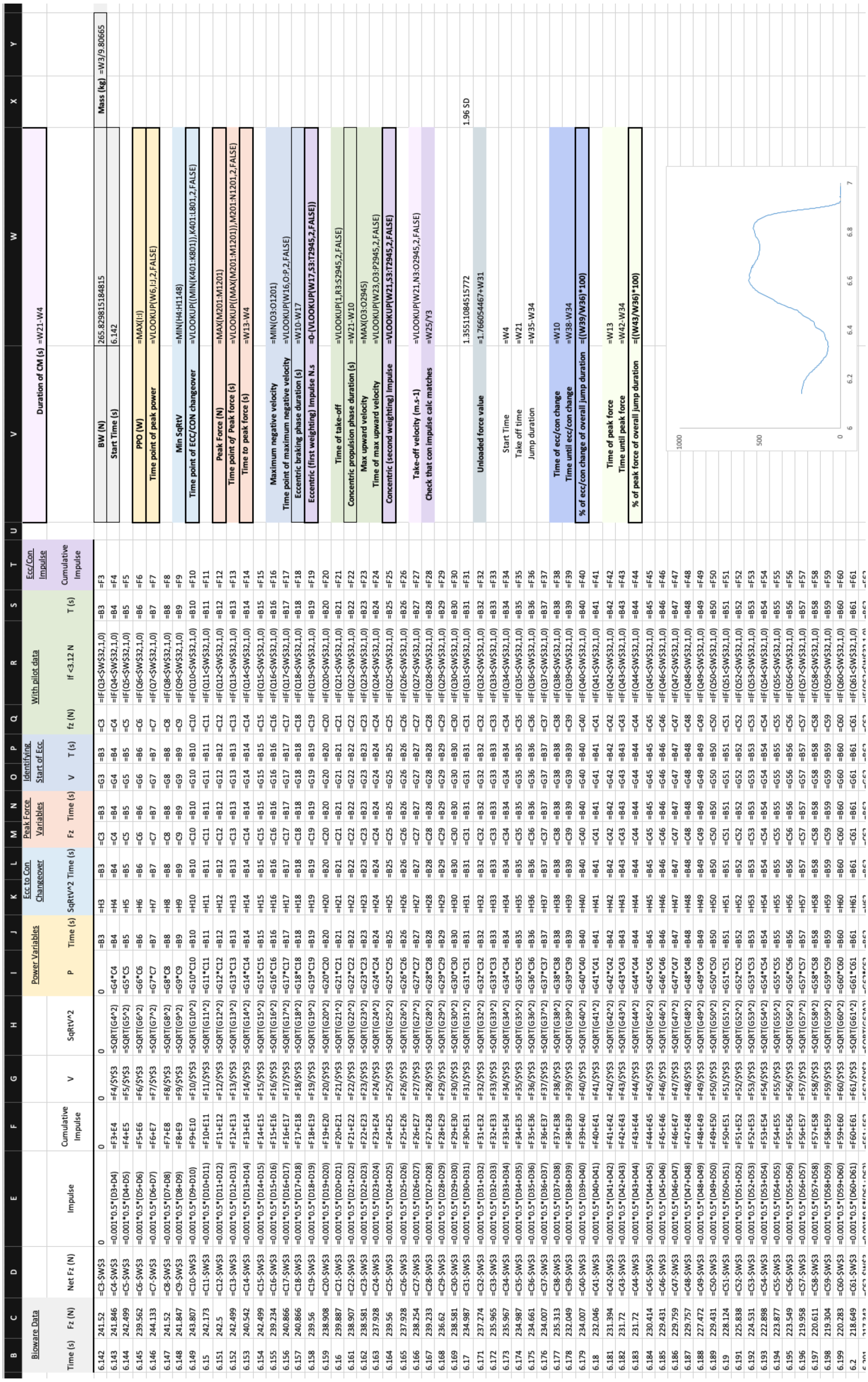
Pilot study investigating the mean unloaded force/ system noise exhibited by the force platform during the flight phase of 50 CMJs. The results of this pilot study are provided in Table G.1 The mean and standard deviation (SD) were then calculated, and the value of mean ± 1.96 SD for the 50 jumps was applied to the current study. This definition of take-off (t_{to}) was tested on the 50 jumps and, from close visual inspection alone, appeared to be a strong identifier of the instant of t_{to}. Within the current study, the instant of t_{to} was identified using IF and VLOOKUP functions in Excel (to identify the values below 3.12 N and their corresponding time intervals).

Table G.1. *Results of Pilot study investigating force platform noise (mean unloaded force) during the flight phase of a countermovement jump in order to help identify a method for defining take-off.*

Jump No.	Mean unloaded force (N)	Jump No.	Mean unloaded force (N)
1	1.49	26	2.24
2	2.43	27	0.07
3	1.73	28	2.42
4	2.31	29	2.55
5	1.65	30	0.37
6	1.00	31	1.74
7	1.45	32	2.05
8	1.04	33	0.85
9	1.01	34	1.01
10	1.85	35	0.48
11	1.05	36	2.15
12	1.09	37	2.01
13	2.66	38	1.45
14	2.09	39	3.02
15	1.89	40	1.11
16	2.90	41	1.87
17	1.15	42	2.67
18	0.65	43	2.18
19	2.27	44	1.51
20	1.93	45	2.22
21	2.10	46	2.68
22	2.36	47	2.56
23	1.53	48	1.50
24	1.97	49	1.18
25	2.61	50	2.20

Mean ± SD of 50 jumps	1.766 ± 0.691
1.96 SD	1.355 N
Mean + 1.96 SD	3.121 N

Appendix H: Custom spreadsheet for assessment of Neuromuscular performance variables.



Appendix I: Custom Spreadsheet for the assessment of countermovement jump performance and bilateral asymmetry. Formula for the calculation of neuromuscular variables.

BW (N)	265.829815184815	Mass (kg)	=W3/9.80665
Start Time (s)	6.142		
PPO (W)	=MAX(I:I)		
Time point of peak power	=VLOOKUP(W6,I:J,2,FALSE)		
Min SqRtV	=MIN(H4:H1148)		
Time point of ECC/CON changeover	=VLOOKUP((MIN(K401:K801)),K401:L801,2,FALSE)		
Peak Force (N)	=MAX(M201:M1201)		
Time point *of* Peak force (s)	=VLOOKUP((MAX(M201:M1201)),M201:N1201,2,FALSE)		
Time *to* peak force (s)	=W13-W4		
Maximum negative velocity	=MIN(O3:O1201)		
Time point of maximum negative velocity	=VLOOKUP(W16,O:P,2,FALSE)		
Eccentric braking phase duration (s)	=W10-W17		
Eccentric (first weighting) Impulse N.s	=0-(VLOOKUP(W17,S3:T2945,2,FALSE))		
Time of take-off	=VLOOKUP(1,R3:S2945,2,FALSE)		
Concentric propulsion phase duration (s)	=W21-W10		
Max upward velocity	=MAX(O3:O2945)		
Time of max upward velocity	=VLOOKUP(W23,O3:P2945,2,FALSE)		
Concentric (second weighting) Impulse	=VLOOKUP(W21,S3:T2945,2,FALSE)		
Take-off velocity (m.s-1)	=VLOOKUP(W21,N3:O2945,2,FALSE)		
Check that con impulse calc matches	=W25/Y3		
	1.35511084515772	1.96 SD	
Unloaded force value	=1.766054467+W31		
Start Time	=W4		
Take off time	=W21		
Jump duration	=W35-W34		
Time of ecc/con change	=W10		
Time until ecc/con change	=W38-W34		
% of ecc/con change of overall jump duration	=((W39/W36)*100)		
Time of peak force	=W13		
Time until peak force	=W42-W34		
% of peak force of overall jump duration	=((W43/W36)*100)		

Appendix J: Bland and Altman analysis for postural stability variables in all populations. The bias and LOA are reported as raw values and as a percentage of the mean.

