NOVEL ALGORITHMS TO CAPTURE KINEMATIC VARIABLES WITH DEPTH-SENSING TECHNOLOGY

The Development of a Reliable, Valid and Practical Movement Assessment Tool

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Patent and Publications

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Publications

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Key Abbreviations

2D – 2-dimensional

3D – 3-dimensional

ACL – Anterior cruciate ligament

AMAT – Athletic movement analysis tool

CI – Confidence interval

CL – Confidence limits

CoM – Centre of mass

LoA – Limits of agreement

ML – Medio-lateral

PA – Posterior-anterior

SDK – Software development kit

SD – Standard deviation

SI – Superior-inferior

Abstract

Movement assessment tools are widely used in practice to monitor injury risk factors in athletes. However, an issue with these tools is the trade-off between the reliability and validity on one hand and the practicality on the other hand. The Windows Kinect has been proposed as an addition to current movement assessment tools because it has depth-sensing technology and it can collect 3-dimensional kinematic data of anatomical landmarks during dynamic movements via markerless tracking. Therefore, this thesis aimed to show the development of a reliable, valid, and practical movement assessment tool (athletic movement analysis tool, [AMAT]) that makes use of depth-sensing technology and a laptop (i5 processor or higher, Windows operating system) to collect kinematic data of athletes during dynamic movements. To that purpose, the first study discussed the strengths and weaknesses of the Windows Kinect. Moreover, it described the development of algorithms to improve the kinematic data collection with the Kinect. Foot markers with retroreflective markers were developed and an algorithm was developed to track these markers with the Kinect. Moreover, an algorithm was developed to calculate the position of the centre of mass and an algorithm was developed to determine the frame of initial contact. In the next study, the foot markers and normal retroreflective markers were placed on 18 different positions within the camera view to determine the static reliability and validity of the foot marker tracking algorithm when compared to Vicon. The technological error of the AMAT ranged from 1.36 millimetres on the positions closest to the camera to 3.30 millimetres on the positions furthest away from the camera. The mean difference between the marker positions as measured with AMAT and Vicon system ranged from -4.51 to 16.23 millimetres. These outcomes imply that the foot marker tracking is reliable and valid to track the markers in static situations,

but it was shown that the AMAT is less reliable when collecting data on positions further away from the Kinect. In the third study, the foot marker tracking algorithm and a tape measure were used to measure jump distances, to determine whether the AMAT is able to track the foot markers during dynamic situations. The mean differences between the AMAT and manual measurements were trivial (-1.69 to 2.41 millimetres). This implies that the foot marker tracking algorithm is valid to track the foot markers during dynamic movements. In the fourth study, the algorithm that calculates the centre of mass position (centre of mass algorithm) was validated by comparing the outcomes of this algorithm with the centre of mass position collected with Vicon during horizontal jumps. The correlations were *moderate* to *extremely high* in the medio-lateral axis (0.65 to 1.00), *extremely high* in the posterior-anterior axis (0.99 to 1.00) and *trivial* to *extremely high* in the superior-inferior axis (-0.08 to 0.98). In the last study of this thesis, the standing broad jump performance and the ability to maintain balance of adolescent female soccer players were monitored during a full season with the AMAT. Here it was found that the within-subject variability of the jump performance was *small* to *moderate* and that a *substantial* improvement in jump performance was found over the season. The within-subject variability of the ability to maintain balance ranged from moderate to extremely large. This implies that the jump performance can be reliably collected with the AMAT, whereas the ability to maintain balance cannot be collected in a reliable manner. In total, this thesis showed that the AMAT can reliably and validly track the foot markers and determine the position of the centre of mass. However, more research on other algorithms remains necessary to determine whether kinematic data of other anatomical landmarks can also be collected in a reliable and valid manner.

1.1 Introduction

Injuries are a large problem in youth soccer, especially in players around the peak height velocity and in the oldest age groups (Price *et al.,* 2004; van der Sluis *et al.,* 2014; 2015; Kemper *et al.,* 2015; Pfirrmann *et al.,* 2016; Read *et al.,* 2017a; Tears *et al.,* 2018). As such, clubs use movement assessment tools to collect information on injury risk, diagnosis, treatment and rehabilitation (Alentorn-Geli et al., 2009; McCall *et al.,* 2014; Light *et al.,* 2018; Read *et al.,* 2018a). Most clubs use these tools two to three times per season, generally during pre-, mid- and at the end of the season (Read *et al.,* 2018a). The data collected with these tools are used to monitor changes in movement performance and movement quality over time and can be used to develop individually-based training programs to reduce injury risk of athletes (Myer et al., 2004; Gokeler et al., 2017; Read *et al.,* 2018a).

An issue with current movement assessment tools is the trade-off between the reliability and validity on one hand and practicality on the other hand (Elliot & Alderson, 2007; Ehrenbrusthoff *et al.,* 2016; Colyer *et al.,* 2018). Depth-sensing technology could be the solution for this issue (Clark *et al.,* 2012; Dutta, 2012; Bonnechere *et al.,* 2014). Depth-sensing technology uses time of flight and/or speckled infrared patterns to determine the 3-dimensional (3D) position of pixels from the camera image in real time (Lachat *et al.,* 2015). Consequently, it becomes possible to collect 3D kinematic data of athletes with this technology. In addition, this technology is low in cost and cameras equipped with this technology are portable. As such, this technology is suitable for use in practice.

Movement assessment tools are used in soccer because of the relationship between the sensorimotor system and lower extremity injuries in soccer (i.e. sensorimotor risk factors of lower extremity injuries [Read *et al.,* 2016a; 2016b]). For example, it has been found that movement strategies used during dynamic movements (Read *et al.,* 2016a; 2016b), (training) load, fatigue and recovery (Brink *et al.,* 2010; Bowen *et al.,* 2017), and maturation and age (van der Sluis *et al.,* 2014; 2015 Kemper *et al.,* 2015) were related to lower extremity injury risk. Interestingly, Read *et al.* (2015) argued that the sensorimotor system is related to all those injury risk factors. More specifically, it has been argued that changes in movement strategies are due to alterations in the sensorimotor system (Alentorn-Geli *et al.,* 2009a). In addition, fatigue causes delayed muscle recruitment (De Ste Croix *et al.,* 2015) and decreased feedforward mechanisms (Mello *et al.,* 2007), which have both been described as part of the sensorimotor system (Read *et al.,* 2015). An important issue that occurs during maturation is that the body segments grow in an earlier stage than the muscles and that this results in a higher position of the centre of mass and greater forces working on the body (Tanner, 1978; Myer *et al.*, 2008a). However, due to the underdeveloped muscles, it becomes more difficult for the body to maintain balance (Tanner, 1978). Read *et al.* (2015) argued that this might affect the movement strategies used by adolescents which might consequently increase injury risk.

The sensorimotor system uses feedback and feedforward mechanisms to activate muscles to keep the body and individual joints in balance (Winter, 1995; Riemann & Lephart, 2002a; Horak, 2006). The sensorimotor system can be divided into the sensory system, the central nervous system and the neuromuscular system (Riemann & Lephart, 2002a). The sensory system is responsible for sensing changes in the body posture relative to the environment, the central nervous system is responsible for processing the information provided by the sensory system and for deciding which muscles need to be recruited, and the neuromuscular system is responsible for activating the correct muscles at the right time to maintain joint stabilisation and to keep the body in balance (Riemann & Lephart, 2002a).

Multiple measurement tools have been developed to quantify sensorimotor risk factors of lower extremity injuries. For example, measurement tools have been developed to quantify the sensory system by assessing the working of the proprioceptive, vestibular and visual systems (Riemann & Guskiewicz, 2000; Riemann *et al.*, 2002). Similarly, tools that quantify the neuromuscular system have also been described in the literature, such as the measurement of muscle activation patterns to determine when specific muscles get activated, the muscle force output, the collection of biomechanical data during dynamic movements to assess movement quality and the ability to maintain balance (Riemann *et al.*, 2002). These measurement systems can be used in prospective studies to determine the relationship between sensorimotor risk factors and lower extremity injuries (Riemann *et al.*, 2002; Hewett *et al.*, 2012).

The measurement tools described by Riemann & Guskiewicz (2000) and Riemann *et al.*, (2002) are laboratory based. As such, these tools are not useful for practitioners, because these laboratory-based systems are expensive (Bardid *et al.*, 2018), suitable for small sample sizes only (Bardid *et al.*, 2018) and often not practical in use (Read *et al.*, 2017b; 2017c; Bardid *et al.*, 2018; Blair *et al.*, 2018; Colyer *et al.*, 2018). The collection of biomechanical data during dynamic movements is more practical and can also provide insight in sensorimotor risk factors of lower extremity (Hewett *et al.*, 2004; 2005; Myer *et al.*, 2005; 2006; Ford *et al.*, 2005; Fransz *et al.*, 2016; Read *et al.*, 2017b). A benefit of biomechanical data collection is that the outcome measures can be relatively easy translated into practice, because the biomechanical data can be directly explained in terms of movements (Hewett *et al.*, 2010). As such, the biomechanical data can be used to develop prevention strategies that focus on the improvement of movements and coordination.

The biomechanical analysis of dynamic movements should provide information on the movement performance (e.g. jump height, jump distance) and on the movement quality (e.g. joint angles, joint moments) (Paterno *et al.,* 2012; Hildebrandt *et al.,* 2015; Wilk, 2015; Gokeler *et al.,* 2017). For example, measuring the jump distance of horizontal jumps and hops provides information about the muscle power of the hip flexors and plantar flexors (Noyes *et al.,* 1991; Markovic *et al.,* 2007). In addition, determining the position of the centre of mass or centre of pressure relative to the base of support during the landing can provide information on the ability to maintain balance (Winter, 1995; Horak, 2006; Goffredo *et al.,* 2006; Allin *et al.,* 2008; Fransz *et al.,* 2015). Similarly, several types of vertical jumps can be used to assess knee extensor and plantar flexor strength (Spagele *et al.,* 1999; Bradley *et al.,* 2007; Chappel *et al.,* 2007; De Ruiter *et al.,* 2007; Earp *et al.,* 2010) and the landing of these movements is frequently used to quantify anterior cruciate ligament injury risk via the assessment of knee abduction movement, trunk movement and positioning of the feet during the landing (Hewett *et al.,* 2005; 2006a; Myer *et al.,* 2008b; 2010; Padua *et al.,* 2009; Dingenen *et al.,* 2015c; Read *et al.,* 2017b).

There are currently two main methods used by practitioners to collect biomechanical data during dynamic movements (Read *et al.,* 2018a). The first main method is the combination of laboratory-based marker 3D systems and force plates. This method is the gold standard for biomechanical data collection and as such often used for this purpose (Smith *et al.,* 2008; Azevedo *et al.,* 2009; Padua *et al.,* 2009; van Diest *et al.,* 2014; Augustus & Smith, 2015; MacPherson *et al.,* 2016; Augustus *et al.,* 2017). There are several issues with the use of marker-based systems. Namely, markers can cause physical and/or psychological constraints on the participant during certain movements, the attaching of the markers causes longer preparation times and there is a potential for erroneous marker placement (Ceseracciu *et al.,* 2014; Colyer *et al.,* 2018). Moreover, laboratory-based systems have limited portability, require complex set-ups, are constrained to small test areas and are confined to one testing location per system (Blair *et al.*, 2018). This is reflected in the fact that only a minority of youth soccer academies have access to these systems (Read *et al.*, 2018a). Nevertheless, force plates are often recommended in practical movement assessment tools to quantify the ability to maintain balance (Dingenen *et al.*, 2016a; Fransz *et al.*, 2016; Read *et al.*, 2017b).

The second main method to assess movements of athletes in a practical environment is by using human raters (Myer *et al.*, 2004; Padua *et al.*, 2009; Gokeler *et al.*, 2017; Read *et al.*, 2017b; Welling *et al.*, 2018a). The human rater observes the movement and uses a scoring sheet to assess the movement (Padua *et al.*, 2009; Read *et al.*, 2017b). Here, the human rater can either score movements in real time (i.e. while the movement is performed [Cook *et al.*, 2006a; 2006b; Padua *et al.*, 2011]) or record the movement and analyse the movements afterwards (Padua *et al.*, 2009; Read *et al.*, 2017b). An issue with movement assessment tools that are scored in real time is that they collect only a few variables per movement (e.g. the Functional Movement Screen [Cook *et al.*, 2006a; 2006b]; the Landing Error Scoring System real-time [Padua *et al.*, 2011]). This might result in a low sensitivity to changes (Wright *et al.*, 2018). The use of video analysis can provide more detailed information about the movement when compared to a real-time score. However, it is more time consuming compared to real time assessment because the practitioner needs to re-watch all movements. Moreover, an issue with the use of video analysis to assess the movement strategies during movements is that the wrong plane of movement can be used due to the use of 2-dimensional (2D) joint positions (Colyer *et al.*, 2018). Namely, movements outside of the principle plane of motion are not included in the analysis which can lead to miscalculations of the principal plane (Colyer *et al.*, 2018). As such, the validity of this 2D video analysis is questionable.

Depth-sensing technology might overcome the methodological issues of current movement assessment tools. The Windows Kinect is a camera with depth-sensing technology and has been recommended as part of a movement assessment tool (Clark *et al.,* 2012; Dutta, 2012; Bonnechere *et al.,* 2014). Benefits of the Windows Kinect are its software development kit, which makes it easy to incorporate other software, for example to make the data collection more convenient (Bujang *et al.,* 2015), its price compared to other high-speed camera systems used for kinematic data collection and compared to Vicon (Pueo, 2016), and its ability to collect the 3D position of 25 anatomical landmarks without the use of any markers (i.e. markerless tracking [Wang *et al.,* 2015]).

The possibilities to collect kinematic data during dynamic movements with the depth-sensing technology of the Kinect have been explored widely (Kharazi *et al.*, 2015; Mentiplay *et al.,* 2015; Auvinet *et al.,* 2017; Eltoukhy *et al.*, 2017). As such, it might be possible in the future to collect kinematic data with depth-sensing technology that is both reliable and valid, practical, and able to quantify the ability to maintain balance, the movement strategies used and the muscle force output (Clark *et al.*, 212; Dutta, 2012; Bonnechere *et al.*, 2014; Colyer *et al.*, 2018). However, two issues exist with regards to the Kinect.

At first, several studies have reported validity issues with regards to the markerless tracking of the anatomical landmarks. For example, Van Diest *et al.* (2014) and Wang *et al.* (2015) reported that the data collection of the feet is not reliable. In addition, peak angles of lower extremity joints collected with depth-sensing technology during walking are not valid (Kharazi *et al.,* 2015; Mentiplay *et al.,* 2015; Auvinet *et al.,* 2017; Eltoukhy *et al.*, 2017). Furthermore, data of the lower extremity is not collected in a valid manner during daily activity tasks (Otte *et al.,* 2016). This shows that the kinematic data collection with depth-sensing technology is not valid to be used in a movement assessment tool. The second issue is that there are currently no practical movement assessment tools that make use of depth-sensing technology. Several studies have described novel methods to use depth-sensing technology as part of a movement assessment tool (e.g. Paolini *et al.,* 2014; Giblin *et al.,* 2016; MacPherson *et al.,* 2016; McGroarty *et al.* 2016), but none of these studies have described a system that is practical in use. As such, more research is necessary to develop a movement assessment tool that makes use of depth-sensing technology.

1.2 Aims and Objectives

This thesis is part of a Knowledge Transfer Partnership (Innovate UK) project where Teesside University cooperated with Pro Football Support to develop the athletic movement analysis tool (AMAT). The AMAT uses the Windows Kinect v2 to collect kinematic data during dynamic movements. The AMAT was developed with the aim to assess movements in a reliable, valid, objective and practical manner. The assessment of these movements could then be related to sensorimotor risk factors of injuries and to develop individual training programs to improve performance and reduce injury risk of athletes. The aim of this thesis was to develop the AMAT and to determine the reliability and validity of the AMAT. Due to time constraints, this thesis will only focus on the ability of the AMAT to collect kinematic data of the feet and the centre of mass during horizontal jumps.

Following this introduction, the thesis can be divided into three sections. The main aims of the first section are to create a rationale behind the development of the AMAT (Chapter 2) and to describe the development of the AMAT (Chapter 3). This includes explanations of the depth-sensing technology used, the movements included in this tool, the discussions with practitioners to improve the AMAT and the development of new algorithms to improve the kinematic data collection with the depth-sensing technology.

The second section consists of Chapter 4, 5, 6 and 7. The main aim of this section is to determine the reliability and validity of three algorithms developed in Chapter 3 during horizontal jumps. More specifically, in Chapter 4 the static reliability and validity of the foot marker tracking algorithm is determined. The dynamic validity of the foot marker tracking algorithm is determined in Chapter 5. This chapter also determined the validity of the algorithm that calculates the landing position. In Chapter 6 the dynamic validity of the algorithm that collects the position of the centre of mass is determined. In Chapter 7 the AMAT is used to monitor the standing broad jump performance throughout a full soccer season (August – April). Here, the within-subject variability and seasonal variability of the jump distance and ability to maintain balance in female soccer players are determined.

The third section consists of Chapter 8, the discussion chapter. Here, all findings are summarized, strengths and weaknesses of the thesis are discussed and recommendations for future research and usage of the AMAT are made. The individual aims and objectives per chapter are displayed in Table 1.1

Table 1.1. Aims and objectives of this thesis.

Aim 1	To develop a new movement assessment tool that makes use of depth-sensing technology and that can be used in a practical setting.	*Chapter 3*
Objective 1	New algorithms were developed to collect 3D kinematic data of anatomical landmarks reliably and validly. Moreover, an app was developed for easy use of the tool in a practical environment.	
Aim 2	To determine the reliability and validity of the foot marker tracking algorithm of the AMAT in a static setting.	*Chapter 4*
Objective 2	The 3D positional data of foot markers were collected with the AMAT to determine the reliability of the foot marker tracking algorithm, and the validity of this algorithm when compared to a laboratory-based marker tracking system.	
Aim 3	To determine the validity of the AMAT to track the foot markers in a dynamical setting.	*Chapter 5*
Objective 3	Participants performed five different types of jumps. The jump distance measured with the AMAT was compared with the jump distance measured manually.	
Aim 4	To determine the validity of the centre of mass algorithm of the AMAT	*Chapter 6*
Objective 4	The centre of mass data of 2 participants was collected during horizontal jumps with AMAT and with a laboratory-based marker tracking system and the positional data collected with both systems was compared.	
Aim 5	To use the AMAT to determine the within-subject and seasonal variability of female adolescent soccer players in standing broad jump performance and ability to maintain balance.	*Chapter 7*
Objective 5	The standing broad jump performance and the ability to maintain balance were collected with the AMAT during nine sessions throughout the season.	

2.1 Introduction

Movement assessment tools are widely used in soccer to determine whether players have an increased injury risk (McCall *et al.*, 2014; Light *et al.*, 2018; Read *et al.,* 2018a). Based on the information collected with these tools, individually-based training programs can be developed to reduce the injury risk of players (McCall *et al.,* 2014; Read *et al.,* 2018a). Youth players sustain on average 0.4 – 2.2 injuries per season and are absent approximately 20 – 30 days per year due to injuries (Price *et al.,* 2004; Le Gall *et al.,* 2006; Deehan *et al.,* 2007; Brito *et al.,* 2012; Read *et al.,* 2017a; Tears *et al.,* 2018). Injuries can lead to reduced performance (Drew *et al.,* 2017), reduced motor skills and reduced muscle force output (Bullock-Saxton *et al,* 1994; Hurley, 1997; Konradsen *et al.,* 1998; Croisier & Crielaard, 2000; Osternig, 2000; Friel *et al.,* 2006; Lee *et al.,* 2009; Thomas *et al.,* 2013; 2015) and due to the relationship between the number of training hours and performance and development (Newell & Rosenbloom, 1981, Ericsson *et al.,* 1993, Elferink-Gemser *et al.,* 2011), injuries might also affect the performance and development in the long term.

An issue with current movement assessment tools is the trade-off between the reliability and validity on one hand and the practicality on the other hand (Elliot & Alderson, 2007; Ehrenbrusthoff *et al.,* 2016; Colyer *et al.,* 2018). In addition, there is a discrepancy between the movement assessment tools recommended by Read *et al.* (2017b) and the tests used in youth academies as found by Read *et al.* (2018a). As such, the use of movement assessment tools in practice is not optimal and more research is necessary to improve the reliability, validity and practicality of these tools. The *Sequence of Prevention* model developed by van Mechelen *et al. (*1987; 1992) could be a useful guideline to improve the usage of movement assessment tools in practice. At first because it provides an overview of the current injury problem in youth soccer. Secondly, it can also help to determine whether a new movement assessment tool has contributed to the injury prevention process in youth soccer.

The first step of the sequence of prevention model aims to determine the extent of the injury problem. Players around the peak height velocity (van der Sluis *et al.,* 2014; 2015; Kemper *et al.,* 2015) and players in the oldest age groups (Price *et al.,* 2004; Read *et al.,* 2017a; Tears *et al.,* 2018) have the highest injury risk in youth academies. Determining the injury burden, the number of days absence divided by 1000 hours of exposure, has been proposed as the best way to determine which injuries are most severe in a specific population (Bahr *et al.,* 2018). In youth soccer, overuse injuries (e.g. Osgood Schlatter [between the ages of 10 and 13] and Sever's disease [between the ages of 11 and 15]), hamstring injuries and knee and ankle ligament injuries are the injuries with the highest injury burden (Price *et al.,* 2004; Read *et al.,* 2017a; Tears *et al.,* 2018). A similar trend exists in professional soccer, with hamstring, knee ligament (medial collateral and anterior cruciate ligament) and ankle ligament (anterior talofibular ligament) injuries having the highest injury burdens (Bahr *et al.,* 2018). Besides the short-term effects of injuries on performance, an additional issue with ligament injuries are the long-term effects on health and performance outcomes. Namely, ACL injuries can result in the ending of a career or lower level of play after rehabilitation (Ardern *et al.,* 2011, Waldén *et al.,* 2016). Moreover, knee ligaments often result in cartilage damage (Potter *et al.,* 2012), which can cause osteoarthritis in the long term (Lohmander *et al.,* 2007; Maffulli *et al.,* 2010a; 2010b). Ankle ligament injuries can result in reduced ankle range of motion and chronic ankle instability, which has been related to reduced performance parameters and increased injury risk (Hertel, 2000; Mattacola & Dwyer, 2002).

The second step of the sequence of prevention model aims to determine the mechanisms of the injuries. Based on the large number of ligament injuries in soccer and the long-term effects on health and performance outcomes, it is decided to focus on the mechanisms of ligament injuries. Several studies have described how changes in the sensorimotor system could affect injury risk (Hewett, 2000; Alentorn-Geli *et al.,* 2009b; 2015; Hewett *et al.,* 2016a; 2016b). In addition, Read *et al.* (2015) discussed that *growth and maturation*, *movement skill*, *fatigue* and *injury history* are the four largest injury ligament risk factors in youth soccer players and related these factors to the sensorimotor system. In a systematic review, Dallinga *et al.* (2012) determined the reliability, validity, sensitivity and specificity of screening tools for injury prediction in team sports. They showed that general joint laxity might be a predictive measure for injuries to the lower extremities. Moreover, knee hyperextension, increased valgus motion and increased valgus moment were related to increased anterior cruciate ligament injury risk and postural sway was related to ankle injury risk (Dallinga *et al.,* 2012). Similarly, Read *et al.* (2017b) related the knee valgus motion and postural sway amongst the largest injury risk factors in youth soccer players.

Based on the risk factors found in step 2, the sequence of prevention model recommends to introduce preventive measures in the third step. Assessing movement strategies with movement assessment tools is recommended to determine the risk of individuals to sustain ligament injuries, because the movements assessed with these tools have been associated with sensorimotor risk factors of lower extremity injuries (Dallinga *et al.,* 2012; Read *et al.,* 2017b). Accordingly, many youth academies screen their players for injury risk and include prevention programs as part of their training sessions (Read *et al.,* 2018a). However, as mentioned previously in this chapter, there is a trade-off between reliability and validity on one hand and the practicality on the other hand of movement assessment tools (Elliot & Alderson, 2007; Ehrenbrusthoff *et al.,* 2016; Colyer *et al.,*

2018) and there is a discrepancy between the movement assessment tools recommended by Read *et al.* (2017b) and the tests used in youth academies as found by Read *et al.* (2018a).

In the fourth and last step of the model it is recommended to assess the effectiveness of the prevention methods. The number of injuries (Read *et al.,* 2017b) and injury burden (Tears *et al.,* 2018) have increased in youth soccer over the last few years and Pfirrmann *et al.* (2016) recently mentioned the high number of injuries in youth soccer also. This shows that current prevention methods are not optimal to prevent injuries. According to the sequence of prevention model, it is recommended to go back to step 2 of the model and focus on the injury risk factors to eventually develop new prevention methods in step 3.

Based on the previous paragraphs, it becomes clear that the sensorimotor system has been frequently related to ligament injury risk and that movement assessment tools can be used to screen for sensorimotor factors of injury risk (Hewett *et al.,* 2006a; 2016a; 2016b; Alentorn-Geli *et al.,* 2009b; Dallinga *et al.,* 2012; 2015; Read *et al.,* 2015). However, there is a discrepancy between the movement assessment tools recommended by Read *et al.* (2017b) and the tests used in youth academies as found by Read *et al.* (2018a) and there is a trade-off between the reliability and validity of movement assessment tools on one hand and their practicality on the other hand (Ehrenbrusthoff *et al.,* 2016). This shows that current movement assessment tools are not optimal for use in practice. To understand the issues of current movement assessment tools in more detail, section 2.2 will discuss the working of the sensorimotor system. Thereafter, section 2.3 will show how the sensorimotor system is related to several injury risk factors. This is followed by an overview on how current movement assessment tools screen for injury risk in section 2.4. The information of these three sections could then be used to develop a new movement assessment tool that is reliable, valid and practical in use.

2.2 Sensorimotor System

The sensorimotor system is a dynamical system that uses feedback and feedforward mechanisms to activate muscles to keep the body and individual joints in balance (Lephart *et al.,* 2000; Riemann & Lephart, 2002a). The sensorimotor system can be viewed as a control system. A control system "is an arrangement of physical components connected or related in such a manner as to command, direct, or regulate itself or another system" (Di Stefano *et al.,* 1976).

Control systems use one or more *input* variables to produce a specific response (Di Stefano *et al.,* 1976). This response is also defined as the *output* of the system (Di Stefano *et al.,* 1976). The way the input and output interact with each other is dependent on the type of control system. In an *open-loop* control system, the relation between the input and output needs to be calibrated to get the desired outcome (Di Stefano *et al.,* 1976). In contrast, in a *closed-loop* control system the difference between the expected and desired output is used as a feedback mechanism to adjust the input, to eventually get the desired outcome (Di Stefano *et al.,* 1976; Leigh, 2004). This implies that over time, a closed system will approach a stable state where almost no action occurs (Di Stefano *et al.,* 1976). However, it should be noted that the range of adjustments is usually constrained, which implies that not all original inputs can always result in a stable outcome (Jacobs, 1996).

The sensorimotor system is a closed-loop system. To explain the working of the sensorimotor system in a clear way, an analogy will be drawn with a thermoregulation system. In a thermoregulation system, the temperature is the input and is registered by the temperature sensor and the temperature is translated into a binary number (Di Stefano *et al.,* 1976; Leigh, 2004). The input of the sensorimotor system includes the acquisition of all sensor stimuli, the conversion of the stimuli to neural signals and the transmission of these signals via afferent pathways to the central nervous system (Lephart *et al.,* 2000;

Horak, 2006). In the thermoregulation system, a computer determines whether the actual temperature is higher, equal or lower than the desired temperature (Di Stefano *et al.,* 1976; Leigh, 2004). In the sensorimotor system, the central nervous system receives signals from the sensory system and determines the position of the body relative to the environment and decides which muscles need to be activated (Lephart *et al.,* 2000). The last step in the thermoregulation system is to increase or reduce the flow of hot water to the radiators based on the difference between the actual and the desired temperature (Di Stefano *et al.,* 1976; Leigh, 2004). The last step in the sensorimotor system is that the information received by the neuromuscular system results in the activation of different muscles at the correct time and with the correct intensity to maintain balance (Lephart *et al.,* 2000; Riemann & Lephart, 2002a). In both systems, this process is dynamical and ongoing, for the thermoregulation system to keep the temperature stable and for the sensorimotor system to keep the body in balance. The next subsections will give an overview of the input (i.e. sensory system), the processing (i.e. central nervous system) and the output (i.e. neuromuscular system) of the sensorimotor system. Thereafter, it will be described how biomechanical variables are involved in this process.

2.2.1 Sensory system.

The input of the sensory system is based on feedback and feedforward mechanisms (Riemann & Lephart, 2002a). The feedback mechanisms work continuously and get input from the somatosensory (i.e. tactile, pain, temperature, proprioception), vestibular and visual systems (Ghez, 1991; Riemann & Lephart, 2002a). Healthy persons rely for approximately 70% on the somatosensory, 20% on the vestibular and 10% on the visual system in a well-lit environment with a firm base of support (Peterka, 2002; Horak, 2006).

Tactile sensing has been defined as a system that can measure a given property of an object or contact event, through physical contact between the system and the object (Lederman, 1982; Dargahi & Najarian, 2004). Touch, tickle, pressure and vibration are viewed as part of the tactile system by Riemann & Lephart (2002a), whereas Dargahi & Najarian (2004) also include temperature and pain as part of the tactile system. To discuss whether pain and temperature are part of the tactile system or not is beyond the scope of this literature review. However, it should be noted that specific receptors exist for the touch, tickle, pressure and vibration (i.e. mechanoreceptors), for pain (i.e. nocioreceptors) and for temperature (i.e. thermoreceptors) (Dargahi & Najarian, 2004). All these receptors are located in the skin throughout the full body and send signals to the central nervous system when activated (Dargahi & Najarian, 2004).

Proprioception encompasses the joint position sense, the sense of resistance and the sense of movement (i.e. kinaesthesia) (Riemann & Lephart, 2002a). Different types of proprioceptors are located in muscles, tendons, ligaments and capsules (Sherrington, 1906; Riemann & Lephart, 2002a). For example, *Ruffini receptors* are based in ligament and capsular tissue and can provide information about the static joint position and amplitude and velocity of joint rotations. *Pacinian corpuscles* are located in the capsular tissue of joints and are sensitive to accelerations and decelerations. *Golgi tendon organs* are located in musculotendinous tissue and give information about muscle tension during contraction. M*uscle spindles* are wrapped around muscle fibres and give information about the muscle length and rate of changes in length (Johansson *et al.,* 2000; Riemann & Lephart, 2002a).

The vestibular system is located in the inner ear and consists of three semi-circular canals in three different axes (medio-lateral [ML], superior-inferior [SI] and posterior-anterior [PA]) and two otolith organs (utricle and saccule) (Agrawal *et al.,* 2009). It senses angular and linear accelerations of the head and uses this information to determine the movement of the head in space (Minor, 1998; Agrawal *et al.,* 2009; Lopez & Blanke, 2011). Moreover, it plays an important role in the movement of the eyes (i.e. oculomotor control) and in postural control (Lopez & Blanke, 2011).

The visual system can be used to detect the motion of oneself or some form of motion in the environment (Sheldon, 1963; Clement *et al.,* 1983; Redfern *et al.,* 2001). The visual system is mainly used during very slow movements (Dichgans *et al.,* 1976; Lestienne *et al.,* 1977; Redfern *et al.,* 2001). Although the visual system is in general the least important sensory system, several studies have shown that changes in vision affect the ability to maintain balance (Dingenen *et al.,* 2015a; 2015b; 2016b).

The feedforward mechanisms are anticipating on actions that will occur and can disrupt the balance of the body but are not detected by any sensory feedback mechanism yet (Ghez, 1991; Johansson & Magnusson, 1991; Riemann & Lephart, 2002a). This implies that feedforward mechanisms work intermittently until the feedback mechanisms take over (Ghez, 1991; Collins & De Luca, 1993; Riemann & Lephart, 2002a). An example of a feedforward mechanism is the stabilisation of muscles before initiating the movement to remain stable (Page, 2006).

All feedback and feedforward systems transform their input into neural signals and are sent to the central nervous system via afferent pathways (Horak, 2006). The processes involved in the central nervous system will be described in the next section.

2.2.2 Central nervous system.

In the central nervous system, all signals are merged together and processed to understand the current interaction between the environment and the body and to determine what actions should be taken to maintain balance (Horak, 2006). This process occurs in three different levels of the central nervous system, namely the spinal cord, the lower brain regions and the cerebral cortex (Biedert, 2000). The spinal cord is the lowest level of motor control and is responsible for quick motor and sympathetic reflexes to maintain joint stabilisation and for elementary patterns of motor coordination (Biedert, 2000; Riemann & Lephart, 2002a). The brain stem is the second level of motor control and it contains the circuits that are responsible for postural equilibrium and the automatic and stereotype body movements (Ghez, 1991; Matthews, 1997; Mihailoff *et al.*, 1997; Riemann *et al.*, 2002). The third level of motor control is the motor cortex, located in the cerebral cortex (Riemann & Lephart, 2002a). The motor cortex consists of three areas, namely the primary motor cortex, the premotor area and the supplemental motor area. They are responsible for (1) the reception of peripheral afferent information and the encoding of muscles to be activated, (2) the reception of afferent information and the organization and preparation of motor commands and (3) the programming of complex sequences of movements that involve groups of muscles, respectively (Riemann & Lephart, 2002a). After information is processed in the nervous system, signals are sent to motor units via efferent pathways.

2.2.3 Neuromuscular system.

The neuromuscular system uses the signals it received from the central nervous system to activate motor units, to eventually maintain balance (Riemann & Lephart, 2002a). A motor unit consists of one motor neuron, located in the central nervous system, a motor axon and multiple muscle fibres (Enoka, 2008). When a motor unit is recruited via an action potential, all fibres within the unit are activated (Rosenbaum, 2009; Winter, 2009). The first motor units that are activated have in general the smallest muscle fibres and are the least forceful. Over time, larger and more forceful motor units will be activated (Henneman *et al.*, 1965; Rosenbaum, 2009). The motor neurons need to fire at a higher firing rate (i.e. overcome a threshold) before the larger motor units become active (Enoka, 2008). This process is similar in all muscles throughout the body that aid in maintaining balance.

When a perturbation of the body (i.e. a lack of balance) is detected by the sensory system, the body has three different strategies to regain balance (Winter, 1995; Horak, 2006). For small balance perturbations, ankle plantar or dorsi-flexors become active, for bigger balance perturbations, hip flexors or hip extensors also become active and for the largest perturbations, the person can shift a foot to maintain balance (Winter, 1995; Horak, 2006). Other strategies can aid in the process to maintain in balance. For example, the *pre-activation* of muscles occurs for the muscles to be activated on time (Jones and Watt, 1971). This is because an *electromechanical delay* (i.e. time between onset of electrical activity and measurable tension of muscle) exists in all muscles (Cavanagh & Komi, 1979). Due to this electromechanical delay, muscles cannot respond on time to aid in maintaining balance, whereas this pre-activation already anticipates on the movement that is coming (Jones and Watt, 1971; Horita *et al.,* 2002). A strategy involved with the muscle pre-activation is muscle stiffness (Serpell *et al.,* 2014). Muscle stiffness is defined as the ratio of change in force per change in length (McNair *et al.,* 1992; Johansson & Sjolander, 1993; Riemann *et al.,* 2002) and is closely related to joint stiffness, which includes the stiffness of all structures over the joint (Johns & Wright, 1962; Sinkjaer *et al.,* 1988; Helliwell, 1993; Riemann *et al.,* 2002). Stiffer joints are better able to resist sudden joint displacement and muscle stiffness can aid in this process (Grillner, 1972; Louie & Mote, 1987; McNair *et al.,* 1992; Johansson & Sjolander, 1993; Riemann & Lephart, 2002b). Another strategy to maintain balance is co-contraction, where the antagonist muscles are also activated during a movement to increase the stability of the joint (Zhang & Wang, 2001; Frost *et al.,* 2002).

2.2.4 The output of the sensorimotor system.

To maintain balanced, the centre of mass must remain within the base of support (Winter, 1995; Horak, 2006). For humans this implies that when the feet are in contact with the ground, the centre of mass must remain inside the surface area that the feet create on the floor. As such, the main aim of the sensorimotor system is to keep the centre of mass within the base of support. An issue with maintaining balance is the bipedal gait Homo Sapiens use (Wittmann & Wall, 2007). This is more difficult than quadrupled movement because of a smaller base of support and a higher centre of mass position. This smaller base of support results in the centre of mass being relatively quickly outside the base of support. In addition, a high centre of mass results in more linear centre of mass displacement by a similar angular displacement (Figure 2.1). Figure 2.2 is adopted from Winter (1995) and shows how the centre of pressure is moved relatively to the centre of mass to change the direction of the angular acceleration during standing still to maintain balance. The different sensory systems are used to determine the movement of the centre of mass, whereas the muscles are used to control the movement of the centre of mass by adjusting the centre of pressure (Winter, 1995; Horak, 2006). This shows that obtaining the centre of mass movement relative to the base of support and the centre of pressure displacement can provide valuable information about the ability of one to maintain in balance and thus about the working of the sensorimotor system.

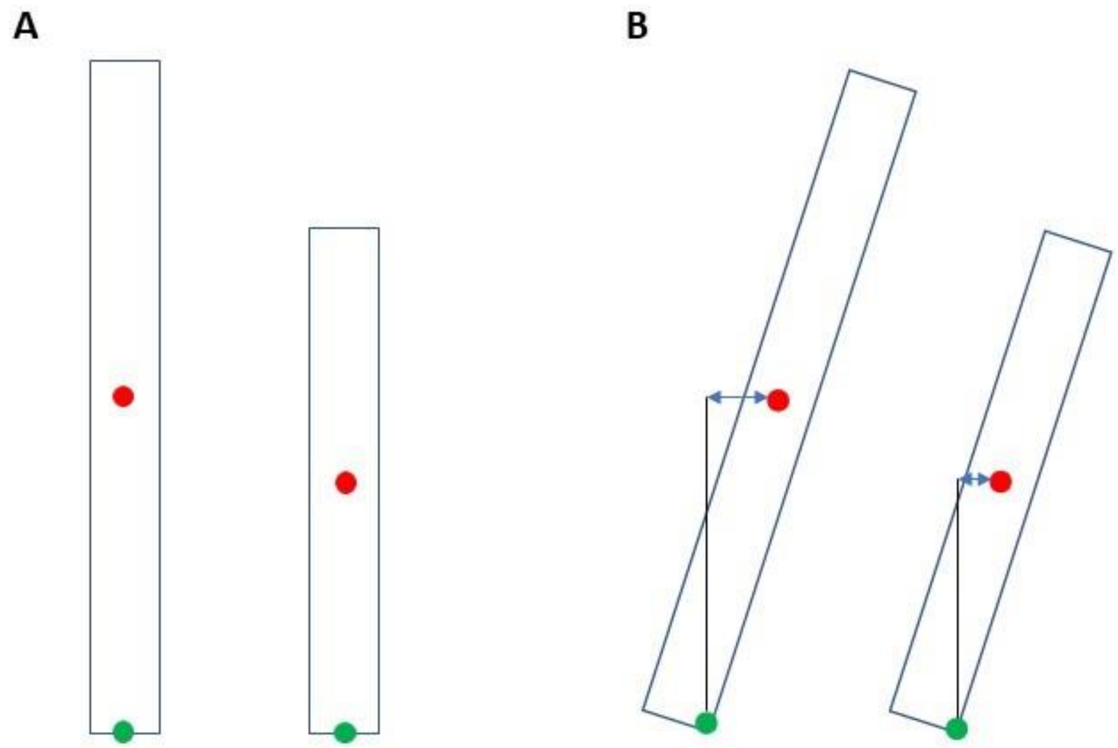

Figure 2.1. Two rectangles with a different height. The red dot represents the centre of mass and the green dot represents the centre of the base of support. Figure 2.1A. Two rectangles standing upright. Figure 2.1B. Both rectangles rotated 18 degrees. The centre of mass displacement of the taller rectangle is larger than the centre of mass displacement of the shorter rectangle.

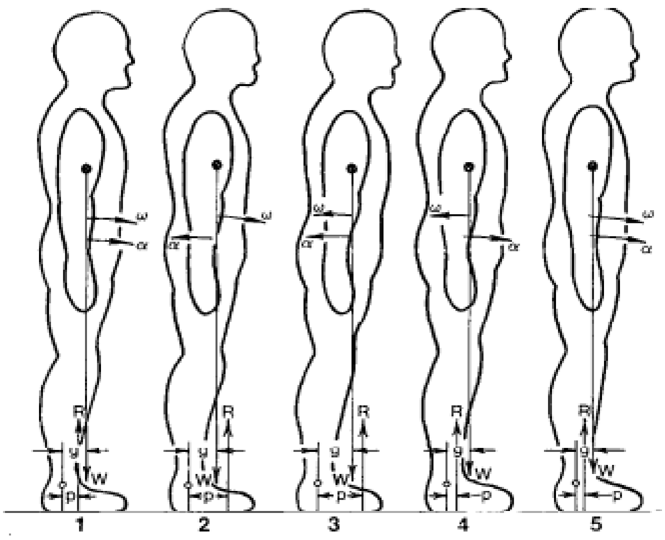

Figure 2.2. The strategy of a person to remain in balance while standing quiet on a force platform during five different points in time. On the first time point, the angular velocity (ω) and angular acceleration (α) are assumed to be clockwise, because the centre of mass (W) is anterior to the centre of pressure (p) and vertical ground reaction force (R). On the second time point, the centre of pressure has been shifted anterior. This results in the vertical ground reaction force being anterior of the centre of mass. As a result, the angular acceleration becomes counter-clockwise. At time point 3, the angular velocity has also become counter-clockwise. Time point 4 shows the response of the body, by shifting the centre of pressure posterior. This results in the vertical ground reaction force moving posterior of the centre of mass and the angular acceleration becomes clockwise. Time point 5 is then the same as time point 1, where the angular velocity also is clockwise. (Adopted from Winter, 1995).

To constantly adapt to changes in the position of centre of mass and changes in the environment, it is useful that variability exists in the recruitment of motor units and movements to maintain balance (Schmidt *et al.,* 1979; Stein *et al.,* 2005; Bartlett *et al.,* 2007; Stergiou & Decker, 2011). This variability exists not only within, but also between humans where each human finds a unique solution to a task (Clark, 1995; Bartlett *et al.,* 2007). Generally, experts display less movement variability than novices, but are better able to use variability in movement to functionally perform a task with changing constraints (Bartlett *et al.,* 2007). Due to the complexity of the sensorimotor system, it is not exactly known how the differences in movement strategy can be explained (Zajac & Gordon, 1989; Gottlieb *et al.,* 1990; Kuo & Zajac, 1993). However, it is known that the sensorimotor system plays a large role in this process. For example, any type of visual or vestibular impairment, the processing of the information and the recruitment of the muscles are important factors in determining which movement strategy to use (Horak, 2006). In addition, the size of different body segments and muscle strength also play an important role in the process of selecting a movement strategy (Horak, 2006).

Multiple studies have related certain movement strategies during dynamic movements to an increased injury risk (Bahr & Krosshaug, 2005; Hewett *et al.*, 2006a; Hewett & Myer, 2011; Read *et al.*, 2017b). This has to do with the way that not the force itself, but the torque (or moment) acts on an object. The torque is calculated via: $\tau = F * c$ where τ is the torque, F is the force and c is the moment arm (perpendicular distance) between the object and the force (Enoka, 2008). This shows that movement strategies that result in a large moment arm or a high force on a specific joint might be related to higher injury risk, because this increases the risk that certain body tissues in that joint cannot cope with the torque that is working on the joint.

The next paragraphs show how certain movement strategies (i.e. the outcome of the sensorimotor system) are related to ACL injury risk and how this is related to certain factors of the sensorimotor system. It was chosen to use ACL injuries as an example due to the devastating consequences of ACL injuries (Ardern *et al.*, 2011) and the large amount of studies on ACL injuries (e.g. Malinzak *et al.*, 2001; Hewett *et al.*, 2006a; Hewett *et al.*, 2008) , so a detailed description of the movement strategies and the related sensorimotor factors could be given.

Knee abduction displacement has been related to increased risk of ACL injury occurrence (Hewett *et al.*, 2006a). Females display larger knee abduction moments than males during landings and this has been related to their higher ACL injury risk (Hewett *et al.*, 2006a). Hewett *et al.* (2006a) argued that factors including anatomic, hormonal and sensorimotor were associated with the difference in movement strategies during landing between males and females. For example, the larger knee abduction displacement and the higher ACL injury risk in females might be related to the relative strength and recruitment of the quadriceps relative to the strength and recruitment of the hamstrings ([i.e. quadriceps: hamstring ratio] Hewett *et al.*, 2006a; Read *et al.*, 2017b). Namely, females display less hamstring strength and recruitment compared to males (Malinzak *et al.*, 2001; Hewett *et al.*, 2008; Holm & Vollestad, 2008) and females that sustained an ACL injury had less hamstring strength but did not differ in quadriceps strength compared to healthy peers (Myer *et al.*, 2009). This quadriceps dominance has been related to increased knee abduction (Hewett *et al.*, 1996; 2005) and a lack of compression of the knee joint and anterior motion of the tibia, which can result in an increased anterior shear force and increased load on the ACL (Skelly & DeVita, 1990; More *et al.*, 1993; Hewett *et al.*, 1996; Morgan *et al.*, 2014).

Movement of the trunk has been related to ACL injury risk also (Hewett & Myer, 2011). Namely, lateral trunk motion has been related to increased knee abduction motion (Hewett *et al.*, 2009) and lateral trunk motion results in lateral motion of the centre of mass and of the vertical ground reaction force, which can result in a greater moment arm relative to the joint centre of the knee (Hewett & Myer, 2011). Accordingly, Read *et al.* (2017b) argued that especially during dynamic movements, the pre-activation and co-contraction of trunk and lower extremity muscles is important to maintain the trunk and the body segments in correct positions to reduce injury risk. Trunk, pelvic and hip muscles are important in stabilising the spine, because the passive structures in the spine cannot maintain the spine in a stable position (Morris & Lucas, 1962; Willson *et al.*, 2005; Kibler *et al.*, 2006). Therefore, the muscles are responsible for returning the spine back to equilibrium after perturbations (Willson *et al.*, 2005) and three mechanisms are used to maintain a stable spine (Willson *et al.*, 2005; Kibler *et al.*, 2006). At first, contraction of the abdominal muscles, the diaphragm and the pelvic floor muscles result in an increased intra-abdominal pressure (Willson *et al.,* 2005; Kibler *et al.,* 2006). The increased intra-abdominal pressure causes the abdomen area to become a rigid cylinder and increases the lumbar stiffness and decreases the load on the spine muscles (Willson *et al.,* 2005; Kibler *et al.,* 2006). Second, co-contraction of the trunk flexors and trunk extensors cause spinal compressive forces, which also aids in the stability of the spine (Willson *et al.,* 2005). At third, hip and trunk muscles can be pre-activated based on planned movements (Willson *et al.,* 2005; Kibler *et al.,* 2006). This has two advantages, namely the spine will be more stable during perturbations (Willson *et al.,* 2005) and a stable core will provide a stable base for the limbs to move (Kibler *et al.,* 2006).

No muscle contributes more than 30% to the total stability of the spine (Cholewicki & Van Vliet, 2002) and this implies that coordination between the muscles is necessary to maintain a stable spine. For example, it is important that the quadratus lumborum co-contracts on both sides of the spine to create lumbar stiffness (Willson *et al.,* 2005). In addition, contraction of individual muscles such as the hamstrings, gluteus maximus and rectus abdominis results in movements, whereas co-contraction of these muscles can result in lumbar stiffness (Willson *et al.,* 2005). In addition, the activation of the multifidi results in more efficiently working multi-joint muscles in the trunk to control spine motions (Kibler *et al.,* 2006). The way different muscles are recruited is dependent on the movement made (Bobbert & Van Zandwijk, 1999). For example, the activation of the gluteus maximus during the landing after vertical jumps is highly related to the ground reaction force (Bobbert & Van Zandwijk, 1999). This is necessary to counteract the downward acceleration of the centre of mass when landing (Bobbert & Van Zandwijk, 1999). Similarly, the trunk extensors (Willson *et al.,* 2005; Kibler *et al.,* 2006) and hamstrings (Skelly & DeVita; 1990; More *et al.,* 1993; Hewett *et al.,* 1996) are important to counteract the deceleration of the centre of mass during horizontal jumping and landing.

2.2.5 Conclusion.

The sensorimotor system is responsible for keeping the body and the individual joints in balance. To that purpose, input about the position of the body relative to the environment and the position and movement of the different body segments is constantly collected by the sensory system. The sensory system transforms the input into neural signals and these signals travel via afferent pathways to the central nervous system where all information is processed. The central nervous system then sends signals via efferent pathways to the muscles with the purpose to recruit specific motor units. To maintain in balance, the centre of mass needs to stay within the base of support and as such, the muscles are recruited in such a way that the centre of mass stays within the base of support. Different solutions can be used to maintain joint stability and to keep the centre of mass within the base of support. These so-called movement strategies are unique in each person and are based on multiple factors, such as the working of the sensorimotor system, muscle strength and anatomical characteristics of the individual. It has been proposed that changes in the sensorimotor system affect movement strategies and can consequently result in increased injury risk. For example, the pre-activation and co-contraction of trunk, hip and pelvic muscles is important to maintain the trunk in balance and a lack of balance can result in increased ACL injury risk. Similarly, a lack of hamstring recruitment during landings can result in increased knee abduction, which has also been related to increased ACL injury risk. This shows that factors related to the sensorimotor system are associated with lower extremity injury risk. As such, the next section will provide more information on the relationship between the sensorimotor system and lower extremity injury risk factors.

2.3 The Relationship Between Injury Risk Factors and the Sensorimotor System

In a narrative review, Read *et al.* (2015) mentioned *growth and maturation*, *movement skill*, *fatigue* and *injury history* as the four largest injury risk factors in youth soccer. Interestingly, these factors seem to be highly related to the sensorimotor system (Read *et al.*, 2015). As mentioned in the previous section, the working of the sensorimotor system affects the movement strategies used and the movement strategies used by an athlete are related to injury risk. Therefore, the next subsections will describe how the four largest injury risk factors are related to the sensorimotor system and how they can affect the movement strategies used. This information could then be used to determine what types of movements a movement assessment tool should screen for to determine possible injury risk factors in youth soccer players.

2.3.1 Growth and maturation.

Movement awkwardness is an often-used term to describe adolescents with reduced motor control around their peak height velocity (Beunen & Malina, 1988; Philippaerts *et al.*, 2006). This reduced motor control is not apparent in all adolescents during their peak height velocity, but a minority of athletes have reduced motor performance on specific tasks such as jumping and plate tapping (Beunen & Malina, 1988). Accordingly, Read *et al.* (2017c) found that hop performance of youth soccer players increased from 14-year olds onwards (effect sizes: 0.65 – 0.84) but the 13-year olds scored worse than the 11- and 12-year old, although no effect sizes were given (Read *et al.*, 2017c). The worse performance of the under 13 group might be due to some sort of movement awkwardness (Read *et al.*, 2017c).

It has been suggested that movement awkwardness is related to the increased injury risk in adolescent athletes (Quatman-Yates *et al.*, 2012) and as such, it is important to understand what the possible causes of this movement awkwardness are. Radnor *et al.* (2018) wrote a review of the literature on factors related to the sensorimotor system that change throughout childhood and adolescence and that can affect movement performance. They argued that the increased muscle size is not solely responsible for the improved muscle strength, but that changes in the sensorimotor system also aid in increasing the force production throughout maturation (Radnor *et al.,* 2018). For example, the motor unit recruitment (Belanger & McComas, 1989; Grosset *et al.,* 2008), pre-activation of muscles (Lazaridis *et al.,* 2010; Oliver & Smith; 2010; Lloyd *et al.,* 2012), short-latency stretch reflex (spinal involuntary command from 30 to 60 milliseconds after initial contact [Radnor *et al.,* 2018]) (Lazaridis *et al.,* 2010; Oliver & Smith; 2010; Lloyd *et al.,* 2012) and the ability to quicker recruit higher-threshold motor units (Falk *et al.,* 2009; Waugh *et al.,* 2013) increased throughout adolescence and are higher in adults than in children. These four changes were associated with increased muscle force production or performance in movements with a stretch shortening cycle, such as repeated hops (Radnor *et al.,* 2018). In addition, muscle co-contraction decreased throughout adolescence (Frost *et al.,* 1997) and is lower in adults than in children (Lambertz *et al.,* 2003; Grosset *et al.,* 2008). Muscle co-contraction can aid in providing balance (Frost *et al.,* 2002), but also reduces the net force output (Malina *et al.,* 2004). The motor unit recruitment, pre-activation of muscles and co-contraction of the muscles have all been related to the ability to maintain balance. Changes in these factors might be related to the movement awkwardness, but currently no longitudinal studies are available to show how these factors might be related to movement awkwardness (Radnor *et al.,* 2018).

Other factors can also be related to movement awkwardness. For example, during a brief period in puberty, the muscles have not yet reached their full size, whereas the trunk length has increased relatively to the leg length (Tanner, 1978). This implies that the vertical position of the centre of mass increases. As described in Section 2.2.4, when the centre of mass of an object increases, the same angular velocity brings the centre of mass quicker outside the base of support. As such, it is more difficult to maintain balance.

Moreover, the relative lower muscle strength makes it more demanding for the body to change the centre of pressure and to maintain balance (Tanner, 1978). In addition, Williams *et al.* (2012) mentioned that asynchronous growth of muscles, changes in the point of inertia of different body segments and the difference in timing of muscle and bone growth might be related to the movement awkwardness. However, more longitudinal studies are necessary to confirm those findings (Williams *et al.,* 2012).

2.3.2 Fatigue.

Injuries occur more towards the end of each half and towards the end of the game in youth (Price *et al.,* 2004; Cloke *et al.,* 2009) and adult (Hawkins *et al.,* 2001; Ekstrand *et al.,* 2011) soccer. Fatigue affects several factors related to the sensorimotor system. For example, after five minutes of uphill running on a treadmill, the central processing of proprioceptive signals in the knee joint was reduced (Miura *et al.,* 2004). In the same study, they found that a local fatigue protocol on knee flexors and extensors did not affect the proprioceptive signals but did cause a reduced knee flexors and extensors strength (Miura *et al.,* 2004). Three other studies focused on biomechanical variables and it was found that after a fatigue protocol including vertical jumps and 30 metre sprints and after a fatigue protocol on a leg press weight machine, alterations in landing technique occurred (Chappell *et al.,* 2005; Gehring *et al.,* 2009). In addition, a fatigue protocol that included the performance of single leg squats or single leg calf raises caused a reduction in postural control (Reimer III & Wikstrom, 2010). After the calf raises, a higher instability in the posterior-anterior axis occurred, whereas the performance of squats caused more instability in the medio-lateral axis (Reimer III & Wikstrom, 2010). All three studies (Chappell *et al.,* 2005; Gehring *et al.,* 2009; Reimer III & Wikstrom, 2010) concluded that the changes in movement after the fatigue protocol could be linked to an increased injury risk.