			Path Length				Sway Area				Mean Velocity			
			Eyes Open		Eyes Closed		Eyes Open		Eyes Closed		Eyes Open		Eyes Closed	
			Raw	%	Raw	%	Raw	%	Raw	%	Raw	%	Raw	%
All Participants	1st 30 s	Bias	0.00	0.3	0.01	2.4	24.67	31.9	13.28	10.9	0.00	0.4	0.02	2.4
		LOA	0.07	25.8	0.06	15.7	45.29	57.1	58.22	57.8	0.22	25.9	0.20	15.7
	2nd 30 s	Bias	0.00	1.4	0.00	0.9	-4.37	-1.6	-2.54	-0.7	0.01	1.4	0.01	1.0
		LOA	0.06	25.1	0.07	22.0	36.88	56.5	45.88	66.9	0.19	25.0	0.23	22.1
	3rd 30 s	Bias	0.00	1.1	0.01	4.8	0.31	4.1	19.02	24.8	0.01	1.2	0.05	4.8
		LOA	0.05	19.7	0.06	17.6	40.03	77.6	53.24	63.6	0.16	19.8	0.18	17.7
	1st 60 s	Bias	0.00	0.8	0.01	1.7	-7.08	-0.9	3.31	-0.7	0.01	0.8	-0.01	-1.7
		LOA	0.10	21.6	0.12	16.9	67.35	62.0	63.49	54.1	0.17	21.6	0.20	16.9
	2nd 60 s	Bias	0.00	1.2	0.02	2.9	-6.60	-1.3	5.76	7.9	0.01	1.3	0.03	2.9
		LOA	0.10	20.3	0.11	17.2	53.30	63.1	44.86	48.7	0.16	20.3	0.18	17.4
	90 s	Bias	0.01	0.9	0.02	2.2	-11.77	-6.1	4.26	-0.4	0.01	1.0	0.02	2.3
		LOA	0.14	19.1	0.16	16.2	59.27	52.4	49.86	49.1	0.16	19.1	0.17	16.3
Female	1st 30 s	Bias	-0.01	-5.4	0.01	4.4	17.83	25.6	18.90	17.2	-0.04	-5.4	0.04	4.4
		LOA	0.05	21.7	0.03	11.3	38.09	53.7	61.84	60.8	0.16	21.8	0.11	11.4
	2nd 30 s	Bias	0.00	1.9	0.01	2.8	-5.30	-1.4	-9.72	-10.4	0.01	2.0	0.03	2.8
		LOA	0.05	27.0	0.06	21.8	37.79	61.7	37.41	53.8	0.18	27.0	0.20	22.0
	3rd 30 s	Bias	0.00	-0.2	0.02	7.6	-0.92	5.7	27.33	38.8	0.00	-0.2	0.07	7.7
		LOA	0.03	15.1	0.05	18.1	46.26	82.7	60.18	67.6	0.10	15.1	0.18	18.3
	1st 60 s	Bias	-0.01	-1.9	0.02	3.5	-11.10	-2.4	8.61	7.8	-0.01	-1.9	-0.04	-3.6
		LOA	0.09	21.5	0.08	13.9	78.79	66.4	45.75	37.7	0.15	21.5	0.14	14.0
	2nd 60 s	Bias	0.00	0.9	0.03	5.2	-4.79	3.3	9.42	11.3	0.01	0.9	0.05	5.2
		LOA	0.08	19.7	0.10	17.8	58.01	76.6	46.66	47.3	0.13	19.6	0.16	17.9
	90 s	Bias	-0.01	-1.4	0.04	4.6	-17.98	-10.3	10.94	8.9	-0.01	-1.3	0.05	4.7
		LOA	0.11	17.7	0.12	14.0	64.75	64.4	23.75	15.1	0.12	17.6	0.13	14.1
Male	1st 30 s	Bias	0.01	5.4	0.00	0.6	30.82	37.5	8.22	5.3	0.05	5.5	0.00	0.7
		LOA	0.07	25.1	0.07	18.0	47.80	57.8	52.84	52.5	0.24	25.2	0.25	18.1
	2nd 30 s	Bias	0.00	0.8	-0.01	-0.7	-3.53	-1.8	3.92	8.1	0.00	0.9	-0.02	-0.7
		LOA	0.06	23.1	0.08	21.6	35.96	51.5	49.00	72.7	0.19	23.1	0.25	21.8
	3rd 30 s	Bias	0.01	2.4	0.01	2.2	1.42	2.7	11.54	12.2	0.02	2.4	0.02	2.3
		LOA	0.06	22.9	0.05	15.5	33.31	72.6	40.90	47.9	0.19	22.9	0.18	15.6
	1st 60 s	Bias	0.01	3.2	-0.01	0.0	-3.45	0.3	-1.45	-8.4	0.02	3.3	0.01	0.0
		LOA	0.11	20.6	0.14	18.5	54.08	57.6	74.78	61.8	0.18	20.6	0.23	18.6
	2nd 60 s	Bias	0.01	1.6	0.00	0.8	-8.22	-5.4	2.46	4.9	0.01	1.6	0.00	0.8
		LOA	0.11	20.8	0.11	15.6	48.44	46.4	42.14	49.2	0.18	20.8	0.18	15.8
	90 s	Bias	0.02	3.0	0.00	0.0	-6.18	-2.4	-1.74	-8.7	0.02	3.1	0.00	0.1
		LOA	0.16	19.4	0.18	16.8	51.45	37.0	62.64	61.8	0.18	19.4	0.19	16.9

Appendix K: Results of sequential averaging analysis for postural stability measures providing the standard error of the mean (SEM), as a percentage of the mean, as the number of trials included in analysis increases. [9 tables]

Table K.1. *Sequential averaging analysis for variable Path Length (m) in the ALL population.*

EPOCH	No. of trials incl.	Eyes Open Mean	Eyes Open SEM	Eyes Open SEM % of Mean	Eyes Closed Mean	Eyes Closed SEM	Eyes Closed SEM % of Mean
1st 30 s	1	0.259	0.072	27.8	0.353	0.110	31.1
	2	0.251	0.049	19.5	0.338	0.077	22.7
	3	0.249	0.039	15.7	0.334	0.061	18.3
	4	0.245	0.033	13.6	0.334	0.053	15.8
	5	0.248	0.030	12.0	0.338	0.051	15.2
	6	0.248	0.028	11.1	0.338	0.048	14.1
	7	0.246	0.025	10.2	0.335	0.043	13.0
	8	0.245	0.023	9.5	0.331	0.040	12.0
2nd 30 s	1	0.229	0.067	29.1	0.300	0.114	37.8
	2	0.226	0.045	20.0	0.290	0.074	25.5
	3	0.221	0.034	15.5	0.291	0.057	19.5
	4	0.220	0.029	13.2	0.292	0.048	16.4
	5	0.220	0.026	12.0	0.295	0.044	14.8
	6	0.220	0.025	11.3	0.295	0.040	13.5
	7	0.220	0.023	10.5	0.293	0.037	12.5
	8	0.219	0.021	9.6	0.291	0.035	12.0
3rd 30 s	1	0.225	0.056	24.8	0.294	0.095	32.5
	2	0.223	0.042	18.7	0.293	0.067	23.0
	3	0.224	0.033	14.8	0.299	0.057	19.1
	4	0.222	0.028	12.6	0.299	0.049	16.5
	5	0.223	0.025	11.4	0.300	0.046	15.4
	6	0.223	0.023	10.4	0.296	0.041	13.9
	7	0.222	0.021	9.7	0.294	0.038	13.0
	8	0.221	0.020	9.0	0.292	0.035	12.1
1st 60 s	1	0.488	0.134	27.6	0.653	0.209	32.1
	2	0.477	0.090	18.8	0.628	0.141	22.4
	3	0.470	0.070	14.8	0.625	0.111	17.8
	4	0.465	0.059	12.7	0.626	0.095	15.2
	5	0.468	0.053	11.4	0.632	0.091	14.3
	6	0.468	0.050	10.7	0.633	0.084	13.3
	7	0.466	0.046	9.9	0.629	0.077	12.2
	8	0.464	0.042	9.1	0.622	0.071	11.5
2nd 60 s	1	0.454	0.118	26.1	0.594	0.201	33.8
	2	0.449	0.084	18.6	0.583	0.135	23.1
	3	0.445	0.064	14.5	0.590	0.108	18.3
	4	0.442	0.054	12.3	0.591	0.093	15.8
	5	0.443	0.049	11.1	0.595	0.086	14.4
	6	0.444	0.046	10.3	0.591	0.077	13.1
	7	0.442	0.042	9.6	0.588	0.072	12.2
	8	0.440	0.039	8.9	0.583	0.067	11.5
90 s	1	0.713	0.183	25.7	0.941	0.297	31.6
	2	0.700	0.127	18.1	0.917	0.202	22.0
	3	0.694	0.098	14.2	0.919	0.165	17.9
	4	0.687	0.084	12.2	0.921	0.142	15.4
	5	0.690	0.076	10.9	0.929	0.134	14.4
	6	0.691	0.070	10.2	0.927	0.122	13.2
	7	0.688	0.065	9.5	0.921	0.112	12.2
	8	0.684	0.060	8.7	0.912	0.104	11.4

***Table K.2**. Sequential averaging analysis for the variable Sway Area (mm²) in the ALL participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	72.3	44.3	61.3	87.0	56.7	65.2
	2	65.0	28.4	43.7	88.7	35.8	40.3
	3	66.2	23.2	35.1	97.9	35.3	36.0
	4	70.5	24.8	35.2	95.1	31.0	32.5
	5	72.4	22.7	31.4	91.3	25.9	28.4
	6	74.2	25.5	34.4	91.1	22.9	25.1
	7	73.0	22.6	31.0	89.9	20.1	22.4
	8	71.4	20.2	28.3	88.5	18.7	21.1
2nd 30 s	1	47.7	34.3	72.0	63.5	39.8	62.7
	2	49.8	22.1	44.4	60.4	25.7	42.5
	3	46.0	16.3	35.3	61.3	22.2	36.2
	4	47.4	14.5	30.6	61.2	20.9	34.2
	5	47.6	13.6	28.6	63.3	19.4	30.6
	6	49.1	14.6	29.7	63.5	18.5	29.1
	7	48.8	13.7	28.0	62.1	16.7	26.9
	8	49.6	12.8	25.9	62.5	16.6	26.5
3rd 30 s	1	47.0	34.6	73.6	61.3	35.9	58.6
	2	47.2	23.9	50.6	66.5	34.7	52.1
	3	48.9	19.3	39.4	72.3	35.6	49.3
	4	46.6	16.2	34.7	71.7	30.0	41.9
	5	47.2	14.6	30.9	70.3	26.6	37.9
	6	50.1	16.0	32.0	66.7	22.8	34.2
	7	48.4	14.2	29.4	64.3	20.3	31.6
	8	46.5	12.7	27.4	62.2	18.2	29.2
1st 60 s	1	85.8	51.8	60.4	104.7	69.5	66.4
	2	82.2	34.3	41.7	105.8	43.1	40.7
	3	82.1	28.1	34.2	114.2	41.6	36.5
	4	89.3	29.8	33.4	113.1	38.1	33.7
	5	89.6	26.2	29.2	111.5	32.6	29.2
	6	93.2	31.2	33.5	110.9	28.7	25.9
	7	93.7	29.5	31.5	109.5	25.4	23.2
	8	92.8	26.6	28.7	111.4	25.5	22.9
2nd 60 s	1	66.2	62.9	94.9	82.4	48.4	58.7
	2	65.8	35.9	54.5	80.0	34.5	43.1
	3	66.4	26.3	39.7	83.8	31.6	37.7
	4	65.2	22.4	34.3	83.5	27.5	32.9
	5	65.5	20.5	31.3	83.6	24.6	29.4
	6	69.6	23.5	33.8	82.2	21.7	26.5
	7	69.7	22.0	31.6	80.0	19.6	24.5
	8	68.5	19.6	28.7	80.7	20.4	25.3
90 s	1	94.3	74.4	78.9	114.4	64.4	56.3
	2	91.6	45.3	49.5	116.1	44.1	37.9
	3	93.2	34.2	36.7	122.7	40.8	33.2
	4	96.9	30.9	31.9	121.6	36.6	30.1
	5	97.6	27.5	28.2	120.7	32.9	27.3
	6	103.3	32.1	31.0	119.4	28.7	24.1
	7	104.1	30.8	29.6	117.7	25.6	21.8
	8	102.8	27.6	26.9	119.5	25.2	21.1

Table K.3. *Sequential averaging analysis for the variable Mean Velocity (cm.s^{-1}) in the ALL participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	0.857	0.239	27.9	1.169	0.365	31.2
	2	0.832	0.163	19.6	1.121	0.255	22.7
	3	0.826	0.130	15.8	1.109	0.204	18.4
	4	0.812	0.111	13.6	1.108	0.176	15.9
	5	0.820	0.099	12.1	1.120	0.171	15.3
	6	0.820	0.092	11.2	1.122	0.159	14.2
	7	0.814	0.084	10.3	1.111	0.145	13.0
	8	0.810	0.077	9.5	1.098	0.132	12.0
2nd 30 s	1	0.758	0.223	29.4	0.994	0.378	38.0
	2	0.746	0.150	20.2	0.962	0.246	25.6
	3	0.731	0.114	15.6	0.964	0.188	19.5
	4	0.728	0.097	13.3	0.968	0.160	16.5
	5	0.728	0.088	12.1	0.976	0.145	14.8
	6	0.729	0.083	11.4	0.976	0.133	13.6
	7	0.727	0.077	10.6	0.973	0.122	12.5
	8	0.725	0.070	9.7	0.965	0.116	12.0
3rd 30 s	1	0.743	0.186	25.0	0.973	0.318	32.7
	2	0.738	0.139	18.8	0.969	0.224	23.1
	3	0.740	0.110	14.9	0.990	0.190	19.2
	4	0.735	0.093	12.7	0.989	0.165	16.6
	5	0.737	0.084	11.4	0.994	0.154	15.5
	6	0.739	0.077	10.5	0.982	0.137	14.0
	7	0.734	0.071	9.7	0.975	0.127	13.0
	8	0.730	0.066	9.0	0.966	0.117	12.1
1st 60 s	1	0.808	0.224	27.7	1.082	0.348	32.2
	2	0.789	0.149	18.9	1.042	0.234	22.5
	3	0.778	0.116	14.9	1.037	0.185	17.8
	4	0.770	0.098	12.7	1.038	0.159	15.3
	5	0.774	0.089	11.5	1.048	0.151	14.4
	6	0.774	0.084	10.8	1.049	0.140	13.3
	7	0.771	0.077	9.9	1.042	0.128	12.3
	8	0.767	0.070	9.1	1.032	0.119	11.5
2nd 60 s	1	0.751	0.197	26.2	0.984	0.335	34.0
	2	0.743	0.139	18.8	0.966	0.224	23.2
	3	0.735	0.107	14.6	0.977	0.180	18.4
	4	0.732	0.090	12.4	0.979	0.155	15.8
	5	0.733	0.082	11.1	0.985	0.143	14.5
	6	0.734	0.076	10.4	0.979	0.129	13.2
	7	0.731	0.071	9.7	0.974	0.120	12.3
	8	0.728	0.065	8.9	0.966	0.112	11.6
90 s	1	0.786	0.204	25.9	1.040	0.330	31.7
	2	0.773	0.141	18.2	1.014	0.224	22.1
	3	0.766	0.109	14.3	1.016	0.183	18.0
	4	0.758	0.093	12.2	1.018	0.157	15.5
	5	0.762	0.084	11.0	1.027	0.148	14.4
	6	0.762	0.078	10.3	1.024	0.136	13.2
	7	0.759	0.072	9.5	1.018	0.125	12.2
	8	0.755	0.066	8.8	1.008	0.115	11.4

Table K.4*. Sequential averaging analysis for the variable Path Length (m) in the Female participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	0.218	0.023	10.7	0.304	0.074	24.3
	2	0.210	0.026	12.5	0.289	0.046	15.8
	3	0.210	0.026	12.2	0.287	0.039	13.6
	4	0.209	0.022	10.5	0.286	0.034	11.9
	5	0.213	0.021	9.9	0.284	0.028	10.0
	6	0.214	0.019	9.0	0.283	0.026	9.2
	7	0.215	0.018	8.4	0.282	0.024	8.5
	8	0.216	0.017	7.9	0.280	0.023	8.1
2nd 30 s	1	0.196	0.029	14.6	0.250	0.051	20.2
	2	0.197	0.022	10.9	0.253	0.056	22.3
	3	0.193	0.019	9.7	0.256	0.041	15.9
	4	0.197	0.019	9.6	0.255	0.035	13.7
	5	0.196	0.017	8.6	0.255	0.030	11.8
	6	0.195	0.015	7.7	0.254	0.026	10.3
	7	0.194	0.014	7.0	0.254	0.025	9.7
	8	0.195	0.013	6.6	0.251	0.022	9.0
3rd 30 s	1	0.199	0.031	15.7	0.252	0.060	23.9
	2	0.197	0.025	12.8	0.248	0.044	17.8
	3	0.203	0.024	12.0	0.259	0.041	16.0
	4	0.200	0.020	10.0	0.259	0.035	13.6
	5	0.199	0.018	9.0	0.258	0.031	12.1
	6	0.200	0.017	8.4	0.254	0.027	10.8
	7	0.200	0.015	7.6	0.251	0.026	10.2
	8	0.200	0.014	7.2	0.248	0.023	9.3
1st 60 s	1	0.414	0.042	10.2	0.554	0.116	20.9
	2	0.408	0.042	10.3	0.542	0.090	16.7
	3	0.403	0.041	10.1	0.543	0.072	13.3
	4	0.406	0.037	9.2	0.541	0.063	11.6
	5	0.410	0.035	8.5	0.539	0.053	9.8
	6	0.408	0.031	7.7	0.538	0.048	8.8
	7	0.410	0.029	7.2	0.536	0.044	8.3
	8	0.411	0.028	6.8	0.530	0.042	7.8
2nd 60 s	1	0.395	0.056	14.1	0.501	0.109	21.7
	2	0.395	0.044	11.2	0.501	0.097	19.4
	3	0.396	0.040	10.1	0.515	0.078	15.1
	4	0.397	0.035	8.9	0.514	0.067	13.0
	5	0.396	0.032	8.1	0.513	0.058	11.4
	6	0.395	0.029	7.4	0.508	0.051	10.0
	7	0.394	0.026	6.7	0.505	0.048	9.5
	8	0.395	0.025	6.4	0.499	0.044	8.8
90 s	1	0.613	0.070	11.4	0.795	0.162	20.4
	2	0.605	0.064	10.5	0.785	0.128	16.3
	3	0.606	0.061	10.0	0.798	0.108	13.5
	4	0.606	0.053	8.8	0.797	0.093	11.7
	5	0.608	0.050	8.2	0.795	0.080	10.1
	6	0.608	0.045	7.4	0.790	0.071	9.0
	7	0.609	0.042	6.9	0.785	0.067	8.5
	8	0.611	0.040	6.5	0.777	0.062	8.0