Studies where soccer-specific protocols were used found similar results. These protocols are based on information about the intermittent nature of soccer where mainly low intensity running is alternated by short bursts of high intensity running (Bangsbo *et al.,* 2006; Di Salvo *et al.*, 2007). For example, semi-professional soccer players were able to maintain balance over the course of a simulated soccer game that was based on match demands in professional soccer (Greig *et al.,* 2006; Greig & Walker-Johnson, 2007). However, after the protocol a change in movement strategy to maintain balance was found (Greig & Walker-Johnson, 2007). This change in strategy caused more plantar flexion of the ankle which reduces the base of support and increases the risk of ankle sprain injury (Palastanga *et al.*, 2006; Greig & Walker-Johnson, 2007).

Multiple studies have been performed to determine the effect of fatigue on sensorimotor risk factors of lower extremity injuries in adolescent soccer players. Most of these studies used the soccer-specific aerobic field test ($SAFT^{90}$), which has been validated by Lovell *et al.* (2008). For example, De Ste Croix *et al.,* (2015) collected EMG data of 36 female soccer players during knee extension movements on a dynamometer. They found that on average the soccer players had a 52% to 67% longer electromechanical delay of the biceps femoris, semitendinosus and gastrocnemius muscles after this fatigue protocol. Lehnert *et al.* (2017) determined the effect of the $SAFT^{90}$ on several sensorimotor risk factors of lower extremity injuries in elite soccer players with an average age of 14 years. The muscle activation decreased in the rectus femoris, vastus medialis and semimembranosus after the fatigue protocol, whereas no changes in vastus lateralis and biceps femoris were found (Lehnert *et al.,* 2017). In addition, joint stiffness also reduced after the fatigue protocol, which has been related to an increased ACL injury risk (Granata *et al.,* 2002a; Granata *et al.,* 2002b). However, no changes in quadriceps: hamstring ratio were found upon the fatigue protocol (Lehnert *et al.*, 2017).

De Ste Croix *et al.* (2017) determined the joint stiffness during a submaximal hopping protocol before and after the SAFT^{90} in three different age groups of female soccer players. A possible reduced, an unclear change and a very likely increase in joint stiffness were found in the under 13, under 15 and under 17 players, respectively, after the fatigue protocol (De Ste Croix *et al.*, 2017). In a study of De Ste Croix *et al.* (2018), the quadriceps: hamstring ratio was determined before and after the SAFT^{90} in three different age groups of adolescent female soccer players. In this study, an unclear change in quadriceps: hamstring ratio was found in the under 13 age group, the quadriceps: hamstring ratio increased in the under 15 age group and a decrease in quadriceps: hamstring ratio was found in the under 17 age group (De Ste Croix *et al.*, 2018).

Two other studies (Oliver *et al.*, 2008; 2014) used a soccer-specific fatigue protocol that was based on the match demands of only one half (Oliver *et al.*, 2007). Oliver *et al.* (2008) researched the effect of the fatigue protocol on muscle activity during squat, counter movement and drop jumps in non-elite youth soccer players. Total muscle activity decreased after a soccer-specific fatigue protocol in all jumps on average more than 10% but only the decrease in muscle activity during the drop jump was significant (Oliver *et al.*, 2008). Oliver *et al.* (2014) also included non-elite youth soccer players. Here, joint stiffness over the joints in the lower extremity during continuous hops was tested before and after the same fatigue protocol. Approximately half of the participants had an increase in joint stiffness, whereas the other half had a decrease in joint stiffness after the fatigue protocol (Oliver *et al.*, 2014). Moreover, the changes in joint stiffness were also related to centre of mass displacement, with a higher stiffness resulting in less centre of mass displacement (Oliver *et al.*, 2014). Based on this information, they argued that a reduced joint stiffness could be associated with a higher injury risk and that the differences in joint stiffness changes were due to the use of different feedforward and feedback strategies between the different athletes to maintain balance (Oliver *et al.*, 2014).

2.3.3 Injury history.

Youth (Le Gall *et al.,* 2006) and adult (Hägglund *et al.*, 2006) soccer players that sustained a specific injury have more chance of sustaining a re-injury. Read *et al.* (2015) suggested that the high risk of a re-injury might be caused by neuromuscular inhibition (the muscle is being prevented from being fully activated [Rice & McNair, 2010]), which leads to altered movement and stabilisation patterns (Fyfe *et al.,* 2013). In addition, Read *et al.* (2015) discussed multiple papers that found sensorimotor deficits following injuries (Bullock-Saxton *et al.,* 1994; Hurley, 1997; Croisier & Crielaard, 2000; Friel *et al.,* 2006; Lee *et al.,* 2009).

In athletes that had sustained an ankle injury, delayed recruitment of the gluteus maximus (Bullock-Saxton *et al.,* 1994), weaker hip abductor muscles (Friel *et al.*, 2006) and worse ankle positioning sense (Konradsen *et al.*, 1998) were found. Similarly, athletes that returned to sport after an ACL injury had reduced quadriceps activation (Hurley, 1997), had reduced knee flexor and extensor strength in the injured side (Thomas *et al.,* 2013) and had less knee flexion in their injured compared to their uninjured side during a landing after a vertical drop jump (Thomas *et al.*, 2015; Ithurburn *et al.,* 2015). Information about the quadriceps and hamstring strength are important return to sport criteria after an ACL injury, because a higher quadriceps asymmetry results in worse functional recovery one-year post return to sport (Ithurburn *et al.,* 2017). In addition, a higher hamstring asymmetry results in decreased tibial internal rotation during gait and increased tibial external rotation during jogging (Abourezk *et al.,* 2017). Furthermore, athletes that sustained a hamstring injury had lower hamstring muscle output during isokinetic joint moments and had lower quadriceps: hamstring ratios in their injured leg than in their non-injured leg and when compared to healthy peers (Croisier & Crielaard, 2000). As such, Opar & Serpell (2014) suggested that a potential relationship might exist between prior hamstring strain injury and future ACL injury risk. However, to my knowledge, no studies have determined the relationship between prior hamstring strain injury and ACL injury risk.

2.3.4 Movement skill.

As described in Section 2.2, each person develops unique movement strategies that vary per movement. These movement strategies can be affected by impairments of the sensorimotor system, by the size of the different body segments and by muscle strength. Moreover, as mentioned in Sections 2.3.1, 2.3.2 and 2.3.3, growth and maturation, fatigue and injury history can all affect different factors related to the sensorimotor system and consequently affect movement strategies. This is important from an injury prevention perspective, because specific movement strategies have been associated with ankle and knee ligament injuries. For example, approximately 50% of all ankle injuries are non-contact injuries in youth (Cloke *et al.,* 2009) and professional soccer (Waldén *et al.,* 2013) and 65% of these non-contact injuries occur during landing, twisting and turning movements (Woods *et al.,* 2003). During these movements, wrong positioning of the foot can lead to excessive supination or pronation, which can lead to ankle ligament sprains (Fong *et al.,* 2009). Similarly, over half of all ACL injuries in soccer are non-contact injuries (Waldén *et al.,* 2011; Waldén *et al.,* 2015) and there are three main inciting events: pressing on the ball (the player makes a sidestep cut), regaining balance after kicking, and landing after heading with the knee being relative straight or in abduction (Waldén *et al.,* 2015).

The analysis of biomechanical data during dynamic movements can aid in understanding how different movement strategies are related to specific injuries (Hewett *et al.,* 2006a; Read *et al.,* 2016b) and how movement strategies are related to other factors related to the sensorimotor system (Riemann *et al.,* 2002). Laboratory-based marker 3D systems and force plates are the gold standard to collect biomechanical data (Riemann *et al.,* 2002; Padua *et al.,* 2009) and multiple prospective studies have used these systems to relate biomechanical variables to injury risk. For example, Hewett *et al.* (2005) found that knee abduction angle and knee abduction moment were larger in female athletes that sustained an ACL injury than in females that did not sustain such an injury during a drop vertical jump. In contrast, Krosshaug *et al.* (2016) did not find a relation between biomechanical variables during a drop vertical jump and ACL injury occurrence. In the study of Paterno *et al.* (2010), 56 athletes that had undergone ACL-reconstruction performed vertical drop jumps. Four variables, namely the non-reconstructed limb hip internal rotation moment impulse, more frontal plane knee range of motion (knee abduction), asymmetries in sagittal plane knee moments at initial contact (knee flexion) and postural control deficits of the reconstructed limb (measured with the Biodex Balance System) were all associated with a re-injury of the ACL. Zazulak *et al.* (2007a) found that male and female athletes that sustained a knee, a knee ligament or an ACL injury had more lateral trunk displacement after the sudden force release compared to athletes that did not sustain such an injury.

2.3.5 Conclusion.

This section was based on a literature review of Read *et al.* (2015) where *growth and maturation*, *fatigue*, *injury history* and *movement skill* were mentioned as the four largest injury risk factors in youth soccer. This section showed that growth and maturation, fatigue and injury history are related to the sensorimotor system and to alterations in movement strategies that have been linked to increased injury risk, such as reduced balance, different positioning of the feet with more plantar flexion, and less knee flexion. Moreover, the section on movement skill showed that for two of the injuries with the highest injury burden in youth football, ACL injuries and ankle ligament injuries, the type of movement strategy used is related to the risk of sustaining such an injury. This shows that movement assessment tools can indirectly be used to assess the four largest injury risk factors in youth football by assessing certain movements such as jumps and balance movements. As such, the next section will discuss how current movement assessment tools quantify movement strategies linked to certain injuries and how these movement assessment tools should be interpreted to develop individually-based training programs.

2.4 Measuring Factors Related to the Sensorimotor System

The tools that measure sensorimotor risk factors of lower extremity injuries can be divided into laboratory- and field-based assessment tools. Riemann *et al.* (2002) wrote a paper on different laboratory-based assessment systems that measure factors related to the sensorimotor system. A large benefit of these laboratory-based assessment tools is their high reliability and validity and the variety of variables they can measure. Riemann *et al.* (2002) mentioned tools that measure proprioception, the integrity of the afferent pathways to the central nervous system, the use of electric stimulation to determine the status of the efferent pathways and the use of electromyography to measure the muscle activity. Also, the use of force plates, marker 3D systems, and combinations of the above-mentioned measurement tools are described by Riemann *et al.* (2002). As such, they can give detailed insight into possible sensorimotor deficits of an individual. However, an issue with these tools is that they cannot be used in a practical environment, such as the youth academy of a professional soccer club, because these measurement tools are expensive and time consuming in use (Padua *et al.*, 2009; Allen *et al.*, 2018; Bardid *et al.*, 2018; Blair *et al.*, 2018; Colyer *et al.*, 2018; O'Donnell *et al.*, 2018). Therefore, more practical tools were developed to measure sensorimotor risk factors of lower extremity injuries.

Practical movement assessment tools make generally use of human raters to assess the movements (Read *et al.*, 2018a). A distinction can be made between tools where the movement assessment occurs in real-time and where the movement assessment occurs via video analysis. With the Functional Movement Screen, the movement is assessed in real-time. This movement assessment tool is frequently used in professional soccer clubs to determine the injury risk of soccer players (Read *et al.*, 2018a). It is a practical movement assessment tool and was developed to assess specific fundamental movements of athletes (Cook *et al.,* 2006a; 2006b). It has been argued that imbalances of the sensorimotor system can be identified when these fundamental movements cannot be performed correctly (Cook *et al.,* 2014a; 2014b). However, an issue with the Functional Movement Screen is that it uses a composite score on a scale from zero to three to assess each movement, whereas the assessment of movements should not be unidimensional (Moran *et al.,* 2017). In addition, a single score on a scale from zero to three results in a low sensitivity to changes (Wright *et al.,* 2018).

Movement assessment tools that make use of video analysis can assess the movement in more detail. Read *et al.* (2016a; 2016b; 2017b) reviewed the literature to develop a movement assessment tool that measures the sensorimotor risk factors of lower extremity injuries. They argued that quadriceps dominance, assessment of leg asymmetry, assessment of frontal plane knee control (knee abduction), trunk dominance, and dynamic stability are the sensorimotor risk factors of lower extremity injuries (Read *et al*., 2016a; 2016b; 2017b). *Quadriceps dominance* is defined as an imbalance between quadriceps and hamstring recruitment patterns (Myer *et al.*, 2004) where an increase in knee extensor moments is preferred over an increase in knee flexor moments during movements with high lower extremity joint torques (Hewett *et al.*, 1996). *Leg asymmetry* can relate to an imbalance in strength, coordination and/or control between the left and right leg (Myer *et al.,* 2004; Read *et al.*, 2016a). *Frontal plane knee control* relates to the knee moving into abduction (Read *et al.,* 2016a). *Trunk dominance* is defined as an imbalance between the inertial demands of the trunk and the ability of the 'core' to resist perturbations to the centre of mass (Hewett *et al.,* 2010b; Myer *et al.,* 2011; Read *et al,* 2016a). *Dynamic stability* (or balance) implies that the sensorimotor system keeps the centre of mass within the base of support (Read *et al.,* 2016a).

Based on the definitions given in the previous paragraph, Read *et al.* (2016b; 2017b) described several field-based tests that can measure biomechanical variables to quantify the sensorimotor risk factors of lower extremity injuries. In addition, several other studies have also described practical movement assessment tools that can be used to quantify the sensorimotor risk factors of lower extremity injuries (Fransz *et al.,* 2013; Dingenen *et al.,* 2014; 2016a; Gokeler *et al.,* 2017) An overview of the different movement assessment tools and which sensorimotor risk factors of lower extremity injuries they are supposed to measure is displayed in Table 2.1.

Table 2.1. Movement assessment tools that measure sensorimotor risk factors of lower extremity injuries.

Quadriceps dominance	*Leg Asymmetry*	*Frontal plane knee control*	*Trunk dominance*	*Dynamic stability*
	HFD[1]	Nomogram[1]	LESS[1]	TTS after SLH[4]
	Triple HFD[2]	LESS[1]	2D video SL DVJ[3]	Y & Star Excursion BT[1]
	Side HFD[2]	2D video SL DVJ[3]	10 s tuck jump[1]	DLS to SLS[5]
	Single leg CMJ[1]			
	Y & Star Excursion BT[1]			
	10 s tuck jump[1]			

HFD: Hop for distance; CMJ: Countermovement jump; BT: Balance test; LESS: Landing error scoring system; SL DVJ: Single leg drop vertical jump; TTS: Time to stabilization; SLH: Single leg hop; DLS: Double leg stance; SLS: Single leg stance
1. Read *et al.,* 2017b; 2. Gokeler et al., 2017; 3. Dingenen et al., 2014; 4. Fransz et al., 2015; 5. Dingenen et al., 2016a.

2.4.1 Practical movement assessment tools to quantify sensorimotor risk factors of lower extremity injuries.

Previous studies have already shown that many practical assessment tools are reliable and have some sort of construct validity (Dallinga *et al.,* 2012; Read *et al.,* 2017b). Hence, these tools are used to monitor the return to sport status of players during their rehabilitation (Gokeler *et al.*, 2017), to monitor performance variables in a youth academy (Lloyd *et al.,* 2015) and to screen for injury risk in professional and youth soccer (McCall *et al.,* 2014; Read *et al.,* 2018a). However, a lack of information exists on the practical ability of these tools and whether they can actually quantify the sensorimotor risk factors of lower extremity injuries. As such, this will be done in the next subsections.

2.4.1.1 Quadriceps dominance.

The quadriceps: hamstring ratio can be used to determine whether quadriceps dominance exists (Read *et al.,* 2016a). Read *et al.* (2017b) did not discuss any movement assessment tools that measure the quadriceps: hamstring ratio. Namely, the handheld dynamometer, force plates and the Nordic hamstring strength assessment are proposed as practical tools to measure the hamstring and quadriceps strength (Read *et al.,* 2017b). However, Read *et al.* (2017b) mention that there is a lack of knowledge on the reliability and validity of these tests in youth male soccer players and that more research is warranted. As discussed previously in this chapter, the quadriceps: hamstring ratio has been related to the knee abduction displacement (Hewett *et al.,* 2006; Read *et al.,* 2017b). This implies that measuring the frontal plane knee control (Section 2.4.1.3) might be used to identify athletes that have quadriceps dominance.

2.4.1.2 Leg asymmetry.

The limb symmetry index, which is defined as the performance with the injured/non-dominant leg divided by the performance with the non-injured/dominant leg, is used to determine whether leg asymmetry exists (Noyes *et al.,* 1991). Tests such as the single, triple and side hop for distance, the single leg countermovement jump and the Star Excursion- and Y-balance test can provide information about leg asymmetry (Hertel *et al.,* 2006; Gribble *et al.,* 2012a; Read *et al.,* 2017b). This asymmetry can be caused by a lack of muscle strength, but also by a lack of coordination and control (Noyes *et al.,* 1991). Most studies only focus on the movement performance, for example by comparing jump distances of both feet or the performance on the balance test which each leg (jumps: Hewit *et al.,* 2012; Wellsandt *et al.,* 2017; Leister *et al.,* 2018 balance: Dallinga *et al.,* 2012; Plisky *et al.*, 2012). However, this implies that the leg asymmetry is based on only one variable.

A recent study went into more detail to assess leg asymmetry during the single leg hop (Welling *et al.*, 2018a). Namely, they showed that although most ACL reconstructed patients have a normal limb symmetry index when it comes to hop distances, knee flexion angles throughout the landing of a single leg hop were reduced in the injured compared to the non-injured leg (Welling *et al.*, 2018a). These findings show that the collection of additional kinematic data can provide more insight about the possible existence of leg asymmetry. In the study of Welling *et al.* (2018a), a camera recorded the movement in the sagittal plane and specific software was used to calculate the knee flexion angles. This might be an issue, because the use of normal cameras to measure joint angles can result in measuring the movement of the wrong plane due to the use of 2D joint positions (Colyer *et al.*, 2018). This implies that this 2D analysis can affect the validity of a movement assessment tool negatively.

2.4.1.3 Frontal plane knee control.

A knee abduction displacement occurs due to a combination of hip internal rotation and tibial external rotation (Krosshaug *et al.*, 2007), in combination with decreased knee and hip flexion angles and pronation at the subtalar joint (Brophy *et al.*, 2010; Read *et al.*, 2016a). Knee abduction displacement has been associated with ACL injuries (Hewett *et al.*, 2005; Walden *et al.*, 2015) and consequently, multiple movement assessment tools have been developed that measure the knee abduction displacement during dynamic movements. However, the movement assessment tools use different methods to determine the knee abduction displacement. For example, with the nomogram, participants have to perform a double-legged vertical drop jump and the medial knee displacement from initial contact to maximal knee abduction displacement is determined (Myer *et al.*, 2010). The landing error scoring system also uses a double-legged vertical drop jump, but the rater has to make a dichotomous decision (yes or no) whether knee abduction displacement occurred during the landing (Padua *et al.*, 2009). With the tuck jump assessment, the rater also has to make a dichotomous decision whether knee abduction displacement occurred during landing, but the participant has to perform repetitive tuck jumps for this test (Myer *et al.,* 2008b). With the 2D video analysis, the participant has to perform a single leg vertical drop jump and special software is used to calculate the knee abduction angle (Dingenen *et al.,* 2015c). It should be noted that other variables are also measured with these tools. For example, foot position during landing, knee flexion angle, stance width, and trunk movement are also named in the different tools (Myer *et al.,* 2008b; 2010; Padua *et al.*, 2009; Dingenen *et al.*, 2015c). An issue with all these movement assessment tools is that they make use of video analysis. As discussed in Section 2.4.1.2, this might reduce the validity of these tools (Colyer *et al.*, 2018).

2.4.1.4 Trunk dominance.

The inability to control forces effectively can result in excessive movement of the trunk and increased ground reaction forces and knee joint torques (Hewett & Johnson, 2010; Read *et al.,* 2016a). Read *et al.* (2017b) mentioned that the most common way to assess the ability to control forces is expensive and lacks ecological validity, because the movement of the trunk is collected during static movements with laboratory-based systems (Zazulak *et al.,* 2007a; 2007b). Similarly, field-based assessments to measure core stability lack ecological validity and have low correlations with several athletic measures (Read *et al.,* 2017b). As such, Read *et al.* (2017b) proposed the use of dynamic movements such as tuck jumps and vertical drop jumps to assess the trunk dominance. As already mentioned in Section 2.4.1.3, the landing error scoring system, the 2D video analysis and the tuck jump assessment include criteria to assess trunk movement. With the landing error scoring system and the tuck jump assessment, a dichotomous choice has to be made whether excessive trunk flexion was visible (Myer *et al.,* 2008b; Padua *et al.,* 2009). With the 2D video analysis, special software is used to calculate the trunk flexion angle (Dingenen *et al.*, 2015c).

2.4.1.5 Dynamic stability.

The double leg stance has been proposed as a method to measure the stability of athletes (Dingenen *et al.*, 2013). To perform this movement, the athlete stands with each leg on a force plate and flexes one hip. Thereafter, the time to stabilisation is being calculated by using the time until the centre of pressure stabilises (Dingenen *et al.*, 2013). The Star Excursion- and Y- balance tests have also been proposed to measure dynamic stability (Shaffer *et al.*, 2013). With these tests, a performance score (i.e. the distance the foot was moved) is used to assess this movement. In addition, movement patterns are assessed during these tests (i.e. loss of balance, the foot is not brought back into the original position, the reach foot is used to gain balance support [Shaffer *et al.*, 2013]) to determine whether the movement is performed correctly. A disadvantage of these movements is that they are relatively static, because the person stands throughout the movement on the same position.

The time to stabilisation after hops and jumps might be a better method to assess the ability of a person to stabilise, because it assesses the stability during a dynamic movement (Fransz *et al.*, 2013; Read *et al.*, 2017b). The time it takes the centre of pressure to stabilise is used to determine the time to stabilisation period (Fransz *et al.*, 2013; 2015). As mentioned in Section 2.2, the centre of pressure is an important variable to assess the sensorimotor system, because it is one of the actual outputs of the sensorimotor system. Moreover, this movement tool also has a high ecological validity due to the dynamic movement. Multiple methods can be used to calculate the time to stabilisation (Fransz *et al.*, 2015). Fransz *et al.* (2015) described how the direction of the ground reaction force, the smoothing of the raw data, the definition when the participant is considered stable, the sampling frequency and the trial length all affect the time to stabilisation values. A disadvantage of the use of the time to stabilisation is that a force plate is necessary, whereas most practitioners do not have access to force plates (Read *et al.,* 2018a). This raises questions whether the time to stabilisation is a practical measure to determine the dynamic stability.

2.4.2 Interpreting the outcome measures of movement assessment tools

Movement assessment tools have been used to relate sensorimotor risk factors of lower extremity injuries to the effect of specific training programs (Myer *et al.,* 2006; Kiesel *et al.,* 2011; Frost *et al.,* 2012), to determine whether differences in sensorimotor abilities exist between players of different performance levels (Paillard *et al.,* 2006; Paillard & Noé, 2006; Hrysomallis, 2011; Okada *et al.,* 2011; Parchmann & McBride, 2011; Lloyd *et al.,* 2015), to determine whether factors of the sensorimotor system are related to injury risk (Hewett *et al.,* 2005; Kiesel *et al.,* 2007; Padua *et al.,* 2009; Lehr *et al.,* 2013) and to determine how factors related to the sensorimotor system change throughout maturation (Barber-Westin *et al.,* 2006; Schmitz *et al.,* 2009; Read *et al.,* 2018b). This shows that a movement assessment tool can be used for different purposes. However, several studies have argued to be cautious when interpreting the data of movement assessment tools.

For example, an issue with monitoring factors related to the sensorimotor system and relating them to performance and injury risk is that these factors can vary over time, purely due to randomness or due to some systematic change (Bahr, 2016; Esmaeili *et al.,* 2018). Systematic changes can be due to fatigue or training (Halson, 2014), but an additional factor in adolescents is the effect of maturation on the sensorimotor system (Hewett *et al.,* 2004; Barber-Westin *et al.,* 2006; Holden *et al.,* 2016; Read *et al.,* 2016a; 2017b). Therefore, it is important to determine the variability in the measurement of physical characteristics throughout a season (Esmaeili *et al.,* 2018) to make informed decisions whether changes in certain variables were related to for example injury risk (Bakken *et al.,* 2016; Vanrenthergem *et al.,* 2017). In addition, it is important to understand the variability within a subject during the testing, because it affects the precision of the estimates of the change (Hopkins, 2000).

Factors related to the sensorimotor system are often collected in prospective studies to relate them to injury risk (e.g. Hewett *et al.,* 2005; Krosshaug *et al.,* 2016). In addition, many clubs determine the injury risk of players by screening them multiple times throughout a season (McCall *et al.,* 2014; Light *et al.,* 2018; Read *et al.,* 2018a). However, Bahr (2016) and Whiteley (2016) both argued that although many studies have found certain injury risk factors, this does not imply that these factors can be monitored to predict injury risk. The next paragraph will explain the issues with screening for injury risk and argue why movement assessment tools should not be used to predict injuries.

Screening focuses on the early detection of a pathological condition to enable an early intervention (Bahr, 2016; Whiteley, 2016). A first issue with screening for injury risk is that the pathological condition is not the injury itself but a specific characteristic of the body such as a specific movement strategy that determines the risk of a person sustaining an injury in the future (Bahr, 2016). Thus, instead of a dichotomous outcome of screening for diseases such as cancer, the outcome when screening for injury is an injury risk between 0% and 100% (Bahr, 2016). This implies that a cut-off value needs to be determined to assess which players have an elevated risk of sustaining a specific injury (Bahr, 2016). This cut-off value is being determined in a prospective study and is based on factors such as sensitivity ("proportion of true positives that is correctly identified by the test"), specificity ("proportion of true negatives that is correctly identified by the test"), positive predictive value ("proportion of patients with positive tests results who are correctly diagnosed") and the negative predictive value ("proposition of patients with negative test results who are correctly diagnosed") (Altman & Bland, 1994a; 1994b; Bahr, 2016). Two issues arise with this cut-off value. At first, a high specificity results in a lower sensitivity and vice versa and second, other prospective studies can often not confirm the initial results (Bahr, 2016). This shows that although movement assessment tools can be used to determine what factors are associated with a certain injury, these tools cannot predict which athletes become injured (Bahr, 2016). This has to do with the likelihood ratios ("value of the test for increasing certainty about a positive diagnosis") of injury screening tools, which are at least 20 times lower than necessary to be useful as a screening tool (Whiteley, 2016).

Bahr (2016) proposes that training programs should be used to reduce the injury risk in specific groups that have a higher injury risk, such as to lower the risk of ACL injury risk in female athletes. For example, the FIFA 11+, a specific injury prevention warming-up, improves sensorimotor risk factors of lower extremity injuries (Impellizzeri *et al.,* 2011) and has shown to reduce injury risk in adolescent females and collegiate male soccer players (Steffen *et al.,* 2013; Barengo *et al.,* 2014; Silvers-Granelli *et al.,* 2015). Similarly, other training programs that focused on improving the sensorimotor system also found a reduction in the number of injuries and improved movement strategies used during dynamic movements (Verhagen *et al.,* 2004; Mandelbaum *et al.,* 2005; Alentorn-Geli *et al.,* 2009b; Lopes *et al.,* 2017). This does not imply that movement assessment tools are worthless. Namely, they can still be used to determine injury risk factors (Bahr, 2016) and they can be used to identify movement deficits, which can then be used to individualise training programs (Hewett, 2016).

2.4.3 Conclusion

Laboratory-based systems are the gold standard to quantify sensorimotor risk factors of lower extremity injuries but are expensive and not practical for use in youth academies. As such, it has been proposed to use practical movement assessment tools that are able to quantify quadriceps dominance, leg asymmetry, frontal plane knee control, trunk dominance and dynamic stability, because they are sensorimotor risk factors of lower extremity injuries. Biomechanical data can be used to quantify those risk factors and as such many studies have described the use of human raters to quantify to risk factors via movement strategies. However, there are questions about the sensitivity of the human raters and more advanced analysis via 2D video analysis might not be valid. Therefore, the outcome measures of the different movement assessment tools should be interpreted with caution. Moreover, although the movement assessment tools can be used to quantify sensorimotor risk factors of lower extremity injuries, they should not be used to predict injuries.

2.5 Conclusion

This literature review discussed the working of the sensorimotor system, showed how the sensorimotor system is related to several injury risk factors and movement strategies, and described the state of the current movement assessment tools that could be used as part of the injury prevention process in sports organizations. The sensorimotor system is a control system and its main task is to keep the body in balance. To that purpose, it uses the sensory stimuli to determine which muscles need to be contracted to keep the centre of mass within the base of support. Injury risk factors such as growth and maturation, fatigue and injury history can affect the sensorimotor system. This can lead to alterations in movement strategies and an increased injury risk. Although it is not possible to determine which of those risk factors might have caused alterations in certain movement strategies, it implies that movement assessment tools can be used to determine whether athletes perform certain movements that increase their injury risk.

Quadriceps dominance, knee abduction, leg asymmetry, the ability to maintain balance and excessive trunk movement have been proposed as the injury risk factors that should be screened for with movement assessment tools. Knee abduction, leg asymmetry, the ability to maintain balance and excessive trunk movement could be quantified with kinematic data during balance movements, vertical jumps and horizontal jumps and hops. Multiple movement assessment tools have been developed that collect kinematic data and link it to these sensorimotor risk factors. However, an issue with current movement assessment tools is the trade-off between reliability and validity on one hand and the practicality on the other hand. Laboratory-based tools are able to collect biomechanical data in a reliable and valid manner, whereas the use of human raters is less reliable and valid but more practical. This implies that there are currently no movement assessment tools available that can collect reliable and valid kinematic data in a practical manner. Therefore, the next chapter will describe the development of a new movement assessment tool that uses depth-sensing technology to collect kinematic data in a reliable, valid and practical manner.

3.1 Introduction

Based on the findings of Chapter 2, it was deemed necessary to develop a new movement assessment tool. This tool should be practical in use and be able to collect kinematic data in a reliable and valid manner to quantify sensorimotor risk factors of lower extremity injuries. More specifically, this tool should have the practicality of current field-based movement assessment tools and have the reliability and validity of laboratory-based systems to collect kinematic data. Therefore, this chapter proposes the use of depth-sensing technology to collect kinematic data.

Depth-sensing technology uses time of flight and/or speckled infrared patterns to determine the 3D position of pixels from the camera image in real time (Lachat *et al.*, 2015). Given the low cost, portability of the camera, and the fact that this camera has the potential to collect kinematic data, the Windows Kinect™ (Kinect for Windows, Microsoft, USA) was proposed as a depth-sensing technology for movement assessment (Clark *et al.*, 2012; Dutta, 2012; Bonnechere *et al.*, 2014). An additional benefit of the Kinect is its ability to collect 3D positional data of 25 anatomical landmarks. As such, multiple studies have determined the ability of the Kinect to function as a movement assessment tool, but the data collection of several anatomical landmarks is not valid (Van Diest *et al.*, 2014; Kharazi *et al.*, 2015; Mentiplay *et al.*, 2015; Otte *et al.*, 2016; Auvinet *et al.*, 2017; Eltoukhy *et al.*, 2017). Moreover, special software is necessary for data collection and data analysis with the Kinect (Bujang *et al.*, 2015). As such, the Kinect is currently not usable as part of a movement assessment tool.

Any measurement system needs to be reliable and valid (Atkinson & Nevill, 1998) and as such, additional algorithms to improve the 3D data collection of anatomical landmarks with the Kinect were developed (Paolini *et al.*, 2014; Gammelgaard, 2015; Motiian *et al.*, 2015; Dolatabadi *et al.*, 2016; Giblin *et al.*, 2016; MacPherson *et al.*, 2016; McGroarty *et al.*, 2016; Wang *et al.*, 2016). In addition, software to make the Kinect practical in use was also developed (Bujang *et al.*, 2015). These studies showed that it is possible to develop a movement assessment tool that makes use of depth-sensing technology. However, collecting data of the different anatomical landmarks is not sufficient. As discussed in Chapter 2, it is important to use the kinematic data to quantify sensorimotor risk factors of lower extremity injuries. For example, the centre of mass displacement relative to the position of the base of support can be used to quantify the ability to maintain balance (Winter *et al.*, 1995). Moreover, the movement of the different body parts provides information about the movement strategies used to maintain balance (Hewett *et al.*, 2012), whereas the movement performance can provide information about muscle force output and leg asymmetries (Noyes *et al.*, 1991; Markovic *et al.*, 2007). However, there are currently no movement assessment tools that collect this information with the Windows Kinect. As such, this study has two aims. The first aim is to give an overview of the current literature on the Windows Kinect to show the strengths and weaknesses of the system. The second aim is to show the development of the athletic movement analysis tool (AMAT), a movement assessment tool that is able to collect kinematic variables with depth-sensing technology.

3.2 The Windows Kinect

The Windows Kinect (hereafter Kinect) is an RGB-D camera, which implies it has a normal colour camera, an infrared camera and depth-sensing technology. The resolution of the colour camera is 1920 x 1080 pixels and the resolution of the infrared camera and depth-sensing technology are 512 x 424 pixels. The frame rate of the Kinect is 30 Hz. The Kinect was originally developed for gaming purposes, but it has some features that might make it usable to collect kinematic variables that can be used to quantify sensorimotor risk factors of lower extremity injuries. Namely, the Kinect is a depth-sensing camera and makes use of the time of flight principle for its depth-sensing technology (Lachat *et al.,* 2015). Cameras that use the time of flight principle to obtain depth values emit infrared light and measure the time before the light is received (Lachat *et al.,* 2015). With the knowledge of the speed of light and the time from emitting light to the collection of the light, it is possible to calculate the depth value. The Kinect uses the indirect time of flight principle to obtain the depth values, because it is cheaper than the direct time of flight principle (Lachat *et al.,* 2015). The difference between the direct time of flight and indirect time of flight principle is the light source used, namely light pulses for direct time of flight, versus amplitude modulated light for the indirect time of flight (Lachat *et al.,* 2015). This implies that with the indirect time of flight, the phase shift in light is measured. Nevertheless, the basic principles to calculate the depth value are similar (Lachat *et al.,* 2015).

There are two additional features that make the Windows Kinect attractive as a tool to collect kinematic variables. At first, the Windows Kinect includes algorithms that can distinguish a person and specific anatomical landmarks of this person from the background due to the differences in depth values (skeletal tracking). The 20 anatomical landmarks used for this project are displayed in Figure 3.1. In combination with the ability to determine the depth value of each pixel, it is possible to obtain 3D information of these landmarks. Second, a software development kit (SDK) was developed for the Kinect. This makes it possible to develop algorithms that make use of the depth-sensing technology and the skeletal tracking features.

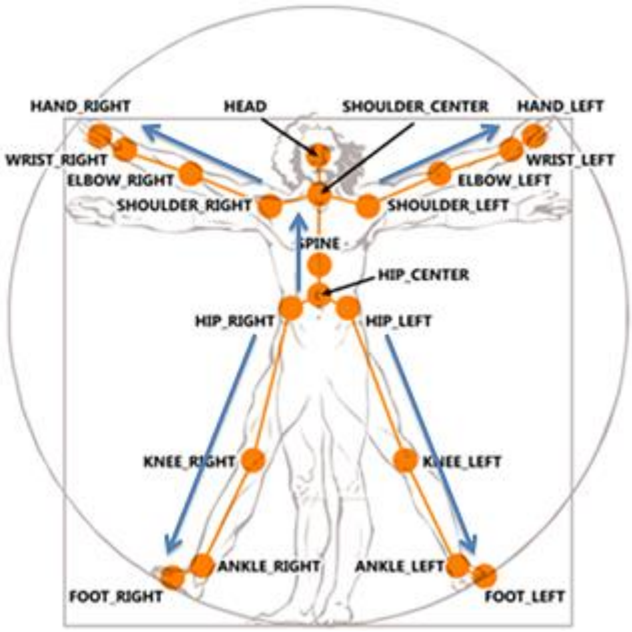

Figure 3.1. The 20 anatomical landmarks used by the AMAT for the analysis of the kinematic data. Source of image (https://msdn.microsoft.com/en-us/library/jj131025.aspx)

Originally, the Kinect was developed as part of the XBOX 360. With the introduction of the XBOX One, a successor of the Kinect (Kinect v1), the Kinect version 2 (Kinect v2), was introduced. The Kinect v2 has better depth-sensing technology compared to the Kinect v1, due to the time of flight principle used (Lachat *et al.*, 2015). In addition, the technological error of the skeletal tracking of the Kinect v2 is lower than the Kinect v1, except for the positional data of the feet (Wang *et al.*, 2015).

Since the release of the Kinect v2 in July 2014, validity studies were performed with the Kinect v2 where it was compared to laboratory-based marker 3D systems. For example, Clark *et al.* (2015) found that the Kinect v2 was valid to determine the postural sway in the posterior-anterior, but not in the medio-lateral axis during several balance tasks. In a study that focused on vertical drop jumps, it was found that hip and knee flexion/extension angles collected with the Kinect were highly correlated ($r \geq 0.96$) with a marker-based camera system, whereas hip and knee abductions and rotations had much lower correlations ($r < 0.3$) when compared to a marker-based camera system (Guess *et al.*, 2017). Moreover, the hip flexion angles measured with the Kinect were on average over ten degrees larger than when measured with the Vicon system and the root mean square errors between the joint angles as measured by both systems were larger than 7.5 degrees (Guess *et al.,* 2017).

Multiple studies assessed the validity of the kinematic data collection of the Kinect v2 during gait and all found that the peak angles of the lower extremities were not valid (Kharazi *et al.,* 2015; Mentiplay *et al.,* 2015; Auvinet *et al., 2015;* 2017; Eltoukhy *et al.,* 2017). For example, knee and hip flexion angles measured with the Kinect v2 throughout a gait cycle correlated with Vicon (r > 0.95), but the peak angles of the ankle and hip flexion were on average overestimated by 20 and 6 degrees, respectively, whereas peak knee flexion was underestimated on average by 10 degrees (Kharazi *et al.,* 2015). However, the Kinect v2 is valid to determine spatio-temporal variables of the gait (Mentiplay *et al.,* 2015; Eltoukhy *et al.,* 2017) and to detect gait asymmetries (Auvinet *et al.,* 2015*;* 2017).

Otte *et al.* (2016) collected kinematic data of participants during six activities of daily living, such as different types of walking and sitting down and standing up. The validity of the kinematic data, when compared to a laboratory-based marker system, was highest in the posterior-anterior axis (Pearson correlation range 0.64 to 0.99), followed by the medio-lateral axis (0.47 to 0.90) and the superior-inferior axis (-0.03 to 0.80). Moreover, the validity of kinematic data of the lower extremity was lower than the validity of kinematic data of the upper body. The outcomes of this study are in accordance with other studies performed on the Kinect v1 and Kinect v2. For example, Bonnechere *et al.* (2014) found that the lower extremity kinematic data of the Kinect v1 was not collected in a valid manner during functional movements. In addition, the Kinect v1 overestimated the foot movement during gait (van Diest *et al.,* 2014) and the collection of kinematic data of the feet is not reliable with the Kinect v2 (Wang *et al.,* 2015).

The previous paragraphs show that the kinematic data collection of the lower extremity is not valid with the Kinect. The lower extremity data is important for collecting kinematic variables during dynamic movements (Read *et al.,* 2017b). This implies that the Kinect v2 on its own cannot be used to collect kinematic data for a movement assessment tool. As such, several possibilities have been proposed to improve the kinematic data collection of the Kinect v2. For example, Clark *et al.* (2013a; 2013b) mounted the Kinect in a set position and calibrated the Kinect to align its field of view with the directions of the movement. This reduced the absolute error of the lateral trunk angle substantially when compared to a laboratory-based marker 3D system.

Other studies described the addition of algorithms to improve the kinematic data collection of the Kinect. For example, Mentiplay *et al.* (2018) manually selected the pixel that captured specific anatomical landmarks during single leg squats and drop vertical jumps. Although joint angles were collected validly with this method when compared to a laboratory-based marker 3D system, it is questionable whether this method is practical. Gammelgaard (2015), Giblin *et al.* (2016) and McGroarty *et al.* (2016) described the use of voxel data to create a point cloud that could be used to improve the kinematic data collection of the Kinect. A voxel is defined as "(in computer-based modelling or graphic simulation) each of an array of elements of volume that constitute a notional three-dimensional space, especially each of an array of discrete elements into which a representation of a three-dimensional object is divided" (Oxford Dictionary) and a point cloud is a set of voxels in space. Gammelgaard (2015) reported correlations larger than 0.8 for the hip flexion/extension and internal/external rotational angle and for the knee flexion/extension angle when the new algorithms were compared with a laboratory-based system. In addition, peak angle differences of the hip inter/external rotation and knee flexion/extension angle were smaller than four degrees when the kinematic data of the new algorithm was compared with the kinematic data of the laboratory-based system, which shows acceptable validity. In addition, Giblin *et al.* (2016) and McGroarty *et al.* (2016) also showed small peak angle differences (-3.8 to 2.6 degrees) between their new algorithms and a laboratory-based system for knee flexion/extension angles. These differences are lower compared to a 2D video analysis tool discussed by Dingenen *et al.* (2015c), where knee flexion angles differed approximately 20 degrees between their system and a laboratory-based marker tracking system. However, the mean difference in peak angles of knee abduction between their new system and a Vicon system were 15.4 and -5.4 degrees before and 9.9 and -6.3 degrees after the jump, for the left and right leg respectively (McGroarty *et al.,* 2016). McGinley *et al.* (2009) discussed that a tool to collect kinematic data is only valid if the difference in the measurement of joint angles when compared to a gold standard are smaller than 2 degrees. Moreover, differences in joint angles between 2 and 5 degrees should be interpreted with caution. Based on these definitions, it could be concluded that the method to collect knee abduction data as described by McGroarty *et al.* (2016) is not valid.

An advanced method compared to the methods described in the previous paragraph was developed by Bauer *et al.* (2017). Based on the original tracking of anatomical landmarks and the original point cloud of the Kinect, algorithms were developed that were able to determine the dimensions of the different body segments. To that purpose, a calibration measurement was performed where the participant first stands straight, then has to flex his/her arms and thereafter has to flex his/her legs. Based on the algorithms they developed, it became possible to determine the 3D sizes of the different body segments. To determine the validity of their method, they measured the lengths of the limb segments of two participants during upright standing and compared this with the length of the limb segments as measured with MRI. The limb segments as measured with the new method were on average 1.5% smaller than the limb segments as measured with the MRI. This implies that this method can improve the validity of the kinematic data collection.

However, the calibration takes 45 – 90 seconds, which raises questions about the practicality of this method.

Motiian *et al.* (2015), Dolatabadi *et al.,* (2016) and Wang *et al.,* (2016) developed new artificial intelligence algorithms to improve the data collection during gait. The raw data of the Kinect were used and based on the recognition of certain patterns in the data, it was found that their methods improved the collection of several gait parameters. Similarly, Segura *et al.* (2016) concluded that the addition of an Extended Kalman Filter and a Generic Algorithm made the Kinect v2 usable to identify the effect of tendon and muscle stiffness at the ankle joint.

Two other studies did not develop additional algorithms to improve the skeletal tracking of the Kinect but developed algorithms to track markers with the Kinect. MacPherson *et al.* (2016) described an algorithm that was able to track retroreflective markers with the infrared camera of the Kinect. These markers were placed on the back of nine participants to track the pelvic and trunk region, because the skeletal tracking of the Kinect does not measure any anatomical landmarks in this area. The positional data of the markers were collected with the Kinect v2 and with a laboratory-based system while the participants were running on a treadmill. Correlations between the two systems were all equal or larger than 0.87 and limits of agreement were all smaller than ten millimetres and 4.6 degrees for different velocities of gait (MacPherson *et al.,* 2016). In addition, Paolini *et al.* (2014) described algorithms that were able to track markers in a specific colour attached to the feet of the participant. To that purpose, a normal camera was linked with a Kinect. The colour information of the normal camera was used to determine where a certain marker was located, based on the colour information of the normal camera. Then, the Kinect would be used to collect the 3D position of this marker. The error of the foot detection ranged from zero to ten millimetres, which is better than the original foot data collection of the Kinect v1 and v2 (van Diest *et al.,* 2014; Wang *et al.,* 2015). Bujang et al. (2015)

described a last addition to the Kinect. They described that the Kinect v2 is not practical in use. As such, they developed software to make it easy for any practitioner to collect kinematic data with the Kinect v2 (Bujang *et al.,* 2015).

3.3 Technological Development

Section 3.2 showed the strengths and weaknesses of the Kinect v2 and the possible solutions proposed and developed by several researchers to improve the kinematic data collection with the Kinect. This study adds multiple of these functionalities together and shows the development of algorithms to make the Kinect suitable to be part of AMAT. The development of the AMAT was performed in cooperation with Pro Football Support (Huddersfield, United Kingdom), a company that focuses on athletic development in youth soccer players. As such, not all parts of the development were solely performed by the author. The next subsections will explain the development of the different parts of the AMAT.

3.3.1 The development of a system to mount the Kinect in a set position.

The set-up of the AMAT consists of a horizontal (length: 97.7 centimetres, width 61 centimetres) and vertical wooden board (height: 96 centimetres, width: 49 centimetres at the base) flanked by rubber mats on each side (length: 300 centimetres, width: 60 centimetres) (Figure 3.2). The Kinect is positioned 375 centimetres away from the vertical wooden board, it is 185 centimetres above the ground and it has a 30-degree angle with the horizontal (Figure 3.3). A custom-developed frame was placed in between the wooden boards and the box of the Kinect camera to have a consistent distance between those two objects. In the centre of the frame, four small wooden blocks were placed for calibration purposes. They are visible in Figure 3.2B and in Figure 3.3. How these blocks were used for calibration purposes will be discussed in Section 3.3.3.

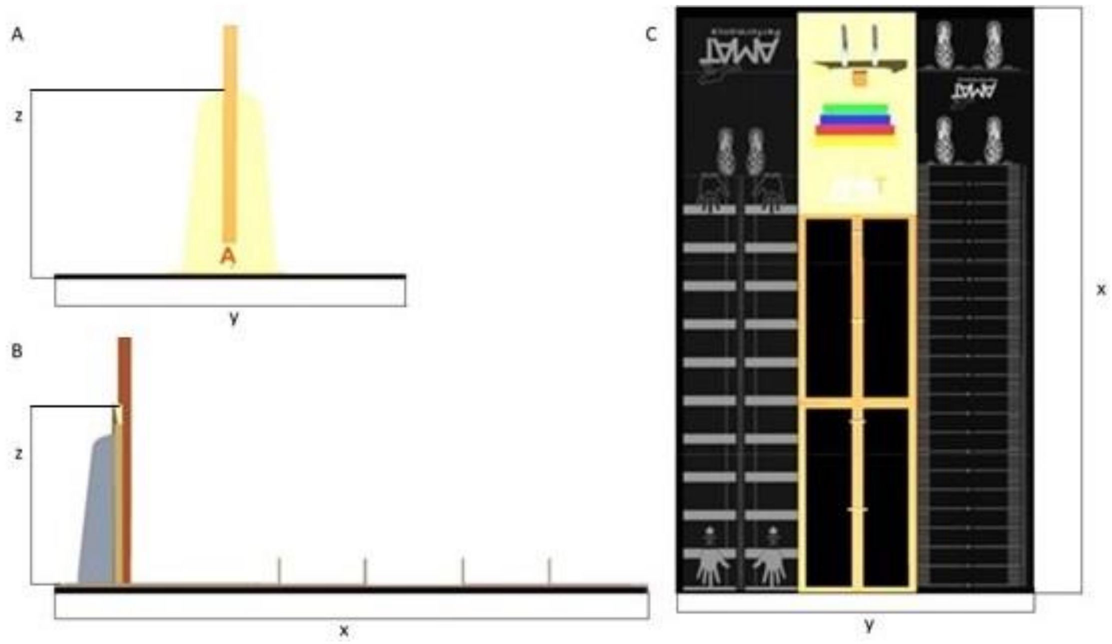

Figure 3.2. The set-up of the AMAT from three different viewing points. Figure 3.2A. Frontal view of the system, with the vertical wooden board visible. Figure 3.2B. Sagittal view of the system. On the left the vertical wooden board is visible. In the centre and on the right, four small wooden blocks are visible that are used for calibration purposes. Figure 3.2C. Transverse view of the system. On the left and the right side of the horizontal wooden board, the mats are visible on which several movements are performed. x: 300 cm, y: 181 cm, z: 96 cm.

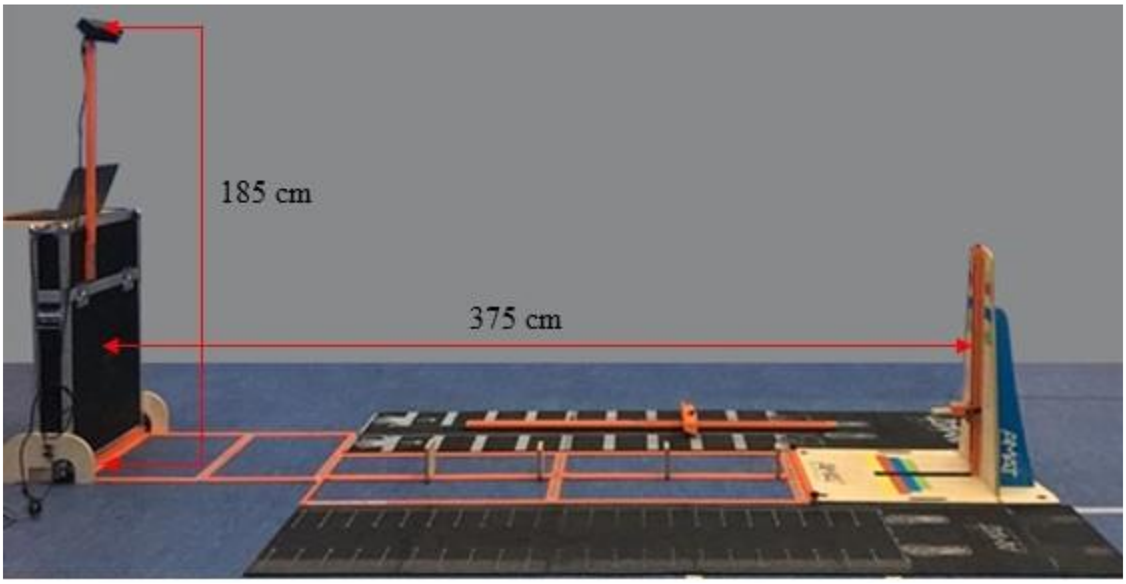

Figure 3.3. Sagittal view of the AMAT with the Windows Kinect camera. The Windows Kinect camera is 185 centimetres above the ground and 375 centimetres from the vertical wooden board. The angle of the Windows Kinect with the horizontal is 30 degrees. The orange part between the wooden board and the Kinect camera is the frame that makes sure the system is always set up correctly.

3.3.2 Movement assessment

The practitioners of Pro Football Support selected five movements that were included in the AMAT, namely standing horizontal jump, anterior balance movement, the back- and overhead-squat and crawling on hands and feet. Some of the movements are similar to the movements proposed by Read *et al.* (2017b). It was not possible in the timeframe of this thesis to develop algorithms and test the algorithms for all these movements. Therefore, this thesis will focus on the horizontal jumps, because it is a dynamical movement that is often used in practice as part of the monitoring of athletes (Read *et al.*, 2018a). In addition, it can be used to assess dynamic balance, trunk dominance, frontal plane knee control and leg asymmetry (Read *et al.*, 2017b). The practitioners opted for two different types of horizontal jumps to be included. One where the focus was on a controlled landing (control jump) and one where the focus was on jumping as far as possible without the necessity of a controlled landing (maximal jump). In addition, the control and maximal jump could be performed with two feet, the standing broad jump (Figure 3.4A) and by setting off on one leg and landing on the other leg, the single leg stride (Figure 3.4B).

A B

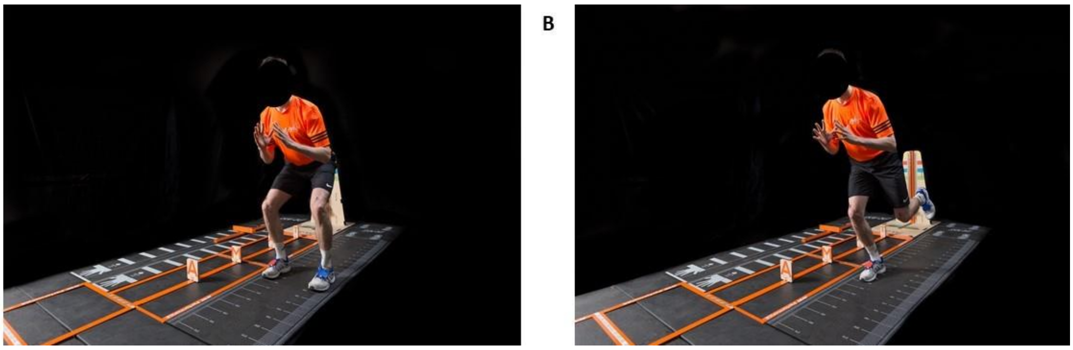

Figure 3.4. The jumps that were included in the AMAT. Figure 3.4A. Double-leg standing broad jump. The person pushes off and lands on both legs. Figure 3.4B. Single leg stride. The person pushes off on one leg and lands on the other leg.

Based on the recommendations by Paterno *et al.,* (2012), Hildebrandt *et al.* (2015), Wilk *et al.,* (2015), Gokeler *et al.*, (2017) and Welling *et al.,* (2018b), it was intended to assess the movement based on movement performance and movement quality. The distance jumped was used as the performance measure. The jump distance can be used to quantify leg flexor strength during the double-leg standing broad jump (Markovic *et al.,* 2007; Nagano *et al.,* 2007), whereas the leg asymmetry of the leg flexor strength can be quantified during the single leg stride (Noyes *et al.,* 1991). To assess the movement quality, the position of the centre of mass relative to the position of the base of support can be used to quantify dynamic balance and trunk dominance, whereas the knee displacement during the landing can be used to quantify frontal plane knee control.