***Table K.5**. Sequential averaging analysis for the variable Sway Area (mm^2) in the Female participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	70.5	49.1	69.6	88.2	63.8	72.4
	2	58.8	27.8	47.3	94.0	41.1	43.8
	3	56.9	19.4	34.1	105.8	41.1	38.9
	4	61.7	17.5	28.3	97.7	33.1	33.9
	5	63.5	17.4	27.5	92.2	27.8	30.2
	6	64.3	17.1	26.6	92.2	24.7	26.8
	7	65.5	16.4	25.0	90.4	21.8	24.2
	8	65.3	15.1	23.1	88.3	20.3	23.0
2nd 30 s	1	40.7	18.4	45.3	46.7	22.5	48.2
	2	41.7	13.5	32.4	51.8	20.9	40.5
	3	38.0	10.0	26.3	57.1	23.0	40.2
	4	42.5	12.1	28.6	52.8	18.5	35.0
	5	41.0	10.9	26.6	54.4	16.3	29.9
	6	41.3	10.0	24.2	57.8	18.3	31.7
	7	42.7	11.1	26.1	57.1	16.3	28.5
	8	45.2	11.3	25.1	57.7	15.2	26.4
3rd 30 s	1	38.4	20.1	52.3	62.8	31.4	50.1
	2	39.8	14.3	36.0	64.3	31.7	49.3
	3	44.8	14.9	33.2	74.0	41.4	56.0
	4	44.0	14.1	32.2	71.7	33.6	46.9
	5	43.5	13.1	30.2	66.4	27.9	42.1
	6	47.2	16.1	34.0	62.8	23.8	37.9
	7	46.4	14.1	30.4	60.4	21.3	35.2
	8	44.4	12.6	28.3	58.0	19.1	32.9
1st 60 s	1	92.2	58.7	63.6	91.5	63.2	69.1
	2	80.1	38.5	48.0	102.7	44.4	43.3
	3	74.1	26.4	35.6	116.9	39.8	34.0
	4	80.8	23.9	29.6	109.3	33.2	30.3
	5	78.9	21.0	26.6	104.3	27.9	26.7
	6	81.4	22.5	27.7	105.3	25.8	24.5
	7	84.8	24.5	28.9	103.8	22.9	22.0
	8	86.3	22.9	26.5	105.0	22.4	21.3
2nd 60 s	1	51.6	24.2	47.0	68.3	29.7	43.6
	2	51.1	14.8	29.0	70.6	28.1	39.8
	3	57.4	15.2	26.4	83.3	35.1	42.1
	4	59.6	17.3	29.1	78.7	28.5	36.2
	5	57.1	15.3	26.9	75.9	24.0	31.6
	6	61.0	18.0	29.6	75.6	21.8	28.8
	7	62.1	17.4	28.1	74.6	19.5	26.2
	8	62.0	15.6	25.1	74.0	18.1	24.5
90 s	1	86.4	54.3	62.8	97.8	62.3	63.7
	2	78.0	33.0	42.4	109.0	44.8	41.1
	3	82.0	25.8	31.5	124.2	41.9	33.7
	4	86.7	24.3	28.0	117.3	34.4	29.4
	5	85.6	21.6	25.3	111.2	28.9	26.0
	6	92.1	24.0	26.1	110.6	25.8	23.3
	7	95.3	25.4	26.6	110.2	23.4	21.2
	8	95.7	23.1	24.2	111.8	23.4	20.9

Table K.6. *Sequential averaging analysis for the variable Mean Velocity (cm.s^{-1}) in the Female participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	0.721	0.077	10.7	1.008	0.246	24.4
	2	0.696	0.087	12.6	0.957	0.151	15.8
	3	0.694	0.085	12.3	0.951	0.130	13.7
	4	0.693	0.073	10.6	0.948	0.113	11.9
	5	0.705	0.070	9.9	0.941	0.094	10.0
	6	0.706	0.064	9.1	0.938	0.087	9.3
	7	0.711	0.060	8.5	0.933	0.080	8.5
	8	0.713	0.057	8.0	0.926	0.076	8.2
2nd 30 s	1	0.648	0.096	14.8	0.827	0.168	20.4
	2	0.653	0.072	11.0	0.837	0.188	22.4
	3	0.639	0.062	9.8	0.847	0.136	16.0
	4	0.651	0.063	9.7	0.846	0.117	13.8
	5	0.649	0.056	8.7	0.844	0.100	11.9
	6	0.643	0.050	7.7	0.842	0.088	10.4
	7	0.642	0.045	7.0	0.841	0.082	9.7
	8	0.644	0.043	6.7	0.830	0.075	9.0
3rd 30 s	1	0.656	0.103	15.8	0.833	0.200	24.0
	2	0.652	0.084	12.9	0.822	0.147	17.9
	3	0.670	0.081	12.1	0.858	0.138	16.1
	4	0.660	0.067	10.1	0.856	0.118	13.7
	5	0.658	0.060	9.1	0.854	0.104	12.2
	6	0.661	0.056	8.5	0.841	0.091	10.9
	7	0.660	0.050	7.6	0.832	0.086	10.3
	8	0.661	0.048	7.2	0.821	0.077	9.4
1st 60 s	1	0.684	0.070	10.3	0.917	0.193	21.0
	2	0.674	0.070	10.3	0.897	0.150	16.8
	3	0.667	0.068	10.2	0.899	0.120	13.4
	4	0.672	0.062	9.3	0.897	0.105	11.7
	5	0.677	0.058	8.6	0.893	0.088	9.9
	6	0.675	0.052	7.7	0.891	0.079	8.9
	7	0.677	0.049	7.2	0.887	0.074	8.3
	8	0.679	0.046	6.8	0.878	0.069	7.9
2nd 60 s	1	0.652	0.093	14.3	0.830	0.182	21.9
	2	0.652	0.074	11.3	0.829	0.162	19.5
	3	0.655	0.067	10.2	0.853	0.129	15.2
	4	0.656	0.059	9.0	0.851	0.111	13.1
	5	0.654	0.054	8.2	0.850	0.098	11.5
	6	0.653	0.048	7.4	0.842	0.085	10.1
	7	0.651	0.044	6.7	0.836	0.080	9.6
	8	0.653	0.042	6.4	0.826	0.073	8.8
90 s	1	0.675	0.077	11.5	0.878	0.180	20.5
	2	0.667	0.071	10.6	0.866	0.142	16.4
	3	0.668	0.067	10.1	0.882	0.120	13.6
	4	0.667	0.059	8.9	0.881	0.104	11.8
	5	0.670	0.055	8.2	0.878	0.089	10.2
	6	0.670	0.050	7.5	0.872	0.079	9.1
	7	0.671	0.046	6.9	0.867	0.075	8.6
	8	0.673	0.044	6.6	0.858	0.069	8.0