3.3.3 The development of algorithms for the Kinect v2.

The process of the development of different algorithms to improve the reliability and validity of kinematic data collection is described in this section. The development of these algorithms was based on the information about the sensorimotor risk factors of lower extremity injuries described in Chapter 2, the use of algorithms in other studies described in Section 3.2 and the movements and movement assessment criteria described in Section 3.3.2.

Developing the algorithms to improve the reliability and validity of the kinematic data collection with the Kinect v2 was a process of trial and error. The first stage was to come up with new ideas to solve existing issues, for example to come up with a calibration algorithm (Clark *et al.,* 2013a; 2013b), to improve the kinematic data collection of the lower extremity (Wang *et al.,* 2015; Otte *et al.,* 2016) and to collect the centre of mass data. When a new idea for an algorithm was born, it had to go through multiple development stages before it could be used in a practical setting. The first stage was to translate the idea of an algorithm into writing the algorithm in programming language. These algorithms were written in the C# programming language and were automatically connected to the software development kit of the Kinect. Consequently, it was possible to test the newly developed algorithms in real time.

The testing of the algorithms can be divided into the testing of algorithms that were used for static (i.e. non-moving) tracking of objects and for dynamic tracking of objects. The testing of the static algorithms was important for calibration purposes and occurred in a laboratory-based setting. The testing of these algorithms was done by comparing the position of the objects as calculated by the static algorithms and the position of these objects in the real world. By doing this on different positions within the camera view, it could be determined whether the static algorithms worked correctly. The first part of testing the dynamic algorithms was identical to the process described above. After it was confirmed that a dynamic algorithm worked correctly in static settings, it was tested in different settings where the algorithm had to track a moving object. The start of the dynamic testing occurred in a laboratory-based setting with slow movements and the speed of the movements increased over time. This was done because failures of the tracking were very valuable to improve the algorithm and it was easier to understand why the tracking of a slow movement failed compared to why the tracking of a fast movement failed, because more datapoints were collected during a slow movement when an object travelled the same distance.

The second part of the dynamical testing was done at the youth academies of professional football organizations. The clubs where the AMAT was tested ranged from League 2 to Premier League clubs in England and one club on the highest level in the Netherlands and one in Belgium. This testing at different football clubs had multiple benefits. For example, the training grounds at the youth academies ranged from the use of classrooms in school buildings to state-of-the-art training grounds with large gyms and indoor artificial grass pitches. This resulted in different backgrounds and different types of light exposure, which in some occasions affected certain tracking algorithms. Also, the different types of movements of athletes and the type of shoes and clothing also affected certain tracking algorithms. Hence, the testing in the practical settings also provided valuable information in improving the marker tracking algorithms.

An extra benefit of the testing at youth academies was the interaction with the athletes and the practitioners (coaches, physical therapists, strength & conditioning coaches). An important part of the interaction with the athletes was observing their behaviour during the different testing sessions. It became apparent that they were interested in the video feedback provided and that they always tried to improve their own scores and tried to beat their peers. Practitioners immediately recognized the benefits of AMAT over their current measurement methods and were especially impressed by two factors of the AMAT. At first, the practicality of the system to collect, process and analyse kinematic data. It would save them time because all outcome measures were collected automatically and were sent to a server, which means they did not have to write all data down during data collection and afterwards enter this data into a spreadsheet on the computer. Second, the ability to display the video of the movement straight after each movement. Moreover, the practitioners also provided valuable feedback in what type of movements they would like to include, such as the drop vertical jump, and what type of movement strategies they were mainly interested in, such as tracking of the knee movement, the movement of the feet during landing and the movement of the displacement of the trunk throughout all movements.

It was possible to spent approximately one year on the development and testing of the different algorithms before the data collection for this thesis had to start. As such, much data was collected during this period to improve the different movements. However, due to time and practical constraints, it was not possible to store all this data and as such, it is not possible to give a detailed description of the development of each algorithm or to present any pilot data. The next sections discuss the development of the algorithms until the moment the data collection for this thesis started. The state of the algorithms described in the next sections is also the state in which the algorithms were used throughout the rest of this thesis.

3.3.3.1 Calibrating the Kinect v2.

The first step in developing calibration algorithms was the creation of a point of origin of the 3D-coordinate system of the Kinect v2. It was chosen to take the bottom centre of the vertical wooden board as the origin of this coordinate system (Figure 3.5). An algorithm semi-automatically determined the position of the origin. This algorithm worked as follows: the approximate position of the vertical wooden board in the camera view was known. Around this position, a search area was created. This search area could best be represented as a collection of pixels within a certain area of the camera view (Figure 3.6A). The depth value of each pixel in this search area was collected during one frame. Based on the depth value of all pixels and the known distance between the camera and the wooden board, the pixels that captured the wooden board were determined (Figure 3.6B). The horizontal (x)-coordinate of the pixels that captured the wooden board were saved. By averaging these x-coordinates, the x-coordinate of the centre of the vertical wooden board could be calculated (*x-PixelOrigin*). A flowchart of this algorithm is displayed in Figure 3.7.

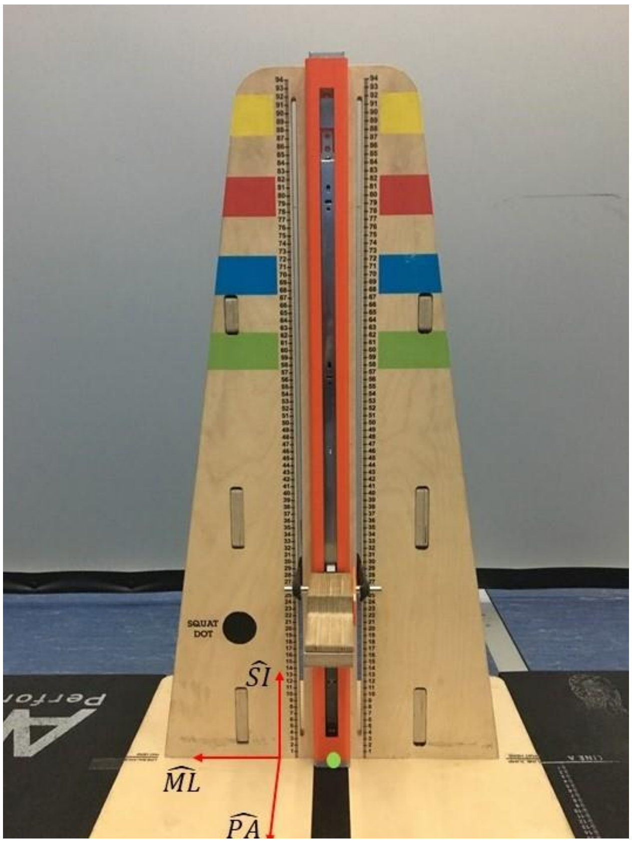

Figure 3.5. The origin of the coordinate system displayed on the vertical wooden board of the set-up of the AMAT. The green dot on the bottom of the orange slider displays the origin of the coordinate system (0,0,0). The three red arrows show the directions of the axis in the coordinate system (ML, medio-lateral axis; PA, posterior-anterior axis; SI: superior-inferior axis).

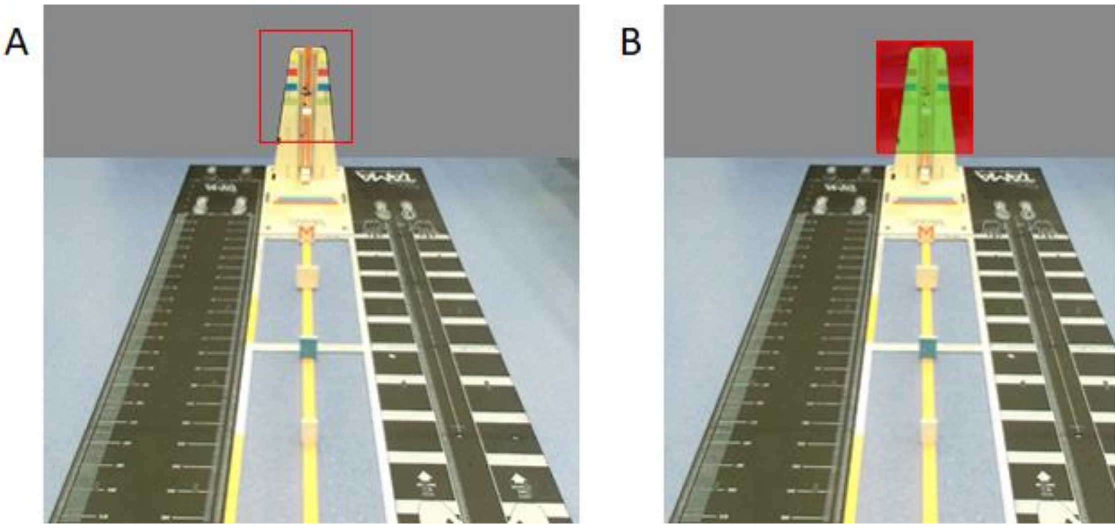

Figure 3.6. The camera view of the Kinect, with the rectangle representing the search area to determine the centre of the set-up. Figure 3.6A. The red square represents the search area. The depth value of all pixels within this search area are collected. Figure 3.6B. The pixels of which the x-value is included (green) or excluded (red).

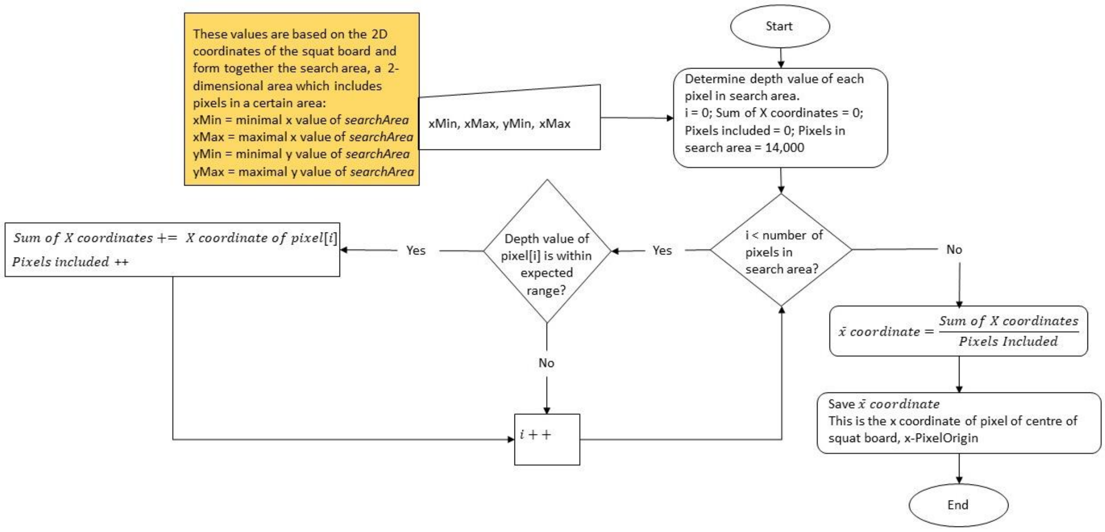

Figure 3.7. Flowchart of the algorithm to determine the x-coordinate of the centre of the squat board (*X-PixelOrigin*). X represents the horizontal coordinates of the camera view and Y represents the vertical coordinates of the camera view in 2D.

The next step was to determine the vertical (y)-coordinate of the pixel that represents the origin of the coordinate system. This was done by creating a scanline over all 424 pixels that had the x-coordinate *x-PixelOrigin* (Figure 3.8). Of all these 424 pixels, the 3D-coordinates were calculated from the 2D pixel coordinates, based on the original transformation algorithm of the SDK. This scanline was projected against a criterion line (axes of real world) in the sagittal and transverse plane (Figure 3.8). Based on the visual information of the scanlines in the two planes (Figure 3.8a), the axes of the Kinect coordinate system could be equalled with the axes of the real-world coordinate system (Figure 3.8b), via rotation and transformation matrices. These rotation and transformation matrices where thereafter added to the original 2D to 3D transformation algorithm of the Kinect SDK. With these additions, the origin of the Kinect coordinate system was set at the bottom of the vertical squat board and aligned with the directions of the different movements. This coordinate system is displayed in Figure 3.5.

A

B

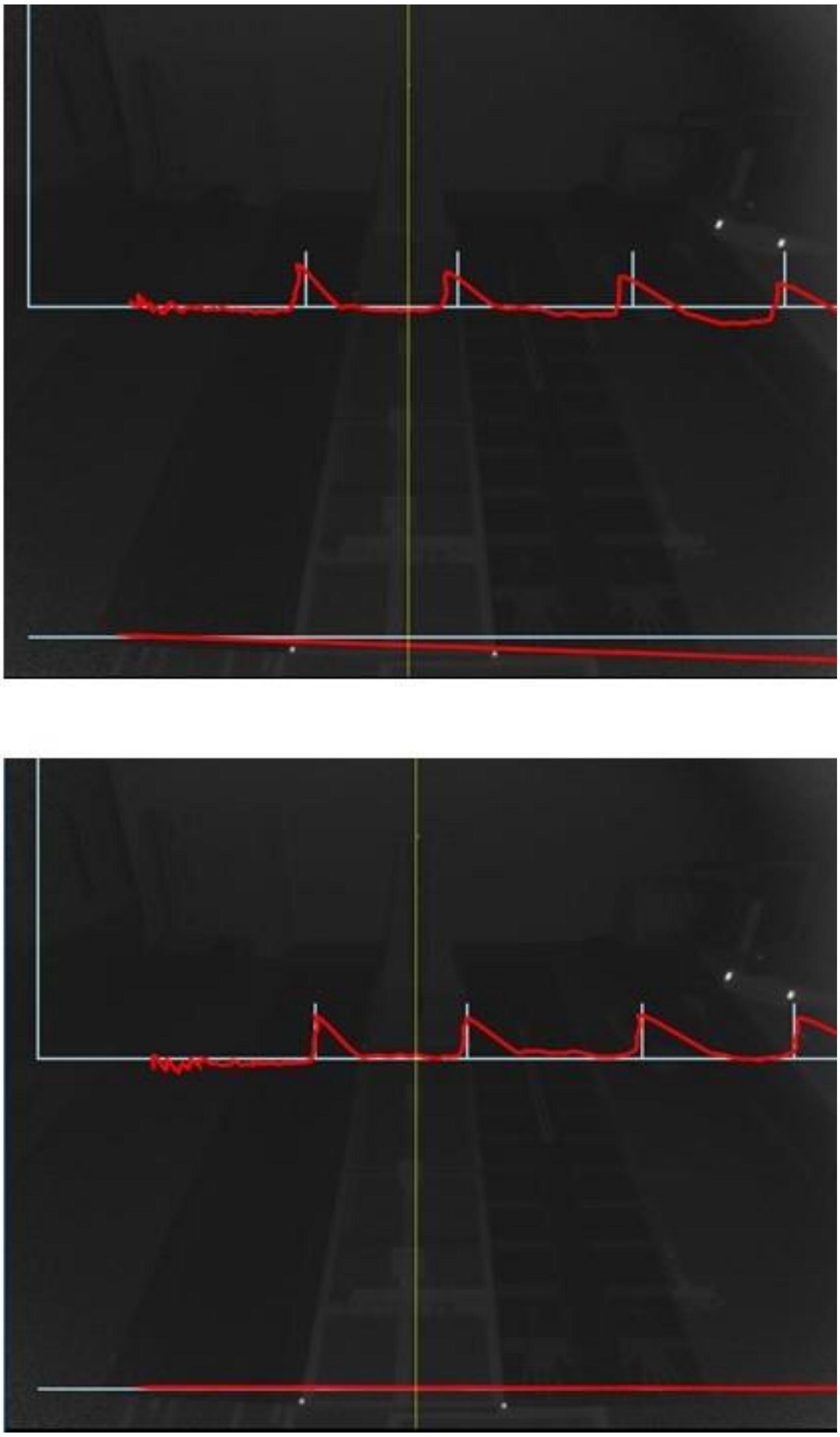

Figure 3.8. The view of the user with the scanlines that are used to optimize the calibration. The yellow line is the scanline with the x-coordinate *x-PixelOrigin*. The blue lines represent the real-world 3D coordinate system. The red lines represent the 3D coordinate system of the Kinect. The top graphs in Figures 3.8A and 3.8B represent the values of the pixels on the yellow line in the sagittal plane (x-axis: PA values; y-axis: SI values) and the bottom graphs represent the values of the pixels on the yellow line in the transverse plane (x-axis: PA values; y-axis: ML values), respectively. Figure 3.8A: Before the calibration. A clear distinction between the red and blue lines show that the calibration of the Kinect is not optimal. Figure 3.8B: After the calibration. The red and blue lines are aligned with the added rotational and transformation matrices, which implies that the real-word and Kinect coordinate systems are aligned.

3.3.3.2 Foot marker tracking.

To improve the positional data collection of the feet with the Kinect v2, a foot marker tracking algorithm was developed based on the foot marker tracking algorithms described by Paolini *et al.* (2014) and MacPherson *et al.* (2016). At first, the algorithm described by MacPherson *et al.* (2016) was almost literally copied. Namely, retroreflective infrared markers were placed on the toe of each shoe (Figure 3.9). To track these markers, a specific algorithm was developed that created a search area around the ankle joint. Thereafter, the infrared values of all pixels within this search area were determined. The highest infrared value within this search area was expected to be the foot marker (Figure 3.10). An issue with the Kinect v2 is that the depth value of pixels with a high infrared value cannot be determined. Hence, once the foot marker was detected, the pixel above was used to determine the position of the toe. A flowchart of this algorithm is displayed in Figure 3.11.

Figure 3.9. A trainer with the original retroreflective infrared marker (red string) attached to it. The idea for such a marker was based on MacPherson *et al.* (2016).

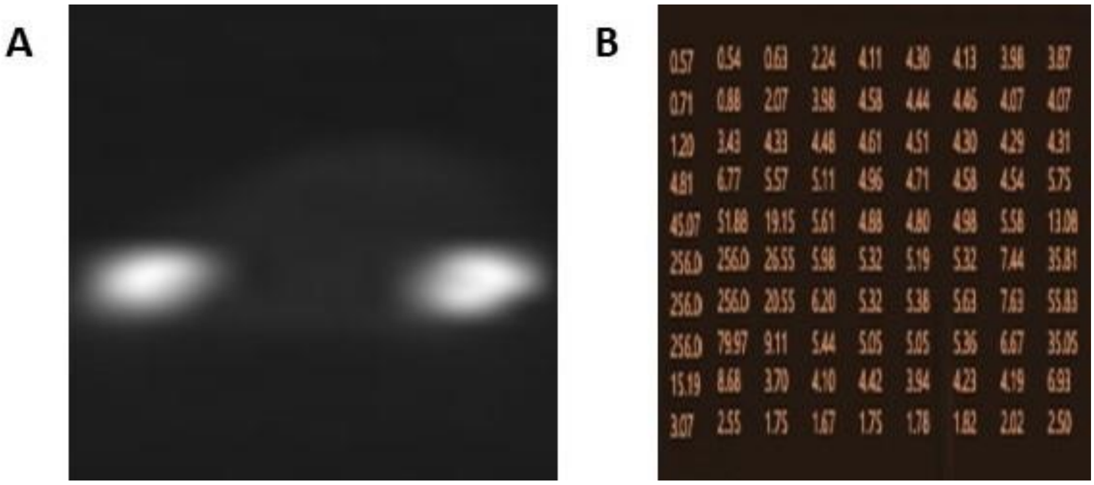

Figure 3.10. Two retroreflective infrared markers as observed by the Windows Kinect. Figure 3.10A: The view of the infrared camera, with two retroreflective markers lighting up. Figure 3.10B: The infrared values on and around one retroreflective marker, where each value represents the infrared value of one pixel. On the left side of this figure, pixels with an infrared value of 256.0 are found. These pixels have captured the retroreflective infrared marker.

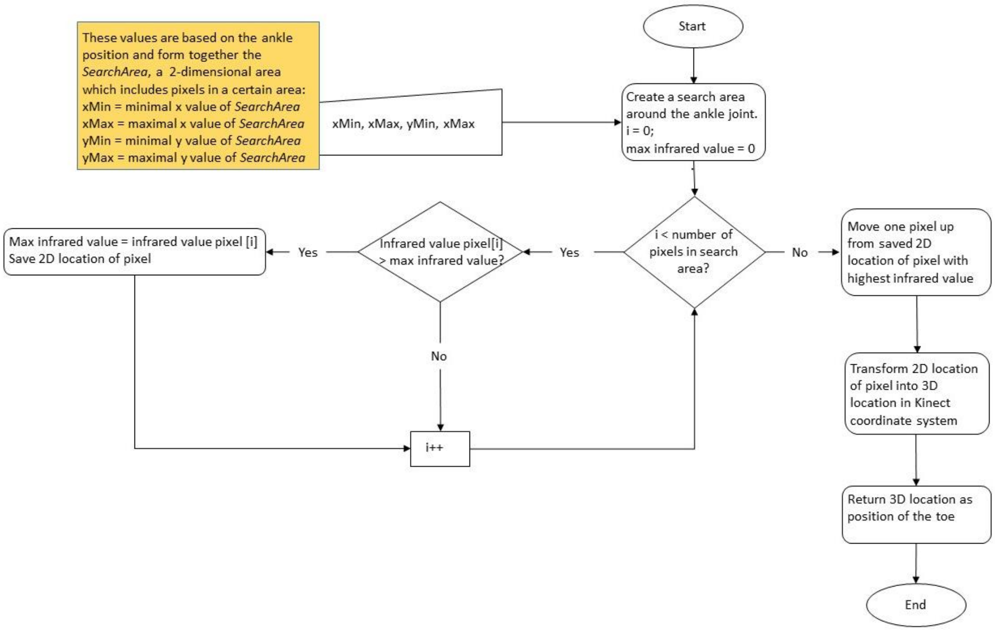

Figure 3.11. Flowchart of the first algorithm to track the foot marker. X represents the horizontal coordinates of the camera view and Y represents the vertical coordinates of the camera view in 2D.

This algorithm should have made it convenient to track the foot position accurately. However, there were certain issues with this concept. Namely, the marker was only visible when it was approximately facing the camera straight and a twisted foot caused higher errors in the foot data. Moreover, the high infrared values of the retroreflective marker reduced the validity of the foot position captured with the new algorithm. Other solutions where larger markers were placed on the toes failed too, because these markers often detached from the shoe during high impact movements such as landings. An additional issue was that there was no way to verify whether the correct marker was detected, because the markers on both feet were detected with exactly the same algorithm. This resulted in situations where the foot markers were not tracked correctly and as such, this algorithm was not reliable and valid.

To improve the marker tracking, larger markers with two small retroreflective markers attached to it were developed and were attached to the laces of the shoes (Figure 3.12). The working of this algorithm is similar to the algorithm displayed in Figure 3.11. Some additions were made for the new algorithm that focused on improving the certainty that the correct marker was identified. To that purpose, it was checked that the two retroreflective markers were not further than 10 pixels away from each other, it was determined whether the foot marker had a realistic 3D position when compared to the ankle joint and the colour of the marker was determined. Usage of the colour of the marker to identify the correct marker was previously successfully demonstrated by Paolini *et al.* (2014), as described in Section 3.2. The difference in colour (red for right, blue for left) is used to distinguish between the two feet. However, this colour information is only available when the foot is moving slowly or standing still. A flowchart of this algorithm is displayed in Figure 3.13.

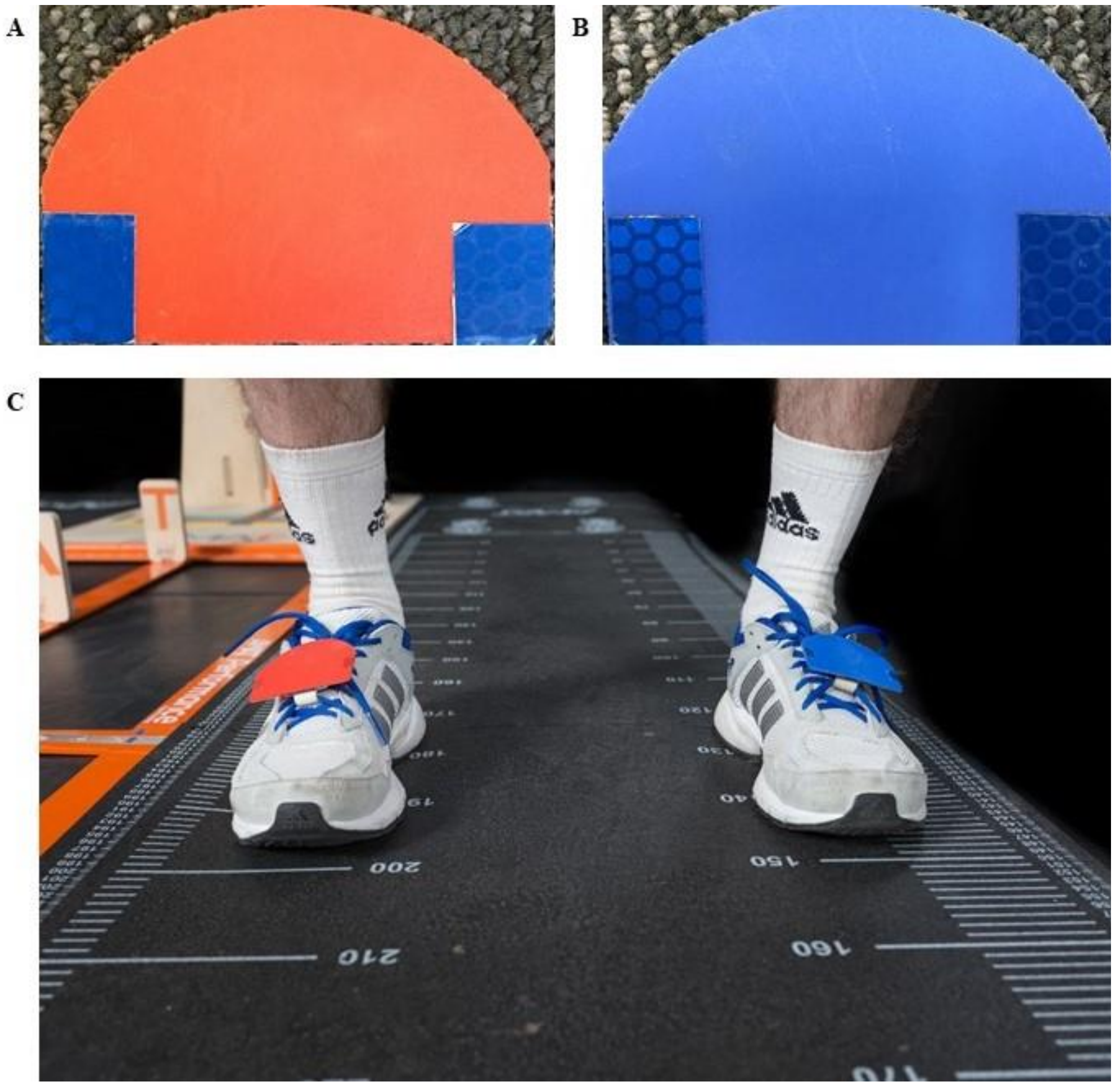

Figure 3.12. The foot markers of the AMAT. Figure 3.12A: The red foot marker with two blue retroreflective infrared markers. Figure 3.12B: The blue foot marker with two blue retroreflective infrared markers. Figure 3.12C: The red marker is attached to the right foot and the blue marker is attached to the left foot.

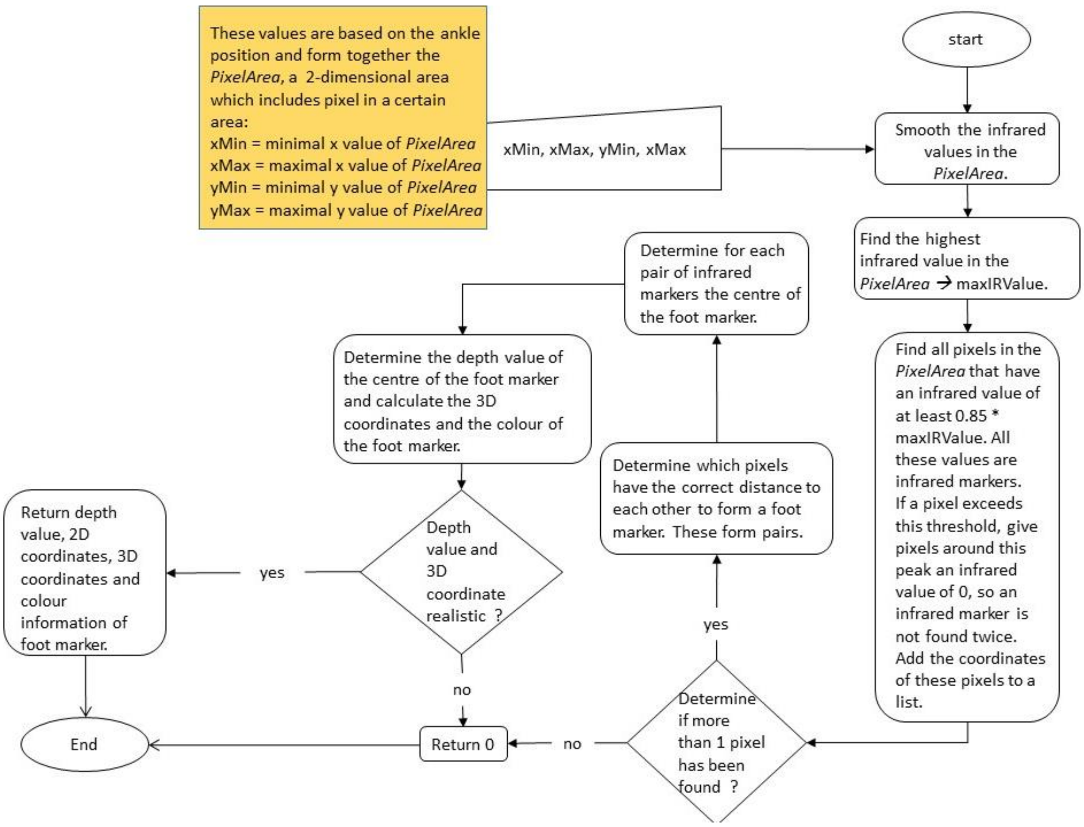

Figure 3.13. The flowchart of the improved algorithm to track the foot markers. Based on the colour data that is being captured, it can be found whether the marker is attached to the right or the left foot. X represents the horizontal coordinates of the camera view and Y represents the vertical coordinates of the camera view in 2D.

3.3.3.3 Knee marker tracking.

The tracking of anatomical landmarks or body parts is important to determine the movement strategy of an athlete. As discussed in Chapter 2, the knee abduction displacement has been related to ACL injuries and to quadriceps dominance. As such, valid tracking of the knees is vital. However, as mentioned in Section 3.2, the Kinect v2 is not valid to track the anatomical landmarks of the lower limbs. Hence, a knee marker tracking algorithm was developed to track the knees more accurately.

The development of this algorithm can be split into three different parts, namely the type of the marker, the position of the marker for optimal tracking and the marker tracking algorithm. Similar to the development of the foot marker and the foot marker tracking algorithm, the development of the knee marker and the knee marker tracking algorithm were based on MacPherson *et al.* (2016). The shape of the knee marker was constrained by the fact that the marker should not limit the movement of the athlete in any way. As such, we did not come up with other solutions than the type of retroreflective marker depicted in Figure 3.14. It was chosen to place the knee marker on top of the patella, because on this position the marker was best visible for the Kinect. In addition, as mentioned previously, pixels that capture a retroreflective infrared marker have an unreliable depth value and cannot be used to determine the position. Therefore, obtaining the position of the pixel below the knee marker would approximate the 3D position of the centre of the patella. With this information, it would be possible to determine the position of the knee relative to the foot and the movement of the knee over time during dynamic movements.

The algorithm works as follows: a search area is created around each knee, based on the skeletal tracking of the Kinect. In this search area, the pixel with the highest infrared value is found. This pixel has captured the retroreflective infrared marker. As mentioned previously, pixels that capture a retroreflective infrared marker have an unreliable depth value and cannot be used to determine the position. Therefore, a pixel below this original pixel was used to determine the position of the knee. This algorithm is very similar to the original algorithm used for the detection of the foot markers. Hence, similar to the issue with the foot markers, the left and right knee marker could not be distinguished from each other during dynamic movements. A flowchart of this algorithm for the knee detection is displayed in Figure 3.15.

Figure 3.14. Retroreflective knee markers placed on top of each patella.

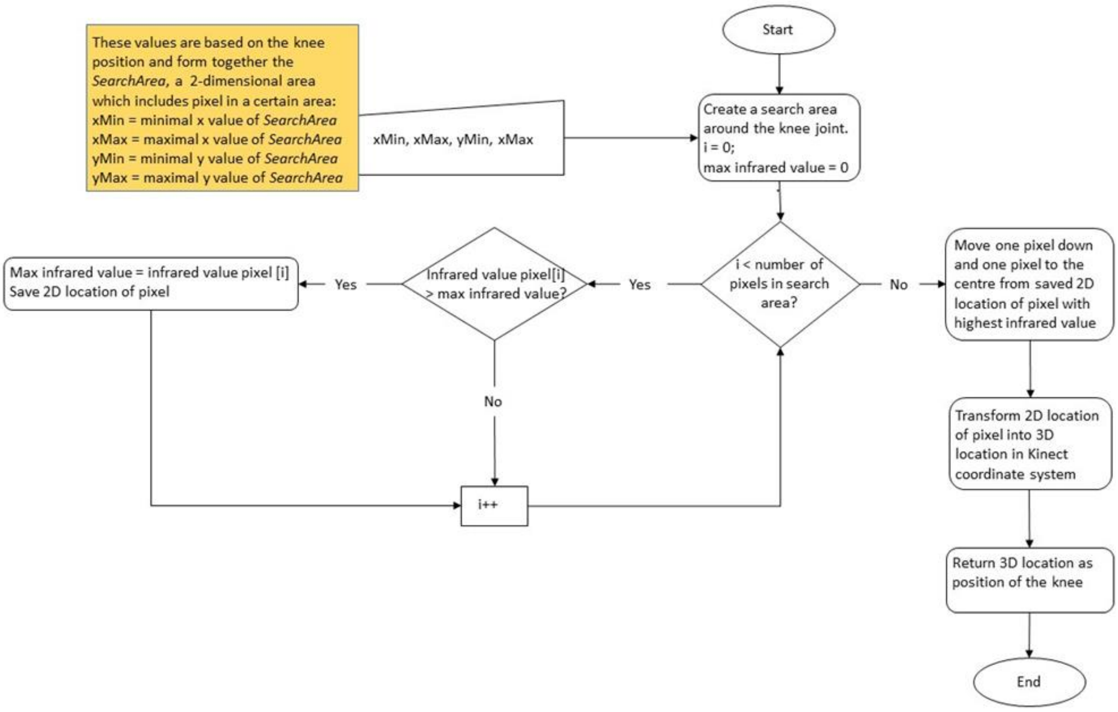

Figure 3.15. Flowchart of the algorithm to determine the position of the knee. X represents the horizontal coordinates of the camera view and Y represents the vertical coordinates of the camera view in 2D.

3.3.3.4 Determining the position of the centre of mass (centre of mass algorithm).

The collection of the centre of mass position was based on voxel data of the athlete captured with the Kinect and this method was based on multiple studies. The first studies it was based on calculated the 2D position of the centre of mass in a valid manner by using the pixels within a camera view that track the body of a human (Goffredo *et al.*, 2006; Allin *et al.*, 2008). In addition, the use of the voxel data collected with the Kinect to create a point cloud to improve the kinematic data collection has also been described (Gammelgaard, 2015; Giblin *et al.*, 2016; McGroarty *et al.*, 2016). To determine the 3D position of the centre of mass, a combination of those methods was developed.

To that purpose, an algorithm was developed that collects the voxel data of all pixels that captured the body in the camera view. The pixels within a radius of 40 pixels around each anatomical landmark, as captured with the Kinect, were selected during each frame. This implies that no markers were used to determine the positions of the different anatomical landmarks. The low validity of the Kinect to collect the position of the anatomical landmarks did not cause any issues for this algorithm. Namely, the anatomical landmarks captured with the Kinect were always positioned on the body and as such, the depth values of these anatomical landmarks were sufficiently accurate. To make optimal use of the anatomical landmarks but also take the limitations of the kinematic data in account, for each of the collected pixels it was determined whether the distance to an anatomical landmark was less than 50 centimetres and whether the depth value of the pixel was similar to the depth value of the closest anatomical landmark.

All pixels that met these requirements were included in a collection. Thereafter this collection was filtered to make sure that each pixel was only included once. Based on this logic, it was expected that all pixels that captured a part of the body were collected. Thereafter, the average 3D position of all these pixels was collected to determine the position of the centre of mass of the body during each frame. A flowchart of this algorithm to collect these pixels is shown in Figure 3.16.

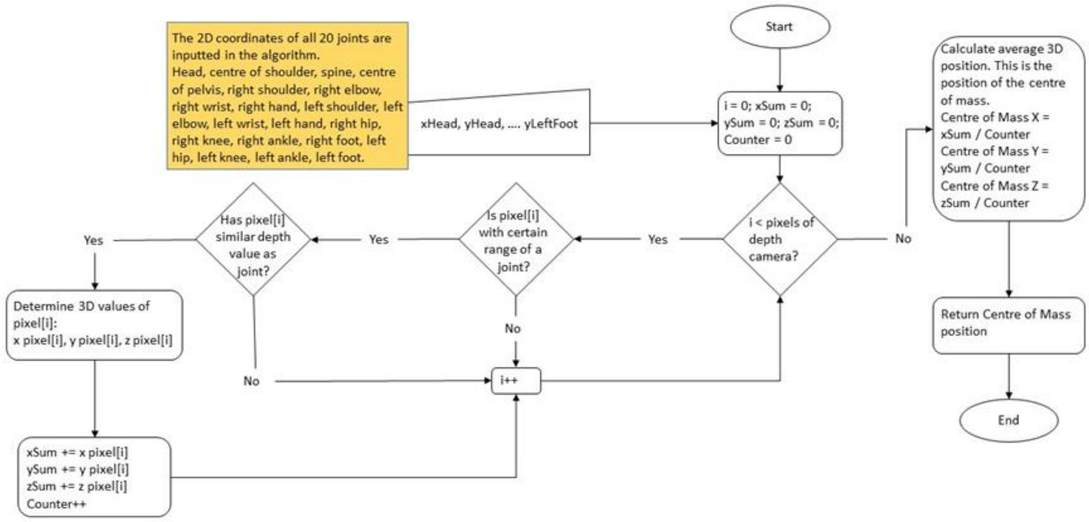

Figure 3.16. Flowchart of the centre of mass algorithm. X: medio-lateral axis. Y: superior-inferior axis. Z: posterior-anterior axis.

3.3.3.5 Additional calibration to improve the accuracy of performance outcomes.

A systematic error in the coordinate system of the Kinect v2 was found after the calibration described in Section 3.3.3.1. This systematic error is displayed in Figure 3.17 and it implies that the movement performance (i.e. jump distance) could not be determined in a valid manner. Namely, the jump distance could only be collected in a valid manner if there was no systematic or random error. As such, additional calibration was necessary to improve the reliability and validity of the AMAT.

Figure 3.17. Surface area plot prior to inclusion of the second calibration algorithm. The different colours in the figure represent the difference in centimetres between the real position and the position obtained by the Kinect^{TM} in the posterior-anterior axis. The legend displays the difference in centimetres every colour represents. The measure unit of the axis and the legend is centimetres.

The additional calibration algorithm did not focus on lining up the coordinate system of the Kinect with the real-world coordinate system. Instead, it focused on equalling the values of the posterior-anterior axis of both coordinate systems on the area where the athletes had to perform their jumps. This was done by using the coordinate system of the real world as the criterion measure. Foot markers were placed on 0, 50, 100, 150, 180, 200, and 250 centimetres from the origin in the posterior-anterior axis of the real-world coordinate system (Figure 3.18). Once these markers were placed on each position, the position of the markers in the coordinate system of the Windows Kinect were collected. Then, the difference between the expected value (i.e. position of markers in real world) and the measured value (i.e. position of markers as measured with the AMAT) of the two coordinate systems was calculated. Thereafter, the outcome of this calculation was extrapolated between the positions where data were collected to determine the difference between the two coordinate systems in the posterior-anterior axis over the full length of the mat. Based on this outcome, the difference between the two coordinate systems was added to the marker position. This resulted in the coordinate system of the real world and of the AMAT being aligned perfectly in the posterior-anterior axis.

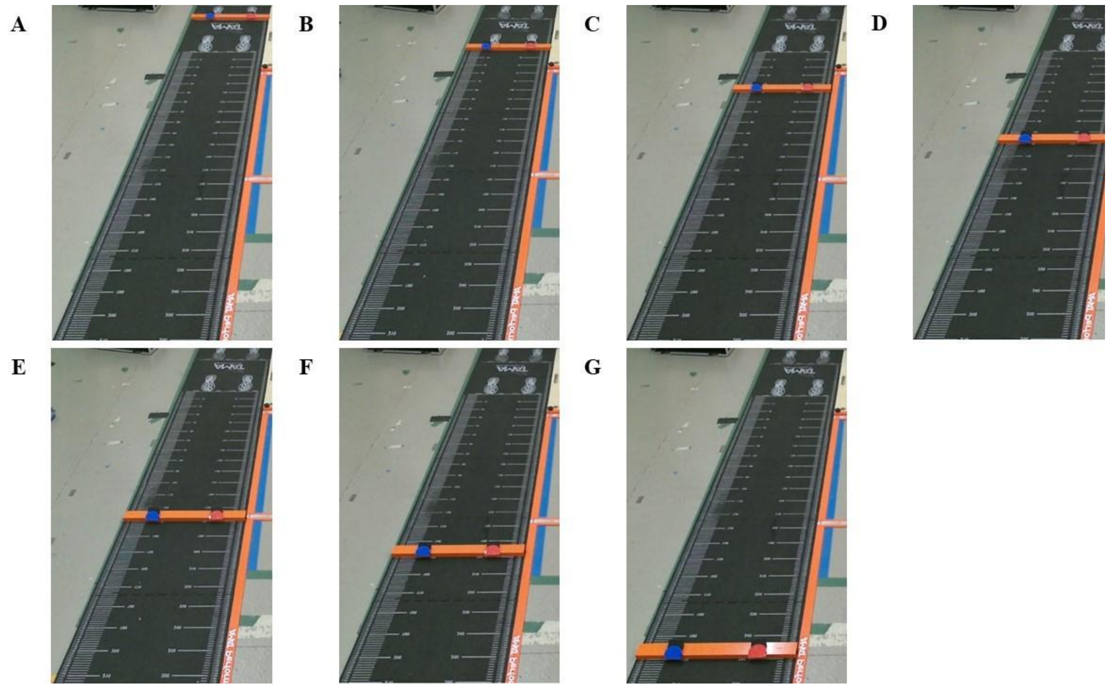

Figure 3.18. The different positions on the rubber mat where the foot markers were placed for the additional calibration. The positional data of the markers were adjusted based on the known value of the real-life coordinate system. Afterwards, this adjustment was interpolated between the positions where the markers were placed to optimize the coordinate system of the Kinect on the whole mat. Figure 3.18A: 0 cm from origin; Figure 3.18B: 50 cm from origin; Figure 3.18C: 100 cm from origin; Figure 3.18D: 150 cm from origin; Figure 3.18E: 180 cm from origin; Figure 3.18F: 200 cm from origin; Figure 3.18G: 250 cm from origin;

3.3.3.6 Determining landing position.

Anterior cruciate ligament injuries occur within the first 50 milliseconds after initial contact (Krosshaug *et al.,* 2007) and the moment of initial contact is therefore often used to measure joint angles (Hewett *et al.,* 2005; 2006b). In addition, the moment of initial contact can also be used to determine the foot position to measure jump and hop distances. Hence, an algorithm was developed to determine the frame of initial contact after jumps and hops. The first algorithm that was developed to determine the landing position made use of positional data of the feet only, because it was expected that the feet have a recognizable movement during horizontal jumps (Figure 3.19).

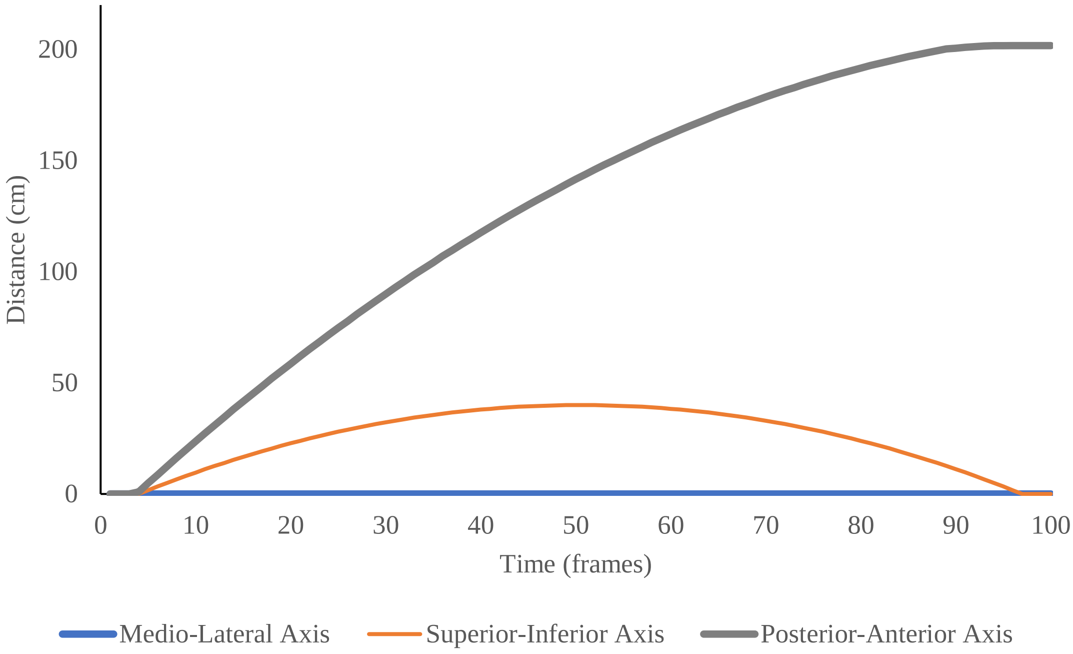

Figure 3.19. A schematic display of the expected 3D movement of the feet during a jump. Around frame 4, the athlete sets off. Around frame 100, the athlete lands.

The development of this algorithm sounds straightforward, but there are a few glitches. Namely, Figure 3.19 is an expectation of the feet movement during a jump, whereas the exact movement strategy during jumps is different for each person. Also, an issue with the foot marker tracking algorithm is that it only works correctly when there is none or minor foot movement. Hence, during the flight phase of the jump, the foot marker position can only be estimated. This implies that the positional data of the foot can be measured accurately during take-off and after the landing only. Moreover, as described in Section 3.3.3.5, the calibration described in Section 3.3.3.1 was not perfect. Thus, Section 3.3.3.5 described additional algorithms to improve the reliability and validity for the collection of the foot marker position in the posterior-anterior axis. However, no additional calibration was performed in the superior-inferior or medio-lateral axis. This implied that the position of the feet in the superior-inferior axis during the landing was not the same as before the jumps and as such, this positional data could not be used to determine the moment of initial contact. These issues became apparent because during several testing sessions the jump distance was not measured accurately. As such, it was concluded that this algorithm was not valid to determine the position of the foot at initial contact.

The development of the new algorithms to determine the landing position ran similar with the development of the new half circle foot markers discussed in Section 3.3.3.2. As mentioned in that section, these new markers made the tracking more reliable and valid, for example due to the use of a blue (for the left foot) and a red (for the right foot) marker. The colour of these markers was only captured correctly by the Kinect camera when the markers were only moving slightly or not at all. Therefore, the colours of the markers could not be determined during the flight phase of the jump but could be determined before (during the push-off phase) and after (during the landing) the flight phase. As such, this colour information was used in the new algorithm to determine the stage of the movement (i.e. push-off, flying through air, landed). Moreover, the difference in colour of the feet markers could also be used to distinguish between the left and the right foot. The centre of mass movement (discussed in Section 3.3.3.4) was also used in the new algorithm to determine the stage of the movement. How these variables are used in this algorithm is displayed in a flowchart (Figure 3.20).

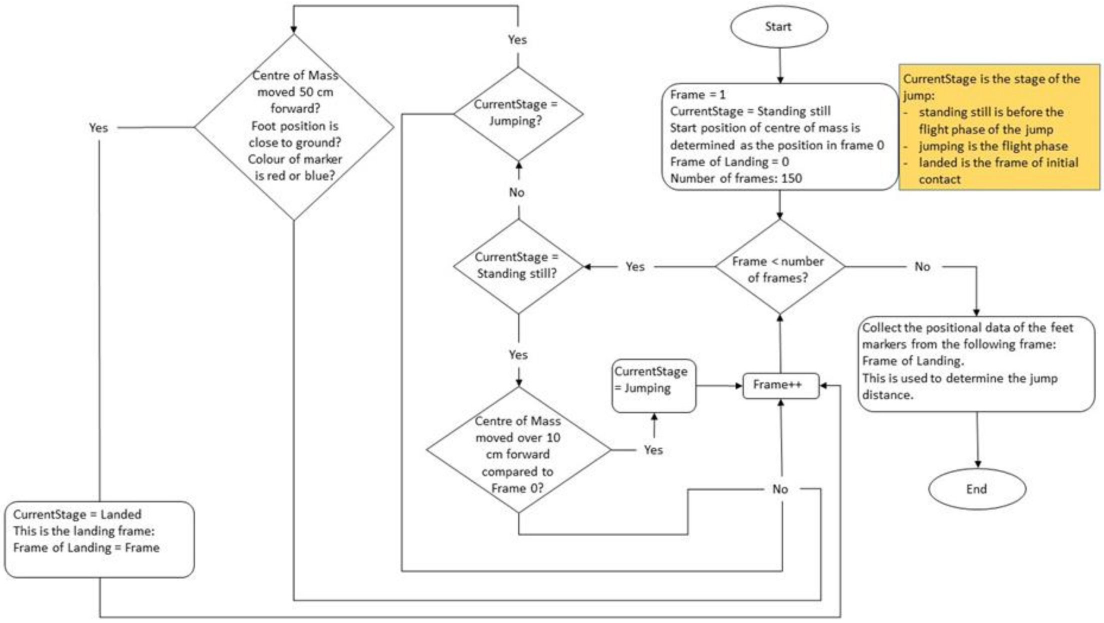

Figure 3.20. Flowchart of the algorithm used to determine the frame of landing.

3.3.4 The development of an app to make the Kinect v2 convenient in use.

As mentioned previously in this chapter, the SDK of the Kinect v2 is beneficial from a developer's perspective because it can be used to create new algorithms and collect data easily. However, this SDK is not practical in use for practitioners. As such, applications were developed to make it convenient for practitioners to use the AMAT to collect kinematic data.

The Kinect v2 only works when it is connected to a USB 3.0 port. This implies that the Kinect v2 needs to be connected to a laptop or computer, because tablets do not have USB 3.0 ports. However, the use of a laptop or computer is not always practical. Namely, a practitioner is often on his own with multiple athletes. The practitioner might want to talk to the athletes, give them instructions, or show them how a certain movement works. If the practitioner needs to walk back to the laptop/computer afterwards to start the data collection, this takes extra unnecessary time. Hence, two apps were developed.

The first app runs from the laptop. This app uses the algorithms from the Windows Kinect SDK to collect the data captured by the Kinect v2: the data of the different anatomical landmarks, the normal camera data of all pixels, the infrared value of all pixels and the depth value of all pixels. Moreover, the algorithms described in Section 3.3.3 were added to this app to collect the positional data of the different markers and objects. The app is constantly collecting the data, but it only stores the data if it gets a signal from another app that it should record the data. The second app works from a tablet. Its main function is to make it easy for the practitioner to select which player is going to perform a movement and which movement has to be performed.

Due to the automatic positional data collection of the anatomical landmarks, the feet and knee markers and the centre of mass, it should be possible to automatically quantify the different factors related to the sensorimotor system. To this purpose, algorithms were developed that quantified these factors based on the kinematic data collection. Because of this automatic quantification, it became possible to give direct objective feedback to the participants.

3.3.5 The development of an app to display video feedback directly after a movement.

Video analysis might be an effective feedback tool to improve movement (Gokeler *et al.*, 2013; Benjaminse *et al.*, 2015; 2017; Welling *et al.*, 2016; 2017). In Section 3.3.3.4 it was explained that the voxel data is used to approximate the position of the centre of mass by collecting the 3D positional data of each pixel. In addition, this voxel data can also be used to create a video of a movement. This works as follows. Besides the 3D position data of each pixel, the Kinect is also able to collect normal camera data of each pixel during each frame. This implies that of all pixels that cover the body, the 3D position and the colour of that pixel are known. This data is saved and sent to the tablet app.

The tablet app is built with Unity3D, a gaming engine tool. One of the features of Unity3D is that a particle system can be created. A particle system exists of many small images that together form a certain fluid image. In Unity3D, the particle system can be given certain characteristics. In this case, each particle got the characteristics of a voxel. Thus, each particle's position was identical to the positional data of a voxel. In addition, this particle was given the same colour as the voxel. When doing this for the data of each frame, a video of that person is created, as depicted in Figure 3.21. Moreover, due to the 3D data of each pixel, the video itself is also displayed in 3D. As such, the practitioner can view the video from different angles in the app by simply swiping his finger over the screen. To make the video look realistic, the video is displayed in an augmented reality system that represents the AMAT.

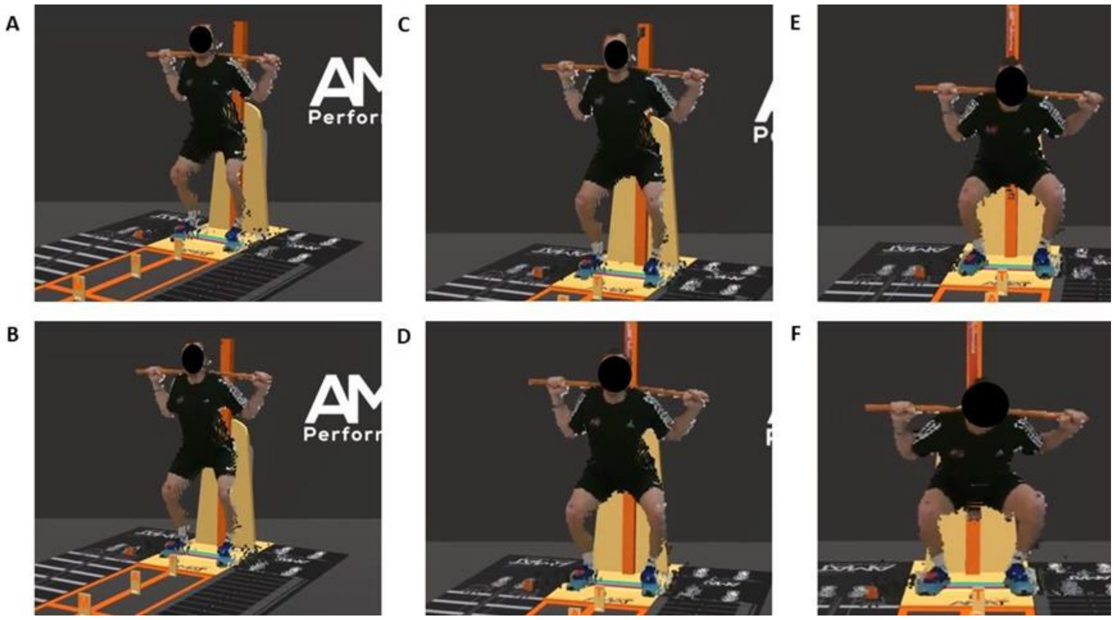

Figure 3.21 Six screenshots of an athlete performing a back squat captured with the video feedback used in the app. Figures 3.21 A – F show a squat performance over time. The practitioners can zoom in and zoom out and can change the view to take a closer look on specific parts of the body. A black object is placed in front of the face for anonymization purposes.

3.4 Discussion

Based on Chapter 2, it was found necessary to develop a practical movement assessment tool that is able to collect kinematic data in a reliable and valid manner to quantify sensorimotor risk factors of lower extremity injuries. The Kinect, a depth-sensing technology, has been proposed as part of a movement assessment tool to collect kinematic data (Clark *et al.*, 2012; Dutta, 2012; Bonnechere *et al.*, 2014). Therefore, this study aimed to determine the strengths and weaknesses of the Kinect and based on this, show the technological development of a movement assessment tool that is able to collect kinematic variables, the AMAT. The strengths of the Kinect are the depth-sensing technology and the infrared camera, its ability to collect 3D data of several anatomical landmarks and the SDK to develop new algorithms to improve the working of the Kinect v2. A weakness is that the kinematic data of the lower extremity anatomical landmarks is not valid and as such, the Kinect v2 cannot be used to collect kinematic variables (Van Diest *et al.*, 2014; Wang *et al.*, 2015; Otte *et al.*, 2016; Eltoukhy *et al.*, 2017). In addition, the Kinect v2 is not practical in use (Bujang *et al.*, 2015). Therefore, this study showed the development of new algorithms for the Kinect v2 to improve the kinematic data collection with the Kinect v2 and the development of an app to improve the practicality of the Kinect v2.

The structure of this study was completely novel. Based on previous studies on the Kinect, new algorithms were developed that can improve data collection of kinematic variables. For example, based on the idea proposed by Clark *et al.* (2013a; 2013b) to calibrate the Kinect, calibration algorithms were developed to align the coordinate system of the Kinect v2 with the directions of the different movements. In addition, algorithms were used to collect foot, knee and centre of mass positional data. Marker tracking algorithms were developed to determine the position of the feet and knees. These algorithms were based on previous research of Paolini *et al.* (2014) and MacPherson *et al.* (2016). More specifically, the foot marker tracking algorithm is a combination of the use of the retroreflective markers used by MacPherson *et al.* (2016) and the use of coloured markers described by Paolini *et al.* (2014). In contrast, the knee marker tracking algorithm was solely based on MacPherson *et al.* (2016), because no suitable methods were found to place coloured markers on the knees that could be tracked accurately. The centre of mass algorithm was based on Goffredo *et al.* (2006) and Allin *et al.* (2008) who also used pixel data to calculate the position of the centre of mass and based on Gammelgaard (2015), Giblin *et al.* (2016) and McGroarty *et al.* (2016) who used the voxel data of the Kinect to determine the position of anatomical landmarks. Moreover, the algorithm that determines the landing frame can be used to assess the movements during landing. However, the Kinect only has a sampling frequency of 30 Hz, which implies that only a few frames of high speed can be collected during the landing. This might be an issue, for example because anterior cruciate ligament injuries occur within the first 50 milliseconds after a landing (Krosshaug *et al.*, 2007).

The development of these algorithms to determine the position of the feet, knees and centre of mass implies that the minimal requirements are available to quantify the ability to maintain balance and that the movement of the knee can be determined, for example to calculate the knee abduction angle. However, the development of the algorithms could only last for approximately a year and it was not possible in this timeframe to develop a knee marker tracking algorithm that collected the position of the knees in a reliable and valid manner during high speed movements. Moreover, it was not possible in the timeframe of this thesis to develop algorithms for all movements described in section 3.3.2 Therefore, future studies should focus on improving the knee marker tracking algorithm and focus on capturing kinematic data during for example squats and balance movements. The next chapters will focus on determining the reliability and validity of the foot marker tracking algorithm and centre of mass algorithm during horizontal jumps, because those were the only two algorithms that were fully functioning before the start of the data collection. It was chosen to focus on horizontal jumps because more speed is generated during this movement compared to the other movements discussed in Section 3.3. This implies that if the foot marker tracking algorithm and the centre of mass algorithm are reliable and valid during the horizontal jumps, the AMAT will probably also be able to track the kinematic data in a reliable and valid manner during other movements.

4.1 Introduction

The foot marker tracking algorithm of the AMAT described in Chapter 3 is important because the positional data of the feet is the least reliable anatomical landmark that is collected with the skeletal tracking of the Kinect. Namely, the errors of the foot tracking are ranging from 36-61 and 32 – 62 millimetres with the Kinect v1 and v2, respectively (Van Diest *et al.*, 2014; Wang *et al.*, 2015; Otte *et al.*, 2016). In addition, the positional data of the feet is necessary to determine the position of the base of support, which is vital to quantify the ability of one to maintain balance when no kinetic data can be collected (Winter, 1995). Moreover, the positional data of the feet are used in multiple movement assessment tools, such as to measure hop and jump distances (Almuzaini & Fleck, 2008; Meylan *et al.*, 2009; Porter *et al.*, 2012), to assess movement strategies during a landing (Onate *et al.*, 2005; Myer *et al.*, 2008b; Padua *et al.*, 2009), to measure squat performance (Butler *et al.*, 2010; Mauntel *et al.*, 2013) and gait assessment (Kharazi *et al.*, 2015; Mentiplay *et al.*, 2015; Eltoukhy *et al.*, 2017). However, there is currently no information available about the reliability and validity of this foot marker tracking algorithm.

To use any type of measurement system, it should be reliable and valid (Atkinson & Nevill, 1998). *Reliability* is the ability of a measurement system to consistently provide the same measurement when repeated (Baumgarter, 1989). The reliability of a measurement system can be defined as the technological error (Hopkins, 2000). In the case of this study, the technological error is a combination of the error of the foot marker tracking algorithm of the AMAT and the error of the depth-sensing technology of the Kinect. No information is available about the error of the foot marker tracking algorithm, but the error in the depth-sensing camera ranges from less than two millimetres to more than four millimetres, dependent on the position in the camera view (Yang *et al.*, 2015). The error is smallest closer and right in front of the camera, whereas further away from the camera and more to the sides of the camera view, the error increases. *Validity* is the ability of a measurement system to measure what it is designed to measure (Atkinson & Nevill, 1998). The AMAT is designed to measure distances between markers, for example to collect performance measures and kinematic data during jumps, hops and squats. To determine whether the foot marker tracking algorithm of the AMAT is valid, it should be compared with a laboratory-based marker tracking system, because these systems are considered the gold standard to collect kinematic data via marker tracking (Padua *et al.*, 2009; MacPherson *et al.*, 2016).

No information is currently available about the reliability and validity of AMAT or about individual algorithms of the AMAT. However, the AMAT should only be used in a practical or scientific setting if it is reliable and valid. To determine the pure reliability and validity of the foot marker tracking of the AMAT, it is important to exclude other sources that can add error, to purely measure the error of the technological system (Hopkins, 2000). In the case of the AMAT, this implies that a first reliability and validity study should be performed during static situations. The jumps developed for the AMAT system all occur in the posterior-anterior axis and as such, this study aims to determine the reliability, i.e. the within- and between-trial technological error, of the foot marker tracking of the AMAT during static measurements in the posterior-anterior axis. In addition, it aims to determine the validity of the foot marker tracking of the AMAT during static situations in the posterior-anterior axis when compared to Vicon, a gold standard to collect marker data of anatomical landmarks in a biomechanical setting.

4.2 Methods

4.2.1 Participants.

Four employees and one student (31.8 ± 3.7 years old) of Teesside University participated in this study. They had prior experience with data collection in a sports science setting. These participants were selected because it was known that they could collect the data for this study in a reliable and valid manner. Moreover, these participants were selected to increase the objectivity of this study compared to when the author would have collected all data on his own. Ethical approval was obtained from the ethics committee at Teesside University, School of Social Sciences, Business and Law (Appendix 1). Informed consent of all participants was collected prior to data collection.

4.2.2 Study design.

An observational study was performed with all participants coming in on two separate days, with one week between the two data-collection moments. Data were collected on six different days. Two newly developed foot markers of the AMAT system (Section 3.3.3.2) were attached to blocks and were placed on an aluminium bar (Figure 4.1). The position of these foot markers in the posterior-anterior axis was determined with the foot marker tracking algorithm of the AMAT, as described in Section 3.3.3.2. A retroreflective marker was placed on the back of each block (Figure 4.1) and the positional data in the posterior-anterior axis of these markers were determined with Vicon (Vicon MX13 and Vicon Nexus 1.7, Vicon Motion Systems, UK).

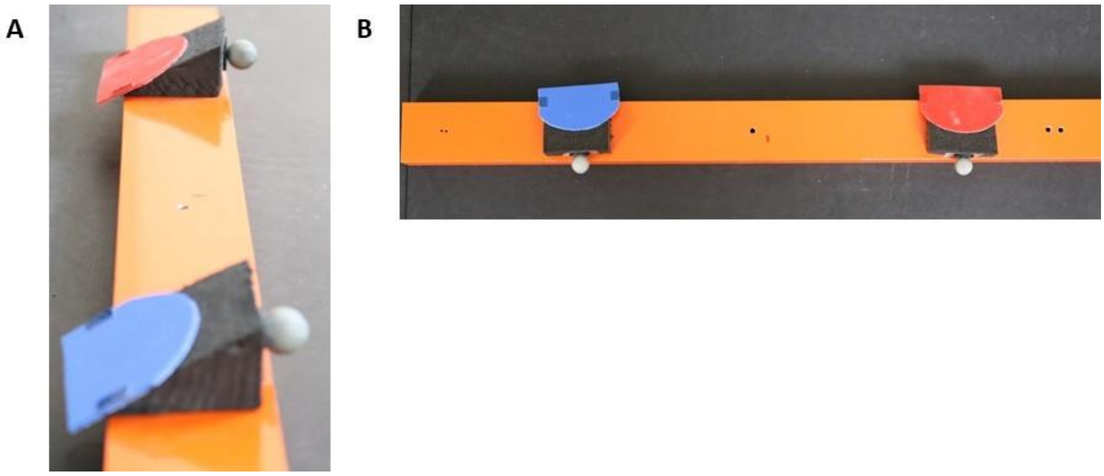

Figure 4.1. The aluminium bar with the foot and Vicon markers attached to the blocks. Figure 4.1A. Sagittal view of the aluminium bar with the markers attached to it. Figure 4.1B. Transverse view of the aluminium bar with the markers attached to it.

The set-up of the system is described in Section 3.3.1. Additionally, six Vicon cameras were placed behind the system (Figure 4.2). The coordinate systems of both measurement systems are explained in Figure 4.2. The marker tracking of the AMAT was used as the practical measure and the marker tracking of Vicon was used as the criterion measure.

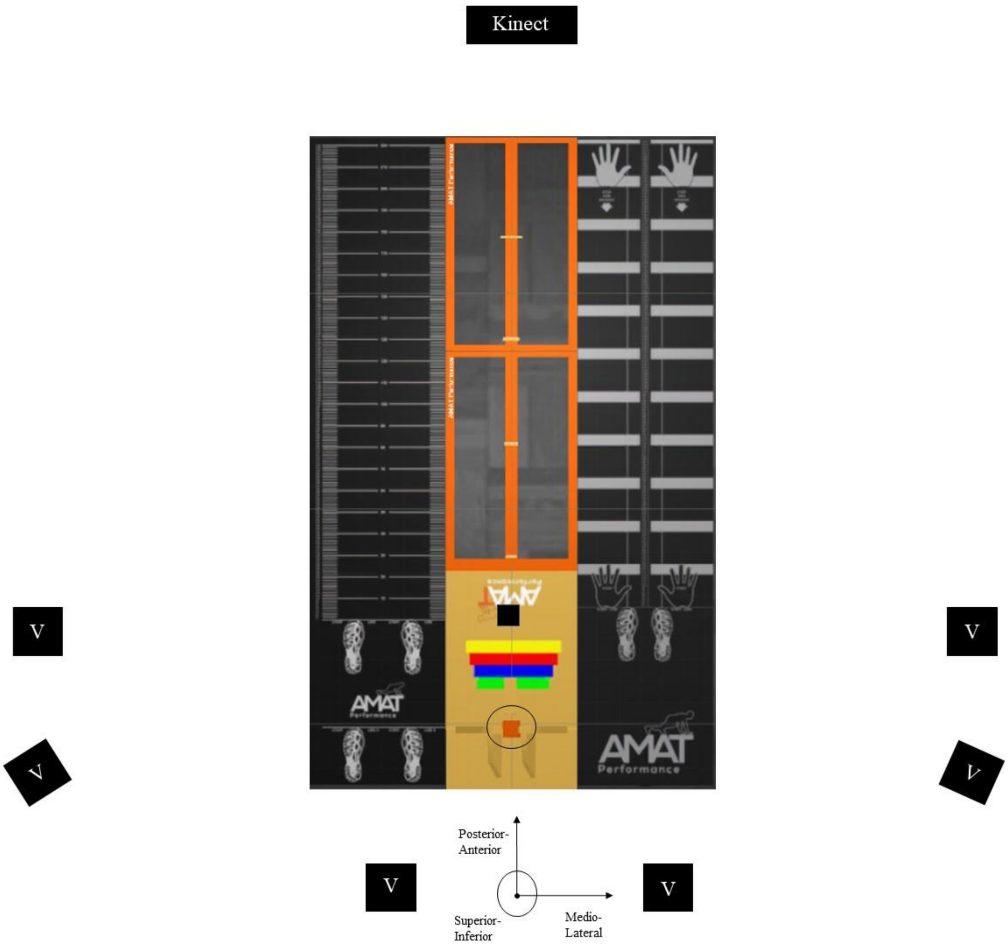

Figure 4.2. The set-up of the system, with the Kinect camera in front of the set-up and the six Vicon cameras behind the set-up. With this set-up, it was possible to collect the new foot markers with the Kinect camera and the retroreflective Vicon markers with the Vicon camera, without both systems interfering. Similar to what is depicted in Figure 3.5, the origin of the Kinect is at the squat board, here displayed as the encircled orange square at the bottom of the figure. The origin of the Vicon was approximately 50 centimetres in front of this and this origin is displayed with a black square. In the Medio-Lateral axis, positions left of the origin have negative values and positions right of the origin have positive values. In the posterior-anterior axis, the positions above the origin have positive values and the positions below the origin have negative values.

4.2.3 Protocol.

At the start of each testing day, the AMAT system had to be set-up and both the AMAT and Vicon were calibrated. The protocol was the same for each participant on both testing days. At first, the participant received instructions on which positions the bar had to be placed and that it was vital to place the bars on these positions in an accurate manner. The participants had to place the aluminium bar with the two foot markers and two Vicon markers each day six times on eighteen pre-determined positions. All these positions were within the set-up described above and within the camera view of the Kinect camera and Vicon (Figure 4.3). The participant placed the bar subsequently on Bar-Position 1, 2, 3 … 18. Each position was marked with tape so the participants knew exactly where to place the bar. This was done to reduce the within- and between-participant error when placing the bar on the different positions. Every time the bar was placed on a new position, the positional data of all four markers were collected for five seconds.

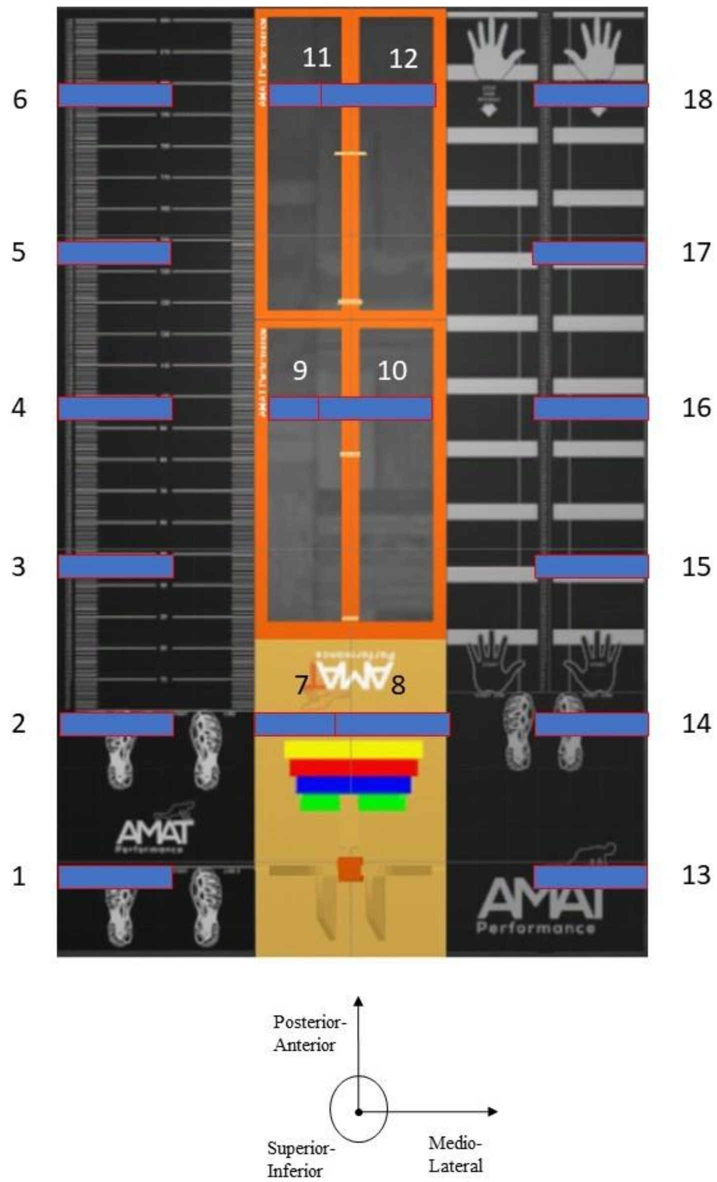

Figure 4.3. The eighteen Bar-Positions where the foot markers were positioned. Bar-Positions 1 – 6 and 13 – 18 were all 500 millimetres away from each other in the posterior-anterior axis. Bar-Positions 7, 9, 11, and Bar-Positions 8, 10, 12 were all 1000 millimetres away from each other in the posterior-anterior axis.

The Vicon system had a sampling frequency of 100 Hz and the Windows Kinect™ camera had a sampling frequency of 30 Hz. This implies that per trial, 500 data points per marker were collected with Vicon and 150 data points per marker were collected with the Windows Kinect™ camera.

4.2.4 Data analysis.

To compare both systems, the same number of data points per trial for both systems were included. Therefore, the data collected with Vicon were reduced from the original sampling frequency of 100 Hz to 30 Hz. This implies that for both measurement systems, 150 data points per marker per measurement were included for further analysis. An issue was that Vicon and AMAT have a right-handed and left-handed coordinate system, respectively. As such, it was decided to multiply the data of the Vicon system with -1 to set the posterior-anterior axis of both systems in the same direction.

4.2.5 Statistical analysis.

Based on the findings of Yang *et al.* (2015) that the error of the depth-sensing technology of the Kinect differs between different parts of the camera view, the error of the foot marker tracking algorithm was determined per marker for each Bar-Position individually. Visual inspection of Q-Q plots of raw data indicated that the data were approximately normally distributed per Bar-Position, although the Vicon data showed signs of a light-tailed distribution (Appendix 2). To compare the Vicon and Kinect data, the raw positional data of each marker are displayed as mean ± standard deviation (SD). Data were not log-transformed because the data could be negative and because the data were measured from an arbitrary reference value (Hopkins, 2015).

To determine the within-trial reliability (technological error) of the AMAT system and Vicon, the standard deviations of the different trials per Bar-Position were averaged with the formula:

$$SD_{bar\ position} = \frac{\sum_{i=1}^{n} SD_i}{n}$$

Here, n is the number of trials per Bar-Position, and SD_i is the standard deviation of an individual trial of five seconds.

To determine the between-trial reliability of the AMAT system and Vicon, the typical error and intraclass correlation between subsequent trials of the participants were calculated per Hopkins *et al.,* (2015). For this calculation, the average position of the markers per trial was used, because the within-trial error (Lachat *et al.,* 2015; Yang *et al.,* 2015) could have affected the between-trial calculation. After the correlations were performed, qualitative inferences were based on the following thresholds: <0.20, *very low*; 0.20 – 0.50, *low*; 0.50 – 0.75, *moderate*; 0.75 – 0.90, *high;* 0.90 – 0.99, *very high*; >0.99, *extremely high* (Malcata *et al.,* 2014).

To determine the criterion validity of the foot marker tracking of AMAT, the difference in distance between adjacent marker positions (± SD) as measured with AMAT and Vicon were calculated to determine the validity of the new foot marker tracking following the formula:

$$\mu_{AMAT-Vicon} = \frac{\sum_{i=1}^{n} \mu_{distance\ between\ markers\ AMAT,i} - \mu_{distance\ between\ markers\ Vicon,i}}{n}$$

Here, n is the number of trials and $\mu_{distance\ between\ markers}$ was calculated via

$$\mu_{distance\ between\ markers} = \frac{\sum_{i=1}^{n} \mu_{BP+1,i} - \mu_{BP,i}}{n}$$

Here, n is the number of trials, $\mu_{BP+1,n}$ is the mean marker position of the nth trial on Bar-Position $_{x+1}$ and $\mu_{BP,n}$ is the mean marker position of the nth trial on Bar-Position $_{x}$.

In addition, to determine the agreement in distance measured between the Bar-Positions between the AMAT and Vicon, a Bland-Altman plot (Bland & Altman, 1986) was created to determine whether this agreement differed between different distances from the camera. For all estimates, the uncertainty was expressed as 90% confidence limits.

4.3 Results

The raw data per Bar-Position of AMAT and Vicon and the distance between the adjacent Bar-Positions as measured with AMAT and Vicon are displayed in Appendix 3. Table 4.1 displays the technological error (expressed as SD) for each Bar-Position for the AMAT and the Vicon system. For the AMAT the technological error ranged from 1.36 to 3.30 millimetres and for Vicon the technological error ranged from 0.05 to 0.15 millimetres.

Table 4.1. The technological error (± SD, in millimetre) in the posterior-anterior axis per Bar-Position of the AMAT and Vicon.

	AMAT	Vicon
Bar-Position 1	2.82 ± 0.97	0.11 ± 0.09
Bar-Position 2	3.30 ± 1.34	0.05 ± 0.04
Bar-Position 3	2.53 ± 0.49	0.05 ± 0.04
Bar-Position 4	2.02 ± 0.35	0.05 ± 0.04
Bar-Position 5	1.50 ± 0.38	0.07 ± 0.06
Bar-Position 6	1.36 ± 0.21	0.09 ± 0.08
Bar-Position 7	3.04 ± 1.51	0.05 ± 0.05
Bar-Position 8	3.22 ± 1.65	0.05 ± 0.06
Bar-Position 9	2.12 ± 0.66	0.05 ± 0.05
Bar-Position 10	2.09 ± 0.45	0.05 ± 0.05
Bar-Position 11	1.24 ± 0.23	0.11 ± 0.09
Bar-Position 12	1.17 ± 0.17	0.15 ± 0.12
Bar-Position 13	2.79 ± 0.86	0.12 ± 0.14
Bar-Position 14	3.10 ± 1.10	0.06 ± 0.05
Bar-Position 15	2.58 ± 0.55	0.05 ± 0.04
Bar-Position 16	1.86 ± 0.48	0.06 ± 0.05
Bar-Position 17	1.44 ± 0.27	0.07 ± 0.07
Bar-Position 18	1.62 ± 0.25	0.13 ± 0.14

Table 4.2 and Table 4.3 display the between-reliability measures. The typical error of the AMAT ranged from 0.96 to 6.57 millimetres and the typical error of Vicon ranged from 0.30 to 0.90 millimetres (Table 4.2). The intraclass correlations between the trials as measured with AMAT ranged from 0.61 (moderate) to 0.94 (very high) (Table 4.3). The intraclass correlations between the trials as measured with Vicon were all very high and ranged from 0.94 to 0.99 (Table 4.3).

Table 4.2. Between-trial typical error (TE, in mm) and 90% confidence intervals (CI) of AMAT and Vicon.

	AMAT		Vicon	
	TE	90% CI	TE	90% CI
Area 1	3.98	3.51 - 4.61	0.47	0.41 - 0.54
Area 2	5.02	4.43 - 5.82	0.47	0.42 - 0.55
Area 3	3.49	3.08 - 4.04	0.41	0.36 - 0.47
Area 4	2.22	1.96 - 2.57	0.65	0.58 - 0.76
Area 5	1.48	1.31 - 1.71	0.67	0.59 - 0.78
Area 6	1.24	1.10 - 1.44	0.72	0.63 - 0.83
Area 7	5.77	5.11 - 6.73	0.30	0.27 - 0.35
Area 8	6.57	5.80 - 7.61	0.90	0.80 - 1.05
Area 9	1.63	1.44 - 1.89	0.41	0.36 - 0.48
Area 10	1.94	1.71 - 2.24	0.34	0.30 - 0.39
Area 11	0.96	0.85 - 1.12	0.40	0.36 - 0.47
Area 12	0.82	0.73 - 0.96	0.36	0.31 - 0.41
Area 13	3.25	2.87 - 3.77	0.71	0.63 - 0.83
Area 14	4.68	4.12 - 5.43	0.81	0.72 - 0.94
Area 15	2.48	2.19 - 2.88	0.78	0.69 - 0.91
Area 16	1.73	1.53 - 2.00	0.80	0.71 - 0.93
Area 17	1.43	1.26 - 1.65	0.66	0.58 - 0.76
Area 18	1.06	0.94 - 1.23	0.83	0.74 - 0.97

Table 4.3. Between-trial intraclass correlations (ICC) and 90% confidence intervals (CI) of AMAT and Vicon.

	AMAT		Vicon	
	ICC	90% CI	ICC	90% CI
Area 1	0.92	0.86 - 0.96	0.99	0.98 - 0.99
Area 2	0.72	0.59 - 0.84	0.99	0.97 - 0.99
Area 3	0.79	0.67 - 0.88	0.99	0.98 - 0.99
Area 4	0.71	0.57 - 0.83	0.97	0.96 - 0.99
Area 5	0.78	0.66 - 0.87	0.97	0.96 - 0.99
Area 6	0.85	0.76 - 0.92	0.98	0.96 - 0.99
Area 7	0.87	0.79 - 0.93	0.97	0.94 - 0.98
Area 8	0.61	0.45 - 0.77	0.94	0.90 - 0.97
Area 9	0.87	0.79 - 0.93	0.98	0.97 - 0.99
Area 10	0.76	0.64 - 0.87	0.99	0.99 - 1.00
Area 11	0.85	0.76 - 0.92	0.99	0.98 - 0.99
Area 12	0.92	0.86 - 0.96	0.99	0.99 - 1.00
Area 13	0.91	0.86 - 0.95	0.98	0.97 - 0.99
Area 14	0.79	0.67 - 0.88	0.98	0.97 - 0.99
Area 15	0.66	0.50 - 0.80	0.98	0.97 - 0.99
Area 16	0.87	0.80 - 0.93	0.98	0.97 - 0.99
Area 17	0.87	0.80 - 0.93	0.99	0.98 - 0.99
Area 18	0.94	0.89 - 0.97	0.98	0.97 - 0.99

Table 4.4 displays the differences in adjacent marker position (± SD) between Vicon and AMAT. The average differences range from -4.51 to 16.23 millimetres and the standard deviations range from 1.71 to 19.53 millimetres.

Table 4.4. Differences in distance measured (AMAT – Vicon [± SD, in millimetres]) in the posterior-anterior axis between the adjacent Bar-Positions.

Bar-Position	AMAT - Vicon
2-1	3.27 ± 16.60
3-2	3.33 ± 11.24
4-3	8.50 ± 7.28
5-4	3.57 ± 2.81
6-5	-2.45 ± 2.36
9-7	14.39 ± 16.03
10-8	13.87 ± 11.39
11-9	1.70 ± 3.65
12-10	1.20 ± 3.33
14-13	-2.45 ± 10.77
15-14	0.96 ± 9.14
16-15	8.57 ± 4.37
17-16	2.44 ± 3.24

Figure 4.4 displays the Bland-Altman plot of the data collected in this study with Vicon and AMAT. The y-axis represents the difference in measurement between AMAT and Vicon and the x-axis represents the posterior-anterior position as measured with the AMAT. The bias was 3.87 millimetres (± 10.37 millimetres) and the 95% limits of agreement (LoA) ranged from -16.46 to 24.20 millimetres. Approximately 6% of all data fell outside the limits of agreement range. From the positions furthest away to the positions closest to the camera, 17%, 13%, 1%, 0% and 0% of the data fell outside the limits of agreements range, respectively.

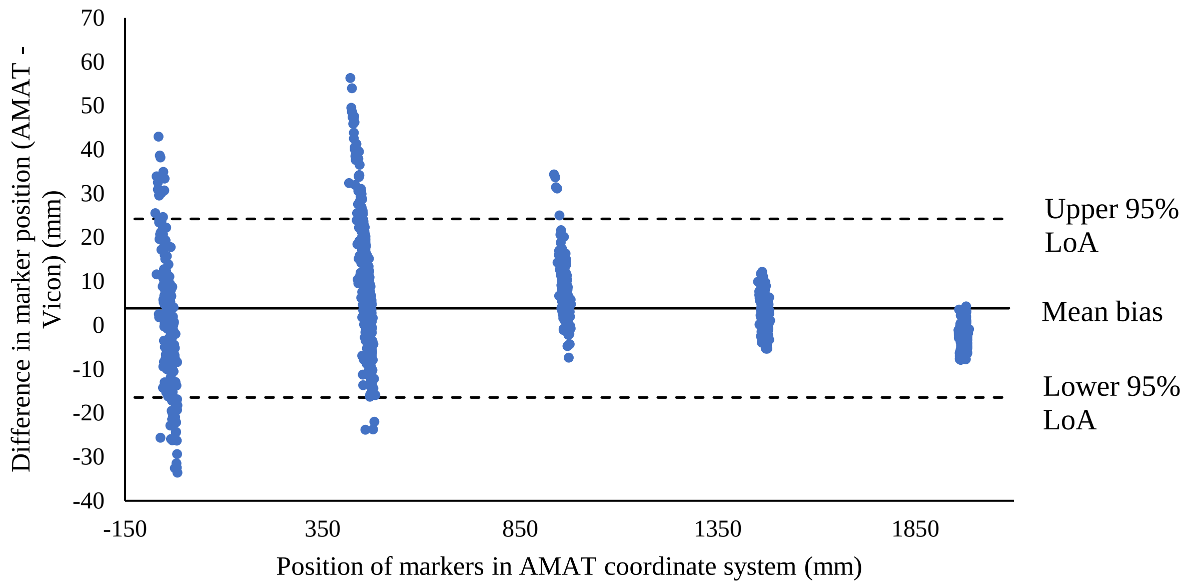

Figure 4.4. Bland-Altman plot to show the agreement in measurements between Vicon and AMAT. The y-axis displays the difference in distance between two Bar-Positions between Vicon and AMAT. The x-axis displays the positions of the markers as measured by the AMAT system, where higher values are closer to the Kinect camera.

4.4 Discussion

Quantifying sensorimotor risk factors of lower extremity injuries has been proposed by multiple studies to determine injury risk in adolescent athletes (Hewett *et al.*, 2006a; Read *et al.*, 2017b). As argued in Chapter 2 and Chapter 3, the ability to maintain balance is the outcome measure of the sensorimotor system. To quantify the ability to maintain balance with kinematic data, it is important that the position of the feet can be determined in a reliable and valid way to determine the position of the base of support. In addition, the position of the feet can also be used to determine jump distances and it can be used to quantify movement strategies. As such, the AMAT was developed with the purpose to collect reliable and valid kinematic data in a practical way. However, it makes use of the Kinect v2 and this camera cannot collect the positional data of the feet in a reliable and valid manner (Van Diest *et al.*, 2014; Wang *et al.*, 2015). Therefore, a foot marker tracking algorithm was developed that is able to automatically collect the position of the feet. Currently no information is available about the reliability and validity of this algorithm. As such, this study had two aims. At first to determine the static reliability of the foot marker tracking algorithm of the AMAT. Secondly, to determine the criterion validity of the algorithm when compared to Vicon, a gold standard to collect marker data of anatomical landmarks in a biomechanical setting.

Foot markers were positioned on eighteen different positions within the camera view and the positional data of these markers were collected 120 times per marker in each position. Here it was found that the within-trial reliability was up to three millimetres larger with the AMAT compared to Vicon. In addition, the between-trial typical errors of the AMAT were up to six millimetres larger compared to Vicon. The intra-class correlations of the AMAT were moderate to very high whereas all intra-class correlations of the Vicon were very high. The validity of the AMAT was determined by calculating differences in distances measured between adjacent Bar-Positions as measured by AMAT and Vicon. The relative differences between AMAT and Vicon showed that on positions further away from the camera the standard deviations of the data collected with AMAT were higher compared to Vicon. This implies that the data were collected in a less reliable and valid manner on these positions with AMAT.

The outcomes of this study are in accordance with the study of Yang *et al.,* (2015), who showed that the depth-sensing technology of the Kinect has a random error that increases on positions further away from the camera. To create 3D data, the depth-sensing technology calculates the distance from the camera to the object captured in each pixel (Lachat *et al.,* 2015). In the algorithms that are part of the AMAT, the data in the posterior-anterior and in the superior-inferior axis are calculated with the depth value. Consequently, the higher technological error of the marker tracking algorithm in the posterior-anterior axis on positions further away from the camera might be due to an error in the depth-sensing technology. It should be noted that for the measurement of the jump distances, the data are mainly collected around Bar-Positions four, five and six (Figure

4.3), on a distance ranging from 1.96 metres to 2.82 metres from the Kinect camera. Here the absolute differences between AMAT and Vicon are less than one centimetre, whereas jump distances of male and female high school and college athletes ranges from 1.0 to 2.1 meters and of elite youth football players ranges from approximately 1.0 to 1.8 meters (Myers *et al.,* 2014; Read *et al.,* 2017c). As such, this error is less than 1% of the total jump distance. In addition, several studies have mentioned that they rounded jump distances to the nearest centimetre (Almuzaini & Fleck, 2008; Meylan *et al.,* 2009) or half inch (Porter *et al.,* 2012). The errors due to the rounding are similar to the errors found with the AMAT and as such, this implies that the AMAT could be used to determine the position of the feet.

The between-trial standard deviations were larger than the within-trial standard deviations for both measurement systems and differed between the different Bar-Positions. Although some variability was expected for the foot marker tracking algorithm, this variability was not expected for the marker tracking of Vicon, because a previous study has not reported any reliability issues (Merriaux *et al.,* 2017). The variability found in this study is probably due to the method used. Five participants moved the markers into different positions to include a large number of trials and as such, a deviation in placing the markers on the correct position between the trials might have affected the between-trial typical error.

The foot marker tracking algorithm used in this study was based on the algorithms used by Paolini *et al.* (2014) and MacPherson *et al.* (2016) and as such, it is important to compare the outcomes of both studies. In the study of Paolini *et al.* (2014), the foot markers were tracked during different types of walking on a treadmill. The Kinect was placed one metre in front of the participants on a height of one metre. When compared to a laboratory-based system, the absolute differences ranged from 0 millimetres in the superior-inferior axis to 9.9 millimetres in the posterior-anterior axis. In addition, the root mean square was smallest in the medio-lateral and largest in the posterior-anterior axis and ranged from 4.9 to 26.5 millimetre. In the study of MacPherson *et al.* (2016), participants ran on a treadmill and the algorithms tracked the markers via a Kinect camera that was placed on 1.5 metres from the participants. They found that the difference in positional data of the four markers was always less than ten millimetres. In the current study, the markers were further away from the camera (~1.96 – 4.11 metre) and the difference in position as measured with the new algorithms and the gold standard ranged from approximately two to seventeen millimetres. The fact that the markers in this study were placed further away from the camera might explain the higher errors relative to Vicon when compared to Paolini *et al.* (2014) and MacPherson *et al.* (2016). However, on the Bar-Positions that were less than 3 metres from the Kinect (Bar-Positions 4-6; 9-12; 16-18), the errors found in this study are similar compared to Paolini *et al.* (2014) and MacPherson *et al.* (2016).

Two differences in the data collection might explain why the errors on Bar-Positions closer to the Kinect were similar to the errors found by Paolini *et al.* (2014) and MacPherson *et al.* (2016), although the markers in this study were further away from the camera. At first, the markers that were developed were specifically designed for the AMAT system. During the development of the AMAT it was found that larger infrared markers and non-smooth surfaces increased the errors of the depth-sensing technology of the Kinect. As such, the relatively large infrared markers used by MacPherson *et al.* (2016) and the collection of the data on a non-smooth surface, i.e. the back of the participants, might have increased the errors found by MacPherson *et al.* (2016). At second, static data were collected in this study whereas dynamic data were collected by Paolini *et al.* (2014) and MacPherson *et al.* (2016). As shown by Timmi *et al.* (2018), dynamic movements result in higher errors on the 3D data of the Kinect. Another possible difference is the use of calibration algorithms in this study, because it was described by

Clark *et al.* (2013a; 2013b) that calibrating the Kinect will improve the data collection. However, no calibration procedures were described by Paolini *et al.* (2014) and MacPherson *et al.* (2016) and as such, it is not certain how and whether they calibrated their systems.

A strength of this study was the study design to determine the reliability and validity of the AMAT system. Most studies determine the reliability and validity of a system that collects kinematic data in a dynamic setting with human participants involved (e.g. Paolini *et al.,* 2014; MacPherson *et al.,* 2016; Timmi *et al.,* 2018). This implies that these studies only provide limited knowledge on the reliability of a system, because they combine the biological error of the humans with the technological error of the assessment system (Hopkins, 2000). In contrast, the data in this study were collected in a static setting and as such no biological error was present in this study. Consequently, the outcomes of this study are not solely applicable on the foot marker tracking algorithm, but also give insight in the technological error of the Kinect. As such, the outcomes of this study can be used to determine whether the technical error of the Kinect is acceptable for any study to track objects.

A possible limitation of this study was the low number of participants, because it has been recommended to use more than 20 to 400 participants for reliability studies (Charter, 1999; Hobart *et al.,* 2012), whereas this study only used five participants. It should be noted to that due to the study design, the number of trials per participant was relatively high due to the participants coming in on two separate days. Consequently, the data that could be used for the within- and between trial calculations doubled from five to ten. Also, on each data collection moment, data was collected six times, which provides extra data points for the within- and between-trial reliability, which reduces the necessary sample size (Walter *et al.*, 1998). Moreover, the sample size recommendations by Charter (1999) and Hobart *et al.* (2012) are based on reliability studies in humans, whereas this study focused on the reliability of the marker tracking of the AMAT. The only human involvement in this study was the placement of the markers in the correct position. These positions were marked and as such it was expected that this effect on the reliability would be minimal. Furthermore, based on previous studies (Paolini *et al.,* 2014; Yang *et al.,* 2015; MacPherson *et al.,* 2016) with the Kinect, high reliability of the foot marker tracking algorithm in a static setting was expected. Nevertheless, it is important to be cautious when interpreting the outcomes of this study.

This study showed that the foot marker tracking algorithm of the AMAT is reliable and valid to capture 3D data of the foot markers, except on positions further away from the camera. This data collection is consistent over multiple measures even after remounting the camera, as long as the orientation of the set-up stays similar. However, it should be noted that this study only determined the static reliability and validity of the foot marker tracking algorithm, whereas it has been recommended to assess movements in a dynamical setting (Read *et al.,* 2017b). The reason behind the use of a static setting was that it was the first study to determine the reliability and validity of the algorithm. Including any type of dynamic movements would have resulted in the use of human participants, due to the nature of the foot marker tracking algorithm. This would have resulted in an additional error caused by the use of humans and as such would have made it impossible to determine the technological error of the AMAT (Hopkins, 2000). Therefore, the next study will determine the validity of the foot marker tracking algorithm in a dynamical setting.

Chapter 5. Criterion Validity of the Foot Marker Tracking Algorithm in a Dynamical Setting

5.1 Introduction

As mentioned in the previous chapter, a vital step in the process of validating the foot marker tracking algorithm of the AMAT is to determine the criterion validity of this algorithm during dynamic movements, because static movements lack ecological validity (Read *et al.,* 2017b). Examples of these dynamic movements are hops and jumps. These movements are often included in movement assessment tools in sport (Halson *et al.,* 2014; Myers *et al.,* 2014; Lockie *et al.,* 2018a) and rehabilitation (Gokeler *et al.,* 2017; Welling *et al.,* 2018a; 2018b) settings, due to the frequent occurrence of these movements in sports and the relatively frequent occurrence of injuries during these movements (Waldén *et al.,* 2015; Read *et al.,* 2016b). As such, it is useful to determine the criterion validity of the foot marker tracking algorithm during hops and jumps.

Several outcome measures are collected during hops and jumps, such as the jump distance (Halson *et al.,* 2014; Myers *et al.,* 2014; Lockie *et al.,* 2018a), kinematic variables during the landing (Dingenen *et al.,* 2015c; Welling *et al.,* 2018a) and the ability of an athlete to maintain in balance after landing (Wright *et al.,* 2016; Fransz *et al.,* 2016; Malmir *et al.,* 2017; Read *et al.,* 2017b). As discussed in Chapter 2 and Chapter 3, these measures can be used to quantify sensorimotor risk factors of lower extremity injuries. To collect these measures, it is vital that the landing position of the jumps can be calculated in a valid manner. The ability of the foot marker tracking algorithm to determine the landing position can be determined by measuring the jump distance. When comparing the jump distance of the AMAT with other measurement systems, the criterion validity of the AMAT to measure jump distances can be determined and this will indirectly show the validity of the foot marker tracking algorithm of the AMAT in dynamic situations. As such, this chapter aims to determine the criterion validity of the AMAT to measure jump and hop distances when compared to measuring these distances with a tape measure.

5.2 Methods

5.2.1 Participants.

33 physically active and healthy male university students (25.7 ± 7.7 years, 177.8 ±7.1 cm, 80 ± 13 kg) participated in this study. Ethical approval was obtained from the ethics committee at Teesside University, School of Social Sciences, Business and Law (Appendix 1). All participants completed a written consent form and medical questionnaire before participating in this study.

5.2.2 Protocol.

The warming-up consisted of 5 minutes cycling on a stationary bike with a power output of 100 – 110 Watts. After the warming-up, foot markers were attached to the laces of the shoes (Figure 5.1). The markers on the laces were tracked by the foot marker tracking algorithm that was validated in Chapter 4. Thereafter, a calibration file was recorded to determine the distance from the centre of the foot marker to the toe, the Marker-to-Toe distance (Figure 5.1). After the calibration file was collected, it was explained to the participant that he had to perform five different types of jumps, namely double-legged standing broad jumps, single leg left and right hops and left to right and right to left strides, five times and that the participant had approximately 30 seconds rest between each jump. It was also explained that the participant should push off after hearing a beep and that the focus of the jump should be on controlling the landing. If the participant was not able to control a landing, he had to redo that trial. After the calibration file was collected, the participant had three familiarisation trials per movement to make himself familiar with the movement. Once the familiarisation trials were completed, the data collection started.

Figure 5.1. A shoe with the foot marker attached to the lace. The Marker-to-Toe distance is displayed by the red arrow and is the distance in the posterior-anterior axis from the centre of the foot marker to the toe of the shoe.

The set-up was the same as described in Chapter 4 and depicted in Figure 4.2. The participant started with the standing broad jumps, followed by the left hop, right hop, left to right stride and right to left stride. Each trial was the same: the participant started with his feet behind Bar-Position 1 (Figure 4.3), to be certain that the toes started on the origin of the posterior-anterior axis. This position was marked to be sure that participants always pushed off from the same position. The researcher [MW] told the participant which jump was expected and started the recording on the customised app of the AMAT. After the customised app was started, a beep sounded which was the signal for the participant to jump. After the landing, the participant had to stick his landing until the manual measurement with the tape measure was taken by measuring the landing position of the toe to the nearest millimetre. If the participant was not able to stick his landing, the jump was repeated. The testing was finished once a participant managed to perform five jumps of each type of jump correctly.

5.2.3 Data analysis.

A total of 1007 jumps were recorded during the testing. 30 standing broad jumps, 53 left leg hops, 37 right leg hops, 29 left to right strides and 33 right to left strides were performed incorrectly and had to be repeated. The participants jumped on average 30.5 times to perform each jump five times correctly (range 27 – 40 jumps). Of the 825 jumps that were performed correctly, four standing broad jumps, two left leg hops, eight right leg hops, one left to right stride and two right to left strides had to be removed from further analysis because the data were not collected correctly with AMAT. Thus, a total of 807 jumps were included for further analysis. The jump distance of each jump was calculated as the position of the foot marker on the frame of landing + the marker-to-toe distance.

5.2.5 Statistical analysis.

Visual inspection of Q-Q plots of raw data indicated that the data were approximately normally distributed (Appendix 4). Therefore, the raw jump data are displayed as mean ± SD. In addition, the data were not skewed and as such no log-transformation was performed. Paired t-tests were performed in SPSS version 24 with mechanistic magnitude-based inference (Batterham & Hopkins, 2006) subsequently applied (Hopkins, 2017b) to determine whether differences in jump distance existed between the AMAT and the manual jump distance measure. An issue with the data collected for this study was that the group was heterogenous in fitness level and as such had large differences in jump distances. This affects the between-subject variability and as such the calculation of the smallest worthwhile change (between-subject SD * 0.2 [Hopkins *et al.,* 2009]). However, no studies were found that described a smallest worthwhile change that was related to performance variables in soccer. Therefore, the between-subject SD of Myers *et al.* (2014) on a large cohort of high school and college athletes was used to determine the smallest worthwhile change. The between-subject SD was approximately 18.78 cm and as such the smallest worthwhile change was set at 0.2 times this distance, 3.76 centimetres. The following scales were used to determine the probabilities that a true difference existed between both systems: 0.5 – 5%, *very unlikely*; 5 – 25%, *unlikely*; 25 – 75%, *possibly*; 75 – 95%, *likely*; 05 – 99.5%, *very likely*; > 99.5%, *most likely* (Batterham & Hopkins, 2006). In addition, to determine the agreement in jump distance measured between the manual measurement and AMAT, for each type of jump a Bland-Altman plot was created (Bland & Altman, 1986). For all estimates, the uncertainty was expressed as 90% confidence limits.

5.3 Results

Table 5.1 displays the average distances jumped on the five different jumps and hops as measured with AMAT and manually. Trivial differences were found in distances measured between the manual tape measure (criterion measure) and the AMAT system.

Table 5.1. The average jump distances (± SD, mm) as measured manually and with the AMAT, the difference in jump distance between both measurement systems (± 90% Confidence Limits [CL], mm) and the qualitative inference of this difference of the five different jumps and hops.

Type of jump (jumps included)	Manual measure	AMAT	AMAT - manual measure	Inference
Standing broad jump (n = 161)	1456 ± 278	1458 ± 282	1.70 ± 1.25	*trivial*°
Left leg hop (n = 163)	1243 ± 247	1244 ± 252	0.66 ± 2.06	*trivial*°
Right leg hop (n = 157)	1273 ± 258	1272 ± 261	-1.69 ± 1.35	*trivial*°
Left to right stride (n = 164)	1424 ± 273	1424 ± 278	-0.66 ± 1.29	*trivial*°
Right to left stride (n = 163)	1404 ± 263	1406 ± 268	2.41 ± 1.90	*trivial*°

° *Most likely*

Figures 5.2 to 5.6 displays the Bland & Altman plots for the standing broad jump, left leg hop, right leg hop, left to right stride and right to left stride, respectively. The bias ranged from -2.54 to 1.90 millimetres for the different types of jumps (± 95% LoA range: 9.62 to 14.99 millimetre). A trend towards a lower difference between the two systems on higher jump distances was detected. Approximately 7% of all data points fell outside the limits of agreement range.

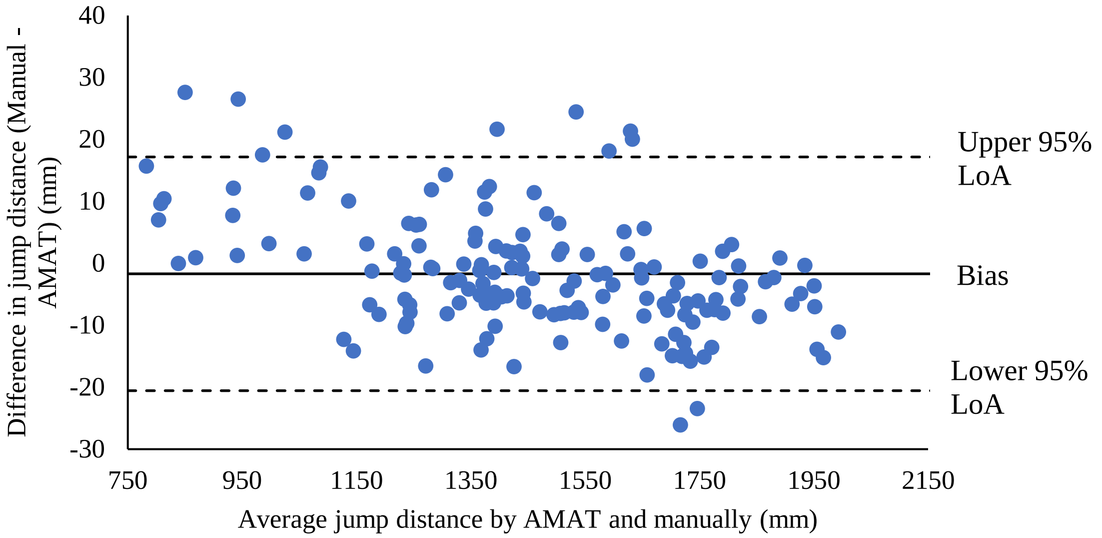

Figure 5.2. Bland-Altman plot to show the agreement in measurements of the standing broad jump distance between the manual measurement and AMAT.

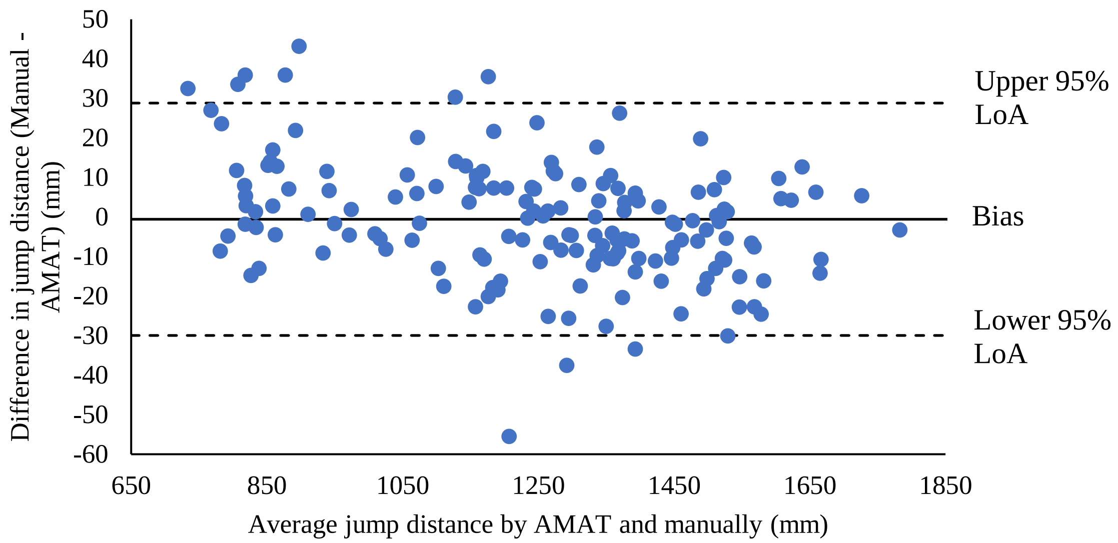

Figure 5.3. Bland-Altman plot to show the agreement in measurements of the left leg hop distance between the manual measurement and AMAT.

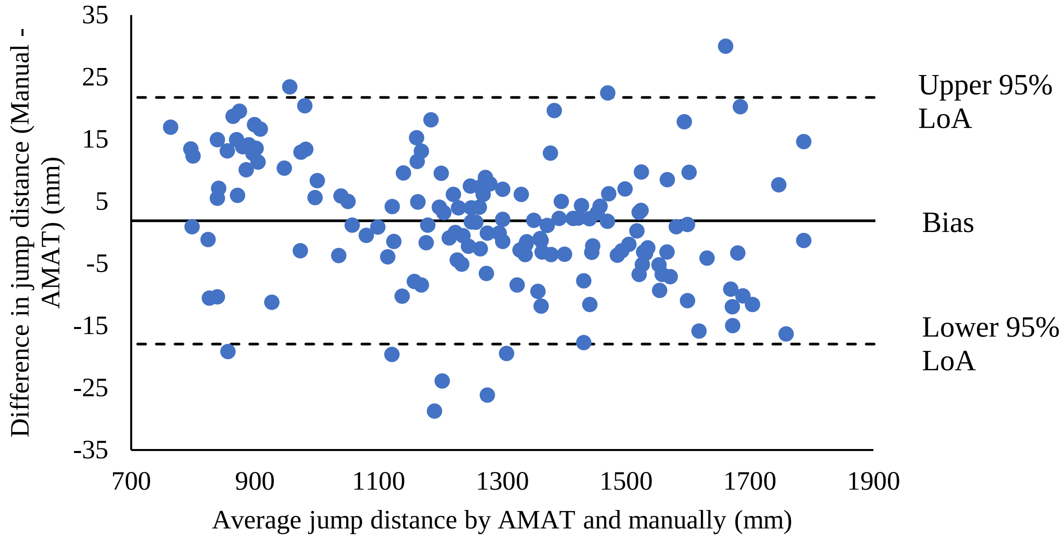

Figure 5.4. Bland-Altman plot to show the agreement in measurements of the right leg hop distance between the manual measurement and AMAT.