***Table K.**7. Sequential averaging analysis for the variable Path Length (m) in the Male participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	0.296	0.081	27.2	0.396	0.118	29.8
	2	0.288	0.052	18.0	0.382	0.088	23.0
	3	0.285	0.038	13.3	0.377	0.067	17.7
	4	0.277	0.033	12.1	0.377	0.057	15.3
	5	0.279	0.030	10.6	0.386	0.058	15.0
	6	0.278	0.029	10.4	0.388	0.056	14.3
	7	0.274	0.026	9.5	0.383	0.049	12.7
	8	0.270	0.024	8.9	0.377	0.044	11.8
2nd 30 s	1	0.259	0.077	29.6	0.345	0.134	38.7
	2	0.251	0.054	21.6	0.324	0.083	25.5
	3	0.245	0.038	15.6	0.322	0.062	19.4
	4	0.241	0.032	13.4	0.325	0.052	15.9
	5	0.242	0.030	12.2	0.330	0.047	14.4
	6	0.243	0.029	12.1	0.331	0.046	13.9
	7	0.243	0.026	10.8	0.329	0.040	12.2
	8	0.241	0.024	9.9	0.328	0.039	11.9
3rd 30 s	1	0.248	0.062	25.1	0.332	0.105	31.7
	2	0.247	0.048	19.5	0.332	0.074	22.4
	3	0.243	0.036	14.8	0.334	0.061	18.3
	4	0.242	0.030	12.6	0.334	0.053	15.9
	5	0.244	0.027	11.1	0.338	0.051	15.1
	6	0.244	0.026	10.6	0.334	0.047	14.1
	7	0.242	0.023	9.7	0.333	0.042	12.5
	8	0.239	0.022	9.1	0.331	0.038	11.6
1st 60 s	1	0.555	0.153	27.6	0.741	0.234	31.5
	2	0.539	0.101	18.8	0.706	0.160	22.7
	3	0.530	0.072	13.6	0.700	0.122	17.4
	4	0.518	0.062	12.0	0.702	0.103	14.7
	5	0.520	0.056	10.8	0.716	0.100	14.0
	6	0.521	0.056	10.7	0.719	0.098	13.6
	7	0.517	0.050	9.6	0.712	0.086	12.0
	8	0.511	0.045	8.9	0.705	0.079	11.2
2nd 60 s	1	0.507	0.133	26.3	0.677	0.227	33.5
	2	0.498	0.098	19.7	0.656	0.148	22.5
	3	0.488	0.071	14.6	0.657	0.117	17.8
	4	0.483	0.060	12.4	0.660	0.100	15.1
	5	0.486	0.054	11.0	0.668	0.093	13.9
	6	0.488	0.053	10.8	0.665	0.088	13.2
	7	0.485	0.047	9.8	0.662	0.078	11.8
	8	0.480	0.043	9.0	0.659	0.073	11.1
90 s	1	0.803	0.206	25.7	1.073	0.328	30.6
	2	0.786	0.139	17.7	1.037	0.220	21.2
	3	0.773	0.105	13.6	1.028	0.184	17.9
	4	0.760	0.089	11.7	1.032	0.154	15.0
	5	0.764	0.080	10.4	1.050	0.148	14.1
	6	0.766	0.075	9.8	1.050	0.136	12.9
	7	0.759	0.070	9.3	1.043	0.124	11.9
	8	0.750	0.065	8.6	1.034	0.115	11.1

Table K.8. Sequential averaging analysis for the variable Sway Area (mm²) in the Male participants population.

EPOCH	No. of trials incl.	Eyes Open Mean	Eyes Open SEM	Eyes Open SEM % of Mean	Eyes Closed Mean	Eyes Closed SEM	Eyes Closed SEM % of Mean
1st 30 s	1	74.0	39.5	53.3	85.9	49.4	57.5
	2	70.6	29.8	42.2	83.9	31.9	38.0
	3	74.5	25.3	33.9	90.9	28.4	31.3
	4	78.4	29.3	37.4	92.8	28.8	31.1
	5	80.4	26.1	32.5	90.5	24.0	26.5
	6	83.1	32.0	38.5	90.2	21.9	24.3
	7	79.7	26.8	33.6	89.6	18.5	20.7
	8	76.8	23.7	30.9	88.7	17.0	19.2
2nd 30 s	1	54.0	43.0	79.7	78.5	45.5	58.0
	2	57.1	27.5	48.2	68.2	29.3	43.0
	3	53.2	19.4	36.5	65.2	21.3	32.6
	4	51.7	16.0	30.9	68.8	22.2	32.3
	5	53.6	15.2	28.3	71.3	21.2	29.7
	6	56.1	17.4	31.0	68.7	19.2	28.0
	7	54.3	15.3	28.3	66.6	16.9	25.4
	8	53.5	13.9	25.9	66.8	17.6	26.3
3rd 30 s	1	54.8	42.2	77.1	60.0	39.5	65.8
	2	53.9	30.2	56.0	68.5	38.8	56.6
	3	52.7	22.3	42.4	70.7	29.4	41.6
	4	49.0	17.7	36.2	71.7	26.4	36.8
	5	50.5	15.6	30.9	73.8	25.3	34.2
	6	52.6	16.5	31.3	70.2	22.9	32.5
	7	50.3	14.3	28.5	67.9	19.4	28.6
	8	48.3	12.9	26.6	65.9	17.2	26.1
1st 60 s	1	80.0	43.8	54.8	116.7	72.7	62.3
	2	84.1	32.0	38.1	108.7	44.1	40.6
	3	89.3	28.9	32.4	111.8	43.2	38.6
	4	96.9	33.8	34.9	116.5	42.0	36.0
	5	99.3	29.4	29.6	118.0	36.1	30.6
	6	103.8	38.3	36.9	115.9	31.7	27.3
	7	101.7	33.1	32.5	114.6	27.3	23.8
	8	98.6	29.5	29.9	117.2	27.9	23.9
2nd 60 s	1	79.5	81.3	102.3	95.2	57.5	60.4
	2	79.1	46.7	59.1	88.6	39.9	45.1
	3	74.4	32.6	43.8	84.3	28.1	33.4
	4	70.2	25.8	36.8	87.9	26.4	30.1
	5	73.0	23.7	32.4	90.6	24.7	27.3
	6	77.3	28.3	36.6	88.1	22.5	25.5
	7	76.5	25.2	32.9	84.8	19.5	23.0
	8	74.3	22.5	30.2	86.7	22.1	25.5
90 s	1	101.4	88.1	86.8	129.4	62.6	48.4
	2	103.8	53.0	51.1	122.6	44.0	35.9
	3	103.3	40.1	38.8	121.3	40.3	33.3
	4	106.0	35.3	33.3	125.4	38.3	30.5
	5	108.4	31.1	28.7	129.2	35.7	27.6
	6	113.4	37.4	33.0	127.3	30.8	24.2
	7	112.1	34.7	30.9	124.4	27.2	21.9
	8	109.1	30.9	28.3	126.3	26.4	20.9

Table K.9. *Sequential averaging analysis for the variable Mean Velocity (cm.s^{-1}) in the Male participants population.*

		Eyes Open			Eyes Closed		
EPOCH	No. of trials incl.	Mean	SEM	SEM % of Mean	Mean	SEM	SEM % of Mean
1st 30 s	1	0.980	0.268	27.4	1.315	0.393	29.9
	2	0.955	0.173	18.1	1.268	0.292	23.0
	3	0.944	0.126	13.3	1.252	0.222	17.8
	4	0.919	0.111	12.1	1.251	0.191	15.3
	5	0.924	0.098	10.7	1.280	0.192	15.0
	6	0.922	0.096	10.4	1.288	0.185	14.4
	7	0.907	0.086	9.5	1.272	0.162	12.7
	8	0.896	0.080	8.9	1.252	0.148	11.8
2nd 30 s	1	0.857	0.255	29.8	1.145	0.445	38.9
	2	0.831	0.181	21.8	1.074	0.275	25.6
	3	0.813	0.128	15.7	1.069	0.208	19.4
	4	0.798	0.108	13.5	1.079	0.173	16.0
	5	0.800	0.098	12.3	1.095	0.158	14.4
	6	0.805	0.098	12.2	1.097	0.152	13.9
	7	0.804	0.087	10.8	1.091	0.134	12.3
	8	0.798	0.079	9.9	1.087	0.129	11.9
3rd 30 s	1	0.821	0.207	25.3	1.099	0.350	31.9
	2	0.816	0.160	19.6	1.102	0.248	22.5
	3	0.803	0.119	14.8	1.109	0.204	18.4
	4	0.802	0.101	12.6	1.109	0.177	16.0
	5	0.808	0.090	11.1	1.120	0.170	15.2
	6	0.809	0.086	10.7	1.109	0.157	14.1
	7	0.801	0.078	9.8	1.104	0.138	12.5
	8	0.791	0.072	9.1	1.097	0.127	11.6
1st 60 s	1	0.919	0.255	27.7	1.230	0.389	31.6
	2	0.893	0.168	18.9	1.171	0.266	22.7
	3	0.878	0.120	13.6	1.161	0.203	17.5
	4	0.859	0.103	12.0	1.165	0.172	14.7
	5	0.862	0.093	10.8	1.188	0.167	14.0
	6	0.864	0.093	10.7	1.193	0.162	13.6
	7	0.856	0.083	9.7	1.182	0.142	12.0
	8	0.847	0.076	8.9	1.170	0.132	11.3
2nd 60 s	1	0.839	0.222	26.5	1.122	0.378	33.7
	2	0.824	0.163	19.8	1.088	0.246	22.6
	3	0.808	0.119	14.7	1.089	0.194	17.8
	4	0.800	0.100	12.5	1.094	0.166	15.2
	5	0.804	0.089	11.1	1.108	0.155	14.0
	6	0.807	0.088	10.9	1.103	0.146	13.3
	7	0.803	0.079	9.8	1.098	0.130	11.8
	8	0.795	0.072	9.1	1.092	0.122	11.2
90 s	1	0.886	0.228	25.8	1.186	0.364	30.7
	2	0.868	0.155	17.8	1.146	0.244	21.3
	3	0.854	0.117	13.7	1.137	0.205	18.0
	4	0.840	0.098	11.7	1.142	0.171	15.0
	5	0.844	0.088	10.4	1.161	0.165	14.2
	6	0.846	0.083	9.9	1.161	0.150	12.9
	7	0.838	0.078	9.3	1.153	0.138	11.9
	8	0.828	0.072	8.7	1.143	0.127	11.1