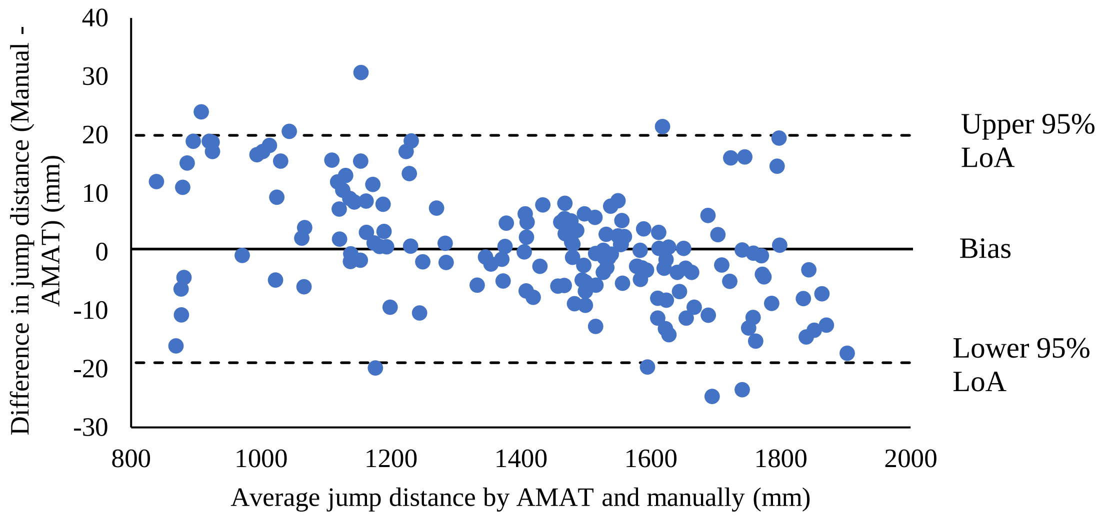

Figure 5.5. Bland-Altman plot to show the agreement in measurements of the left to right stride distance between the manual measurement and AMAT.

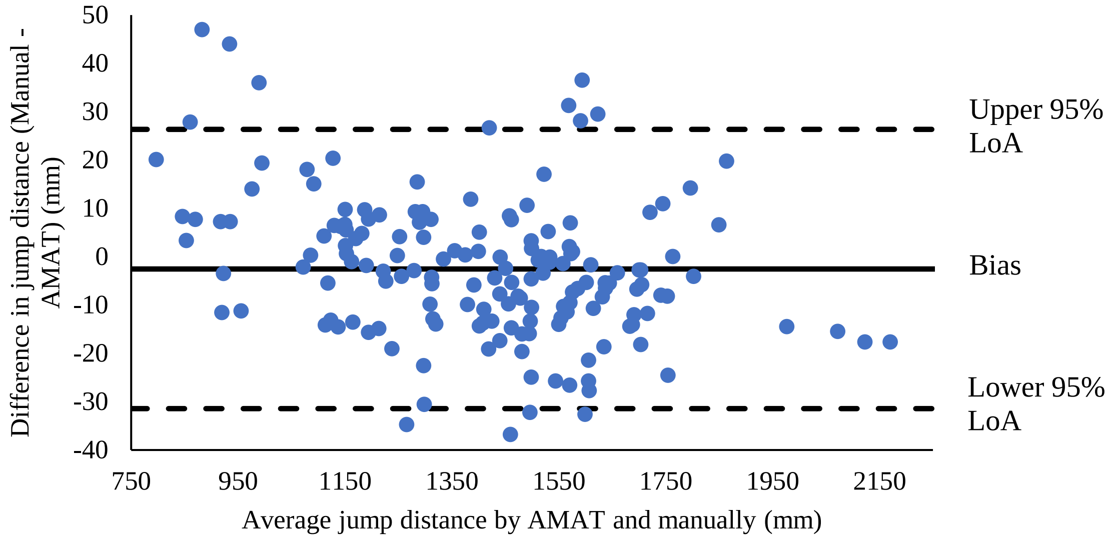

Figure 5.6. Bland-Altman plot to show the agreement in measurements of the right to left stride distance between the manual measurement and AMAT.

5.4 Discussion

The foot marker tracking algorithm described in Chapter 3 is reliable and valid in a static environment. However, no information about the validity of the foot marker tracking algorithm in dynamic situations was available. Important dynamic movements in some movement assessment tool are jumps and hops (Halson *et al.*, 2014; Myers *et al.*, 2014; Gokeler *et al.*, 2017; Lockie *et al.*, 2018a; 2018b; Welling *et al.*, 2018a; 2018b). The position of the feet during jumps and hops is used to measure the distance jumped (Halson *et al.*, 2014; Myers *et al.*, 2014; Lockie *et al.*, 2018a), to measure kinematic variables during the landing (Dingenen *et al.*, 2015c; Welling *et al.*, 2018a) and to measure the ability of an athlete to maintain in balance after landing (Wright *et al.*, 2016; Fransz *et al.*, 2017; Malmir *et al.*, 2017; Read *et al.*, 2017b). These measures can all be used to quantify sensorimotor risk factors of lower extremity injuries (Read *et al.*, 2017b). Therefore, this study aimed to determine the ability of the AMAT to measure jump distances, to indirectly determine the criterion validity of the foot marker tracking algorithm during dynamic movements.

The outcomes of the dependent t-tests and the Bland-Altman plots showed that the foot marker tracking algorithm is valid to track the foot markers during dynamic movements and can be used to measure jump distances. The outcomes of this study are in accordance with MacPherson *et al.* (2016), who found that a similar marker tracking algorithm in combination with the Kinect was valid to track markers when compared to Vicon. However, it should be noted that the data collected in this study was approximately one to two and a half meters further away from the camera compared to MacPherson *et al.* (2016). As shown by Yang *et al.* (2015) and in the previous chapter, data collected further away from the camera is less reliable and valid. It was already assumed in the previous chapter that the calibration algorithm and the foot markers of the AMAT might have resulted in a reduced error compared to MacPherson *et al.* (2016). This study shows that those assumptions are probably true, especially because no other changes in software or hardware exist between this study and the studies of Yang *et al.* (2015) and MacPherson *et al.* (2016).

The outcomes of this study imply, in combination with the outcomes of Chapter 4, that the AMAT can be used to determine the displacement of the feet during dynamical movements. This is an important finding because dynamic movements have the highest ecological validity (Read *et al.*, 2017b), the foot movement is often used to assess the movement performance and movement strategy (Halson *et al.*, 2014; Myers *et al.*, 2014; Lockie *et al.*, 2018a) and the foot position can be used to determine the position of the base of support. In addition, current movement assessment tools often lack validity or practicality to assess movements accurately (McCall *et al.*, 2014; Dorrel *et al.*, 2015; Ehrenbrusthoff *et al.*, 2016; Bonazza *et al.*, 2017; Moran *et al.*, 2017; Read *et al.*, 2017b; Colyer *et al.*, 2018). Moreover, the Kinect is not able to collect positional data of anatomical landmarks of the lower extremity in a reliable and valid manner (Wang *et al.*, 2015; Otte *et al.,* 2016). Therefore, this foot marker tracking algorithm is a valuable addition to collect kinematic data in a practical manner with the Kinect.

A strength of this study was the fact that jump data of 25 jumps were collected per participant. A total of 807 individual jumps were included to determine the validity of the foot marker tracking algorithm, which is a sufficient number of data points to determine the validity (Charter, 1999; Hobart *et al.*, 2012). An additional benefit of assessing the validity during jumps is that the jump movement is a high-speed movement. As explained in Chapter 3, it was more difficult to track the markers during these high-speed movements compared to lower speed movements, for example during balance movements and squats. This implies that it is likely that the foot marker tracking algorithm that was validated in this study can also be used to track the feet in a reliable and valid manner during squat and balance movements.

A limitation of this study was that the dynamic validity was based on the end position of the foot marker, whereas no information about the flight phase was provided. This choice was made based on two reasons. At first, as described in Chapter 3, it was impossible to track the foot marker in a reliable and valid way during high speed movements. This is also reflected in the high errors found in the study of Timmi *et al.* (2018) during high speed movements. Second, from a sensorimotor perspective, the most important part of the movement is the first 50 milliseconds after landing, because of the high ground reaction forces during this period that can work on the joints (Krosshaug, 2007). This is also reflected in the large number of studies that have assessed the movement of athletes during the landing phase (Dingenen *et al.,* 2015c; Wright *et al.,* 2016; Fransz *et al.,* 2016; Malmir *et al.,* 2017; Read *et al.*, 2017b; Welling *et al.*, 2018a), whereas almost no studies focus on the flight phase, because during this part of the jump, besides the gravity no external forces work on the body. As such, the risk of injuries during the flight phase is minimal. This implies that the AMAT might be usable to assess movements during the landing phase, because this study showed that the landing position can be determined in a valid manner. However, more research is necessary to determine what type of movement strategies can be assessed in a reliable and valid manner with the AMAT.

This chapter and the previous chapter showed that the foot marker tracking algorithm of the AMAT is reliable and valid to track the feet during static and dynamic situations. As mentioned in Chapter 2 and Chapter 3, collection of the feet is important to determine jump distances, and it can be used to determine the ability to maintain balance. Information on these sensorimotor risk factors is necessary for any movement assessment tool (Read *et al., 2016b;* 2017b). However, to quantify the ability to maintain balance, information on the position of the centre of mass is necessary also (Winter, 1995). As such, the next chapter will determine the validity of the centre of mass algorithm of the AMAT, because the centre of mass can be used to determine the ability to maintain balance (Winter, 1995).

6.1 Introduction

In Chapter 2 and Chapter 3 it was described that the collection of the centre of mass and base of support position can be used to quantify the ability of one to maintain balance, based on Winter, (1995), Winter *et al.* (1996) and Winter *et al.* (1998). Chapter 4 and Chapter 5 showed that the data collection of foot markers is reliable and valid with the AMAT. This implies that the base of support position can be determined in a reliable and valid manner. As described in Chapter 3, the position of the centre of mass is calculated based on the voxel data of all pixels that capture the athlete within the camera view. Due to the use of a depth-sensing camera, it is possible to calculate the 3D position of the centre of mass during dynamic movements. Therefore, collecting the base of support and centre of mass positional data with the AMAT would be very useful for practitioners, because they can then quantify the ability of athletes to maintain balance. However, currently no information is available about the reliability and validity of the centre of mass algorithm of the AMAT.

During quiet stance a small centre of mass displacement should be expected (i.e. postural sway, Winter, 1995; Winter *et al.*, 1996). Chapter 4 showed how the reliability of the foot marker tracking algorithm was determined without any human interference. This was important to determine the technological error of the foot marker tracking algorithm (Hopkins, 200). However, due to the nature of the algorithm that calculates the position of the centre of mass, it is not possible to determine the reliability of the centre of mass algorithm without any human participants. This implies that only the validity of this algorithm can be determined when using human participants. Therefore, this study aims to determine the criterion validity of the AMAT to determine the position of the centre of mass during jumps and landings when compared to Vicon, a gold standard to capture the positional data of anatomical landmarks during movement (Padua *et al.*, 2009; MacPherson *et al.*, 2016).

6.2 Methods

6.2.1 Participants.

Two healthy males (1: age: 24.5 years old, height: 178 cm, weight: 65 kg, 2: age: 30.4 years old, height: 170 cm, weight: 81 kg) participated in this study. Ethical approval was obtained from the ethics committee at Teesside University, School of Social Sciences, Business and Law (Appendix 1). Upon arrival, both participants completed an informed consent sheet and a medical questionnaire.

6.2.2 Protocol.

Figure 6.1 shows the set-up used for this study. To obtain a full body plug-in gait model with Vicon, retroreflective markers were placed on the following 35 anatomical landmarks of the participant: left and right temple, left and right back head in horizontal plane with temple markers, C7, T10, middle of right scapula, jugular notch where the clavicles meet the sternum, xiphoid process of the sternum, left and right anterior superior iliac spine, left and right posterior superior iliac spine, lateral side of both upper arms, lateral epicondyle of both elbows, medial and lateral side of both wrists, dorsum of hand below the head of the second metacarpal, lateral side of both thighs, lateral epicondyle of both knees, lateral side of both tibias, lateral malleolus of both ankles, on both calcaneus and on the second metatarsal head. To calculate the position of the centre of mass with Vicon, leg length, knee, ankle and elbow width and shoulder offset of the participant were obtained and inputted in Vicon. Vicon automatically calculates the centre of mass of each segment and the centre of mass of the full body is the weighted average of the different segments. In contrast to previous studies, no foot markers of the AMAT were attached to the feet of the participants, due to a possibly interference with the Vicon markers.

Both participants started with a calibration trial where they had to stand still on the landing zones. In addition, the first participant performed three standing broad jumps, three left to right and right to left strides and the second participant performed five standing broad jumps, three left to right and right to left strides. Before participation, participants were explained the movements they had to perform and as a warming-up they performed three practice trials of each movement for familiarization purposes. Participants were instructed to land on the pre-specified landing zones to enhance the data collection of the retroreflective markers with Vicon.

The centre of mass data were collected with the AMAT (sampling frequency 30 Hz), as described in Section 3.3.3.4, and with Vicon (sampling frequency 100 Hz).

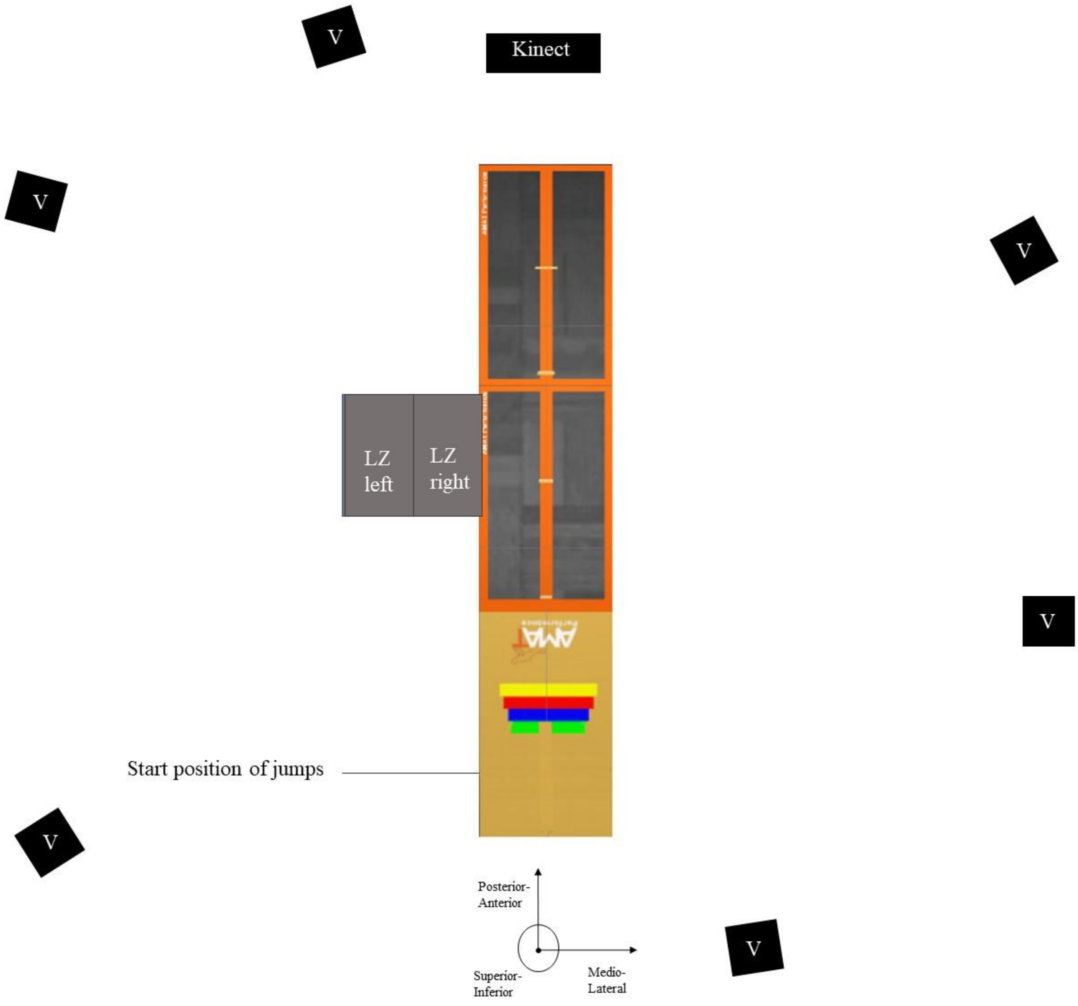

Figure 6.1. Set-up of this study. The six Vicon cameras (V) were set up around the set-up of the AMAT. Instead of the black rubber mats, landing zones (LZ) were displayed to show where the participant had to land with his left and right foot.

6.2.3 Data analysis.

The Vicon and AMAT have a right-handed and left-handed coordinate system, respectively. Therefore, the centre of mass data collected with Vicon was rotated around the superior-inferior axis. In addition, a transformation matrix was performed to align the posterior-anterior and medio-lateral axes. Due to the different sampling frequencies of the systems used in this study, the centre of mass data of Vicon was reduced to a sampling frequency of 30 Hz to compare it with the centre of mass data of the AMAT. It was not possible to start the data collection of both systems at exactly the same time and as such, the data of both systems was synchronized manually afterwards. This was done by plotting the centre of mass position of both systems per axis and visually inspecting the data to fit the data in such a way that it was synchronized in all three axes. Of the data collected with participant 2, the data of two standing broad jumps and of one left to right stride had to be excluded due to an issue of the data collection with Vicon.

6.2.4 Statistical analysis.

To determine the agreement in distance measured between the centre of mass data collection with Vicon and AMAT, a Bland-Altman plot (Bland & Altman, 1986) was created to determine whether this agreement differed between different distances from the camera. In addition, Pearson correlations were performed for each movement individually between the centre of mass positions calculated with AMAT and with Vicon in a custom-made spreadsheet (Hopkins, 2017a). The magnitudes used to interpret the magnitude of the correlation coefficients is as follows: <0.20, *very low*; 0.20 – 0.50, *low*; 0.50 – 0.75, *moderate*; 0.75 – 0.90, *high* 0.90 – 0.99, *very high*; >0.99, *extremely high* (Malcata *et al.*, 2014). For all estimates, the uncertainty was expressed as 90% confidence limits.

6.3 Results

Figures 6.2, 6.3 and 6.4 display the Bland-Altman plot of the centre of mass data collected in this study with Vicon and AMAT in the medio-lateral, superior-inferior and posterior-anterior axis, respectively. The y-axis represents the difference in measurement between AMAT and Vicon and the x-axis represents the average centre of mass position as measured with the AMAT and Vicon.

In the medio-lateral axis, the bias was 3.99 millimetres and the 95% limits of agreement ranged from -53.46 to 45.48 millimetres. In the superior-inferior axis, the bias was -25.65 millimetres and the 95% limits of agreement ranged from -102.95 to 51.64 millimetres.

In the posterior-anterior axis, the bias was -61.67 millimetres and the 95% limits of agreement ranged from -108.91 to -14.42 millimetres. Approximately 93% of all data points fell within the 95% limits of agreement range.

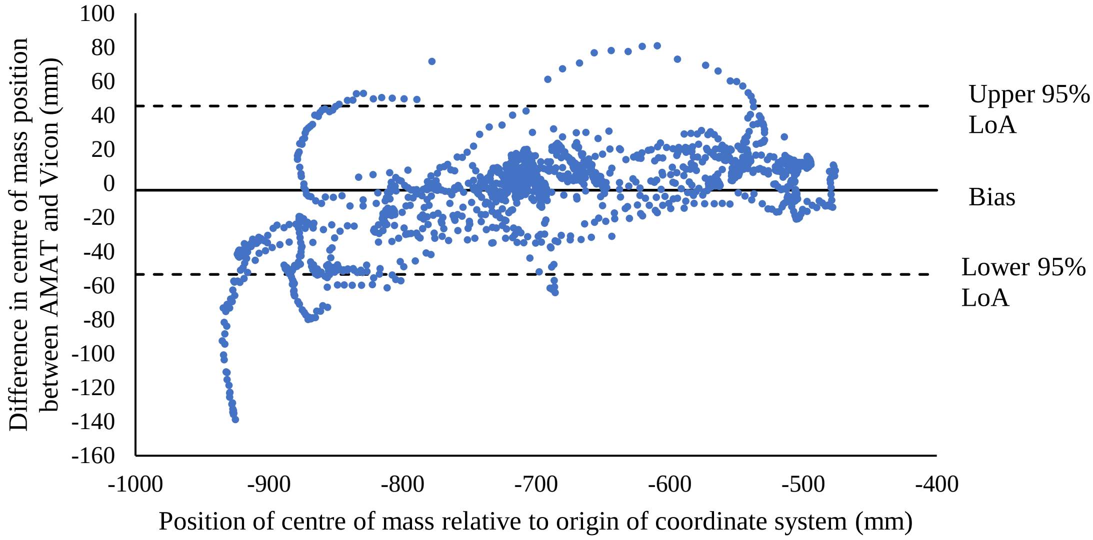

Figure 6.2. Bland-Altman plot to show the agreement in measurements of the centre of mass position between Vicon and AMAT in the medio-lateral axis.

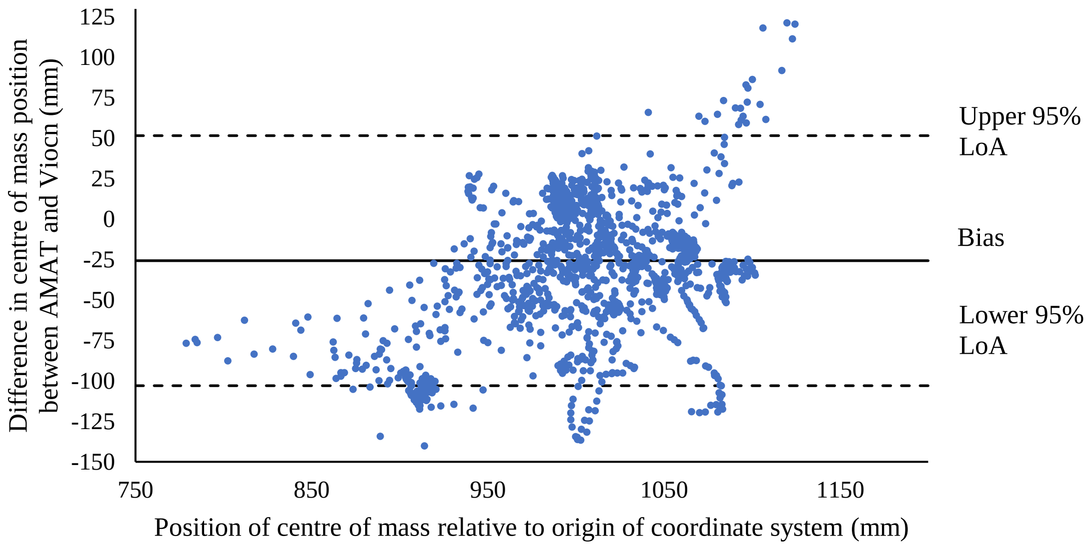

Figure 6.3. Bland-Altman plot to show the agreement in measurements of the centre of mass position between Vicon and AMAT in the superior-inferior axis.

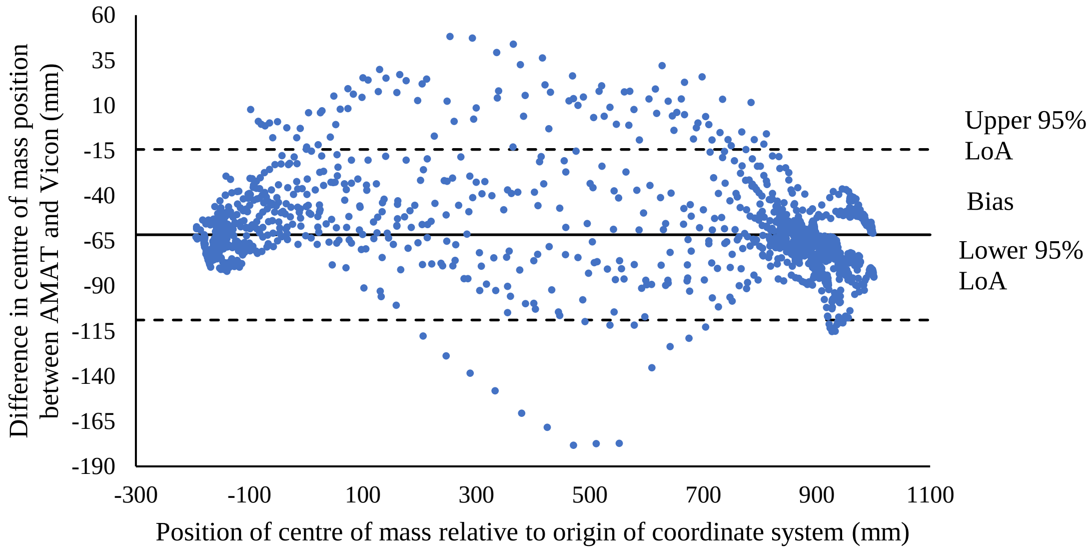

Figure 6.4. Bland-Altman plot to show the agreement in measurements of the centre of mass position between Vicon and AMAT in the posterior-anterior axis.

Table 6.1 displays the correlations between the centre of mass positions collected with both systems. The correlations were *moderate* to *extremely high* in the medio-lateral axis (0.65 to 1.00), *extremely high* in the posterior-anterior axis (0.99 to 1.00) and *trivial* to *extremely high* in the superior-inferior axis (-0.08 to 0.98). The centre of mass position per movement as measured with Vicon and AMAT are displayed in Appendix 5.

Table 6.1. Pearson correlations (*r)* and 90% confidence limits between the positional data of the centre of mass as collected with the AMAT and Vicon for all jumps of both participants.

	Medio-lateral axis		Superior-inferior axis		Posterior-anterior axis	
Participant 1	*r*	90% CL	*r*	90% CL	*r*	90% CL
Standing broad jump	0.88	0.84 to 0.92	0.96	0.94 to 0.97	1.00	1.00 to 1.00
Standing broad jump	0.65	0.53 to 0.74	0.91	0.88 to 0.94	1.00	1.00 to 1.00
Standing broad jump	0.86	0.80 to 0.90	0.97	0.96 to 0.98	0.99	0.99 to 1.00
Left to right stride	1.00	1.00 to 1.00	0.95	0.92 to 0.96	1.00	1.00 to 1.00
Left to right stride	1.00	1.00 to 1.00	0.86	0.80 to 0.90	1.00	1.00 to 1.00
Left to right stride	0.99	0.99 to 1.00	0.98	0.97 to 0.99	1.00	1.00 to 1.00
Right to left stride	0.98	0.98 to 0.99	0.82	0.77 to 0.86	1.00	1.00 to 1.00
Right to left stride	0.99	0.98 to 0.99	0.55	0.42 to 0.66	1.00	0.99 to 1.00
Right to left stride	0.99	0.98 to 0.99	0.78	0.71 to 0.84	1.00	1.00 to 1.00
Participant 2						
Standing broad jump	0.65	0.57 to 0.73	0.79	0.73 to 0.84	1.00	1.00 to 1.00
Standing broad jump	0.99	0.99 to 0.99	0.88	0.84 to 0.91	1.00	1.00 to 1.00
Standing broad jump	0.93	0.91 to 0.95	0.85	0.81 to 0.88	1.00	1.00 to 1.00
Left to right stride	0.99	0.99 to 0.99	0.30	0.18 to 0.42	1.00	1.00 to 1.00
Left to right stride	1.00	0.99 to 1.00	-0.10	-0.25 to 0.07	1.00	1.00 to 1.00
Right to left stride	1.00	1.00 to 1.00	0.64	0.53 to 0.73	1.00	1.00 to 1.00
Right to left stride	0.93	0.90 to 0.96	-0.08	-0.30 to 0.14	1.00	1.00 to 1.00
Right to left stride	0.99	0.98 to 0.99	1.00	1.00 to 1.00	0.52	0.41 to 0.62

6.4 Discussion

The base of support and centre of mass are important variables to quantify the ability of humans to maintain balance (Winter, 1995). Chapter 3 described new algorithms to automatically capture the foot and centre of mass position. Previous chapters showed that the foot position can be determined in a reliable and valid manner with the AMAT. This implies that the base of support position can be determined in a reliable and valid manner. The algorithm that calculates the position of the centre of mass uses the voxel data of all pixels that capture the body, based on algorithms described by Goffredo *et al.* (2006) and Allin *et al.* (2008). However, no information was available about the validity of this centre of mass algorithm of the AMAT. As such, this chapter aimed to determine the validity of the AMAT to calculate the centre of mass position.

To determine the criterion validity of the AMAT to calculate the position of the centre of mass, it was compared with Vicon. Vicon is generally viewed as a gold standard to capture the positional data of anatomical landmarks during movement (Smith *et al.*, 2008; Padua *et al.*, 2009; van Diest *et al.*, 2014; Augustus & Smith, 2015; MacPherson *et al.*, 2016; Augustus *et al.*, 2017). The Bland-Altman plots were used to determine the validity of the centre of mass collection of AMAT when compared to Vicon. These plots showed that approximately 93% of all data points fell within the 95% limits of agreement. In addition, the correlations between the two measurement systems were *extremely high* in the posterior-anterior axis, *moderate* to *extremely high* in the medio-lateral axis and *trivial* to *extremely high* in the superior-inferior axis. All together this shows that the validity of the centre of mass data collection with the AMAT seems to be valid.

The outcomes of this study might be explained by the way the algorithm works. A first factor that might have contributed to the *moderate* to *extremely high* correlations in the medio-lateral axis and extremely high correlations in the posterior-anterior axis is the way the algorithm of the AMAT calculates the centre of mass position. Namely, as described in Chapter 3, the skeletal tracking of the Kinect v2 is used to determine which pixels are included to calculate the position of the centre of mass. However, as discussed in Section 3.2, the skeletal tracking of the Kinect during dynamic movements is not reliable nor valid (Kharazi *et al.*, 2015; Mentiplay *et al.*, 2015; Otte *et al.*, 2016; Auvinet *et al.*, 2017; Eltoukhy *et al.*, 2017). Nevertheless, it approximates the position of the anatomical landmarks and could therefore be used in the newly developed algorithm to collect the voxel data.

The lower correlations in the superior-inferior and medio-lateral axes between the different movements might also be explained by the working of the algorithm of the AMAT. Figure 6.5 displays the voxel data captured during three frames of a movement. The voxel data in the posterior-anterior axis is captured correctly and this probably explains the high validity of the centre of mass position in the posterior-anterior axis (Figure 6.5A). A slight issue occurs in the medio-lateral axis, because more voxels are collected on the side closer to the Kinect camera compared to the side further away from the Kinect camera, because due to a parallax, the side closer to the camera is better visible (Figure 6.5B, Figure 6.5C). An issue with the centre of mass in the superior-inferior axis is that during the jump, players can move partly out of the camera view if they are tall or jump high (Figure 6.5A). This issue might explain the lack of validity in the superior-inferior axis. The centre of mass position in the medio-lateral axis can provide information about the sway to either side, whereas the centre of mass data in the posterior-anterior axis can provide information about the posterior and anterior sway (Winter, 1995; Fransz *et al.,* 2013; 2015). In contrast, the centre of mass position in the superior-inferior axis cannot be used to determine sway and is generally not used to assess the ability to maintain balance (Winter, 1995). As such, the lower correlations in the superior-inferior axis will not affect any calculations regarding body sway.

The Bland-Altman plots might give additional insight in the effect of the position of the body relative to the camera. For example, in the medio-lateral axis it is apparent that the largest differences between the AMAT and Vicon are on the left side of the graph. The left side of the graph is further away from the centre of the AMAT set-up and thus further away from the camera. As such, it might be that when the person moves further to the side, the centre of mass data are collected less validly with the AMAT. In the superior-inferior axis, it is apparent that the largest differences between AMAT and Vicon are found when the centre of mass has the highest positions. As such, similar to what was argued in the previous paragraph, it might be that the AMAT is not able to capture the position of the centre of mass correctly when the person jumps, because the person can jump partly out of the camera view. This is also apparent in the posterior-anterior axis, because here it is visible that the largest differences between the Vicon and AMAT are found during the flight phase of the jumps. However, as argued previously, it is most important that the centre of mass collection is collected during the landing phase of the jumps. As such, based on Figures 6.2, 6.3 and 6.4, it seems that the centre of mass position can be collected in a valid manner during the landing phase of the jumps.

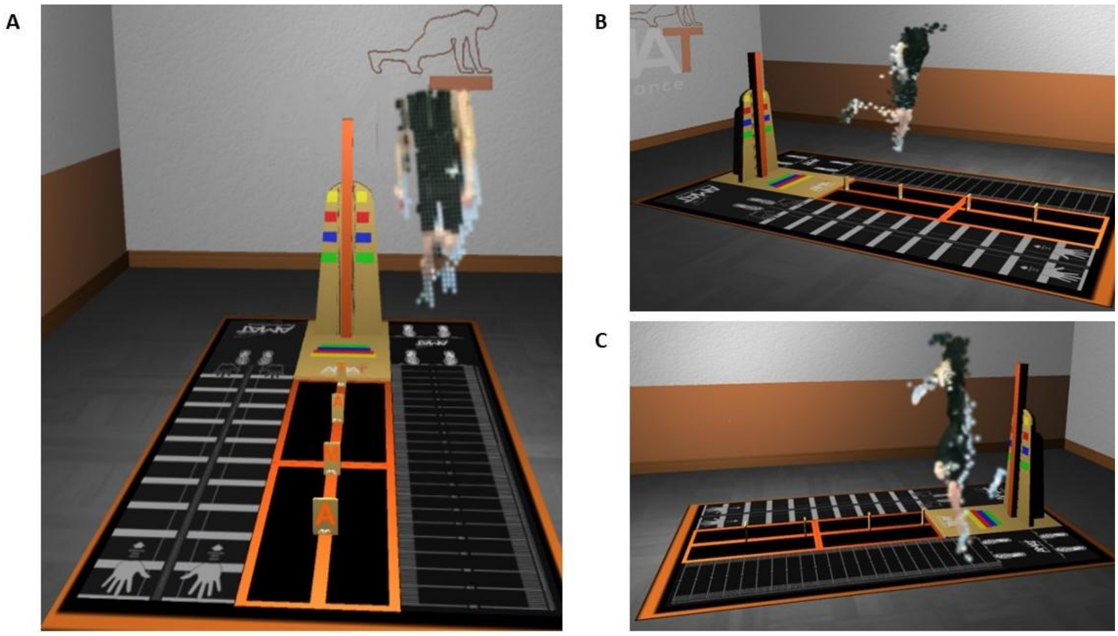

Figure 6.5. Screenshots of a right to left stride by a participant. The voxel data are displayed as a point cloud and is used to calculate the centre of mass. Figure 6.5A: The voxel data in the frontal view (posterior-anterior axis) is correctly collected, except for the position of the head. This might have affected the superior-inferior data of the centre of mass. Figure 6.5B: The voxel data from a sagittal view point. On this side, the arm and a part of the leg is visible. Figure 6.5C: No arm on this side is visible, because the arm is not within the camera view. This might have affected the centre of mass positional data in the medio-lateral axis.

A limitation of this study was that the data of only two participants was included for analysis. This was due to AMAT system interfering with Vicon. Namely, certain parts of the body work of the AMAT system were picked up by the Vicon system as possible retroreflective markers. This was not an issue in previous studies because only two Vicon markers were tracked in those studies, but it caused issues in tracking the full body plug-in gait model correctly. It took several weeks to solve this issue, because the AMAT system was only a few days per month available for testing. Consequently, less time was available for the data collection and hence only two participants were used in this study.

This low number of participants limits the generalisability of this study and as such, research on larger sample sizes of males and females remains necessary.

A second limitation of this study was the absence of data to calculate the technological error of the centre of mass algorithm. As mentioned previously, this was due to the fact that during this study, the algorithm only worked on humans, which implies that a biological error is involved in this type of data collection (Hopkins, 2000). As such, at the time of the data collection, it was impossible to determine the technological error of this system. However, during a pilot study where the centre of mass was collected during 40 jumps, it was found that during each frame approximately 1000 voxels per frame were used to calculate the centre of mass position (Appendix 6). Yang *et al.* (2015) described that the depth-sensing technology of the Kinect has a random error. It is expected that this large number of data points will probably reduce the random error of the centre of mass position. Therefore, based on the outcomes of this study and the data of the pilot study, it is probable that the AMAT can calculate the centre of mass position in the medio-lateral and posterior-anterior axis in a valid manner.

Chapters 4, 5 and 6 showed that the foot marker tracking algorithm and the algorithm that calculates the position of the centre of mass are reliable and valid. However, it should be noted that these studies were performed in a laboratory-based setting. A laboratory-based setting is different from a practical setting because data can be collected in a more structured and valid manner. As such, those studies were perfect to determine the reliability and validity of these algorithms. Nonetheless, information in a practical setting is necessary to determine whether the AMAT can be used in a practical setting and whether there might be biases during practical measurements. As such, the next chapter will collect jump performance data of elite adolescent female soccer players over the course of a full soccer season.

Chapter 7. Within-Subject and Seasonal Variability in Standing Broad Jump Performance and Centre of Mass Displacement During Landings in Elite Youth Female Soccer Players

7.1 Introduction

Female soccer has been professionalised in the last decade and participation has increased (FIFA, 2014; FA, 2018; KNVB, 2018), but the majority of studies in soccer is still performed on males (Milanovic *et al.,* 2017). Female soccer players are viewed as a special population (Rosenbloom *et al.,* 2006) because they cover less distance and cover less distance on high-intensity speed compared to male soccer players (Krustrup *et al.,* 2005). Moreover, there are differences in anatomical, hormonal and sensorimotor factors between male and female athletes (Hewett *et al.*, 2006a) and female soccer players have a higher ACL injury risk compared to their male counterparts (Arendt & Dick, 1995; Arendt *et al.,* 1999; de Loes *et al.,* 2000; Agel *et al.,* 2005; Allen *et al.,* 2016). As such, it is important that more studies on female soccer players are performed that can aid in understanding how performance can be improved and injuries can be reduced.

Cross-sectional and longitudinal studies can be used to enhance the knowledge on female soccer players. In general, longitudinal studies have a benefit over cross-sectional studies because they show how time can affect the variables measured (Caruana *et al.,* 2015). In soccer, performance measures can change over time, due to randomness or due to a systematic change such as maturation (i.e. peak height velocity) or a training effect (Bahr, 2016; Esmaeili *et al.,* 2018). To determine whether these changes over time are substantial, it is important to understand the within-subject variability during each session, because it affects the precision of the estimates of the change (Hopkins, 2000).

The AMAT can be used to collect standing broad jump performance data of female soccer players in a reliable, valid and practical manner and it can collect the centre of mass in a valid manner during jumps. The standing broad jump has been related to the horizontal power output of the lower extremity hip flexor muscles and has high correlations with sprint and acceleration performance (Markovic *et al.,* 2007; Nagano *et al.,* 2007; Salaj & Markovic, 2011; Lockie *et al.,* 2016; 2017). Also, changes in jump distance can provide information about the fatigued state of an athlete (Halson, 2014) and the effect of training programs on the horizontal power output of hip flexor muscles (Markovic *et al.,* 2007).

Several studies have used the position of the centre of mass relative to the centre of pressure (Winter *et al.,* 1998; Corriveau *et al.,* 2000; Lafond *et al.,* 2004; Lee *et al.,* 2006) and the centre of gravity relative to the base of support (Cheung & Azevedo, 2015) to quantify the ability to maintain balance. Quantifying the ability to maintain balance is a sensorimotor risk factor of lower extremity injuries (Read *et al.,* 2016a; 2016b; 2017b) and it has been related to ACL injury risk (Hrysomallis, 2007; Zazulak *et al.,* 2007a). This implies that the AMAT might be able to quantify the ability to maintain balance during the landing after jumps via measuring the position of the centre of mass relatively to the position of the base of support.

As described in Chapter 2, Fransz *et al.* (2015) showed how multiple calculation methods could be used to calculate the time to stabilisation after jumps and hops. Similarly, it is also possible to use different methods to quantify the ability to maintain balance based on the centre of mass and centre of pressure. For example, Winter *et al.* (1998) mentioned that both the amount of body sway and the body sway speed could be used to quantify the ability to maintain balance. Moreover, Fransz *et al.* (2015) showed how the centre of pressure displacement could be calculated in the medio-lateral axis, in the posterior-anterior axis, or a combination of both axes, the resultant axis. Similarly, the centre of mass displacement can also be calculated in these different axes. In addition, Fransz *et al.* (2015) also mentioned that both the raw data, but also filtered data, for example with a sequential average filter, can be used to analyse the data. Furthermore, the start frame of the data analysis to quantify the ability to maintain balance can be set on any position after initial contact also. Usually, the moment of initial contact is used as the starting frame of the data analysis (Fransz *et al.*, 2015). However, analysis of pilot data showed that the centre of mass position is posterior to the base of support position and that the centre of mass moves anterior during the first frames after landing. This anterior movement of the centre of mass is necessary until the centre of mass is within the base of support, because at that point the person is in balance. Therefore, the data analysis can also start from the frame when the centre of mass is within the base of support for the first time.

Currently, no longitudinal studies have been performed with the AMAT and as such, the variability of the outcome measures of the AMAT over time are unknown. Information on the within-subject and between-session variability is necessary to interpret the outcomes during longitudinal studies and to relate this to performance and injury risk factors. As such, in this study the standing broad jump performance and ability to maintain balance of adolescent female soccer players of two different age groups were monitored throughout a season with the AMAT. The aims of this study were (1) to determine the within-subject variability of the standing broad jump performance and of the ability to maintain balance after jumps, (2) to determine the seasonal variability of the standing broad jump performance and ability to maintain balance after jumps and (3) to determine whether differences exist in within-subject variability and variability throughout a season between different methods to quantify the ability to maintain balance.

7.2 Methods

7.2.1 Participants.

A total of 29 elite adolescent female soccer players (14.9 ± 1.5 years old, 165.2 ± 6.5 cm, 55.4 ± 9.1 kg) playing in the under 14 (U14) [n = 16] and under 16 (U16) [n = 13] of an 'English FA Girls Centre of Excellence' participated in this study. Ethical approval was obtained from the ethics committee at Teesside University, School of Social Sciences, Business and Law (Appendix 1). The participants and parents of participants completed a written consent form.

7.2.2 Study design.

An observational study was performed throughout one full season (August – April). Table 7.1 shows the dates when data of the participants were collected. Data were collected prior to strength and conditioning sessions. The first session where jump data were collected was used as the baseline measure. During the first session each participant attended during pre-season, anthropometric data were collected. The set-up during each measurement was the same as described in Chapter 3 and depicted in Figures 3.2 and 3.3.

Table 7.1. Dates when jump data were collected and number of participants that attended the session.

	Pre-season				In-season				
Date	1 Aug	8 Aug	15 Aug	22 Aug	17 Oct	12 Dec	19 Dec	9 Jan	10 Apr
Participants	14	20	20	21	15	21	22	16	14

7.2.3 Protocol.

The protocol during each session was the same. The AMAT, described in Chapter 3, was used to measure the jump distances. When the players came in, foot markers (Figure 3.12) were attached to the laces of the shoes of each player. This was followed by a standardized warming-up. This warming-up consisted of three control and three maximal standing broad jumps and its main aim was to familiarize the athletes with the two different types of jumps described in Section 3.3.2. With the control standing broad jumps, the players had to jump as far as possible but still be able to control the landing and keep their feet in the same position. With the maximal standing broad jumps, the participants were instructed to jump as far as possible without taking the landing technique into account. After the warming-up, one by one the participants performed three control and three maximal standing broad jumps, with approximately 20-30 seconds rest between the jumps to allow for full recovery.

7.2.4 Data analysis.

The jump distances as calculated by the AMAT were used for further analysis. Moreover, the ability to maintain balance was only calculated for the control jumps, because the participants were not instructed to maintain balance after the maximal jumps. The position of the base of support and centre of mass were used to quantify the ability to maintain balance. To determine the position of the base of support, the position of the feet at the frame of initial contact was used. The algorithms to determine the position of the feet and to determine the frame of initial contact have been described in Chapter 3 and have been validated in Chapters 4 and 5. The position of the base of support was the position between the two foot markers at the frame of initial contact. The centre of mass data were collected with the algorithm described in Chapter 3 and validated in Chapter 6.

7.2.4.1 Calculation methods to quantify the ability to maintain balance

Based on Winter *et al.* (1998), Fransz *et al.* (2015) and pilot data collected with the AMAT, it was decided to quantify the ability to maintain balance with 48 different calculation methods. These calculation methods were based on five different aspects:

1) The quantification of the centre of mass. This can be based on body sway and body sway speed (Winter *et al.*, 1998). Here, body sway is defined as the absolute distance between the centre of mass and the base of support in millimetres. Body sway speed is defined as the number of millimetres the centre of mass moves during a frame. For both variables, a lower value implies a better stability.
2) The calculation of the body sway and body sway speed. Both variables were calculated as a maximum and as an average. Here, the maximum body sway and body sway speed were defined as the highest absolute value during any frame

from the first frame included in the analysis. The average body sway and body sway speed were defined as the average value from all frames included in the analysis.

3) The first frame that was included for analysis. The frame of initial contact and the first frame where the centre of mass is anterior of the base of support, ($CoM_{PA} > BoS_{PA}$).
4) The axis used to calculate the body sway and body sway speed. The body sway and body sway speed variables were calculated in the medio-lateral axis, the posterior-anterior axis and the combination of the two axes, the resultant axis.
5) The smoothing of the centre of mass data. The raw and the sequential averaged centre of mass data were included for analysis.

During five landings the centre of mass did not move anterior of the base of support. As a consequence, the ability to maintain balance of these landings could not be quantified. However, upon analysis of these five landings, it became apparent that the centre of mass was stabilised just before it became anterior of the base of support. This implies that the centre of mass stabilisation was excellent during those jumps. As a consequence, the variables during those five jumps were given the same value as the minimum value found for that specific variable, which represents an excellent ability to stabilise.

7.2.5 Statistical analysis.

Visual inspection of Q-Q plots of raw data indicated that the jump data did not violate normality assumptions, whereas the ability to maintain balance variables violated normality assumptions (Appendix 7). Therefore, the raw data of the jump distances are expressed as mean ± SD and the raw data of the methods to quantify the ability to maintain balance are displayed as median + range. All data were log-transformed prior to analysis and back transformed following the analysis to express the changes in percentages (Hopkins *et al.*, 2009). The analysis performed were the same for each dependent variable. A mixed model was performed for each session separately with the athletes as a random intercept to determine the within- and between-subjects' variability during each session (SPSS version 24). The variability was expressed as a typical error by taking the square root of the residual estimate (coefficient of variation [CV, %]; Hopkins, 2000). The between-subject SD was 19.9 cm for the control jump and 17.2 cm for the maximal jump. Thresholds to assess the magnitude of the within-subject were based on standardised thresholds of 0.1, 0.3 and 0.6 times the between-subject SD of the baseline measure for *small*, *moderate* and *large* changes, respectively (Hopkins *et al.,* 2009). Changes in within-subject variability were deemed meaningful when they crossed a magnitude threshold.

A mixed model analysis was performed with the athletes as random intercept and time as a fixed factor to determine changes throughout the season (SPSS version 24). Thresholds for meaningful changes in each dependent variable were based on the between-subject SD of the baseline measurement multiplied by 0.2, 0.6 and 1.2 for *small*, *moderate* and *large* effects, respectively (Hopkins *et al.,* 2009). To determine whether changes in performance were meaningful, the pairwise contrasts of the mixed linear model between the baseline and every other session were performed with mechanistic magnitude-based inference (Batterham & Hopkins, 2006) subsequently applied (Hopkins, 2017b). The following probabilistic terms were used to determine whether the observed change in performance were a true effect: < 0.5%, *most unlikely*; 0.5 – 5%, *very unlikely*, 5 – 24.9%, *unlikely*; 25 – 74.9%, *possibly*; 75 – 94.9%, *likely*; 95 – 99.4%, *very likely*; ≥ 99.5%, *most likely* (Batterham & Hopkins, 2006). The uncertainty in all estimates are expressed as 90% confidence intervals.

7.3 Results

At the start of the season, the average height and weight (± SD) of the participants were 160.9 ± 8 cm and 54.3 ± 11 kg. The baseline measures on the control and maximal standing broad jump (± SD) were 1867 ± 201 mm and 1933 ± 169 mm, respectively. The baseline measurements of the different calculation methods that quantified the ability to maintain balance are displayed in Table 7.2.

Table 7.2. The median (range) of the different calculation methods of the variables that quantify the ability to maintain balance.

First frame	Type of data	Axis	Body sway speed: Maximum speed*	Body sway speed: Average speed*	Body sway: Maximum distance†	Body sway: Average distance†
Initial contact	Raw	ML	14 (4 to 68)	4 (1 to 9)	72 (18 to 168)	38 (8 to 111)
		PA	39 (8 to 90)	8 (3 to 22)	227 (83 to 596)	148 (32 to 386)
		Resultant	42 (14 to 101)	9 (4 to 24)	232 (91 to 602)	157 (46 to 392)
	Sequential average	ML	4 (1 to 21)	1 (0 to 3)	53 (12 to 120)	40 (4 to 101)
		PA	19 (3 to 44)	4 (2 to 10)	180 (65 to 421)	141 (22 to 350)
		Resultant	20 (4 to 45)	4 (2 to 11)	183 (70 to 425)	147 (30 to 355)
CoM_{PA} > BoS_{PA}	Raw	ML	14 (4 to 68)	4 (1 to 9)	72 (18 to 168)	38 (8 to 111)
		PA	39 (8 to 90)	8 (2 to 22)	227 (51 to 596)	148 (31 to 386)
		Resultant	42 (8 to 101)	9 (3 to 24)	232 (74 to 602)	157 (45 to 392)
	Sequential average	ML	4 (0 to 21)	1 (0 to 3)	53 (12 to 120)	40 (4 to 101)
		PA	18 (2 to 44)	4 (1 to 10)	180 (24 to 421)	147 (17 to 350)
		Resultant	18 (2 to 45)	4 (1 to 11)	183 (36 to 425)	152 (3 to 355)

* The speed unit is millimetres per frame.

† The distance unit is calculated as the distance in millimetres from the centre of mass to the base of support during one frame.

7.3.1 Within subject variability

The magnitudes of the typical errors on the control and maximal jump were small to moderate in each session throughout the season (Figure 7.1). Figures with the outcome measures of the within-subject variability of the body sway and body sway speed are displayed in Appendix 8 and Appendix 9, respectively. For the body sway variables, the magnitudes of the typical errors were *extremely large* in the sequential average data of the medio-lateral axis (Appendix 8, Figures A8.1B, A8.1D, A8.4B, A8.4D). For all other body sway (Appendix 8) and for all body sway speed (Appendix 9) variables, the magnitudes of the typical errors ranged from *moderate* to *extremely large.* None of the within-subject typical errors had a *substantial* reduction throughout the season.

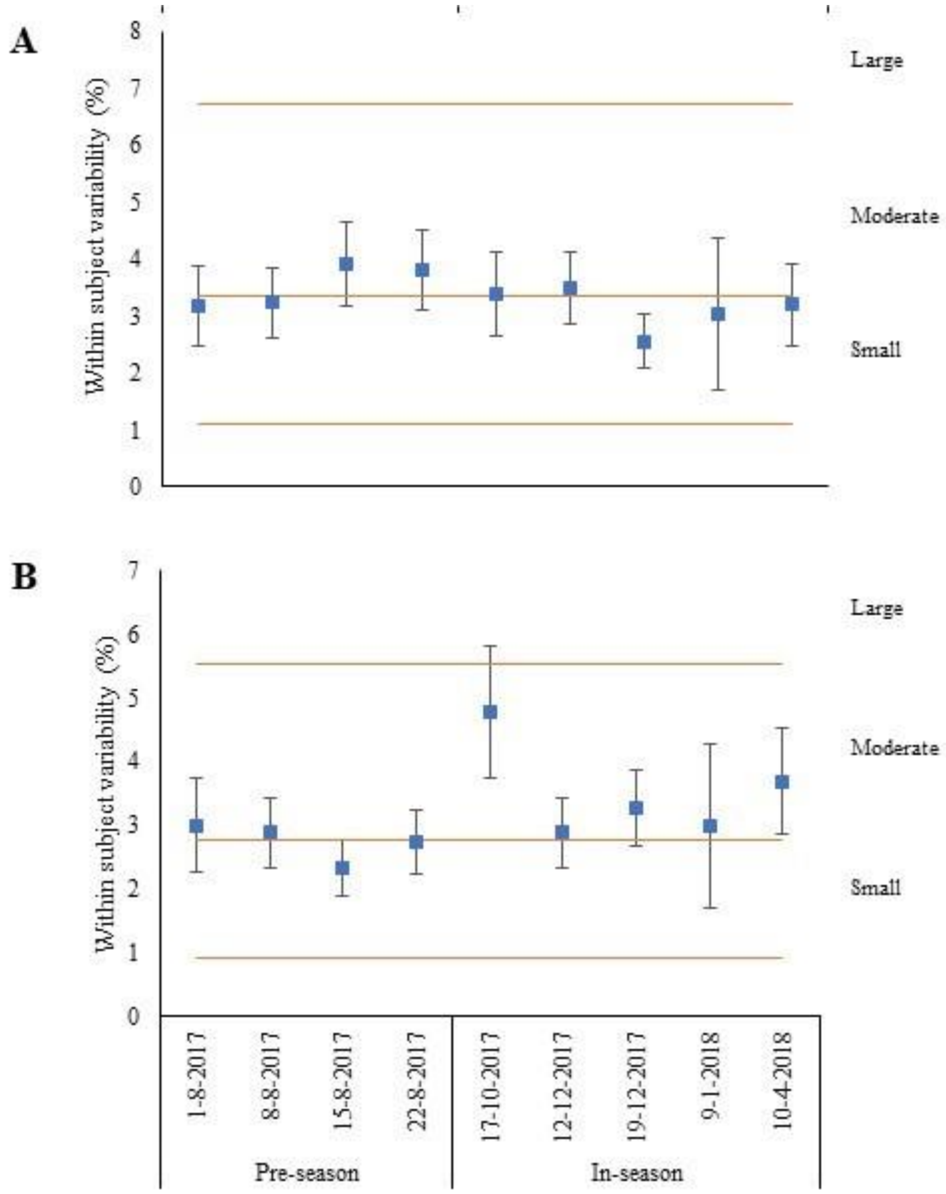

Figure 7.1. The within-subject variability (%) during the different sessions. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure 7.1A: control standing broad jump. Figure 7.1B: maximal standing broad jump.