Appendix L. Descriptive statistics for the individual CMJ trials for each population. [3 tables]

Table L.1. *Descriptive statistics (Mean ± SD) for each of the 5 trials on day 1 and day 2 for each variable.* ***Study Population: All Participants***

		1		2		3		4		5		Mean of all trials	
Variable	Day	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD
Peak Force (N)	Day 1	1685	443	1705	481	1723	492	1706	480	1702	443	1704	465
	Day 2	1691	404	1714	429	1697	437	1688	438	1685	441	1695	428
Eccentric Impulse (N.s)	Day 1	85.8	28.5	90.5	28.2	91.1	30.0	89.3	29.7	88.0	27.0	88.9	26.7
	Day 2	88.1	25.6	96.1	26.8	92.0	28.2	95.7	33.7	91.9	28.5	92.7	25.6
Concentric Impulse (N.s)	Day 1	181.5	58.6	181.5	59.5	181.1	58.7	179.7	58.8	179.0	57.6	180.6	58.5
	Day 2	182.3	58.3	181.2	58.6	182.0	58.9	179.6	56.9	178.9	57.5	180.8	57.9
Peak Mechanical Power (W)	Day 1	3393	1133	3387	1146	3381	1151	3334	1131	3306	1080	3360	1126
	Day 2	3419	1120	3369	1113	3403	1126	3349	1056	3322	1079	3372	1097
Take Off Velocity ($m.s^{-1}$)	Day 1	2.23	0.38	2.23	0.39	2.22	0.39	2.20	0.39	2.20	0.38	2.22	0.38
	Day 2	2.24	0.36	2.22	0.37	2.23	0.37	2.21	0.37	2.20	0.37	2.22	0.37
Ecc/Con Changeover % of jump	Day 1	68.2	3.0	67.9	3.6	68.8	3.5	67.8	3.0	68.9	4.2	68.3	2.9
	Day 2	68.6	3.7	67.8	3.9	68.2	3.7	67.3	4.0	68.4	3.5	68.1	3.2
Peak Force % of jump	Day 1	79.3	11.9	78.3	11.0	78.7	11.0	77.9	11.5	78.0	10.6	78.5	10.1
	Day 2	79.2	10.6	74.9	12.1	82.9	9.0	82.4	9.8	82.4	10.1	80.4	8.8

Table L.2. *Descriptive statistics (Mean ± SD) for each of the 5 trials on day 1 and day 2 for each variable.* ***Study Population: Female***

		1		2		3		4		5		Mean of all trials	
Variable	Day	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD
Peak Force (N)	Day 1	1416	228	1431	234	1437	251	1424	252	1457	258	1433	242
	Day 2	1448	242	1454	239	1421	222	1424	239	1407	226	1431	232
Eccentric Impulse (N.s)	Day 1	74.1	19.2	78.8	19.4	81.2	28.1	74.1	19.4	78.1	24.0	77.2	17.9
	Day 2	81.2	27.0	86.2	24.9	77.1	20.9	86.8	35.0	76.8	19.6	81.6	20.1
Concentric Impulse (N.s)	Day 1	136.4	29.7	136.4	30.7	135.9	28.5	135.3	28.6	134.9	30.2	135.8	29.4
	Day 2	138.3	29.3	135.8	28.1	137.0	29.2	135.7	28.8	134.7	27.5	136.3	28.5
Peak Mechanical Power (W)	Day 1	2485	554	2494	574	2486	562	2455	563	2461	535	2476	555
	Day 2	2553	561	2511	549	2525	562	2521	555	2500	550	2522	554
Take Off Velocity ($m.s^{-1}$)	Day 1	1.93	0.31	1.93	0.34	1.93	0.36	1.92	0.33	1.91	0.34	1.93	0.33
	Day 2	1.97	0.32	1.93	0.32	1.95	0.32	1.93	0.33	1.92	0.32	1.94	0.32
Ecc/Con Changeover % of jump	Day 1	67.5	2.7	68.1	4.1	68.9	3.6	68.6	3.3	69.2	5.1	68.5	3.5
	Day 2	68.1	4.4	67.5	5.0	67.9	4.7	68.0	3.4	68.5	4.5	68.0	4.1
Peak Force % of jump	Day 1	74.8	10.9	77.1	11.5	75.2	11.0	77.9	11.3	78.6	10.8	76.7	10.5
	Day 2	79.1	9.8	71.3	11.2	84.8	6.0	86.2	6.3	86.7	6.2	81.6	5.4

Table L.3. *Descriptive statistics (Mean ± SD) for each of the 5 trials on day 1 and day 2 for each variable.* ***Study Population: Male***

		1		2		3		4		5		Mean of all trials	
Variable	Day	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD	Mean	± SD
Peak Force (N)	Day 1	1927	450	1952	512	1980	513	1959	494	1922	458	1948	482
	Day 2	1909	396	1948	427	1946	433	1926	439	1935	438	1933	424
Eccentric Impulse (N.s)	Day 1	96.3	31.2	100.9	30.7	100.1	28.7	103.1	30.6	97.0	26.5	99.5	28.8
	Day 2	94.3	22.6	105.0	25.4	105.5	27.1	103.6	30.4	105.4	28.5	102.8	25.9
Concentric Impulse (N.s)	Day 1	222.1	47.4	222.1	48.9	221.8	48.1	219.6	49.7	218.6	46.5	220.8	48.0
	Day 2	221.9	48.6	222.0	48.0	222.4	48.6	219.1	45.9	218.6	47.5	220.8	47.6
Peak Mechanical Power (W)	Day 1	4210	867	4191	914	4186	928	4126	905	4066	858	4156	892
	Day 2	4199	904	4141	908	4195	897	4093	819	4062	885	4138	879
Take Off Velocity ($m.s^{-1}$)	Day 1	2.49	0.18	2.49	0.19	2.49	0.19	2.46	0.21	2.47	0.17	2.48	0.18
	Day 2	2.48	0.17	2.48	0.18	2.49	0.19	2.45	0.18	2.45	0.20	2.47	0.18
Ecc/Con Changeover % of jump	Day 1	68.9	3.1	67.7	3.1	68.7	3.3	67.1	2.6	68.6	3.2	68.2	2.3
	Day 2	69.0	2.9	68.0	2.6	68.6	2.4	66.7	4.3	68.3	2.1	68.1	2.1
Peak Force % of jump	Day 1	83.3	11.3	79.4	10.4	81.9	10.0	77.9	11.7	77.5	10.4	80.0	9.6
	Day 2	79.3	11.3	78.1	12.0	81.1	10.7	79.0	11.1	78.5	11.3	79.2	10.8

Appendix M. Sample of the existing literature that have adopted seated measures of postural stability.

Table M.1. *Sample of existing literature adopting the seated method of postural stability assessment.*

Variables / Author	Static/ dynamic conditions	Force platform	Instructions	Leg positioning	Arm position	Duration & Repetitions	Sampling & Filtering Frequency (Hz)	Data Normalised
Cholewicki *et al*. (2000)	Polyester resin hemispheres of various diameters were attached to the bottom of the seat	Yes	Instructed to maintain his/her balance	Seat equipped with leg and foot supports to prevent any lower body movement. Foot support was adjusted to create 90º knee angle. Same foot placement used for all subjects.	Arms crossed Safety railing built around force plate for security if the subject loses - balance.	5 x 7 s Rest: 30 s 1 min practice at each instability level.	No info	No info
Radebold *et al*. (2001)	Polyester hemispheres of varying diameters	Yes	Instructed to maintain balance while sitting upright	Seat equipped with a foot support to prevent any lower body movement	Arms crossed	5 x 7 s Rest: 30 s 1 min practice at each instability level.	1600 & 10	No info
Kerr & Eng (2002)	(1) reach forwards using both hands (2) reach to the side using left/right hand (3) lean backward, both hands in your lap	Yes	Subjects were to move as far and fast as possible and hold the terminal position for 3 s.	80% of the thigh supported Supported foot condition- hips/knees at 90° and feet on two force plates. Unsupported foot condition- raised seat height and feet dangling	Refer to conditions (1), (2) & (3)	5 x 6 s for each condition	600 & No info	Normalised to upper body length
Slota *et al*. (2008)	Wobble chair Adapted from Cholewicki *et al*. (2000)	No	Instructed to sit with an upright posture	Affixed to the seat was an adjustable footrest to limit the motion of the lower limbs during unstable seated balance testing	Arms crossed in front of the chest	6 – 8 practice trials 4 x 60 s 30 min Vibration 3 x 60 s	n/a	No info
Larivière *et al*. (2013)	Wobble chair Reproduction of Slota *et al*. (2008)	Inertial sensors 1 60 s trial used FP		Feet were strapped on an adjustable footrest that was affixed to the seat. Further motions of the lower limbs were hindered with the use of foam cubes.	Arms crossed in front of the chest	2 sessions separated by 2 weeks 3 forward tilting trials 4 practice trials (30 s EO, 30 s EC & two 60 s EC) 3 x 60 s trials	1000 & Decimated to 50 then low pass filtered at 10	COP- n/a Spring positions adjusted to control for confounding effect of body size.

Appendix N: Ethical application and confirmation of approval.

Ethical approval was granted before the consequent changes to the thesis as a result of COVID-19. Therefore, this application relates to the original study design. However, as the current study was an in-depth analysis of two of the original study components, new ethical approval was not required.

N.1

AB

Andrew Bloodworth
Wed 06/11/2019 09:16
To: GROOM J. [redacted]
Cc: Katherine Thomas

We can approve this one immediately Jordan. I'll send you a screening questionnaire that I was sent by mark Waldron yesterday that might be useful for you to use, but it looks to me as though you are covering the same questions. Sorry for the delay and good luck with the research, all the best

Andrew

N.2

APPLICATION FOR ETHICAL COMMITTEE APPROVAL OF A RESEARCH PROJECT

All research with human participants, or on data derived from research with human participants that is not publicly available, undertaken by staff or students linked with A-STEM or in the College of Engineering more widely must be approved by the College of Engineering Research Ethics Committee.

RESEARCH MAY ONLY COMMENCE ONCE ETHICAL APPROVAL HAS BEEN OBTAINED

The researcher(s) should complete the form in consultation with the project supervisor. After completing and signing the form students should ask their supervisor to sign it. The form should be submitted electronically to coe-researchethics@swansea.ac.uk.

Applicants will be informed of the Committee's decision via email to the project leader/supervisor.

1. TITLE OF PROJECT
Working title: Does a relationship exist between a rower's static balance and/ or bilateral asymmetry of the lower limbs and instability of a rowing shell?

2. DATE OF PROJECT COMMENCEMENT AND PROPOSED DURATION OF THE STUDY
01/10/19- 30/09/20

3. NAMES AND STATUS OF THE RESEARCH TEAM *State the names of all members of the research group including the supervisor(s). State the current status of the student(s) in the group i.e. Undergraduate, postgraduate, staff or other (please specify).*
Jordan Groom- Postgraduate MSc student- Main researcher Nick Owen- Staff- Research Project Supervisor Charley Haynes- Researcher

4. RATIONALE AND REFERENCES *Describe in* ***no more than 200 words*** *the background to the proposed project.* *In all sections below that detail your study and its aims please use language suitable for a lay audience.*
Despite the existing research into predictors of performance on ergometers, there is very limited research into how faithfully the ability to perform on rowing ergometers can represent rowing ability on water. Consequently, the correlation of performance on rowing ergometers to actual race performance remains unclear (Mikulić, Smoljanović, Bojanić, Hannafin, & Matković, 2009). Some rowing ergometers that attempt to replicate motion on water do however exist (Domeika, Grigas, Žiliukas, & Vilkauskas, 2016), but a key characteristic of on-water rowing is instability, in particular roll, which is not often considered. There are a number of factors that can be deemed to impact upon the magnitude and frequency of roll within a boat, such as the forces exerted upon the boat by the environment such as drag, air resistance and buoyancy, and the forces exerted upon the boat by the rowers themselves- by means of the oar-blades and/or their distribution of weight and forces within the shell. It is with this that the purpose of the study is to firstly quantify the magnitude of roll of a boat on-water during a training session and secondly assess the relationship between rowers' static balance and bilateral asymmetry on the magnitude and frequency of roll. Bibliography Domeika, A., Grigas, V., Žiliukas, P., & Vilkauskas, A. (2016). Unstable simulator of academic rowing. *Mechanics*, *79*(5), 48-51–51. https://doi.org/10.5755/j01.mech.79.5.15472 Mikulić, P., Smoljanović, T., Bojanić, I., Hannafin, J. A., & Matković, B. R. (2009). Relationship between 2000-m rowing ergometer performance times and World Rowing Championships rankings in elite-standard rowers. *Journal of Sports Sciences*, *27*(9), 907–913.

5. OBJECTIVES *State the objectives of the project, i.e. one or more precise statements of what the project is designed to achieve.*
- Quantify roll and pitch of a rowing shell on water for different crews (male/ female, novice/ senior) and boats (e.g. 2, 4, 8 seated boats)

- Assess whether a relationship exists between an individual's static balance (measured using a Romberg test) and instability of a modified ergometer designed to simulate the roll of a boat (as earlier quantified).
- Assess whether a relationship exists between bilateral asymmetry of the lower limbs and instability of a modified ergometer.

6.1 STUDY DESIGN
Outline the chosen study design (e.g., cross-sectional, longitudinal, intervention, RCT, questionnaire etc)

This will be a cross-sectional study comprising of quantitative research with emphasis on the collection of numerical data, the summary of that data and the drawing of inferences from the data. The purpose of the study will be as fundamental research (as opposed to applied) in order to add to the field of application within rowing training and development.

6.2. STUDY DESIGN
- state the number and characteristics of study participants
- state the inclusion criteria for participants
- state the exclusion criteria for participants and identify any requirements for health screening
- state whether the study will involve vulnerable populations (i.e. young, elderly, clinical etc.)
- state the requirements/commitments expected of the participants (e.g. time, exertion level etc)

- Participants will be students that are current or recent members of Swansea University Rowing Club (SURC) and are/have been members of both the Novice and Senior teams.
- All rowing members (both male and female) will be invited to participate in the first stage of the study- the quantifying of roll on water. Novice members with less than 1 year of experience will not be invited to participate in the further areas of the study as the collection of data on the unstable simulator requires participants to have experience with the use of rowing ergometers.
- The intention for the study is to involve as many of each of the squads as is interested- approximately 15 of each group. The interest of participants is key as those who are not interested are under no pressure to participate.

<u>Inclusion criteria</u>
- Participants must be members of SURC 19/20 squad or must have been previous members from 17/18 or 18/19- Participants may be alumni members.

<u>Exclusion Criteria</u>
- Are members of either of the Novice squads.
- Do not wish to participate in the research
- On the advice of the captain- they are not suitable as a result of an injury that could affect their ability to perform or place them in personal danger. <u>Additionally, a PAR-Q will be used in order to identify any health risks.</u>

- The study will not involve vulnerable populations
- There will be no penalties for not participating in the research.

Commitments/ Requirements

- The study would require participants to give approximately 30 minutes of their time on up to 5 occasions- this would comprise of approximately 10mins of moderate to intense exertion for 2/3 of these sessions with the remaining time for warm-up and cool-down procedures. The first data collection would comprise of some low intensity testing- Romberg test and Minimum 3 countermovement jumps.

6.3. PARTICIPANT RECRUITMENT

How and where will participants be recruited? How will you ensure that these methods of recruitment do not compromise the ability of the research participant to freely consent to and withdraw from the study?

- Participants will be recruited from the students registered as members of SURC.
- The participant will be informed of the study via a post made by myself on the SURC Facebook page which is a closed group. Participants will also be addressed at training sessions as a face to face invitation. In both cases they will be provided with the participation information sheet (see Appendix 1). If in agreement, participants will provide their informed consent to participate using the informed consent from provided in Appendix 2.
- To ensure that these methods of recruitment do not compromise the ability of the research participants to freely consent to and withdraw from the study, all transactions will remain anonymous to others and with Facebook participants informing me of their decisions privately without having to notify other members of the group.