7.3.2 Seasonal variability

The control and maximal standing broad jump performance improved *substantially* throughout the season when compared to the baseline measure (Figure 7.2). Figures with the outcome measures of the seasonal variability of the body sway and body sway speed variables are displayed in Appendix 10 and Appendix 11, respectively. No clear trend was found in any of the methods that calculate the body sway throughout the season. The body sway speed during landing reduced *substantially* throughout the season. This reduction in speed throughout the season was substantially larger in the posterior-anterior and resultant axis than in the medio-lateral axis, *substantially* larger in the method that used the initial contact as the starting point compared to the method that used $CoM_{PA} > BoS_{PA}$ as the starting point and *substantially* larger for the maximum speed than for the average speed calculations.

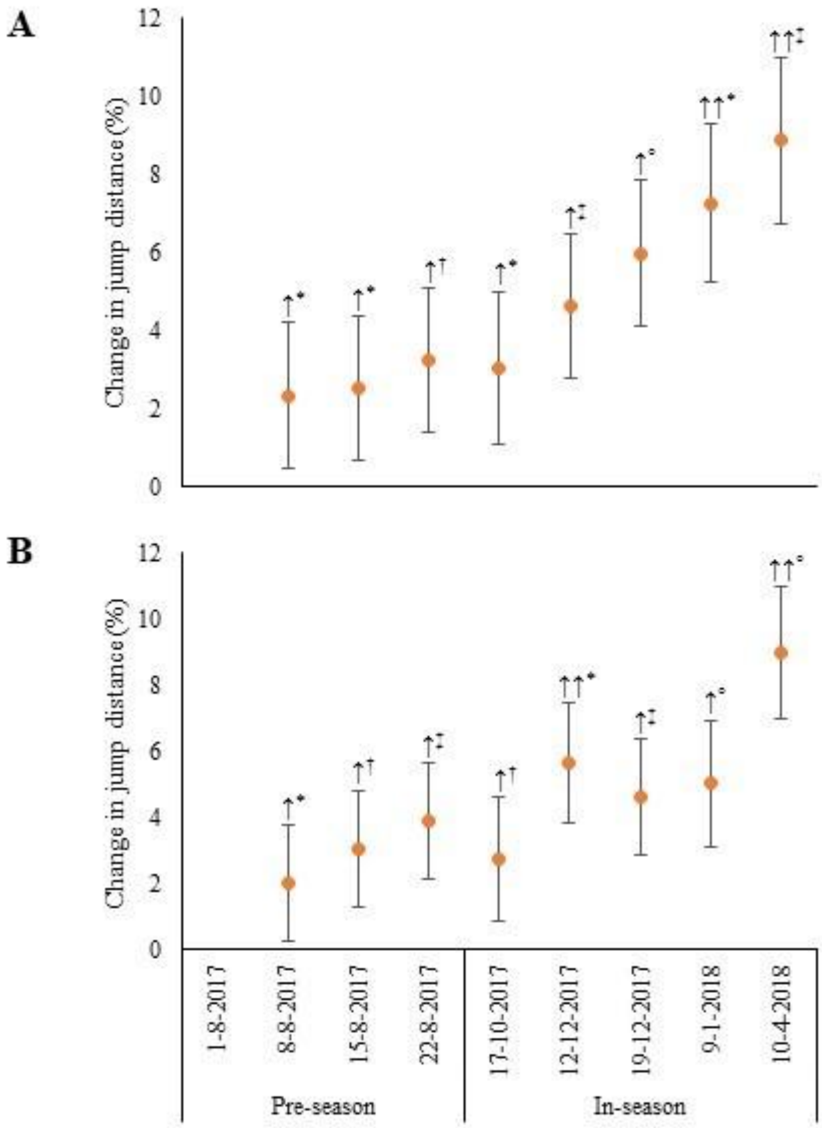

Figure 7.2. The changes in jump performance during the different sessions compared to the baseline measurement. The error bars display the 90% confidence limits. Figure 7.2A: control jump. Figure 7.2B: maximal jump. ↑ and ↑↑ represent small and moderate improvements in jump performance between sessions, respectively. *: possibly, †: likely, ‡: very likely, °: most likely.

7.4 Discussion

Female soccer has been professionalised in the last decade and participation has increased (FIFA, 2014; FA, 2018; KNVB, 2018). However, the majority of studies in soccer is still performed on males and consequently, the characteristics of female soccer players are less well understood (Milanovic *et al.*, 2017). It is important to understand the changes of physical characteristics of athletes throughout a season (Esmaeili *et al.*, 2018) to make informed decisions whether changes in these characteristics are related to performance and injury risk (Bakken *et al.*, 2016; Vanrenthergem *et al.*, 2017). In addition, it is important to understand the variability within a subject during the testing, because it affects the precision of the estimates of the change (Hopkins, 2000). The AMAT could be used to monitor movement performance in female athletes, because it can determine the jump distance in a reliable and valid manner and it can calculate the position of the centre of mass during jumps in a valid manner. Jump performance has been related to the horizontal power output of the lower extremity flexor muscles and has high correlations with sprint and acceleration performance (Markovic *et al.*, 2007; Nagano *et al.*, 2007; Salaj & Markovic, 2011; Lockie *et al.*, 2016; 2017) and the centre of mass can be used to quantify the ability to maintain balance (Winter, 1995), which has been linked to several lower extremity injuries (Hrysomallis, 2007, Zazulak *et al.*, 2007a Read *et al.*, 2016a; 2016b; 2017b). However, no information about the within-subject and seasonal variability of the outcome measures of the AMAT is available. Therefore, this study aimed to determine the within-subject variability of the standing broad jump performance and of the ability to maintain balance after jumps, it aimed to determine the seasonal variability of the standing broad jump performance and ability to maintain balance after jumps and it aimed to determine whether differences exist in within-subject variability and variability throughout a season between the 48 different methods to quantify the ability to maintain balance

7.4.1 Within-subject variability.

During data collection, each participant performed three control and three maximal standing broad jumps. The within-subject variability of each session was calculated via a linear mixed model and expressed as a typical error. The magnitudes of the typical errors of the control and maximal jump distance were *small* to *moderate* during all sessions. The magnitude of the typical errors might be due to the protocol used. Namely, it has been recommended to perform at least three jumps to collect reliable data, because the jump performance stabilises after the third jump (Ageberg *et al.,* 1998; Munro *et al.,* 2011; Griffin *et al.,* 2018). This implies that the within-subject variability decreases after those first three jumps and therefore the familiarization trials were added. As such, it is expected that those familiarization trials reduced the within-subject variability. However, the jump distance was not collected during those familiarization trials and as such it is not known what the exact effect of these familiarization trials on the within-subject variability was.

The magnitude of the typical errors of the ability to maintain balance ranged from *moderate* to *extremely large* and here no trend was apparent, because none of the variables showed a *substantial* in- or decrease in typical error throughout the season. An explanation for these typical errors might be the movement variability. Movement variability is an accepted phenomenon (Schmidt *et al.,* 1979; Stein *et al.,* 2005; Bartlett *et al.,* 2007; Stergiou & Decker, 2011) and partly explains the within-subject variability of the jump performance and ability to maintain balance. However, the ability to maintain balance is also affected by the jump performance. Namely, players that jump further need more time to stabilise their landing (Gribble *et al.,* 2012b). This might be due to the greater forces that need to be controlled when jumping larger distances (Gribble *et al.,* 2012b). If the player can control the forces on the body during the landing, the centre of mass stays within the base of support and the displacement of the centre of mass consequently remains low. However, when the player could not control the forces during the landing, the centre of mass moves outside the base of support and this can result in a larger centre of mass displacement and a possibly higher centre of mass speed. Players were instructed to jump as far as possible but with the ability to control the landing. It was observed that players tried to improve their previous jump performance, which might have resulted in higher forces that were harder to control and as such more difficult to maintain balance. This might have resulted in the *large* to *extremely large* typical errors on the variables that quantified the ability to maintain balance. However, this is only a hypothesis and more research is necessary to determine the cause of this high within-subject variability.

7.4.2 Seasonal variability.

The pairwise contrasts of the mixed linear model were used to determine the seasonal variability over the nine data collection measures throughout the season. The control and maximal standing broad jump performance increased *substantially* throughout the season. Interestingly, this is in contrast with one previous study in female adolescent soccer players (Taylor *et al.,* 2013), who reported decreased acceleration, sprint and agility performance throughout the season.

However, several other studies that monitored physical characteristics of professional (Clark *et al.,* 2008; Silva *et al.,* 2011), semi-professional (Caldwell & Peters, 2009) and male adolescent (Williams *et al.,* 2011) soccer players throughout one or multiple seasons found that the periods where certain physical characteristics increased and decreased differed between the physical characteristics within studies and between studies. Most studies mentioned increased physical characteristics (e.g. sprint, agility, jump, acceleration) from the start of the season to the middle of the season, but the physical characteristics that improved differed between the studies (Clark *et al.,* 2008; Caldwell & Peters, 2009; Silva *et al.,* 2011; Williams *et al.,* 2011).

All authors argued that their findings might be due to the different training strategies, the number of games played and the type of participants included in the study (Clark *et al.,* 2008; Caldwell & Peters, 2009; Silva *et al.,* 2011; Williams *et al.,* 2011; Taylor *et al.,* 2013). This argumentation is in accordance with studies in a controlled environment. Namely, Ageberg *et al.* (1998), Ross *et al.,* (2002), Maulder & Cronin, (2005) and Munro *et al.,* (2011) showed that that the horizontal jump performance in groups without any specific intervention is reliable on the short term (2 – 31 days) without the occurrence of systematic changes in performance. In contrast, plyometric training sessions over a period of 6 to 10 weeks *substantially* improved the horizontal jump performance in adolescent male soccer players and physical education students, whereas the control groups in these studies did not *substantially* improve over this period (Markovic *et al.,* 2007; Ramirez – Campillo *et al.,* 2015). As such, the changes in jump performance in the present study might be explained by the training that the athletes received throughout the season. During pre-season, the strength and conditioning sessions were familiarization sessions to make the athletes familiar with the different exercises. In contrast, the in-season strength and conditioning sessions had a strong focus on improving the muscle power output. However, this is merely a hypothesis because no control group was used in this study.

Another factor that might explain the substantial changes throughout the season is a learning or familiarisation effect. This effect is a systematic bias that causes a trend towards an improvement or decrease in a certain measure (Atkinson & Nevill, 1998; Hopkins, 2000). Before this study took place, performance data of the athletes was collected three times per season. Measuring the standing broad jump performance was included in this testing. Therefore, except for two athletes that were new at the club, all athletes were familiar with this type of testing and as such it was expected that no learning or familiarization effect would be apparent throughout the season. However, the use of AMAT was new for all athletes and this might have caused some sort of learning or familiarization effect that improved the jump performance throughout the season.

No clear trend in body sway movement over the season was found, whereas the body sway speed *substantially* reduced throughout the season. The body sway speed decreased especially during the in-season. These changes in performance might be related to the fact that the strength and conditioning sessions during pre-season were mainly used to familiarise the girls with the concept of strength and conditioning training, whereas the in-season period was used to train muscle power output and core stability. However, no criterion measure was used in this study and as such, it is not possible to determine whether the body sway or body sway speed quantified the ability to maintain balance correctly.

7.4.3 Differences in within-subject and seasonal variability between calculation methods of the ability to maintain balance.

The magnitude of the within-subject typical errors varied between the different calculation methods that quantify the ability to maintain balance. The magnitude of the typical errors of the calculation methods that used the sequential average data of the body sway variables in the medio-lateral axis were *extremely large* (Appendix 8, Figure A8.1B, Figure A8.1D, Figure A8.4B, Figure A8.4D). In contrast, the magnitude of the typical error of all other calculation methods ranged from *moderate* to *extremely large*. The *extremely large* typical errors of the four body sway variables are probably due to the relatively small between-subject values of those four variables. Namely, the magnitude of the typical error is based on the between-subject variability (Hopkins *et al.*, 2009), where a smaller between-subject variability relative to a higher within-subject variability results in higher magnitudes of the typical error. This is the case for those four variables and as such explains the *extremely large* magnitudes. It is not certain why the between-subject variability of those calculation methods was smaller. However, it is hypothesized that it is related to the low centre of mass displacement in the medio-lateral axis and the use of sequential average data, which also reduces the movement of the centre of mass data.

The seasonal variability differed between the body sway and body sway speed variables. Namely, only four body sway variables, the raw data of the average distance for both starting frames in the posterior-anterior and resultant axes (Appendix 10, Figures A10.2A, A10.3A, A10.5A, A10.6A), showed a *substantial* reduced centre of mass displacement throughout the season. In contrast, except for two body sway speed variables, the sequential average data of the average and maximal speed in the medio-lateral axis from the frame $CoM_{PA} > BoS_{PA}$ (Appendix 11, Figures A11.4B and A11.4D), all body sway speed variables showed a *substantial* reduced centre of mass speed throughout the season. Both the body sway and the body sway speed are based on the centre of mass position and as such, it was expected that both would have similar changes throughout the season. It is not clear why the body sway speed showed an improved ability to maintain balance, whereas the body sway did not. An issue is that no criterion measure was used in this study to quantify the ability to maintain balance and as such, no conclusion can be drawn whether body sway or body sway speed quantified the ability to maintain balance correctly. However, based on the improved jump performance and based on the training sessions that focused on muscle force output and core stability, it is expected that the body sway speed correctly identified an improved ability to maintain balance.

7.4.3 Limitations.

The performance on the standing broad jump has been related to sprint and acceleration performance and lower extremity flexor muscle strength (Markovic *et al.,* 2007; Nagano *et al.,* 2007; Salaj & Markovic, 2011; Lockie *et al.,* 2016; 2017). Those latter performance variables are, in contrast to the standing broad jump, directly used in a soccer match and related to the performance of soccer players. However, this study did not measure any of those performance variables and as such, it is not possible to determine the relation between the changes in standing broad jump performance and the changes in sprint and acceleration performance and lower extremity flexor muscles strength. Including those measures to get insight in the construct validity of the standing broad jump performance would have been preferable, but this was not possible due to practical constraints.

Similarly, it was not possible to collect a gold standard measure of the ability to maintain in balance. In an ideal situation, Vicon or force plate sway measures could have been collected to compare the balance measures of these systems with the balance calculations of the AMAT. However, the data were collected in an environment where no Vicon or force plate were available and as such, this data could not be collected.

Another limitation in this study was the lack of a control group. It is expected that the strength and conditioning training received by the players would have contributed to the improvement in jump performance. However, this is merely a hypothesis because no control group was included in this study. Similarly, a control group of peers that received the same training but only performed the jumps at the start and the end of the season could have determined whether a habituation effect was apparent in the group of athletes included in this study.

7.4.4 Implications.

Despite the limitations of this study, there are three implications for practitioners and scientists. At first, this was the first time the AMAT was used to perform research in a practical environment. This study showed that the AMAT can be used to monitor the jump performance in adolescent female soccer players in a reliable manner throughout a season. Secondly, the standing broad jump performance of the athletes improved throughout the season. It is not certain whether this is due to the training received by the athletes or due to other covariates. For example, a habituation effect can improve the jump performance throughout a session (Munro *et al.*, 2011), whereas maturation has been linked to an increased jump performance over a season (Read *et al.,* 2017c). As such, future studies should focus on controlling for possible covariates. At third, the ability to maintain balance has high within-subject typical errors and as such, the balance data should be interpreted with caution.

7.4.5 Conclusion

This study showed that the within-subject variability of the jump performance was *small* to *moderate* and as such it can be concluded that the jump data collected in this study was reliable. In addition, this study showed that adolescent female soccer players *substantially* improved their horizontal double-legged jump performance throughout the season. However, due to the methodology used in this study, it is not certain whether this improvement was due to changes in jump technique, lower extremity muscle force output or due to some habituation or maturation effect. As such, this study cannot give any clarity on the reason why the jump performance improved throughout the season. The different methods used to quantify the ability to maintain balance have *moderate* to *extremely* large typical errors and this implies that this data is not reliable and should be interpreted with caution. Moreover, one of the two main variables to quantify the ability to maintain balance, namely the body sway speed, reduced *substantially* throughout the season whereas the other main variable, the body sway, did not reduce *substantially* throughout the season. As such, it is also not clear whether any of the variables that were used to quantify the ability to maintain balance can be used in a practical setting.

7.4.6 Recommendations

Based on the limitations and conclusions of this study, three recommendations for future research will be made. The first two recommendations are related to the limitations of this study. At first, it is recommended to determine the construct validity of the AMAT over a longer period of time, for example a full soccer season. This can be done by comparing the AMAT with measures such as sprint and acceleration performance, lower extremity muscle force output, and vertical jump height. At second, it is recommended to perform this study with a control group to determine whether a learning or familiarization effect might affect the jump performance over time. At third, although it was not mentioned previously in this chapter, it was noted that the players were interested in the technology, for example to view their own movements (Section 3.3.5) or to compare their scores with their peers. Although this does not have to be a problem, it might affect the within-subject reliability because it can give athletes additional motivation to improve their jump performance. As such, it is recommended to determine what the effect of this feedback is on the jump performance of athletes.

Chapter 8. Conclusion and Future Directions

8.1 Introduction

The chapters of this thesis aimed to show the development of a reliable, valid, and practical movement assessment tool that makes use of depth-sensing technology. Chapter 2 used the *Sequence of Prevention* model (Van Mechelen *et al.,* 1987; 1992) as a guideline to assess the current injury problem in youth soccer. In this chapter, it was discussed how the sensorimotor system is related to injury risk in youth soccer players. Five main sensorimotor factors of lower extremity injury risk were identified, namely leg asymmetry, quadriceps dominance, frontal plane knee control, trunk dominance and dynamic stability. Several moment assessment tools were discussed that can assess one or multiple of these injury risk factors. Moreover, Chapter 2 also discussed the limitations of these movement assessment tools.

Based on the information provided in Chapter 2, Chapter 3 suggested the use of the Kinect as part of a movement assessment tool, because of its ability to collect kinematic data of specific anatomical landmarks, its low cost, the presence of depth-sensing technology, the portability of the system and the existence of an SDK (Clark *et al.,* 2012; Dutta, 2012; Bonnechere *et al.,* 2014). The main aims of Chapter 3 were to show the strengths and limitations of the Kinect to collect kinematic data and to show the development of the AMAT, a movement assessment tool that is able to collect kinematic variables during dynamic movements with the Kinect. A limitation of the Kinect is its unreliable kinematic data collection of the lower extremity anatomical landmarks. As such, as part of the AMAT, algorithms were developed to collect the position of the feet, the knees and the centre of mass in a reliable manner. It was expected that these algorithms could aid in quantifying the sensorimotor factors of lower extremity injury risk described in the previous paragraph.

The AMAT could only be used if the newly developed algorithms were reliable and valid. It was not possible to develop a knee marker tracking algorithm that was reliable and valid in the timeframe of this thesis. Therefore, primary aims of this thesis were to determine the reliability and validity of the foot marker tracking algorithm and the centre of mass algorithm. The static reliability and validity of the foot marker tracking algorithm were determined in Chapter 4. Here it was found that the foot marker tracking algorithm of the AMAT could reliably and validly collect the marker positions, but it should be noted that the within-trial reliability was lower on positions further away from the camera. Subsequently, in Chapter 5 the dynamic validity of the foot marker tracking was determined. Here it was shown that the AMAT was able to measure the jump distance in a valid way, which implies that the AMAT could track the feet markers during dynamic movements. Thereafter, Chapter 6 aimed to determine the validity of the centre of mass algorithm. Here it was shown that this algorithm was valid to determine the position of the centre of mass during jumps when compared to the centre of mass position as calculated with Vicon. The AMAT was developed to be used in a practical setting and therefore, the last aim of this thesis was to determine the within-subject and seasonal variability in jump performance and ability to maintain balance in female adolescent soccer players. The ability to maintain balance was quantified based on the position of the base of support and the position of the centre of mass throughout the landing. The jump performance in adolescent female soccer players increased *substantially* throughout the season and *small* to *moderate* within-subject typical errors were found during each session. The method to quantify the ability to maintain balance had *moderate* to *extremely large* within-subject typical errors, which implies that this data should be interpreted with caution.

8.2 Summary of Findings

This thesis could be split in three separate sections, namely the development of the AMAT, determining the reliability and validity of the AMAT in a laboratory setting and the use of the AMAT in a practical setting. Per section, the findings and implications of these findings will be discussed. In addition, the novelty of each section and its original contribution to knowledge will be described.

8.2.1 The development of new algorithms for the AMAT.

The aim of the kinematic data collection with the AMAT is to eventually quantify sensorimotor risk factors of lower extremity injuries. Therefore, several studies were used to develop a rationale to determine which sensorimotor risk factors could be quantified with the AMAT. Winter (1995) described how the centre of pressure, centre of mass and base of support are the outcome measures of the sensorimotor system to maintain balance. In addition, Read *et al.* (2016a; 2017b) described the, according to them, five most important sensorimotor risk factors of lower extremity injuries and how these factors could be measured with movement assessment tools. Based on these studies, it became clear that the main focus of the AMAT should be on quantifying the ability to maintain balance (Winter, 1995; Horak, 2006; Read *et al.*, 2017b), on quantifying the movement strategies used to maintain balance (Winter, 1995; Horak, 2006; Read *et al.*, 2017b) and to determine movement performance to quantify leg asymmetry (Read *et al.*, 2017b).

To quantify the sensorimotor risk factors of lower extremity injuries, previous studies on the Kinect v2 and own experiences during pilot testing were used to determine what type of algorithms had to be developed. For example, the foot position cannot be measured in a reliable way with the Kinect v2 (Wang *et al.*, 2015). Therefore, foot markers and a foot marker tracking algorithm were developed to improve the collection of the foot position, based on Paolini *et al.* (2014) and MacPherson *et al.* (2016). In addition, the centre of mass cannot be collected with the Kinect v2 and as such, similar to previous studies that used the voxel data to determine the position of joint centres (Gammelgaard, 2015; Giblin *et al.*, 2016; McGroarty *et al.* 2016; Bauer *et al.*, 2017), this point cloud was used to calculate the 3D data of the centre of mass, based on Goffredo *et al.* (2006) and Allin *et al.*, (2008). Moreover, to quantify movement strategies, marker tracking algorithms were developed to determine the position and movement of anatomical landmarks. During jumps, the movement strategies are often associated with injury risk (Hewett *et al.*, 2005; 2006; Walden *et al.*, 2015), especially in the first 50 milliseconds after initial contact (Krosshaug, 2007). Studies frequently use jumps to assess the movement strategies at initial contact (Hewett *et al.*, 2005; 2016; Welling *et al.*, 2018) and as such, it is important to determine the moment of initial contact. In addition, the moment of initial contact can be used to determine jump distances, which can be used to measure leg asymmetry by comparing the jump distances when pushing off with the left and right foot (Read *et al.*, 2017c). Therefore, an algorithm was developed to determine the moment of initial contact.

The algorithms described in the previous paragraph make it possible to collect kinematic data in a reliable and valid manner. This kinematic data collection might be used to quantify several sensorimotor risk factors of lower extremity injuries, such as leg asymmetry, ability to maintain balance, and trunk dominance. Therefore, both the rationale behind the development of these algorithms and the development of these algorithms are original contributions to knowledge. Moreover, this rationale can also be used to develop new movement assessment tools or assessment criteria to quantify other sensorimotor risk factors of lower extremity injuries. An additional benefit of the AMAT is the development of applications for laptop and tablet that make this tool practical in use. This could aid in the process of monitoring sensorimotor risk factors of lower extremity injuries in athletes.

8.2.2 The reliability and validity of the foot marker tracking algorithm and the centre of mass algorithm.

A measurement system should only be used if it is reliable and valid (Atkinson & Nevill, 1998). However, no information was available about the reliability and validity of the AMAT. Therefore, the reliability and validity of the newly developed foot marker tracking algorithm was determined in two studies and the validity of the centre of mass algorithm was determined in one study. This was an original contribution to knowledge and could be used by practitioners to interpret the data they collect with their participants.

8.2.2.1 Static reliability and validity of the foot marker tracking algorithm.

The reliability of the foot marker tracking algorithm was determined in a static setting to obtain the pure error of the algorithm and the Kinect v2. This is important because the technological error of a measurement system should be lower than the smallest worthwhile change of the variable it is going to measure (Hopkins, 2000). The within-trial typical error of the Kinect was smaller than four millimetres on all positions and the between-trial typical error ranged from approximately one to seven millimetres for the Kinect. The smallest worthwhile change is 0.2 * the between-subject standard deviation (Hopkins, 2000). Chapter 7 found a between-subject standard deviation of 19.9 centimetres for the control standing broad jump and 17.2 centimetres for the maximal standing broad jump. This implies that the smallest worthwhile change is approximately 3.5 centimetres, which is higher than the typical errors found in Chapter 4. This shows that the AMAT is able to determine jump distances in a reliable and valid manner. The largest typical errors were found in the positions furthest away from the camera. This can be explained by the process of calculating the 3D data. The data in the posterior-anterior axis are based on the depth-sensing technology (Lachat *et al.*, 2015). Yang *et al.* (2015) described a random error of the depth-sensing technology that increased from two millimetres close to the camera to approximately four millimetres on positions further away from the camera (Yang *et al.*, 2015), which is similar to the outcomes of this study.

When determining the static validity, it was found that measurements of distances between markers were less accurate further away from the Kinect. On positions close to the camera all data points fell within the 95% limits of agreement, whereas on the positions furthest away from the camera only 83% of all data points fell within the 95% limits of agreement. This shows that the validity of the foot marker tracking algorithm is higher closer to the Kinect camera.

8.2.2.2 Dynamic validity of the foot marker tracking algorithm.

The dynamic validity of the foot marker tracking algorithm was based on the ability of the AMAT to measure jump distances. As such, both the foot marker tracking algorithm (Section 3.3.3.2) and the algorithm that determines the frame of landing (section 3.3.3.6) were used in this study. Over 93% of all jump measures were within the 95% limits of agreement and the average difference between the tape measure and AMAT was less than three millimetres for the five types of jumps included in this study.

As shown in Chapter 4, the positions furthest away from the Kinect have the highest technological errors. Those errors were approximately 2 millimetres higher compared to the errors closer to the Kinect. As such, it was important for an optimal validity of the foot marker tracking algorithm and jump distance calculation that the athletes jumped towards the camera. In addition, the origin of the coordinate system of the AMAT in the posterior-anterior axis was set at the push-off position (Bar-Position 1, Figure 4.3) of the jump. By doing this, only the landing distance had to be included to calculate the jump distance, which probably improved the validity of the jump distance measurement.

An issue with the Kinect camera that might have affected the reliability and validity of the foot marker tracking algorithm is the relative low sampling frequency of 30 Hz. This is over three times lower than Vicon (100 Hz) and it implies that the data collection during dynamic movements of the feet might have been affected. This became evident during the development of the algorithm to track the foot markers and to determine the moment of initial contact, because it was not possible to accurately determine the position of the foot markers during the flight phase of the jump. Due to the use of the coloured foot markers, it was possible to determine the moment of initial contact, but it shows that the foot marker tracking is not optimal. Timmi *et al.* (2018) also showed that the error of the marker tracking with the Kinect v2 increased during higher speed dynamic movements when compared to a criterion measure. Nevertheless, the findings of Chapter 4 and Chapter 5 showed that the foot marker can be tracked in a reliable and valid manner during dynamic movements and as such, the base of support position and landing position can be determined. Also, similar to Timmi *et al.* (2018), these markers might also be used to collect kinematic data of other joints, especially because the original knee marker tracking algorithm (Section 3.3.3.3) did not work in dynamic situations.

All data used in these two reliability and validity studies were raw data collected by the AMAT. This shows that the depth-sensing technology of the Kinect should be sufficient to collect 3D data of anatomical landmarks via algorithms that are developed to track retroreflective markers. In addition, applying specific filters on the data might reduce the errors of the system even further. Also, it is expected that future depth-sensing technology will have a reduced random error, due to the quick development of technology in general. This reduced random error will then result in a higher reliability and validity of the AMAT, without any other changes in the algorithms.

8.2.2.3 Validity of the centre of mass algorithm.

The centre of mass algorithm collects all voxels that capture a part of the body. With this information, it calculates the average position of these voxels to determine the position of the centre of mass (section 3.3.3.4). In this thesis it was aimed to determine the validity of the centre of mass algorithm. To that purpose, the centre of mass data was collected with AMAT and Vicon during seventeen jumps of two participants. Bland-Altman plots and Pearson correlations were used to compare both systems. Based on the outcomes of this study, it was concluded that the centre of mass algorithm of the AMAT is probably valid. However, due to practical constraints, these conclusions are preliminary and more research is necessary to confirm the conclusions of this study. The next paragraphs will address two of the main limitations of this study.

At first, it was not possible to determine the technological error of this algorithm, whereas knowledge about the technological error is important to obtain knowledge on the reliability of the measurement system without any biological error (Hopkins, 2000). This is due to the nature of this algorithm, because centre of mass data could only be collected to determine the reliability and validity of this algorithm when humans were involved. However, as explained in Chapter 4, the technological error should be collected in a setting where no humans are involved, to remove any type of biological error. As such, no information about the reliability of the centre of mass algorithm could be obtained, which limits the knowledge about the technological error of the centre of mass algorithm.

At second, only two adult males were recruited for this study. This was due to the fact that certain parts of the body work of the AMAT system were picked up by the Vicon system as retroreflective markers. However, the combination of AMAT system and Vicon was only available for a few days per month and as such, it took approximately two months to solve this issue. Consequently, there was not much time to collect data for this study. This implies that the data collected in this study is not generalisable to the whole population and more studies on males and females of several ages and body postures is necessary to determine the reliability and validity of the centre of mass algorithm in the general population.

8.2.3 Monitoring adolescent soccer players with the AMAT.

The last study of this thesis was performed on adolescent female soccer players in a practical setting. The AMAT was used to measure the standing broad jump performance and the ability to maintain balance of adolescent female soccer players throughout a season. The aims of this study were (1) to determine the within-subject variability of the standing broad jump performance and of the ability to maintain balance after jumps, (2) to determine the seasonal variability of the standing broad jump performance and ability to maintain balance after jumps, and (3) to determine whether differences exist in within-subject variability and variability throughout a season between different methods to quantify the ability to maintain balance.

The magnitudes of the within-trial typical errors of the control and maximal jump were *small* to *moderate* during all sessions. This implies that the jump performance of adolescent female athletes can be measured reliably when three familiarization standing broad jumps are followed by three standing broad jumps that are measured. This is in accordance with previous studies such as Ageberg *et al.* (1998), Munro *et al.* (2011) and Griffin *et al.* (2018) who all found that the jump performance stabilised after the third jump. The within-trial typical errors of the measures that quantified the ability to maintain balance ranged from *moderate* to *extremely large.* This implies that although the centre of mass data can be collected in a reliable manner, the methods used to quantify the ability to maintain balance are not reliable and this data should be interpreted with caution.

The athletes had a *substantial* increase in jump performance throughout the season. Moreover, methods that used the body sway speed to quantify the ability to maintain balance reduced *substantially* throughout the season, whereas the methods that used the body sway to quantify the ability to maintain balance did not find a clear trend throughout the season. Due to the nature of the study, it is not certain whether changes in the different variables are due to a learning effect, due to training, due to maturation or due to another effect. Therefore, future studies should focus on factors that could be associated with enhanced jump performance of these athletes throughout the season. Moreover, future studies should also focus on developing a new method to quantify the ability to maintain balance that is more reliable. In addition, this method should be compared with a gold-standard method, such as the body sway measured with Vicon or with force plates. Due to time and practical constraints it was not possible to use one of these systems during the training sessions of the female athletes. As such, no conclusions can be drawn on what type of method would have the highest reliability and validity to quantify the ability to maintain balance.

8.2.4 Overall implications.

The aim of the AMAT is to collect kinematic data in a reliable, valid and practical manner during dynamic movements, because the kinematic data could be used to quantify sensorimotor risk factors of lower extremity injuries. The foot marker tracking algorithm of the AMAT is reliable and valid and jump distances can be measured in a valid manner with this algorithm. This implies that the leg asymmetry during single leg hops and jumps (Noyes *et al.*, 1991) can be quantified with the AMAT. In addition, it was shown that the centre of mass algorithm of the AMAT is valid. However, it was not possible to quantify the ability to maintain balance based on the centre of mass and base of support data in a reliable manner. This implies that of the five sensorimotor risk factors of lower extremity injuries in youth soccer mentioned by Read *et al.* (2016a; 2016b; 2017b) (quadriceps dominance, leg asymmetry, assessment of frontal plane knee control (knee abduction), trunk dominance, and dynamic stability), only leg asymmetry can be determined in a reliable and valid manner with the AMAT system. This is not interesting from a practical perspective, because measuring the jump distance with a tape measure is more practical than installing the AMAT to measure jump distances.

In Chapter 2 it was argued that the *Sequence of Prevention* model (Van Mechelen, 1987; 1992) could be used to determine whether new movement assessment tools were contributing in the injury prevention process in youth soccer. This thesis did not include a study where the effect of the use of the AMAT on the number of injuries in a youth soccer academy were compared. As such, it is not possible to determine whether the AMAT can aid in the injury prevention process of youth soccer academies. In the previous paragraph it was described that the AMAT in its current state is not interesting for practitioners, because the jump distance is the only performance measure of the AMAT that can be collected in a reliable and valid manner. Based on this information, it can be argued that, based on current information, the AMAT will not aid in the injury prevention process of youth academies. In the future, the AMAT might be useful for the injury prevention process of youth academies, if it becomes possible to collect reliable and valid kinematic data during dynamic movements and if this data can be used to quantify sensorimotor factors of lower extremity injury risk. Moreover, the ability to provide video feedback to athletes is another feature that might make the AMAT an addition in the injury prevention process. However, more development and research on these topics is necessary to determine whether the AMAT can contribute in the injury prevention process of youth academies.

8.3 Strengths and Limitations of the Studies

A strength of this thesis was that the author was involved in the development of the different algorithms. Consequently, algorithms could be adjusted to collect specific data that could not be collected with the original version of the AMAT. For example, Chapter 4 described the static marker tracking of the foot marker tracking algorithm. In the original algorithm, the area where the algorithm searched for the marker was based on the position of the ankle. However, by adjusting the algorithm, the search area was changed to specific locations around the Bar-Positions. Due to this small change, the algorithm still worked in exactly the same way (i.e. it searched for the markers in a specific area) but could now collect static data instead of data that was affected by human movement. Another benefit of the fact that the algorithms were developed by the author is that the data could be outputted in a way specified by the author. This eventually led to the quick analysis of large data sets, because most data were processed instantly and as such the data analysis after data collection was limited. The large datasets in multiple studies made the outcome of these studies more robust.

A second strength of this thesis was the combination of the data collected in a laboratory and in a practical setting. The data collected in the laboratory setting provided information on the reliability and validity of the different variables collected with the movement assessment tool. This information is important when interpreting the outcomes of the measurement system (Hopkins, 2000). In addition, the data collected in the practical setting provided information about the variability of the variables collected with the AMAT. As such, the combination of the data collection in both settings gave a complete overview of the current strengths and weaknesses of the outcome measures of the AMAT. This data can be used by practitioners as reference values when interpreting their own collected data with AMAT.

A limitation of this thesis was the absence of reliability and validity study on the knee marker tracking algorithm and the absence of studies where squat and balance movements were assessed with the AMAT. As such, this thesis does not give a complete overview of the possibilities of the AMAT. This was due to time constraints, because it was not possible to get all algorithms functioning in time and to collect data on them within the time frame of this thesis. The issue with the knee marker tracking was that it could not track the knee in a reliable and valid manner during dynamic movements. With the balance movements, different tracking algorithms had to be developed to track the feet and knee movement and an algorithm had to be developed for the squat movement to determine the depth of a squat. Although a lot of time was spent on these algorithms during this project, they were not finished in time to be part of this thesis.

A second limitation of this thesis was the absence of a method that could be used to determine the technological error of the centre of mass algorithm. This is due to the fact that a human that is instructed not to move already shows a small body sway (Winter, 1995). This body sway will affect the reliability measurement of the algorithm, whereas it is necessary to determine the reliability of the algorithm without any human involvement, the technological error (Hopkins, 2000). However, it was not possible to conduct a similar type of experiment as in Chapter 4, where the foot marker tracking algorithm was collected without the foot marker being attached to a human. This was due to the fact that at the time of data collection, it was only possible to collect the centre of mass of humans based on the anatomical landmarks of the Kinect (Section 3.3.3.4). It was tried to use different types of mannequins and other static objects that looked like humans, but these were not recognized by the Kinect as humans and as such, the centre of mass of these static objects could not be determined with this algorithm. Consequently, it was not possible to determine the technological error of the centre of mass algorithm.

8.4 Future Recommendations

Based on the outcomes of this thesis, the experience of using this tool in practice, the feedback from practitioners, and the existing literature, recommendations will be made for further development of the movement assessment tool, for future research projects and for practitioners.

8.4.1 Recommendations for the development of the AMAT.

The position of the centre of mass relative to the base of support determines whether someone is in balance or not (Winter, 1995). As such, the main aim of the sensorimotor system is to keep the centre of mass within the base of support to keep the body in balance

(Winter, 1995). In Chapter 6 it was shown that the AMAT is able to collect the position of the centre of mass in a valid manner. However, the technological error could not be determined and as such it is recommended to come up with a method that is able to determine the technological error of the centre of mass algorithm. Furthermore, in Chapter 7 it was shown that the methods used to quantify the ability to maintain balance, based on the centre of mass and base of support position, were not reliable. As such, a first recommendation for the development of the AMAT is to come up with a method that makes it possible to quantify the ability to maintain balance in a reliable manner.

A second recommendation is to improve the kinematic data collection of the anatomical landmarks with AMAT. Kinematic data collection, for example to measure knee, hip and trunk displacement during landings, is frequently used to assess movement strategies (Hewett *et al.,* 2005; 2006b; 2016a; Welling *et al.,* 2018a). As already described in Chapter 3, no methods were developed to calculate the joint angles and joint displacements in a reliable and valid manner. Besides the foot marker tracking algorithm and the centre of mass algorithm, this thesis did not come up with any methods to improve the kinematic data collection of other anatomical landmarks in a reliable and valid manner. However, from a research and a practical perspective, the development of a method to determine the joint angles and joint displacements of participants would improve the usability of this tool.

A method to improve the kinematic data collection is the use of Kalman filters. Kalman filters are described in the literature to filter nonlinear data (Julier & Uhlman, 1997; Wan & Van der Merwe, 2000). The movement of the body during movements is non-linear and accordingly, these filters have been used to improve the kinematic data collection of the Kinect (Bo *et al.,* 2011; Eltoukhy *et al.,* 2016; Segura *et al.* 2016). As such, it would be interesting to determine what the effect of a Kalman filter is on the reliability and validity of the AMAT to collect kinematic data.

A third recommendation is to improve the voxel data collected with the Kinect. The voxel data of the athlete captured by the Kinect is used to determine the position of the centre of mass (Section 3.3.3.4) and to display video feedback (Section 3.3.5). Chapter 3 described the study of Bauer *et al.* (2017), who created a 3D benchmark image of the participant by collecting the dimension of the different body segments of the participant during a standardized calibration measurement. They found that their method underestimated the length of the different segments by approximately 1% to 5% when compared to the calculation of the body segments with MRI. This method could be used to improve the centre of mass data collection and in addition it could make the video data look more realistic.

The method described by Bauer *et al.* (2017) might be used to collect other types of kinematic data also. Namely, Giblin *et al.* (2016) and McGroarty *et al.* (2016) described how they used the voxel data of the Kinect to create a point cloud of the different segments. The video data described by Bauer *et al.* (2017) has an improved validity compared to the voxel data and as such, the kinematic data collected with the method by Bauer *et al.* (2017) might have an increased validity compared to Giblin *et al.* (2016) and McGroarty *et al.* (2016). However, a possible issue to implement the method of Bauer *et al* (2017) in this movement assessment tool is the different position of the camera. Bauer *et al.* (2017) mentioned that their camera was positioned straight in front of the person and was placed 50 centimetres above the floor. In contrast, the camera of the movement assessment tool captures the person in an angle of 30 degrees with the horizontal and is approximately 185 centimetres above the floor. Therefore, it is important to examine the effect of the difference in camera position on the ability to improve the point cloud data. In addition, it is important to determine whether it is possible to implement the method by Bauer *et al.* (2017) in this movement assessment tool, because creating a calibration file took 45 to 90 seconds. Based on this long calibration time, it was decided to focus on the development of markers, because this seemed more practical due to the quick attachment of the foot and knee markers compared to the longer calibration time described by Bauer *et al.* (2017).

8.4.2 Recommendations for future research.

It is recommended for future studies to determine the ability of the AMAT to determine the squat and balance performance in a reliable and valid manner. This thesis focused on the reliability and validity of algorithms during horizontal jumps to measure the position of the feet. The use of horizontal jumps has been widely described in the literature, for example to measure training status (Halson, 2014), to measure horizontal muscle power output (Markovic *et al.,* 2007; Nagano *et al.,* 2007; Salaj & Markovic, 2011) and to quantify sprint and acceleration performance (Lockie *et al.*, 2016; 2017). Similarly, balance movements (Gribble *et al.*, 2012a; Coughlan *et al.,* 2014; Chimera *et al.,* 2015; McCann *et al.,* 2015; Ness *et al.,* 2015; Overmoyer *et al.*, 2015) and squats (Butler *et al.,* 2010; Schoenfeld, 2010; Myer *et al.*, 2014; Lloyd *et al.*, 2015) are frequently mentioned in the literature to determine injury risk factors and performance parameters. Algorithms have been developed for the AMAT to determine the performance on the anterior balance movement and the back- and overhead-squat, but currently no information is available on the reliability and validity of these algorithms.

8.4.3 Recommendations for practitioners.

The monitoring of athletes is frequently described in the literature because it can provide information about their current training status and the monitoring can be useful to identify strengths and weaknesses of athletes. For example, Stølen *et al.* (2005), Bangsbo *et al.* (2008), Taylor *et al.* (2013), Halson (2014), Watkins *et al.* (2017) and McLaren *et al.,* (2018) described how physical fitness tests could be used determine the effect of a training session or a period of training sessions on levels of fatigue and physical fitness. It was expected that the AMAT could also be used for monitoring purposes, with the benefit that it could provide information about leg power output via the jump distance, information about the ability to maintain balance via the centre of mass displacement and information on movement strategies via the collection of kinematic data of anatomical landmarks. However, this thesis could only focus on the reliability and validity of the foot marker algorithm and centre of mass algorithm. Therefore, it is recommended that practitioners interpret the data with caution and in addition, they should not relate the data collected with the AMAT to other physical performance measures until it has been demonstrated that the AMAT is valid to indirectly measure those physical performance measures.

8.5 Conclusion

This study showed the development of a new movement assessment tool, the AMAT, that makes use of depth-sensing technology to collect kinematic data of anatomical landmarks. To that purpose, algorithms were developed to collect reliable and valid kinematic data and to develop applications to make the AMAT practical in use. The idea behind the AMAT is to use this kinematic data to quantify sensorimotor risk factors of lower extremity injuries. As such, the most important algorithms of the AMAT are the foot marker tracking algorithm, the centre of mass algorithm and the knee marker tracking algorithm. This thesis focused on the foot marker tracking algorithm and the centre of mass algorithm during horizontal jumps. It was shown that the foot marker tracking algorithm is valid and it can be used to measure the jump performance of adolescent soccer players in a reliable and valid manner throughout a full season. The centre of mass algorithm is valid to calculate the position of the centre of mass, but 48 methods that were used to quantify the ability to maintain balance based on the centre of mass position were not reliable. Also, no reliability and validity studies have been performed on the knee marker tracking algorithm of the AMAT and there is currently no information available about the reliability and validity of the foot marker tracking algorithm and centre of mass algorithm during balance movements and squats. As such, of the five main sensorimotor risk factors of lower extremity injuries discussed in this thesis, leg asymmetry is currently the only one that can be collected in a reliable and valid manner. This implies that the AMAT could be used in practice, but practitioners should be cautious when interpreting the outcome measures, because the current studies do not provide enough information on the reliability and validity of the AMAT. As such, it is recommended for future studies to perform reliability and validity studies on the different algorithms used by the AMAT to collect kinematic data. Subsequently, it can be determined whether the AMAT can be used to quantify sensorimotor risk factors of lower extremity injury risk.

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Appendices

Appendix 1. Ethical clearance approval letters

Approval letter Chapter 4

Professor Paul Crawshaw
Dean
School of Social Sciences, Business & Law
Teesside University Middlesbrough
Tees Valley TS1 3BX UK
T: +44 (0)1642 342851
tees.ac.uk

Teesside University

26 October 2016

Ref: SSSBLREC010

Professor Iain Spears

Dear Iain

School Research Ethics Committee

Project title:	Reliability and validity study of a marker tracking system in a static setting
Researcher(s) Name:	Mark Wijenbergen

The above proposal has received ethical clearance and the project may proceed.

If the research should change or extend beyond the indicated dates, the researcher(s) must report the nature of the proposed changes and the revised end date to the Chair/Secretary of the Research Ethics Committee.

Yours sincerely

M. A. Tayler.

Dr Martin Tayler
Chair
Research Ethics Committee
School of Social Sciences and Law

VAT REG NO. GB 686 4809 81

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POSITIVE ABOUT DISABLED PEOPLE

THE QUEEN'S ANNIVERSARY PRIZES 2013

Approval letter Chapter 5

Professor Paul Crawshaw
Dean
School of Social Sciences, Business & Law
Teesside University Middlesbrough
Tees Valley TS1 3BX UK

T: +44 (0)1642 342851
tees.ac.uk

Teesside University

26 October 2016

Ref: SSSBLREC011

Professor Iain Spears

Dear Iain

School Research Ethics Committee

Project title: Validity of tracking moving markers with a newly developed marker tracker system

Researcher(s) Name: Mark Wijenbergen

The above proposal has received ethical clearance and the project may proceed.

If the research should change or extend beyond the indicated dates, the researcher(s) must report the nature of the proposed changes and the revised end date to the Chair/Secretary of the Research Ethics Committee.

Yours sincerely

M. A. Tayler.

Dr Martin Tayler
Chair
Research Ethics Committee
School of Social Sciences and Law

VAT REG NO. GB 686 4809 81

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Approval letter Chapter 6

Professor Paul Crawshaw
Dean
School of Social Sciences, Humanities & Law
Teesside University Middlesbrough
Tees Valley TS1 3BX UK
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Teesside University

29th March 2018

Matt Portas
Teesside University
Middleborough
Tees Valley
TS1 3BX

Dear Matt

School Research Ethics Committee

Project title(s): Criterion validity of a new technology to calculate the centre of mass

Researcher(s) Name: Mark Wijnbergen

The above proposal(s) have received ethical clearance and the project(s) may proceed. If the research should change or extend beyond the indicated dates, the researcher(s) must report the nature of the proposed changes and the revised end date to the Chair/Secretary of the Research Ethics Committee.

Yours sincerely

Katherine Swainston
Chair
Research Ethics Committee
School of Social Sciences, Humanities and Law

VAT REG NO. GB 686 4809 81

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Approval letter Chapter 7

Professor Paul Crawshaw
Dean
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Teesside University

10th October 2017

Matthew Wright
Teesside University
Middleborough
Tees Valley
TS1 3BX

Dear Matthew

School Research Ethics Committee

Project title(s): Applied sports science support to young high performance

Researcher(s) Name: Matthew Wright

The above proposal(s) have received ethical clearance and the project(s) may proceed.

If the research should change or extend beyond the indicated dates, the researcher(s) must report the nature of the proposed changes and the revised end date to the Chair/Secretary of the Research Ethics Committee.

Yours sincerely

Katherine Swainston
Chair
Research Ethics Committee
School of Social Sciences, Humanities and Law

Appendix 2. Q-Q plots of the data of Chapter 4

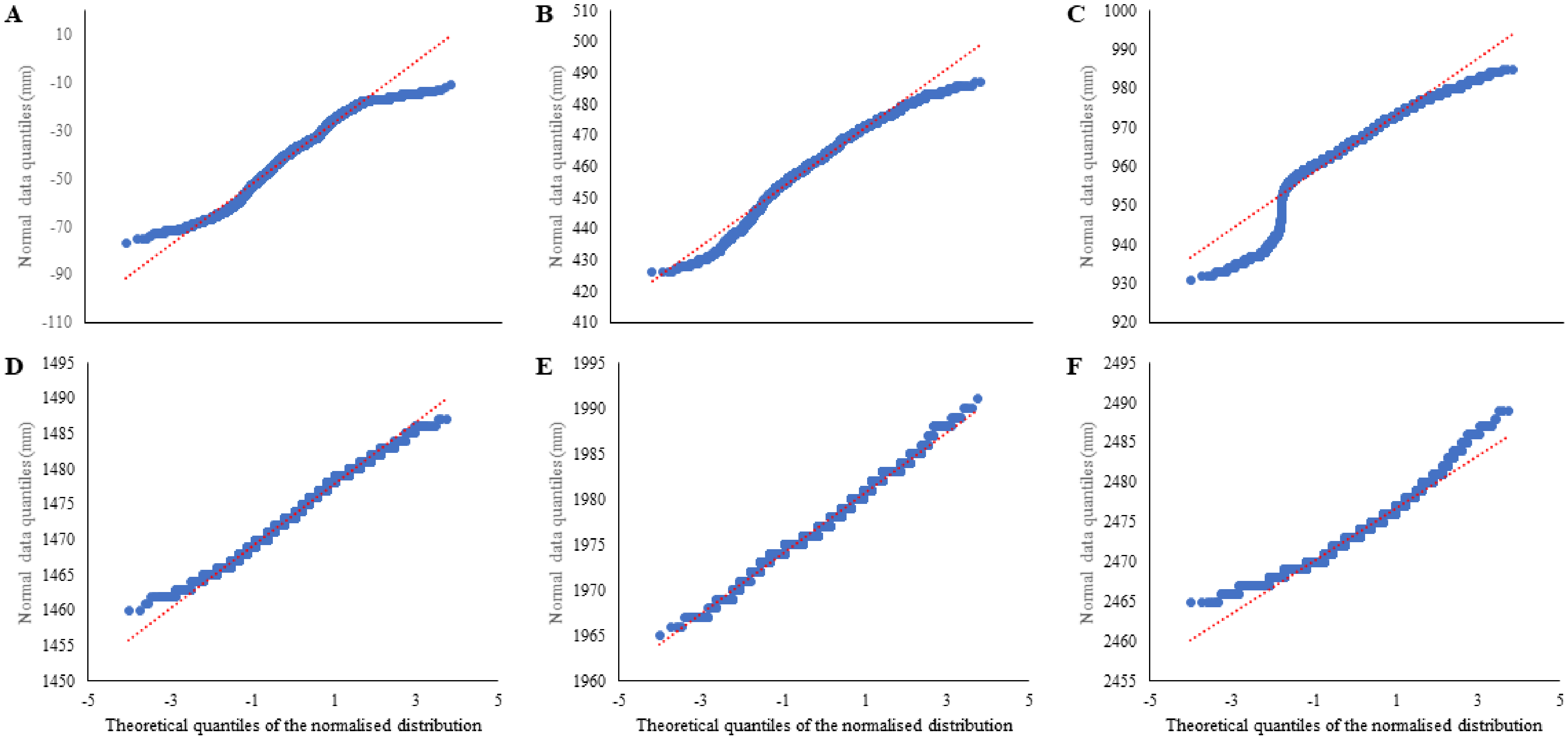

Figure A2.1. Q-Q plots of the marker data collected with the Kinect on Bar-Positions 1 to 6 (Figures A to F). The blue lines represent the data collected with the Kinect, the red line represents the theoretical data of perfectly normal distributed data.

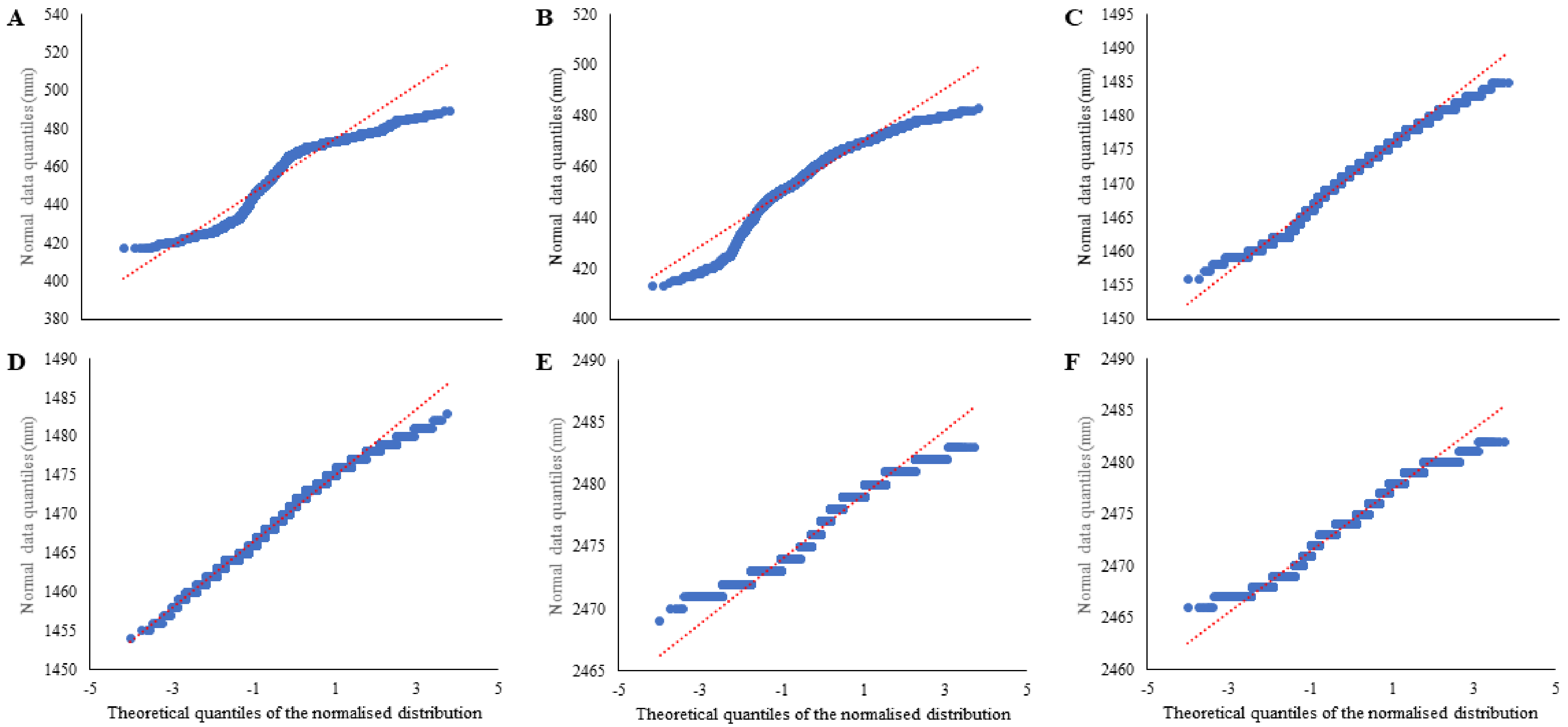

Figure A2.2. Q-Q plots of the marker data collected with the Kinect on Bar-Positions 7 to 12 (Figures A to F). The blue lines represent the data collected with the Kinect, the red line represents the theoretical data of perfectly normal distributed data.

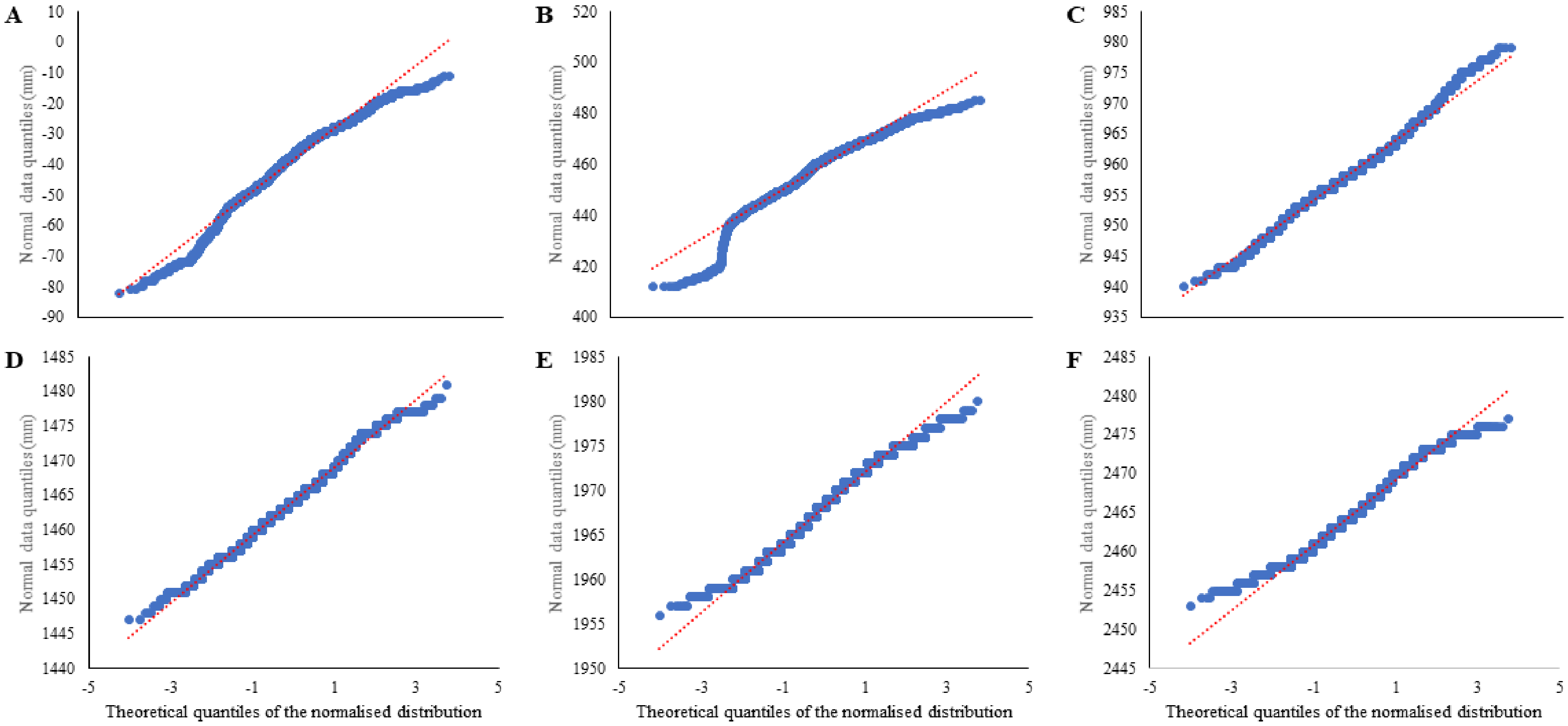

Figure A2.3. Q-Q plots of the marker data collected with the Kinect on Bar-Positions 13 to 18 (Figures A to F). The blue lines represent the data collected with the Kinect, the red line represents the theoretical data of perfectly normal distributed data.

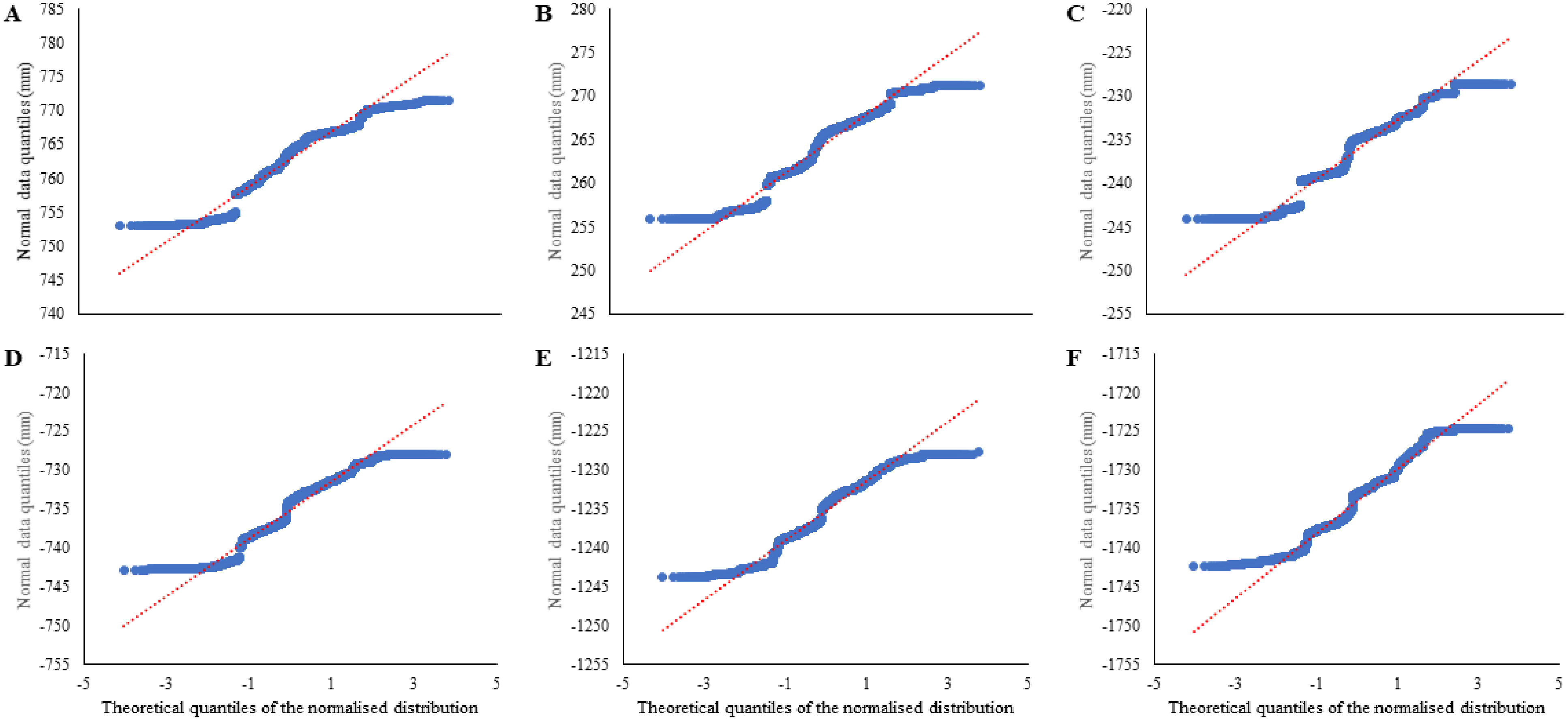

Figure A2.4. Q-Q plots of the marker data collected with the Vicon on Bar-Positions 1 to 6 (Figures A to F). The blue lines represent the data collected with Vicon, the red line represents the theoretical data of perfectly normal distributed data.

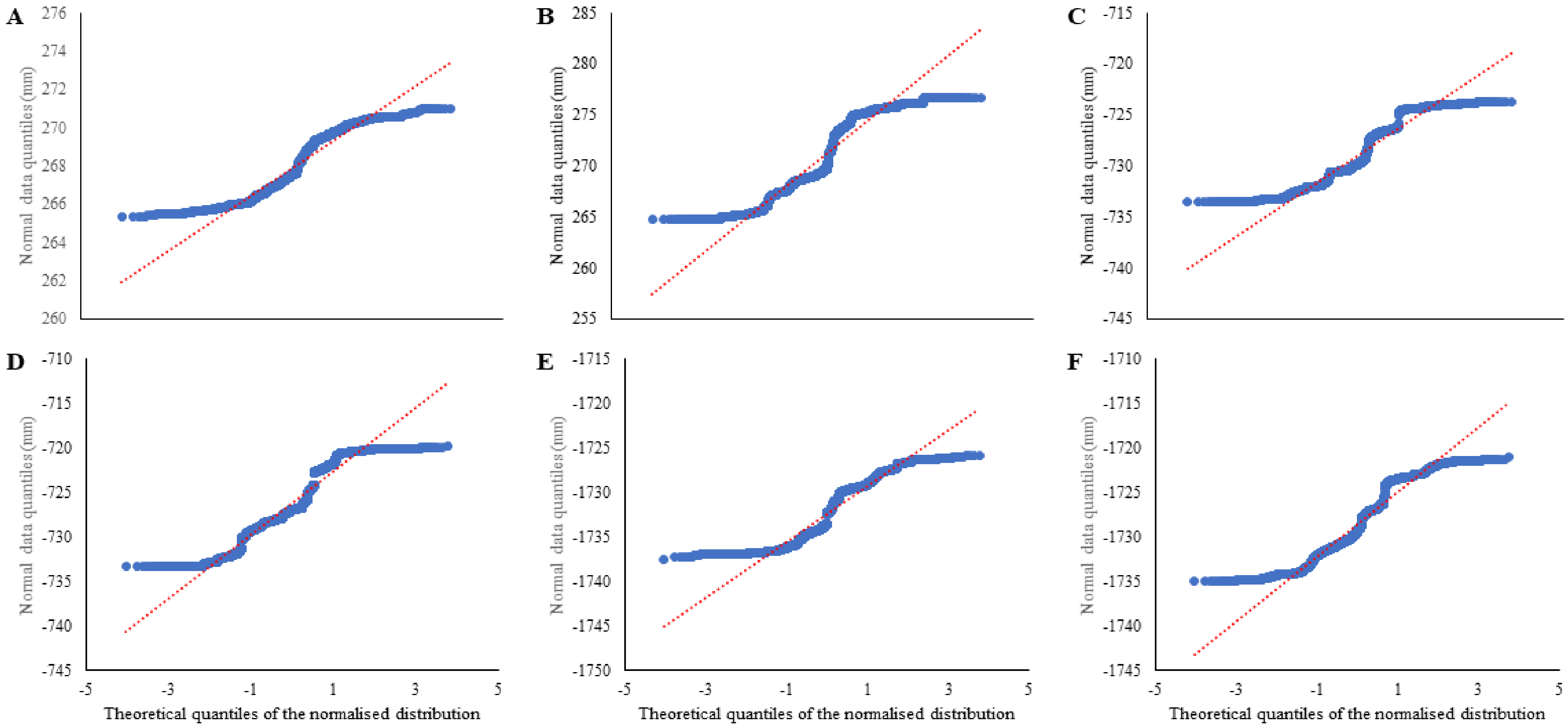

Figure A2.5. Q-Q plots of the marker data collected with the Vicon on Bar-Positions 7 to 12 (Figures A to F). The blue lines represent the data collected with Vicon, the red line represents the theoretical data of perfectly normal distributed data.

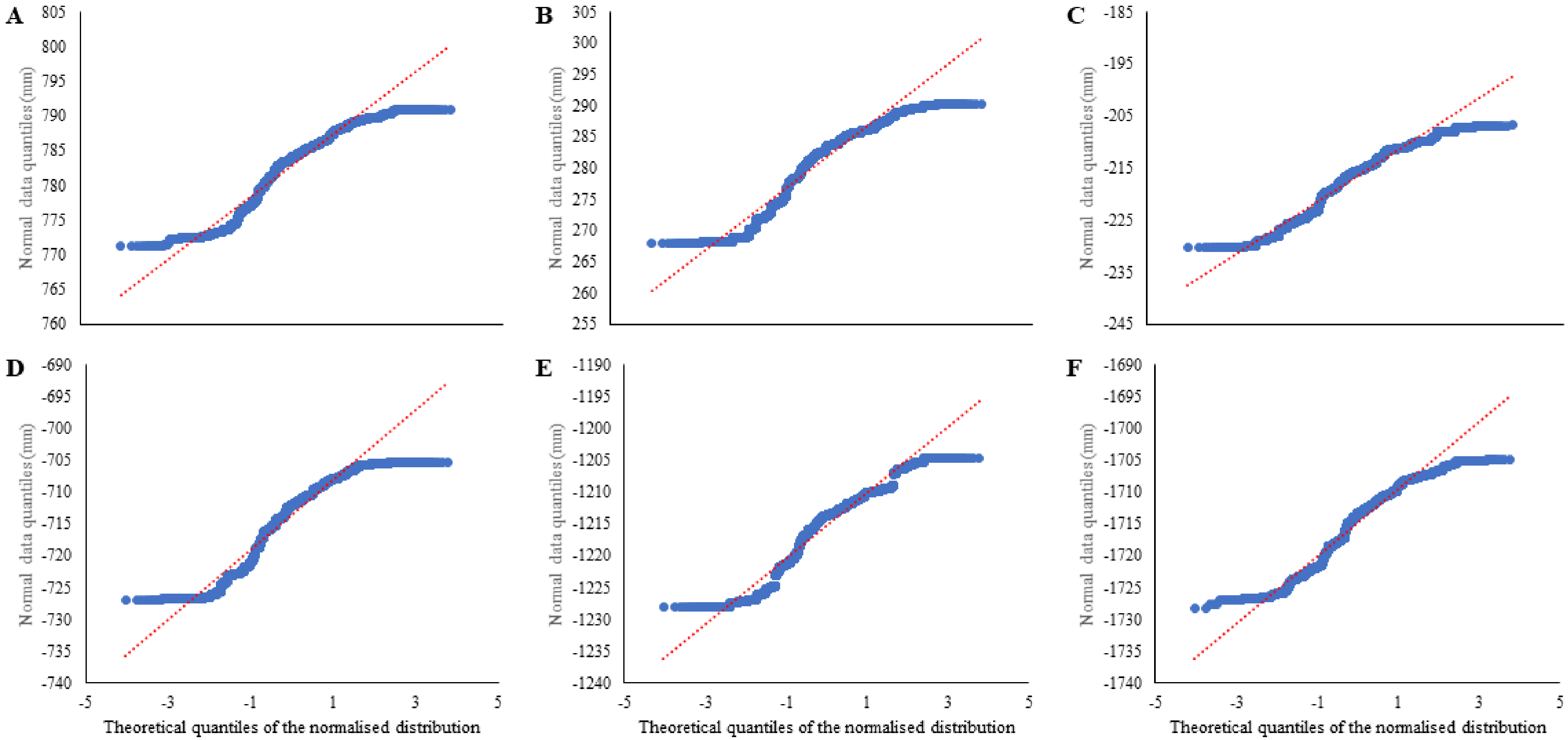

Figure A2.6. Q-Q plots of the marker data collected with the Vicon on Bar-Positions 13 to 18 (Figures A to F). The blue lines represent the data collected with Vicon, the red line represents the theoretical data of perfectly normal distributed data.

Appendix 3. Supplementary Tables of Chapter 4.

Table A3.1. The mean marker position (± SD, in millimetre) in the posterior-anterior axis per Bar-Position for both markers of the AMAT and Vicon.

	AMAT		Vicon	
Bar-Position	Marker 1	Marker 2	Marker 1	Marker 2
1	-35.52 ± 13.96	-43.30 ± 10.93	-760.71 ± 3.87	-765.00 ± 3.59
2	463.53 ± 10.24	461.49 ± 9.34	-262.11 ± 3.12	-266.07 ± 2.86
3	967.44 ± 9.24	965.06 ± 5.47	238.42 ± 2.99	234.29 ± 2.85
4	1473.12 ± 4.60	1473.95 ± 4.19	737.30 ± 3.19	732.96 ± 2.97
5	1976.89 ± 3.36	1977.77 ± 3.27	1237.48 ± 3.39	1233.14 ± 2.98
6	2474.08 ± 3.25	2472.70 ± 3.30	1736.03 ± 3.84	1731.87 ± 3.43
7	458.64 ± 17.74	460.79 ± 12.53	-266.61 ± 0.63	-269.21 ± 0.96
8	457.78 ± 13.18	461.20 ± 7.81	-267.85 ± 1.51	-273.74 ± 2.17
9	1471.64 ± 4.76	1471.30 ± 4.88	731.37 ± 1.08	726.76 ± 2.05
10	1471.55 ± 4.19	1470.00 ± 4.30	729.08 ± 1.95	723.42 ± 2.70
11	2477.76 ± 2.35	2475.44 ± 2.43	1735.41 ± 1.06	1729.48 ± 1.54
12	2475.83 ± 2.64	2472.89 ± 2.55	1731.68 ± 1.61	1725.37 ± 2.36
13	-38.67 ± 12.30	-39.84 ± 9.14	-781.31 ± 4.19	-784.27 ± 4.82
14	458.60 ± 11.57	460.30 ± 8.28	-279.44 ± 5.00	-283.00 ± 5.12
15	958.95 ± 4.26	958.46 ± 5.77	219.00 ± 4.43	214.41 ± 4.98
16	1463.04 ± 5.16	1465.23 ± 4.46	715.90 ± 5.03	711.33 ± 5.30
17	1967.78 ± 3.38	1968.47 ± 4.51	1216.94 ± 4.69	1213.43 ± 5.39
18	2463.72 ± 3.97	2466.25 ± 4.04	1716.67 ± 4.94	1713.02 ± 5.24

Table A3.2. The distance (± SD, in millimetre) between adjacent marker positions in the posterior-anterior axis as measured by AMAT and Vicon for both markers.

	AMAT		Vicon	
Bar-Position	Marker 1	Marker 2	Marker 1	Marker 2
2-1	499.23 ± 19.00	504.85 ± 12.09	498.61 ± 1.11	498.93 ± 1.15
3-2	503.96 ± 11.73	503.59 ± 10.61	500.53 ± 0.74	500.36 ± 0.79
4-3	505.69 ± 8.79	508.85 ± 4.80	498.88 ± 0.87	498.67 ± 1.01
5-4	503.77 ± 3.50	503.75 ± 2.41	500.19 ± 0.90	500.18 ± 1.01
6-5	497.15 ± 2.24	495.06 ± 2.04	498.49 ± 0.98	498.67 ± 1.33
9-7	1012.39 ± 18.14	1010.34 ± 14.52	997.98 ± 0.94	995.97 ± 1.52
10-8	1013.16 ± 13.53	1008.66 ± 9.14	996.93 ± 1.46	997.16 ± 1.28
11-9	1005.99 ± 4.30	1004.16 ± 4.27	1004.04 ± 1.35	1002.72 ± 1.89
12-10	1004.12 ± 3.56	1002.83 ± 3.70	1002.60 ± 1.70	1001.95 ± 1.22
14-13	497.36 ± 11.81	500.95 ± 9.46	501.87 ± 1.52	501.27 ± 1.37
15-14	500.23 ± 10.50	497.48 ± 8.13	498.44 ± 1.16	497.41 ± 1.29
16-15	504.18 ± 4.52	506.79 ± 4.34	496.90 ± 1.21	496.92 ± 1.20
17-16	504.71 ± 3.26	503.31 ± 3.52	501.04 ± 1.33	502.10 ± 1.33
18-17	495.91 ± 2.25	497.69 ± 2.89	499.73 ± 1.58	499.59 ± 1.64

Appendix 4. Q-Q plots of the data of Chapter 5

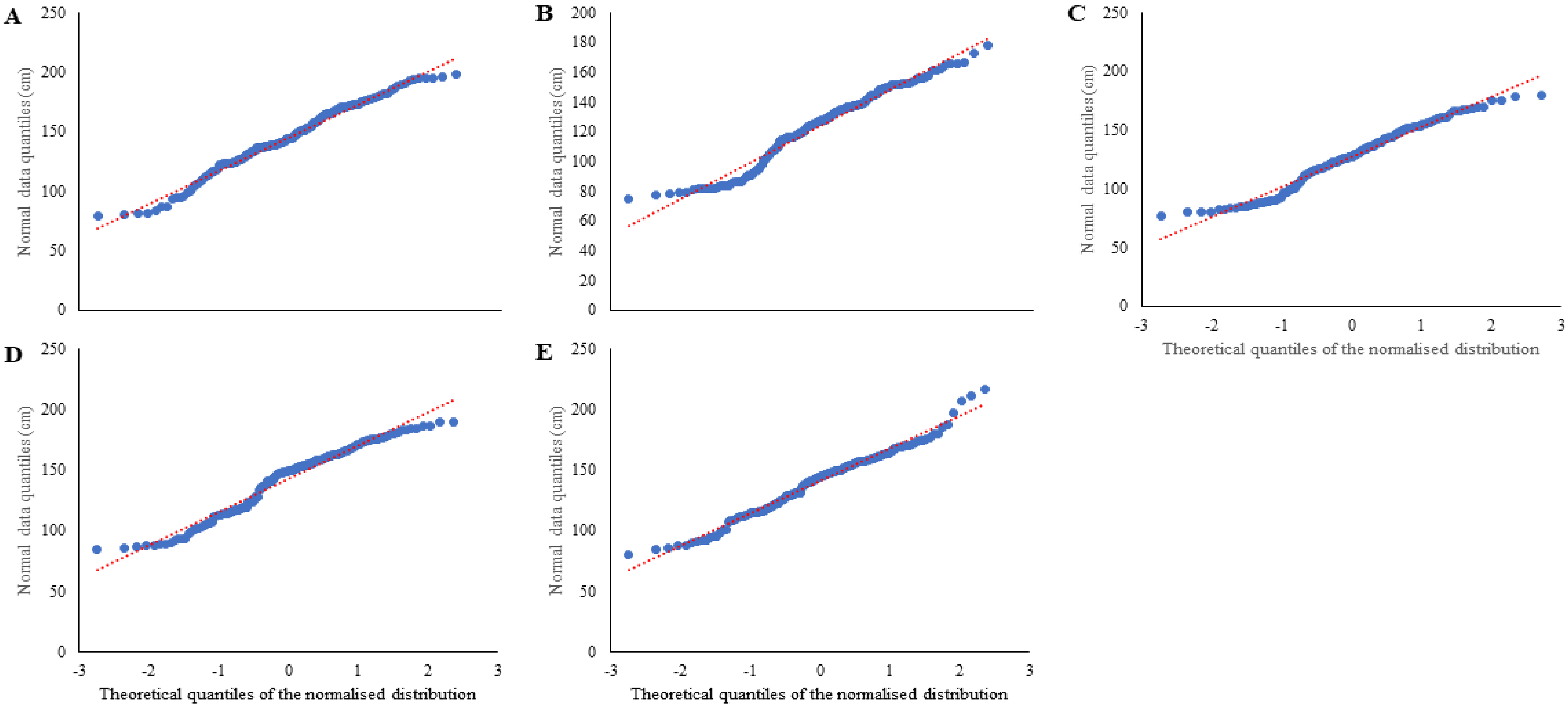

Figure A4.1. Q-Q plots of the jump data collected manually. Figure A4.1A: standing broad jump. Figure A4.1B: left leg hop. Figure A4.1C: right leg hop. Figure A4.1D: left to right stride. A4.1E: right to left stride. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

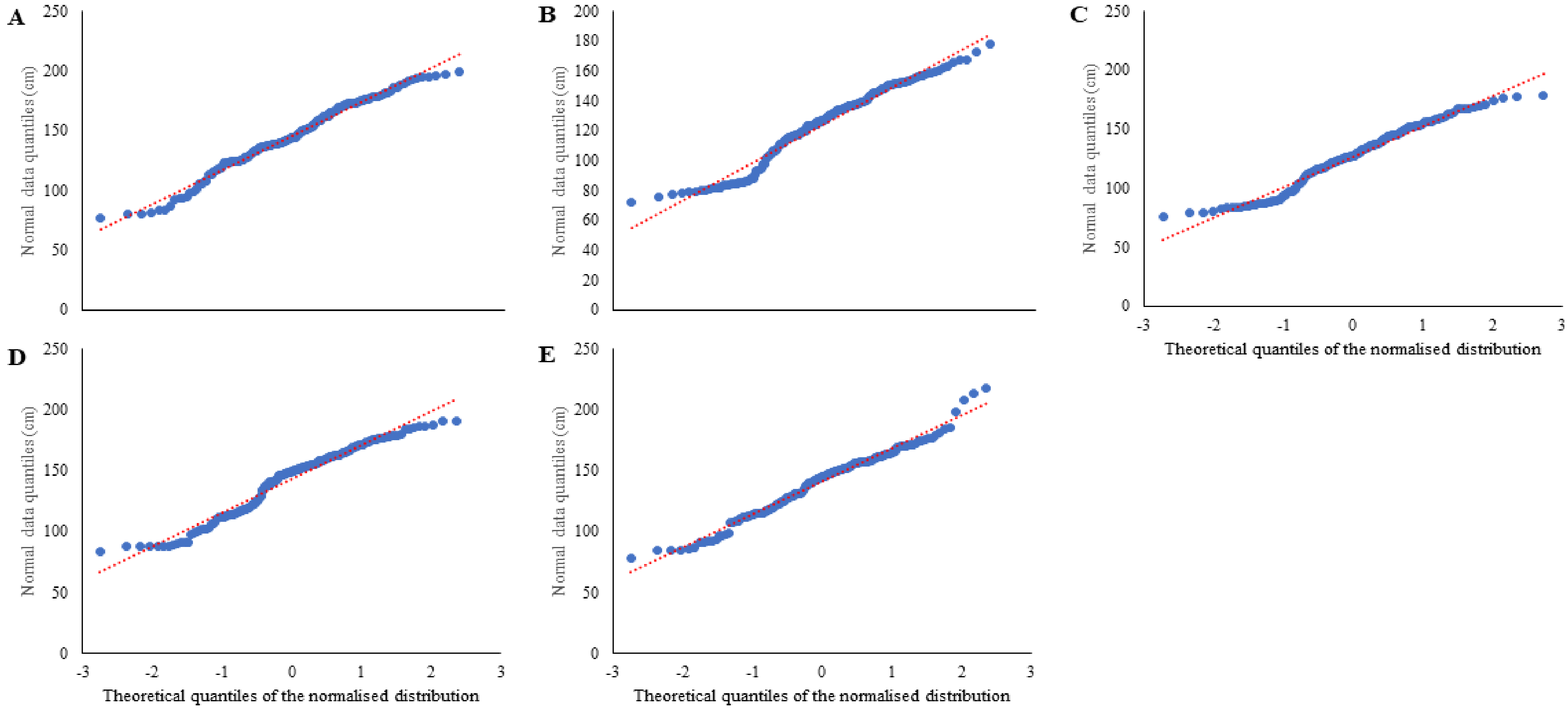

Figure A4.2. Q-Q plots of the jump data collected with the AMAT. Figure A4.2A: standing broad jump. Figure A4.2B: left leg hop. Figure A4.2C: right leg hop. Figure A4.2D: left to right stride. A4.2E: right to left stride. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

Appendix 5. Figures of centre of mass position per jump as measured with Vicon and AMAT

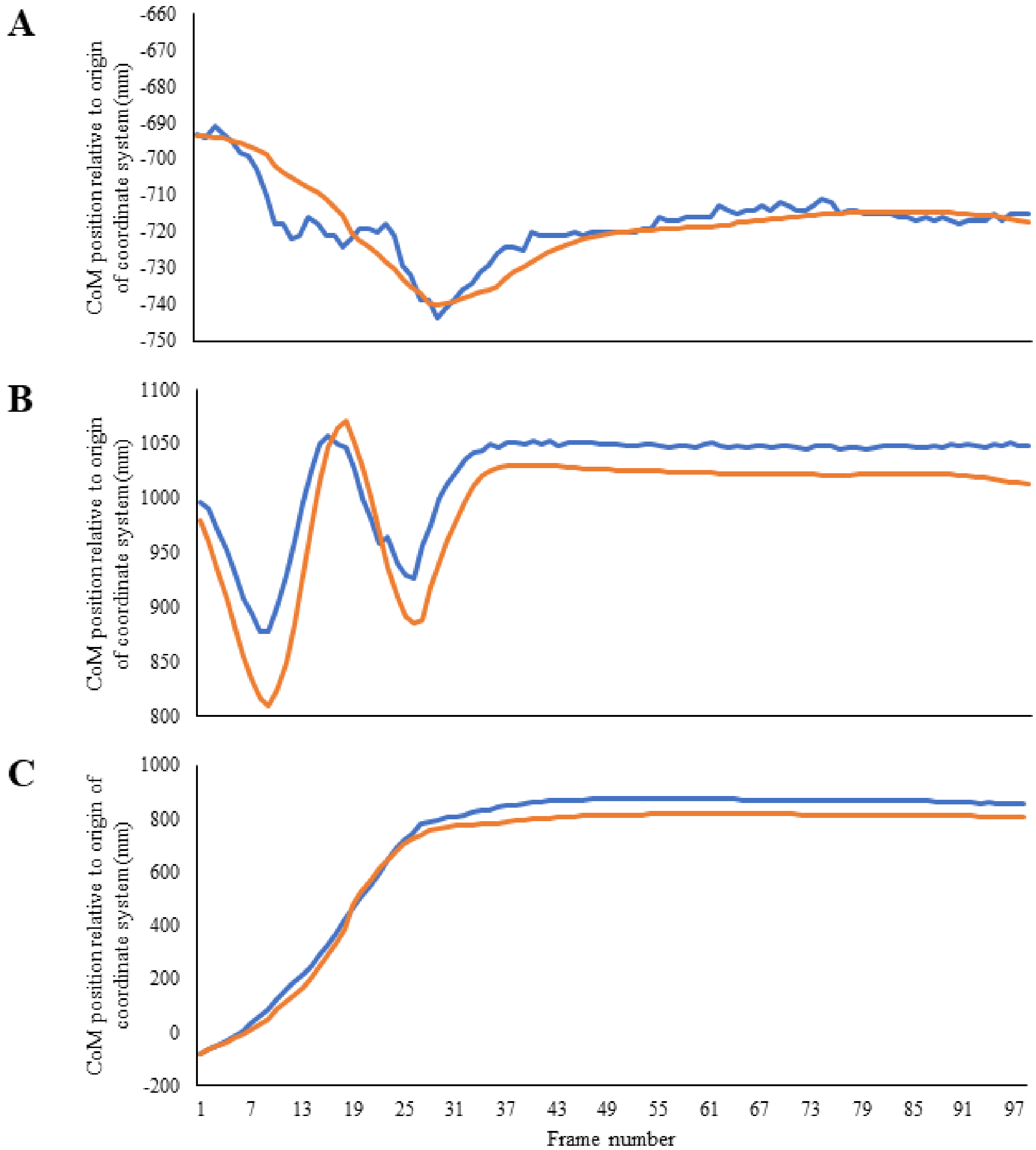

Figure A5.1. Centre of Mass (CoM) position over time (orange: Vicon; blue: AMAT) of participant one during the first standing broad jump. Figure A5.1A: Medio-lateral axis. Figure A5.1B: Superior-inferior axis. Figure A5.1C: Posterior-anterior axis.

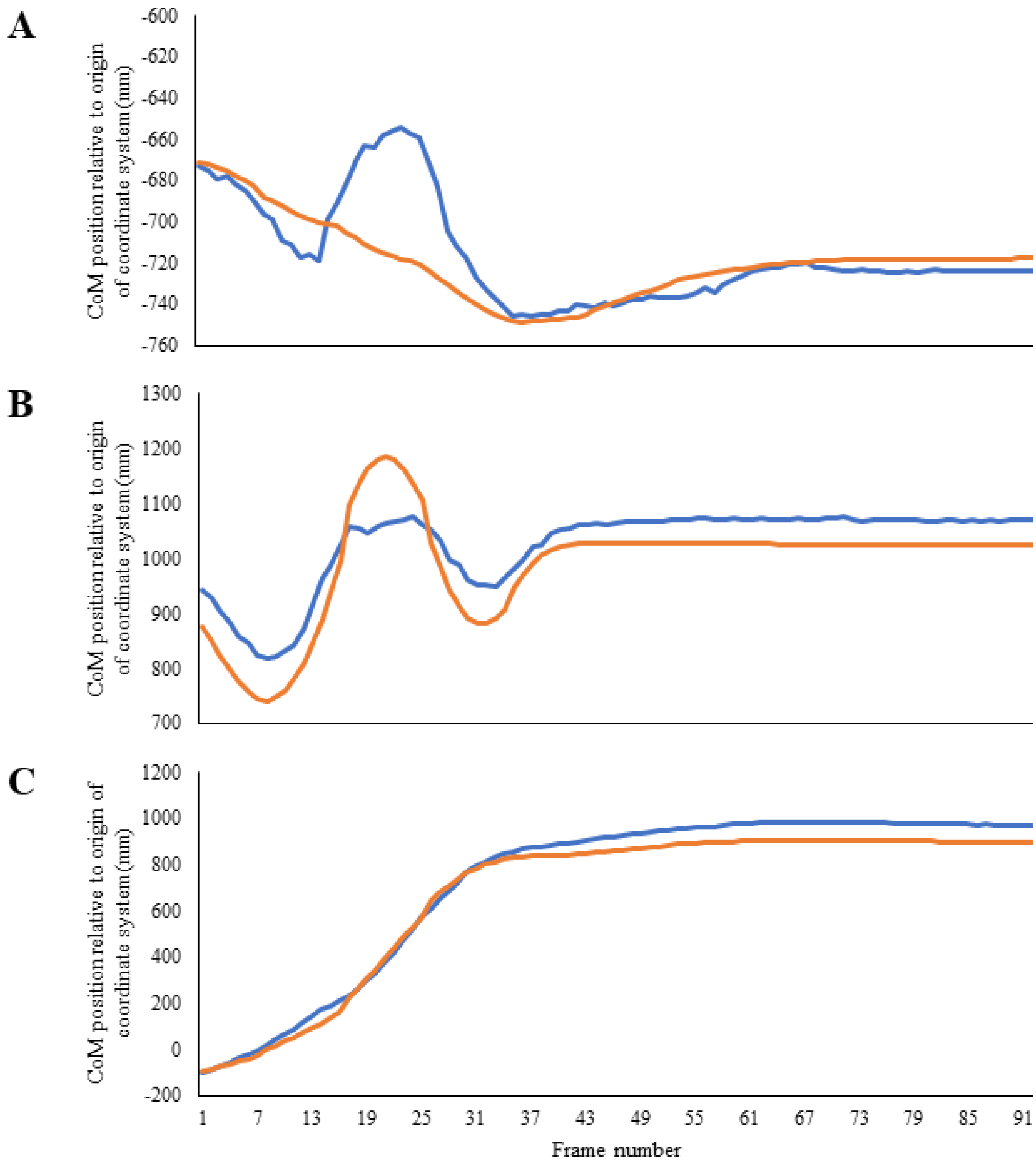

Figure A5.2. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the second standing broad jump. Figure A5.2A: Medio-lateral axis. Figure A5.2B: Superior-inferior axis. Figure A5.2C: Posterior-anterior axis.

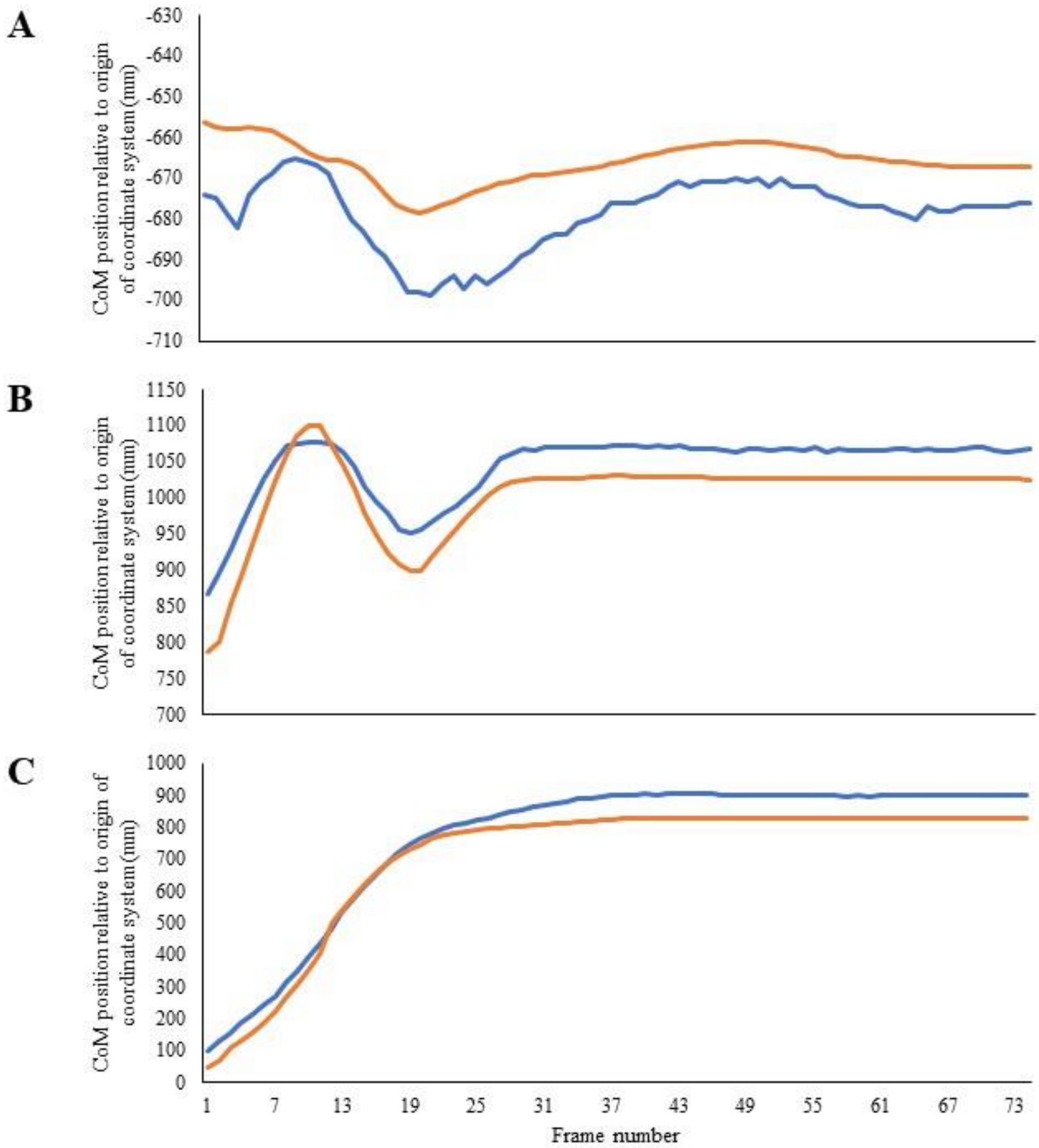

Figure A5.3. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the third standing broad jump. Figure A5.3A: Medio-lateral axis. Figure A5.3B: Superior-inferior axis. Figure A5.3C: Posterior-anterior axis.

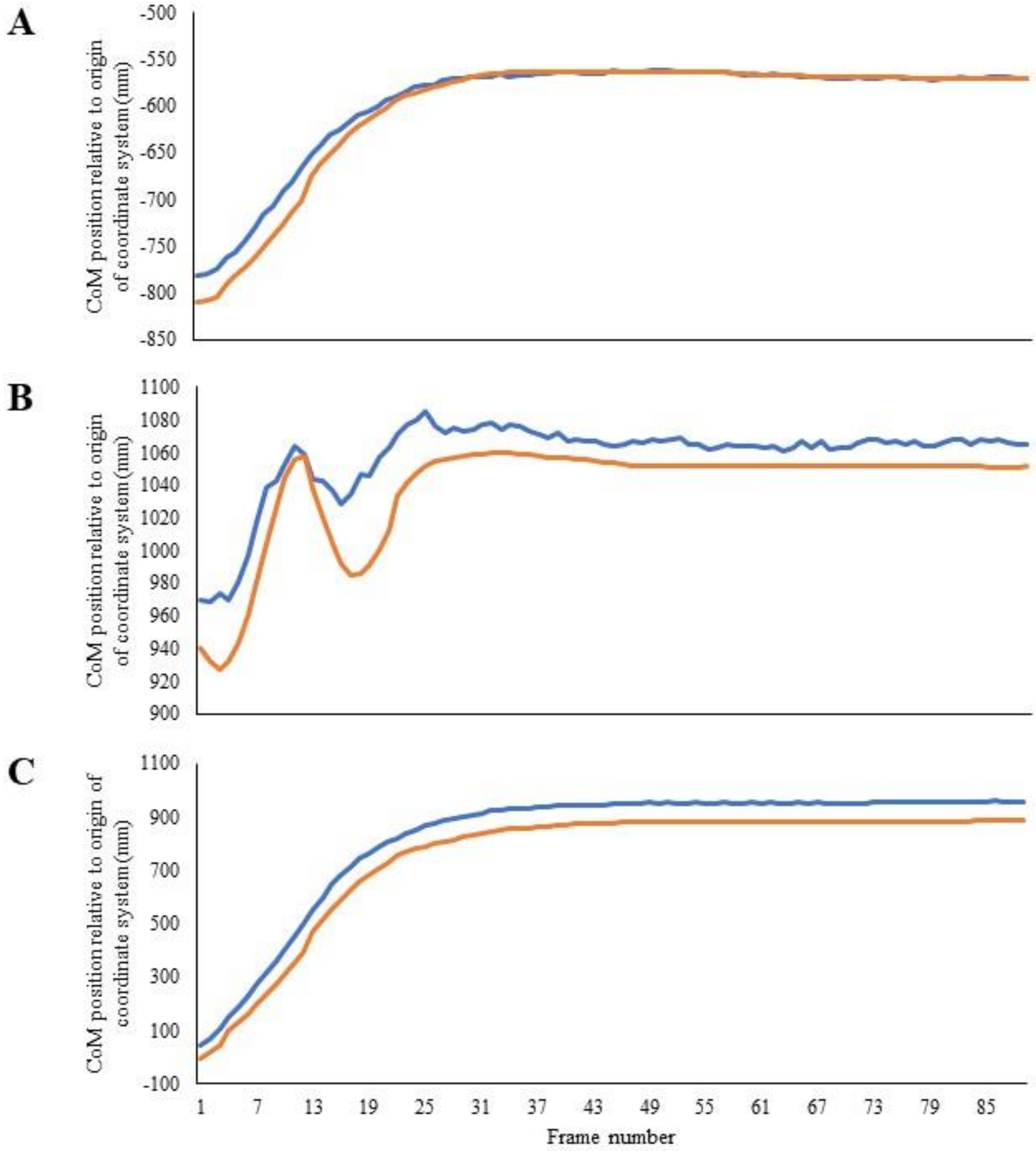

Figure A5.4. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the first left to right stride. Figure A5.4A: Medio-lateral axis. Figure A5.4B: Superior-inferior axis. Figure A5.4C: Posterior-anterior axis.

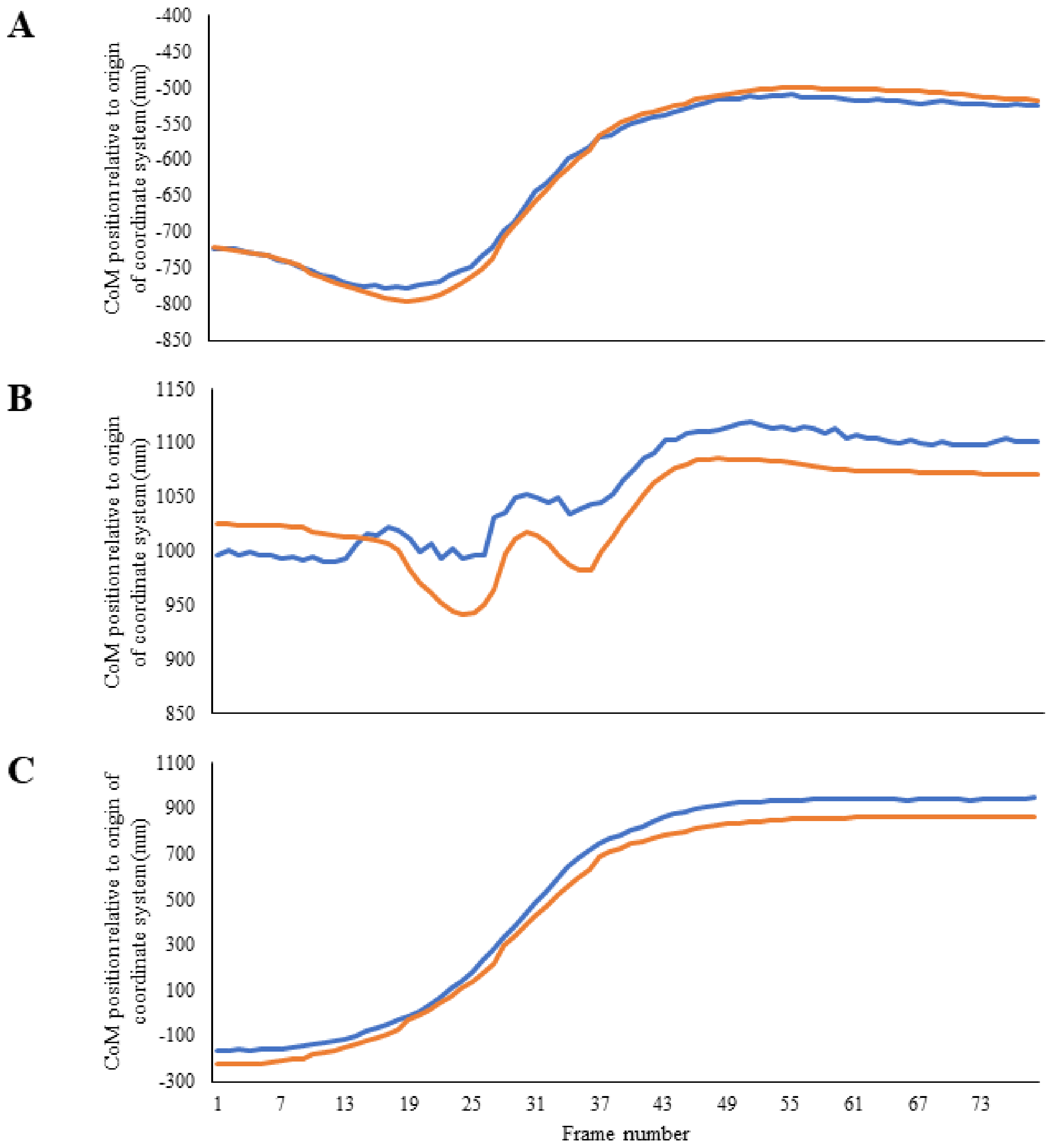

Figure A5.5. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the second left to right stride. Figure A5.5A: Medio-lateral axis. Figure A5.5B: Superior-inferior axis. Figure A5.5C: Posterior-anterior axis.

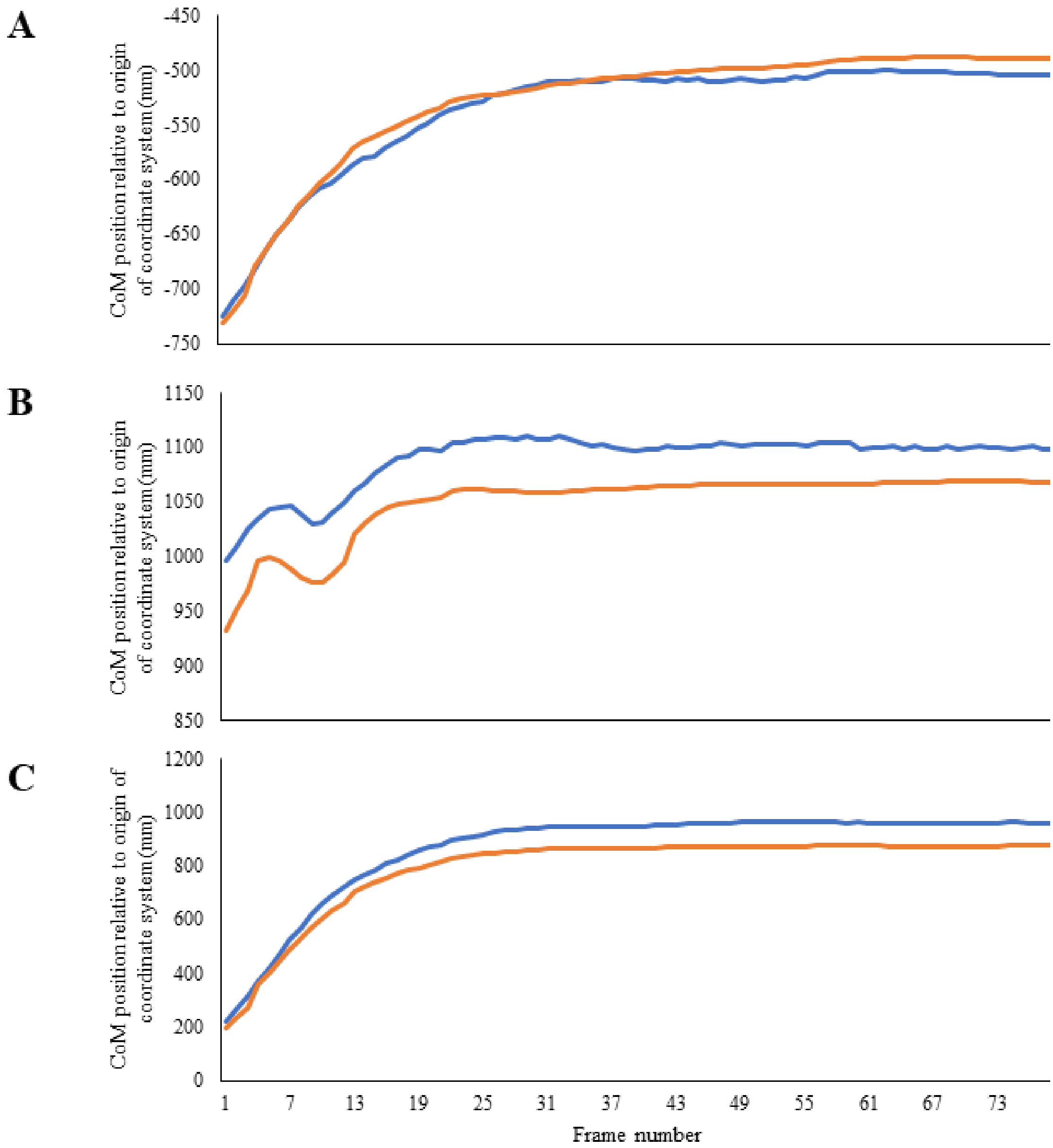

Figure A5.6. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the third left to right stride. Figure A5.6A: Medio-lateral axis. Figure A5.6B: Superior-inferior axis. Figure A5.6C: Posterior-anterior axis.

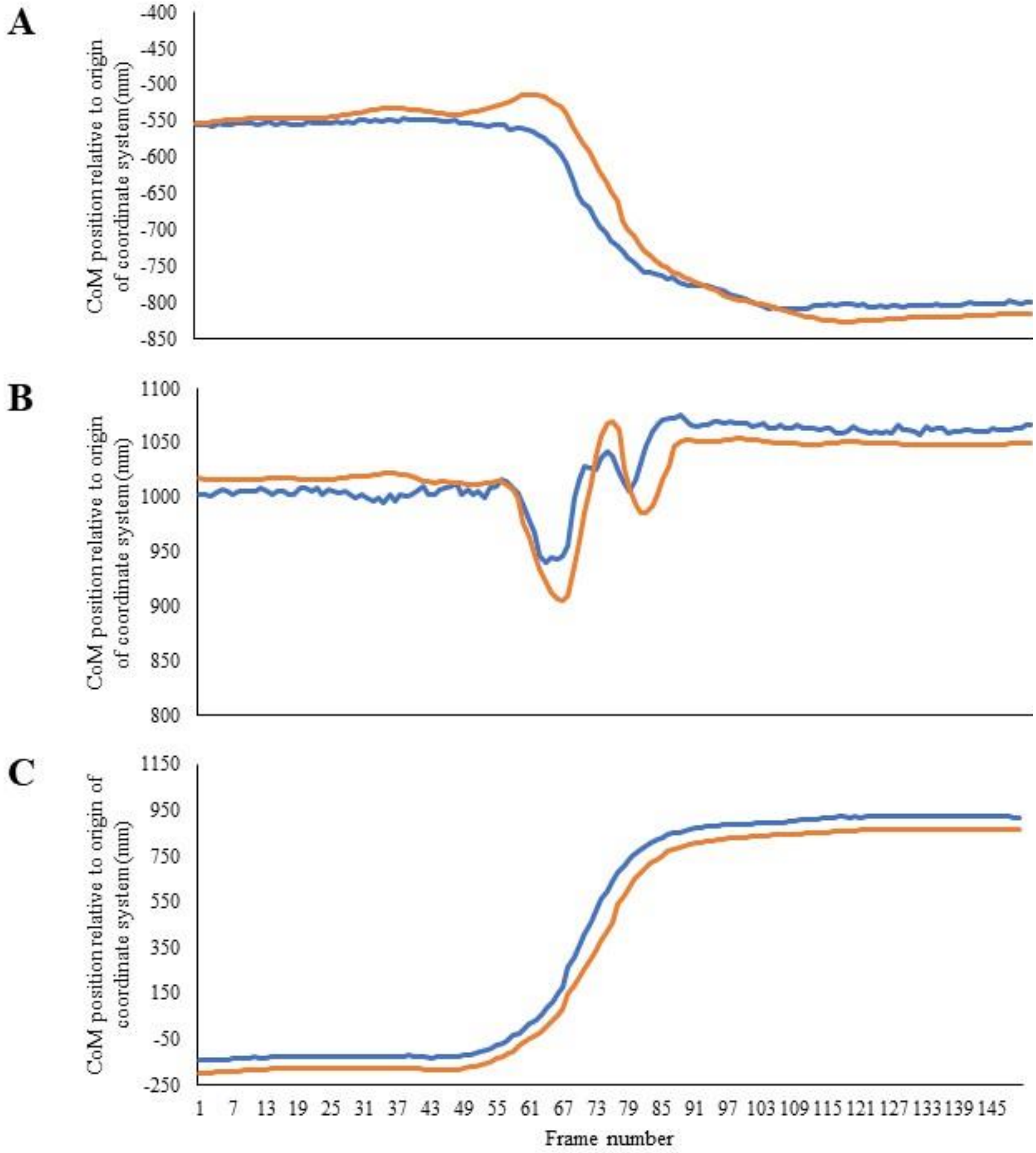

Figure A5.7. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the first right to left stride. Figure A5.7A: Medio-lateral axis. Figure A5.7B: Superior-inferior axis. Figure A5.7C: Posterior-anterior axis.