6.4 DATA COLLECTION METHODS

- describe all of the data collection/experimental procedures to be undertaken
- state any dietary supplementation that will be given to participants and provide full details in Section 6.5
- state the inclusion of participant information and consent forms (and assent forms where necessary in appendices)
- Where you are asking research participants to undertake physical activity consider appropriate health screening processes. Note that the ACSM have updated their guidelines in a consensus statement dated 2015.

Experimental procedures:

- Video footage of rowing boats on water during training sessions
- Romberg test using force platform
- Countermovement jumps on 2 force platforms to assess bilateral asymmetry.

- 2000 m rowing ergometer test on a concept 2 ergometer modified to have some instability (roll).
- Bespoke software will be used to determine neuromuscular variables from the force-time histories of participants performing a vertical countermovement jump.

Performance Measures:

- On-water footage will enable the roll of boats to be quantified during a training session.
- Romberg test will allow static balance to be quantified.
- Peak force and power will be reported from the countermovement jumps for each leg with differences in values being used in order to assess/ quantify bilateral asymmetry.
- In the 2000 m ergometer test, magnitude and frequency of roll will be recorded and measured via footage of the ergometer- tracking the movement of either end of the ergometer's rear or front base support.
- The 2000 m row will also collect power output, split times and total time.

Data Collection:

- Romberg and countermovement jump data will be collected using force platforms in the lab.
- Throughout the 2000 m row part of the study, power output data will be collected using a force transducer, while data points such as the splits and time can be obtained via the ergometer.
- Magnitude and frequency of roll will be obtained from the processing of video footage using Quintec.
- As identified in 6.3, the use of participant information sheets and consent forms will be included in order to ensure all participants have the same level of information regarding the study.

-As the study requires participants to undertake physical activity, all participants will be required to complete a PAR-Q (see Appendix 3) in order to establish whether they are (A) physically able to carry out the study and (B) suitable to be a participant.

6.5 DATA ANALYSIS TECHNIQUES

- describe briefly the techniques that will be used to analyse the data

- ANOVA will be used to measure changes in quantitative variables across the two conditions and correlation analysis will be employed to explore the relationships between the given data.
- Quintec analysis will provide roll data for both the on-water and ergometer sections of the study.

6.6. STORAGE AND DISPOSAL OF DATA AND SAMPLES
describe the procedures to be undertaken for the storage and disposal of data and samples
- identify the people who will have the responsibility for the storage and disposal of data and samples
- identify the people who will have access to the data and samples
- *state the period for which the raw data will be retained on study completion (normally 5 years, or end of award. But data should not be retained for longer than is necessary for the purposes of the research project.)*
- *Please confirm that where data is being stored away from Swansea University (for example on cloud-based services) that procedures are still in line with GDPR legislation.*

- Data will be stored on a password-secured computer with each participant identified by a number rather than by name.
- The data will be accessible by Jordan Groom and Supervisor- Nick Owen.
- Information regarding the numerical identifiers of participants will be stored on a separate spreadsheet on a password protected computer.
- All participants will be made aware that these data are kept in confidence and will be provided with their allocated participant number to use if they wish to withdraw their data.
- Hard copies of consent forms will be stored within a secure filing cabinet, in a locked office at Swansea University. Data will be held for a maximum period of 5 years (following Swansea University requirements) after the completion of the research project, as required by Research Councils. Upon completion of the 5-year period or publication of the data, anonymous electronic data files will be deleted and destroyed by the principal researcher (Jordan). Hard copies of informed consent forms will be destroyed using a confidential waste system.
- All video footage will be stored on a secure memory card that will be privately and securely stored away in a locked room or a password protected computer system. All footage is anticipated to be recorded with no participant identifiers- for example recorded from behind so that participant faces are not shown to maximize anonymity.
- Upon return of the university supplied camera and memory card, all data will be wiped to ensure data protection.

6.7 HOW DO YOU PROPOSE TO ENSURE PARTICIPANT CONFIDENTIALITY AND ANONYMITY?

- Participants' data will remain confidential to the research team and nobody outside of the study, or other participants, will have access to the data.
- No names of participants will be used in any presentations or publications.

7. LOCATION OF THE PREMISES WHERE THE RESEARCH WILL BE CONDUCTED. *- list the location(s) where the data collection and analysis will be carried out* *- identify the person who will be present to supervise the research at that location* *- If a first aider is relevant, please specify the first aider and confirm that they possess the first aid qualifications appropriate for this form of research*
- On-water footage will be collected from the Swansea university rowing club situated adjacent to the river Tawe. Recording locations will be carefully selected to ensure no members of the public are recorded/ included in the footage. - All other data will be collected in the SPEX labs on Bay Campus, Swansea University. Analysis will also be carried out on Bay campus. - Nick Owen or a member of the technical staff will be present to supervise the research.

8. POTENTIAL PARTICIPANT RISKS AND DISCOMFORTS *- identify any potential physical risk or discomfort that participants might experience as a result of participation in the study.* *- identify any potential psychological risk or discomfort that participants might experience as a result of participation in the study.* *- Identify the referral process/care pathway if any untoward events occur*
Physical risks or harms- as this study is physical and requires the participants to exert a significant amount of energy, there is the risk of fatigue, pain or discomfort. In particular, as with any erg or water session, if the rower begins to slack on their technique, they may put themselves at risk of potential injury. In the section of the study using an unstable ergometer- this will be a process that the participants will not have experience with and so the potential for harm is greater than a standard stable ergometer. However, precautions will be made to ensure the rowers have time prior to the testing stage in order to become comfortable with the way the rowing machine will deviate. With this, if the rowers are able to focus on their technique, the risk of harm will be reduced. A suitably trained first aider will be present throughout testing, usually a member of the sports science technical staff. Other data collection methods should pose no threat to participants- there is a period during the Romberg test that will require the participants to close their eyes, however, another individual will be situated close by to ensure they feel comfortable and should they lose balance they can be helped.

9.1. HOW WILL INFORMED CONSENT BE SOUGHT? *Will any organisations be used to access the sample population?* *Will parental/coach/teacher consent be required? If so, please specify which and how this will be obtained and recorded?*

- The sample population is easily accessible- as one of the university sports societies that I myself am a member of. As a result of this, an organisation will not be required.
- No specific parental/ coach/ teacher consent will be required by the participants; however, the rowing club captains have been consulted in order to check that they are happy for their rowers to be involved in the study.
- Participants in the research project will be required to sign a consent form (see Appendix 2) to give evidence of their consent to take part in the research project.

9.2 INFORMATION SHEETS AND CONSENT/ASSENT FORMS

- Have you included a participant information sheet for the participants of the study? YES (Appendix 1)
- Have you included a parental/guardian information sheet for the parents/guardians of the study? N/A
- Have you included a participant consent (or assent) form for the participants in the study? YES (Appendix 2)
- Have you included a parental/guardian consent form for the participants of the study? N/A

10. IF YOUR PROPOSED RESEARCH IS WITH VULNERABLE POPULATIONS (E.G., CHILDREN), HAS AN UP-TO-DATE DISCLOSURE AND BARRING SERVICE (DBS) CHECK (PREVIOUSLY CRB) IF UK, OR EQUIVALENT NON-UK, CLEARANCE BEEN REQUESTED AND/OR OBTAINED FOR ALL RELEVANT RESEARCHERS?.

N/A

11. HUMAN TISSUE ACT

Does your research involve the collection or storage of human tissue samples? Where not relevant please respond N/A. Where appropriate please provide further details. Please note that University ethics committee approval is not sufficient to comply with legislation for the storage of relevant material for research.

N/A

12. STUDENT DECLARATION

Please read the following declarations carefully and provide details below of any ways in which your project deviates from these. Having done this, each student listed in section 2 is required to sign where indicated.

- ***"I have ensured that there will be no active deception of participants.***
- ***I have ensured that no data will be personally identifiable.***
- ***I have ensured that no participant should suffer any undue physical or psychological discomfort (unless specified and justified in methodology).***
- ***I certify that there will be no administration of potentially harmful drugs, medicines***

or foodstuffs.

- ***I will obtain written permission from an appropriate authority before recruiting members of any outside institution as participants.***
- ***I certify that the participants will not experience any potentially unpleasant stimulation or deprivation.***
- ***I certify that any ethical considerations raised by this proposal have been discussed in detail with my supervisor.***
- ***I certify that the above statements are true with the following exception(s):***

Student/Researcher signature: JORDAN GROOM
Date: 26/02/20

Where submitted electronically we will accept the lead supervisor/researcher's email of the application as confirmation that both they and other researchers on the project have discussed and are happy to adhere to the above.