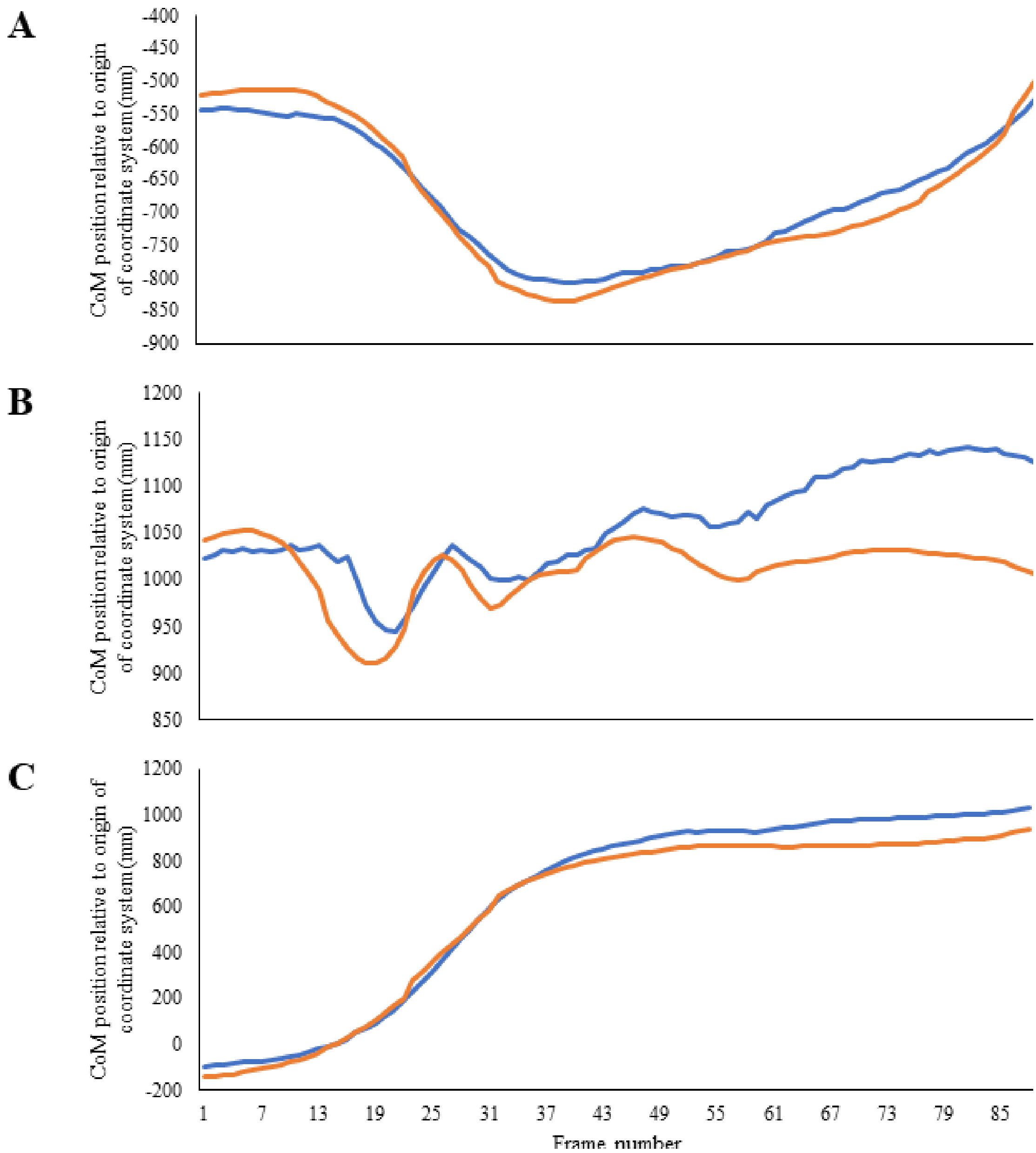

Figure A5.8. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the second right to left stride. Figure A5.8A: Medio-lateral axis. Figure A5.8B: Superior-inferior axis. Figure A5.8C: Posterior-anterior axis.

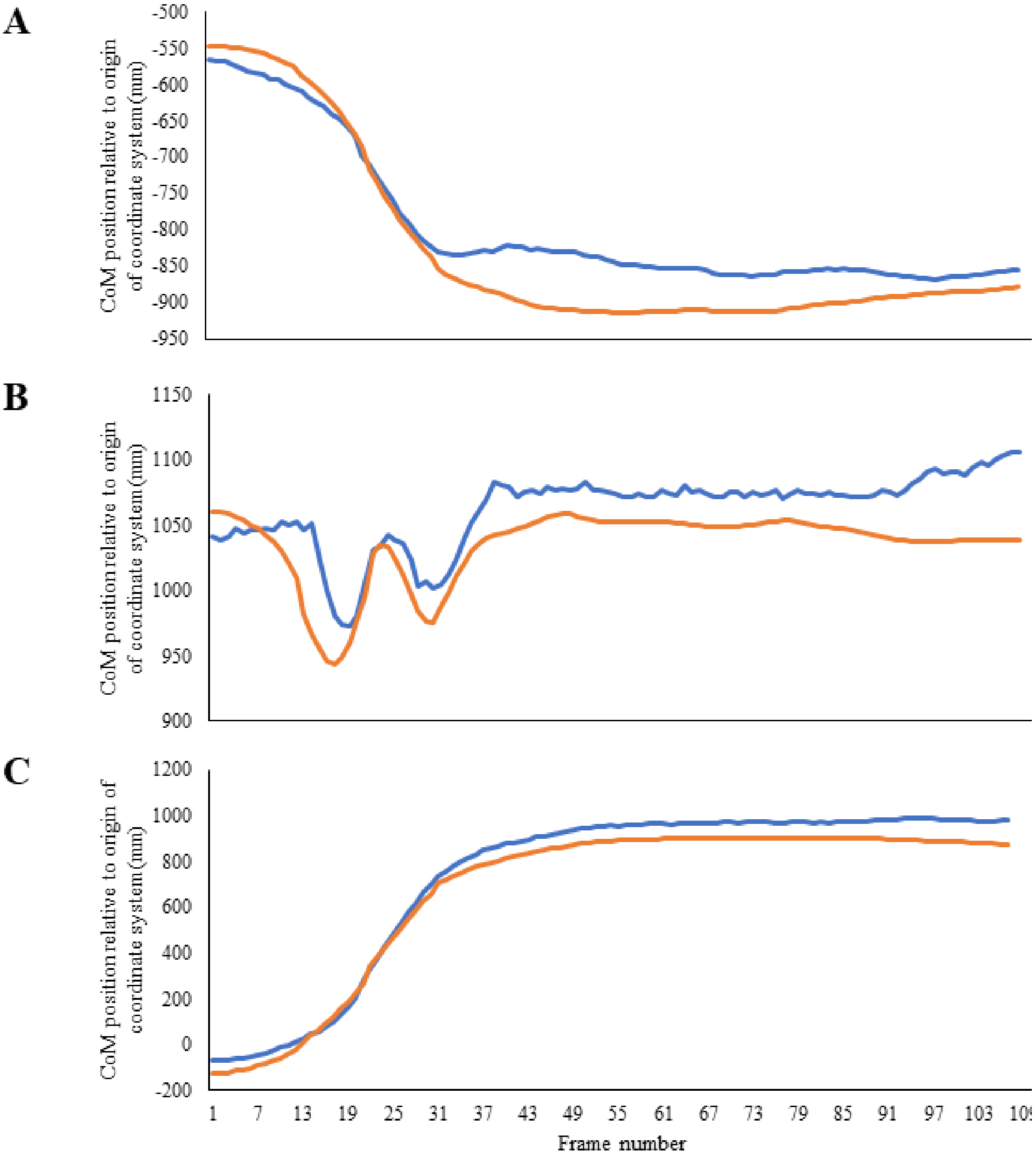

Figure A5.9. CoM position over time (orange: Vicon; blue: AMAT) of participant one during the third right to left stride. Figure A5.9A: Medio-lateral axis. Figure A5.9B: Superior-inferior axis. Figure A5.9C: Posterior-anterior axis.

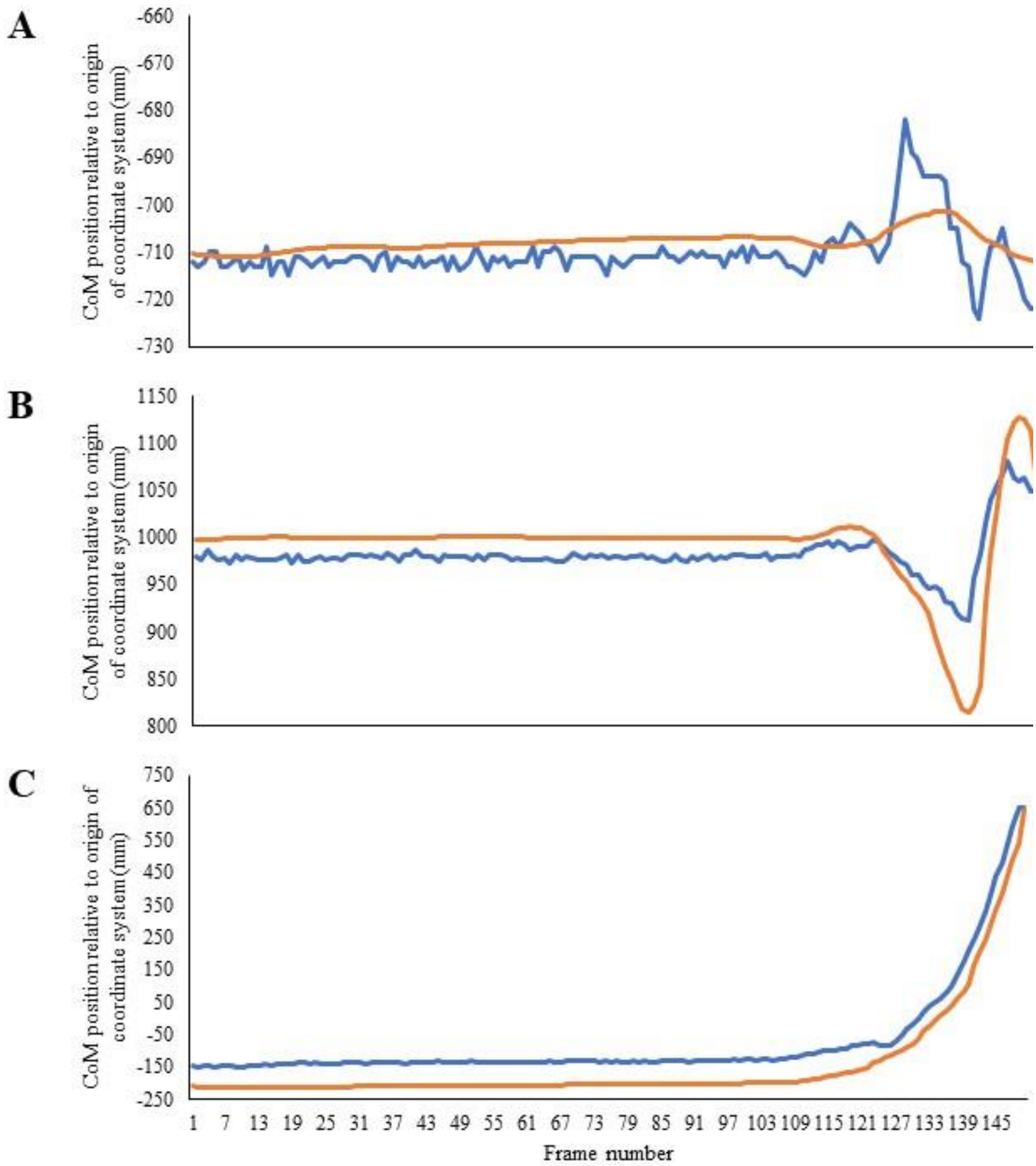

Figure A5.10. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the first standing broad jump. Figure A5.10A: Medio-lateral axis. Figure A5.10B: Superior-inferior axis. Figure A5.10C: Posterior-anterior axis.

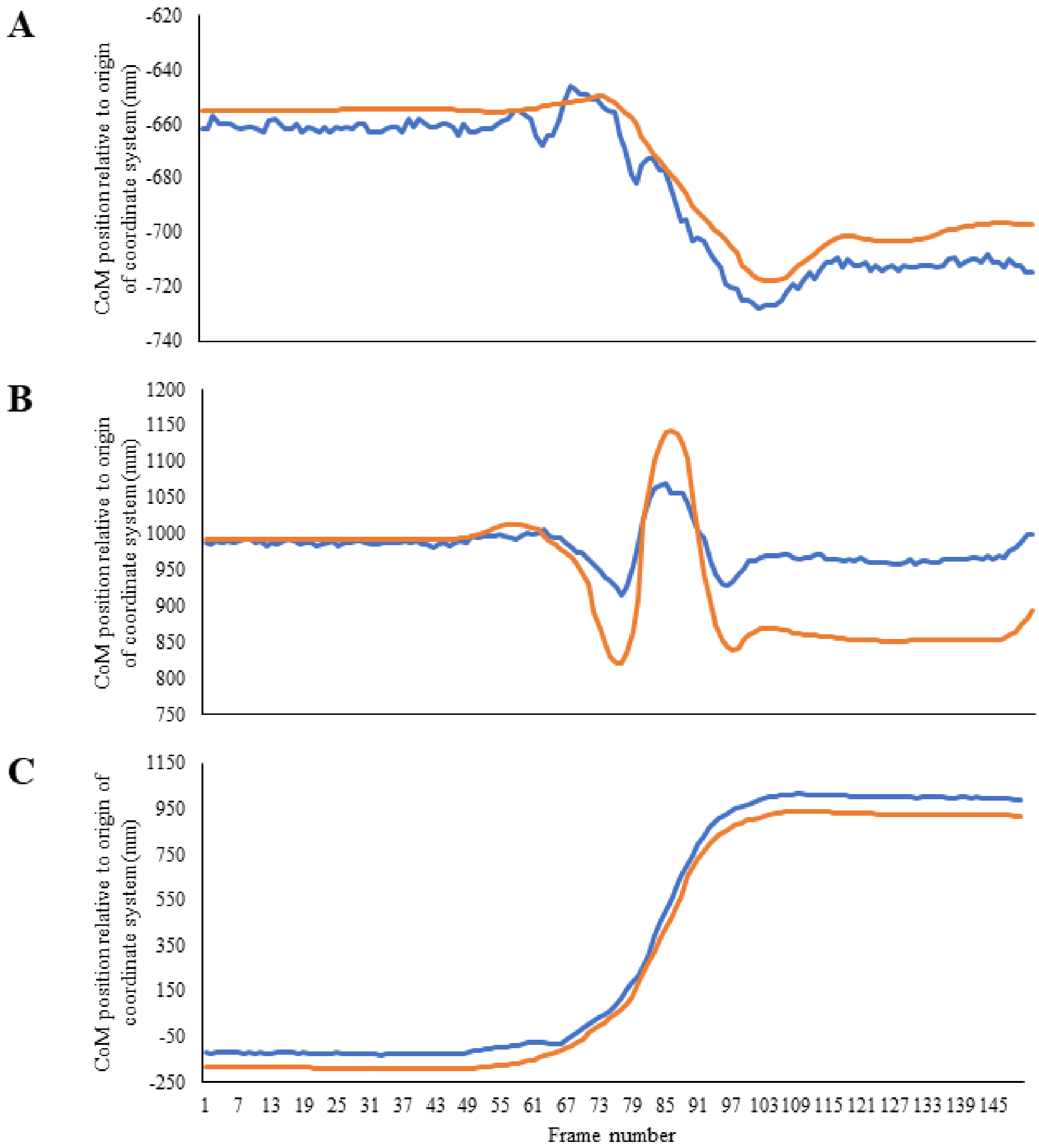

Figure A5.11. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the second standing broad jump. Figure A5.11A: Medio-lateral axis. Figure A5.11B: Superior-inferior axis. Figure A5.11C: Posterior-anterior axis.

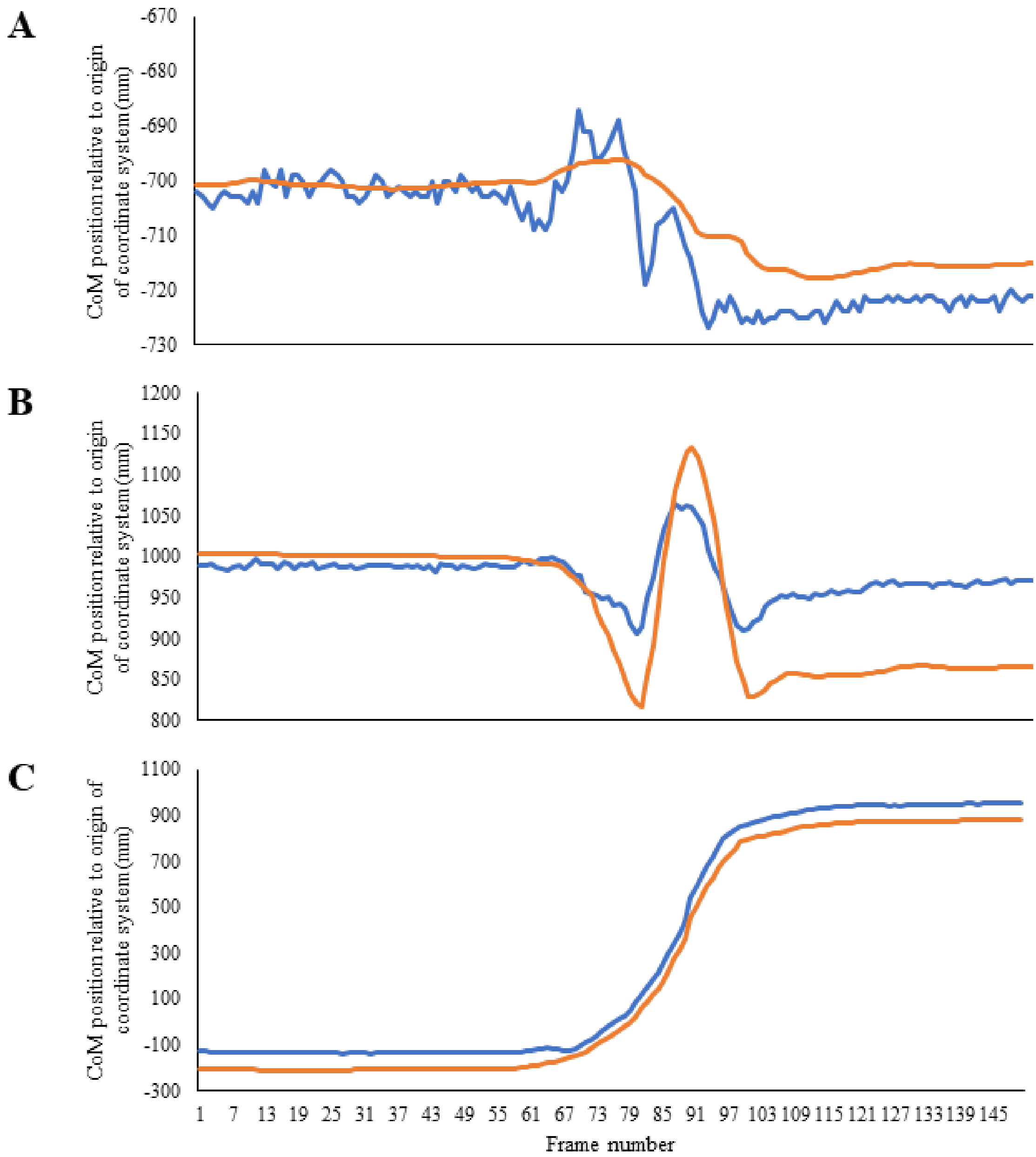

Figure A5.12. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the third standing broad jump. Figure A5.12A: Medio-lateral axis. Figure A5.12B: Superior-inferior axis. Figure A5.12C: Posterior-anterior axis.

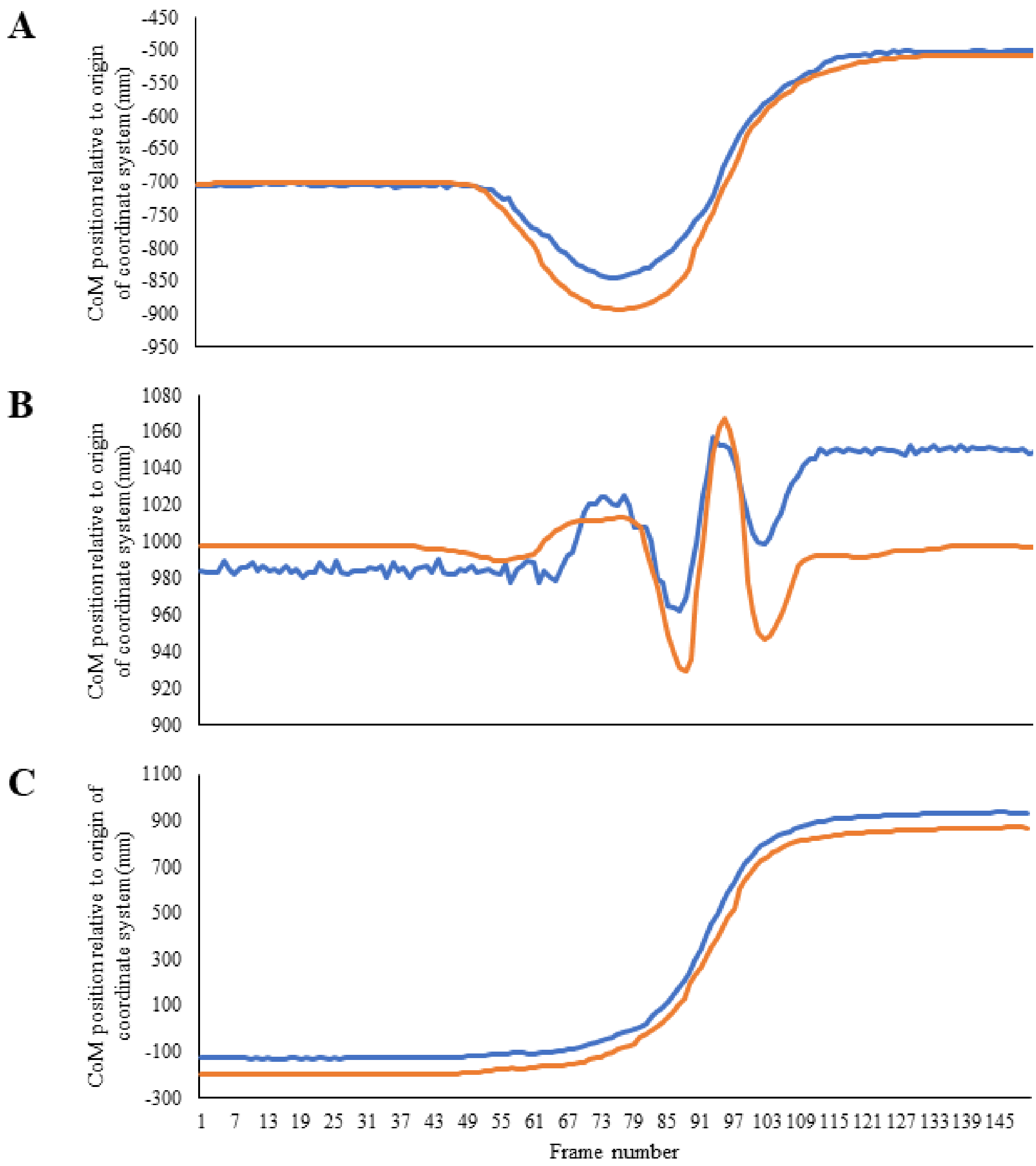

Figure A5.13. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the first left to right stride. Figure A5.13A: Medio-lateral axis. Figure A5.13B: Superior-inferior axis. Figure A5.13C: Posterior-anterior axis.

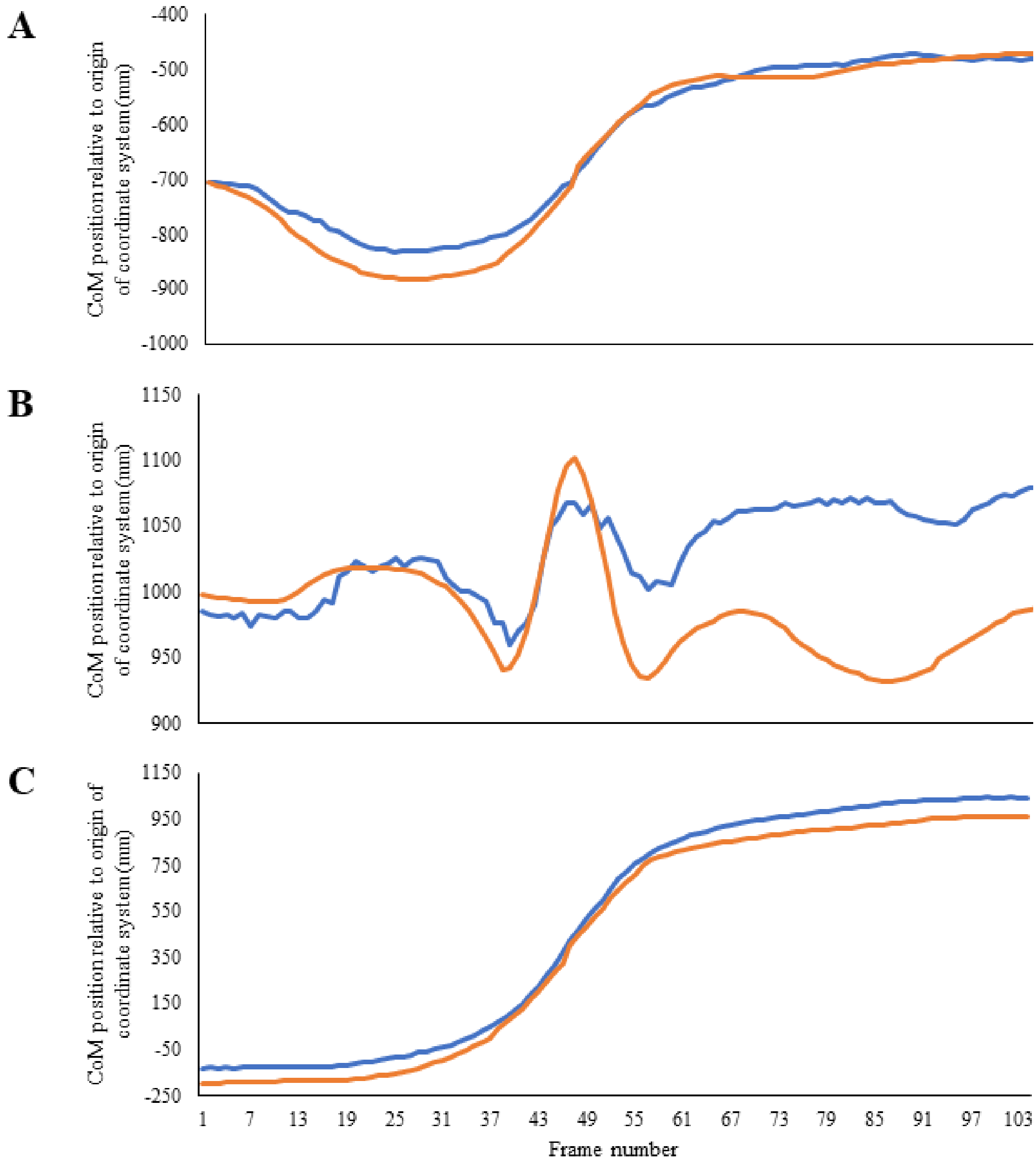

Figure A5.14. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the second left to right stride. Figure A5.14A: Medio-lateral axis. Figure A5.14B: Superior-inferior axis. Figure A5.14C: Posterior-anterior axis.

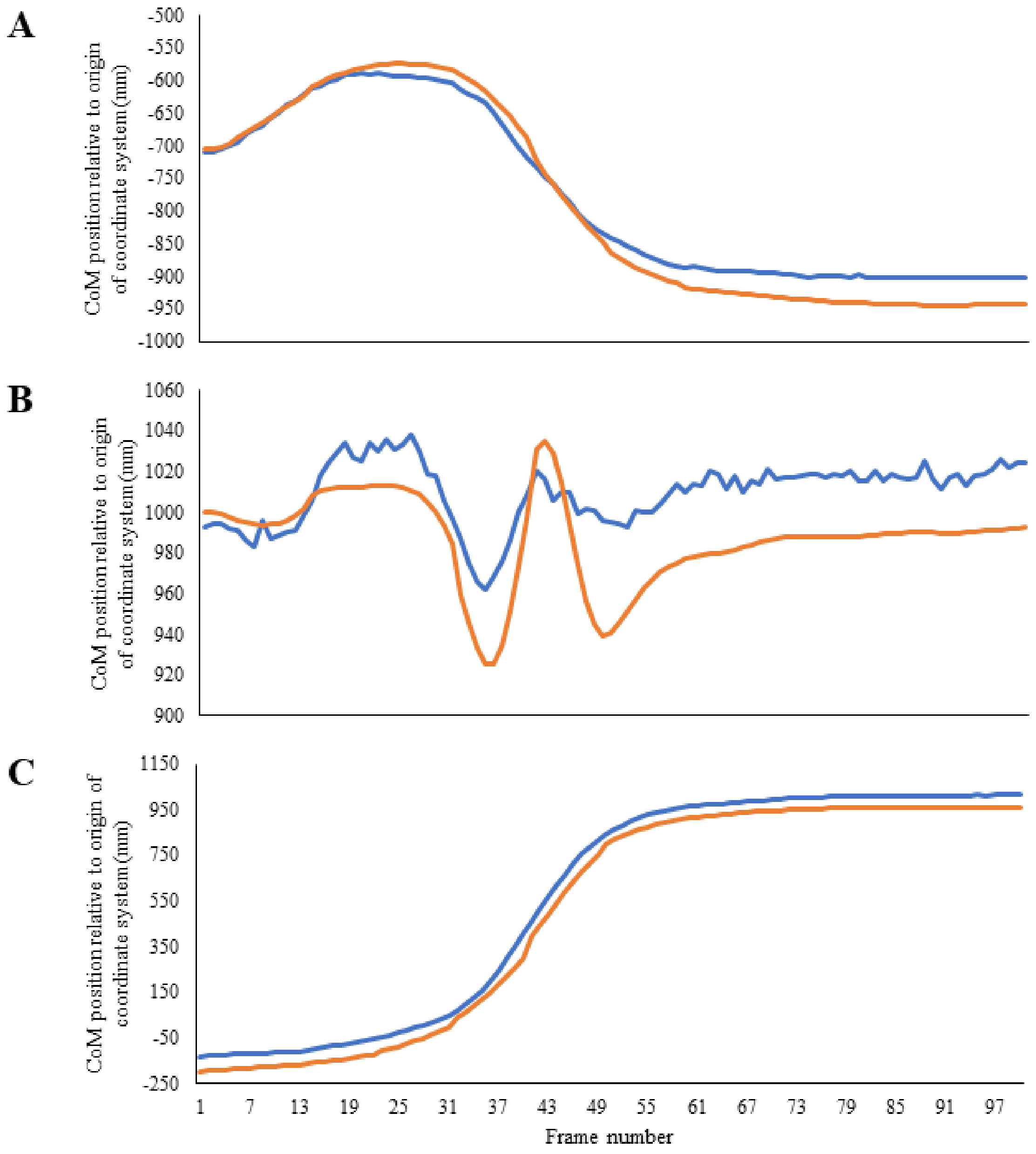

Figure A5.15. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the first right to left stride. Figure A5.15A: Medio-lateral axis. Figure A5.15B: Superior-inferior axis. Figure A5.15C: Posterior-anterior axis.

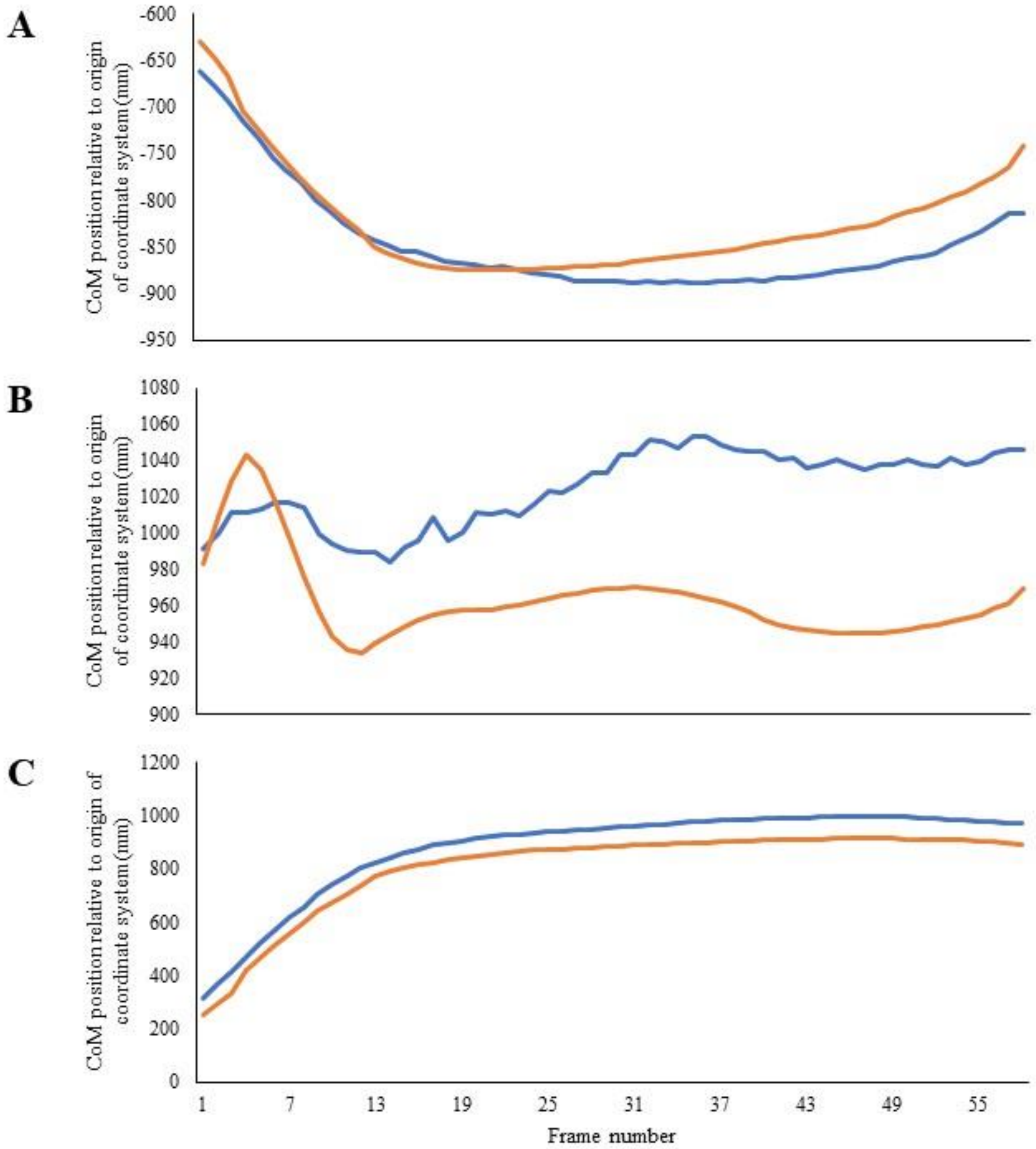

Figure A5.16. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the second right to left stride. Figure A5.16A: Medio-lateral axis. Figure A5.16B: Superior-inferior axis. Figure A5.16C: Posterior-anterior axis.

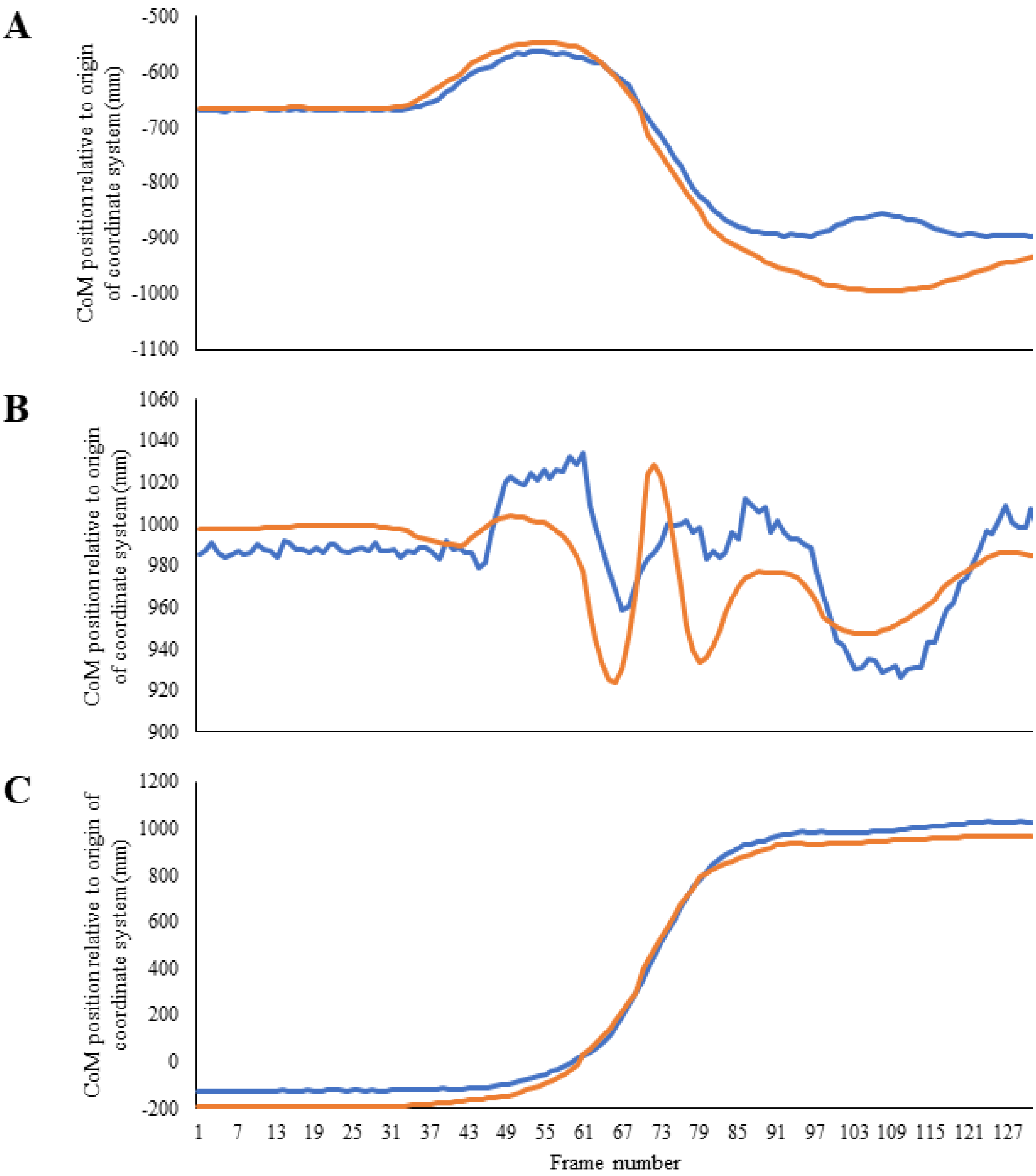

Figure A5.17. CoM position over time (orange: Vicon; blue: AMAT) of participant two during the second right to left stride. Figure A5.17A: Medio-lateral axis. Figure A5.17B: Superior-inferior axis. Figure A5.17C: Posterior-anterior axis.

Appendix 6. Data of the pilot study where centre of mass data was collected.

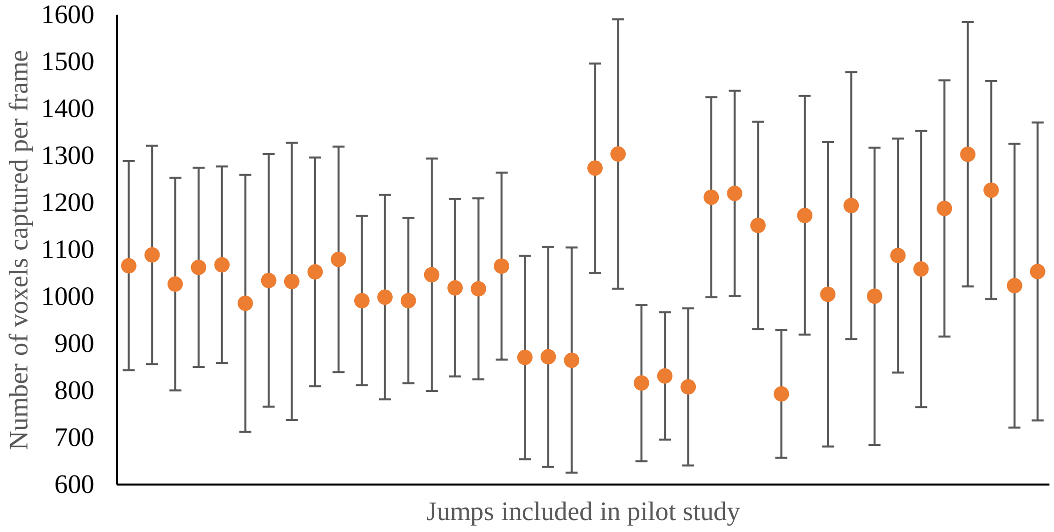

Figure A6.1. The average number of voxels per frame (vertical axis) collected during the 40 jumps of the pilot study (horizontal axis). The error bars display the standard deviation in the average number of voxels captured per frame.

Appendix 7. Q-Q plots of the data of Chapter 6

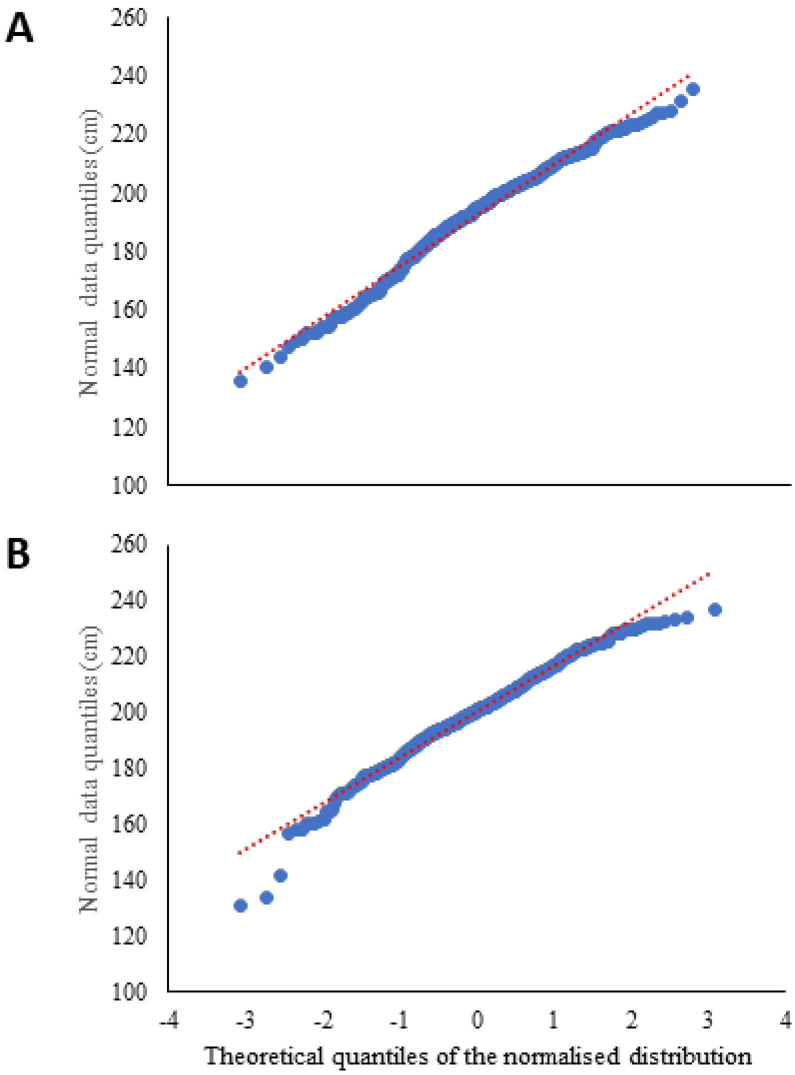

Figure A7.1. Q-Q plots of the jump data throughout the season. Figure A7.1A: control standing broad jump. Figure A7.1B: maximal standing broad jump. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

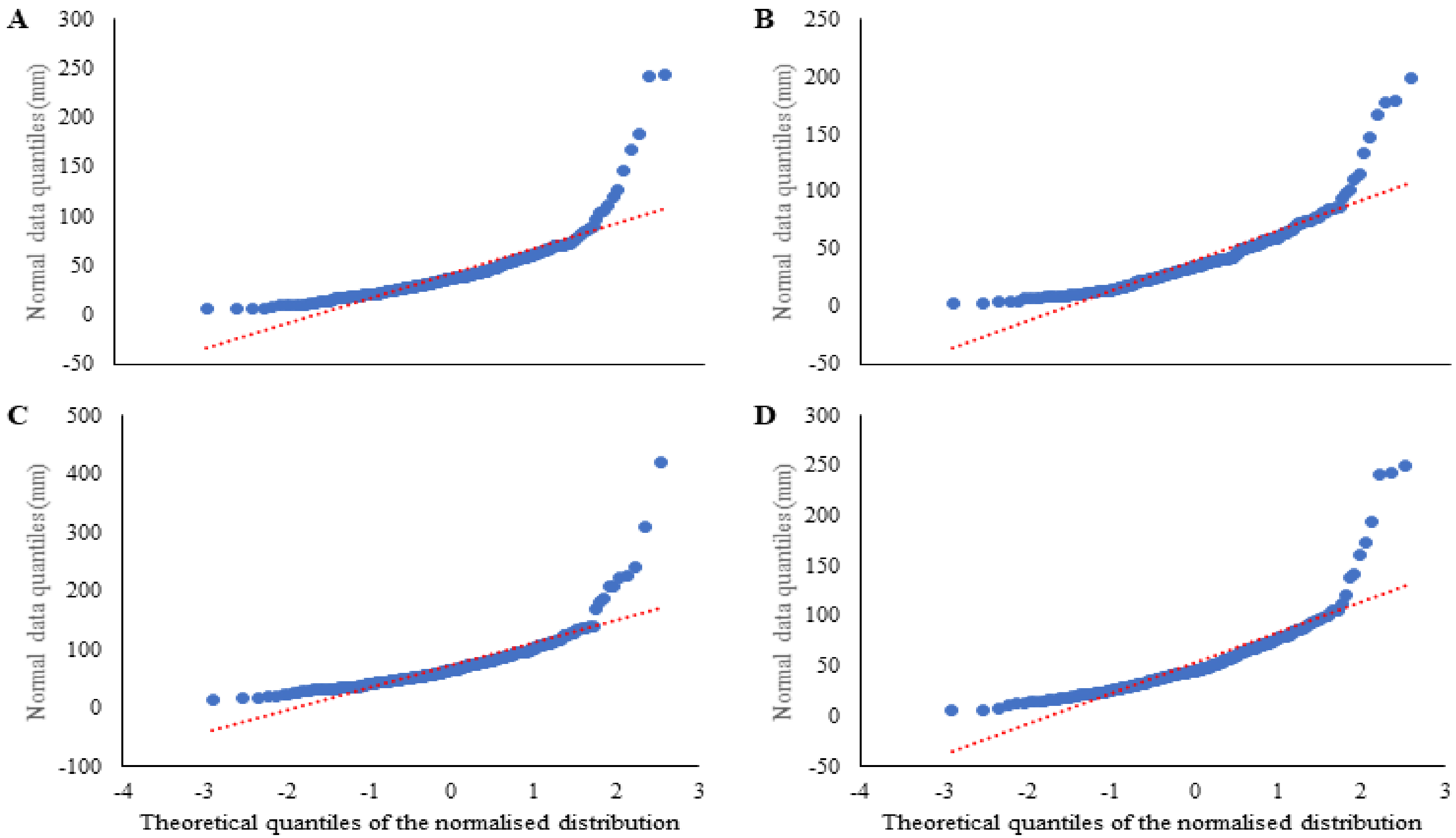

Figure A7.2. Q-Q plots of the body sway variables in the ML axis where the frame of initial contact is the first frame. Figure A7.2A: Average distance raw. Figure A7.2B: Average distance SA. Figure A7.2C: Maximal distance raw. Figure A7.2D: Maximal distance SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

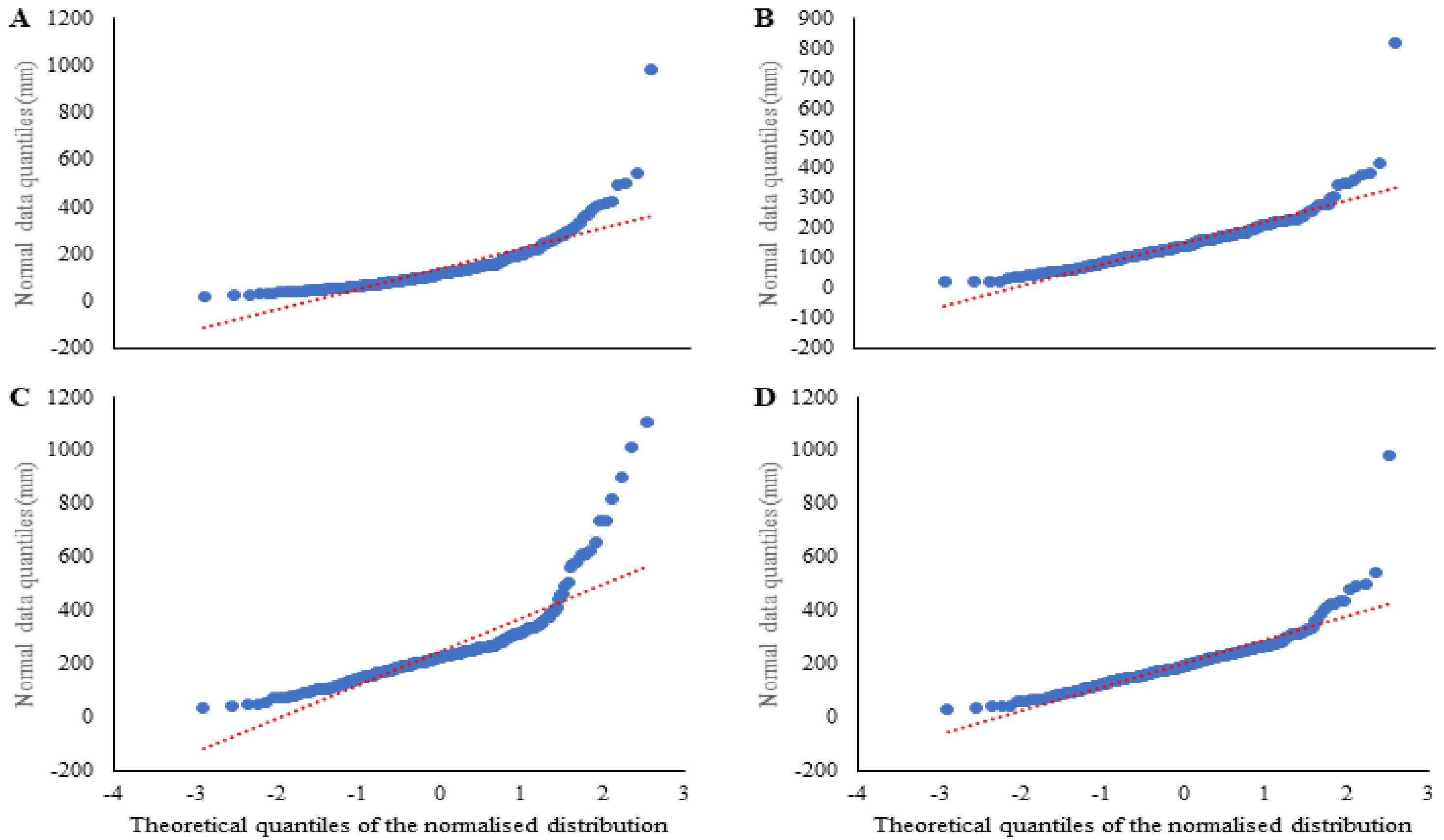

Figure A7.3. Q-Q plots of the body sway variables in the PA axis where the frame of initial contact is the first frame. Figure A7.3A: Average distance raw. Figure A7.3B: Average distance SA. Figure A7.3C: Maximal distance raw. Figure A7.3D: Maximal distance SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

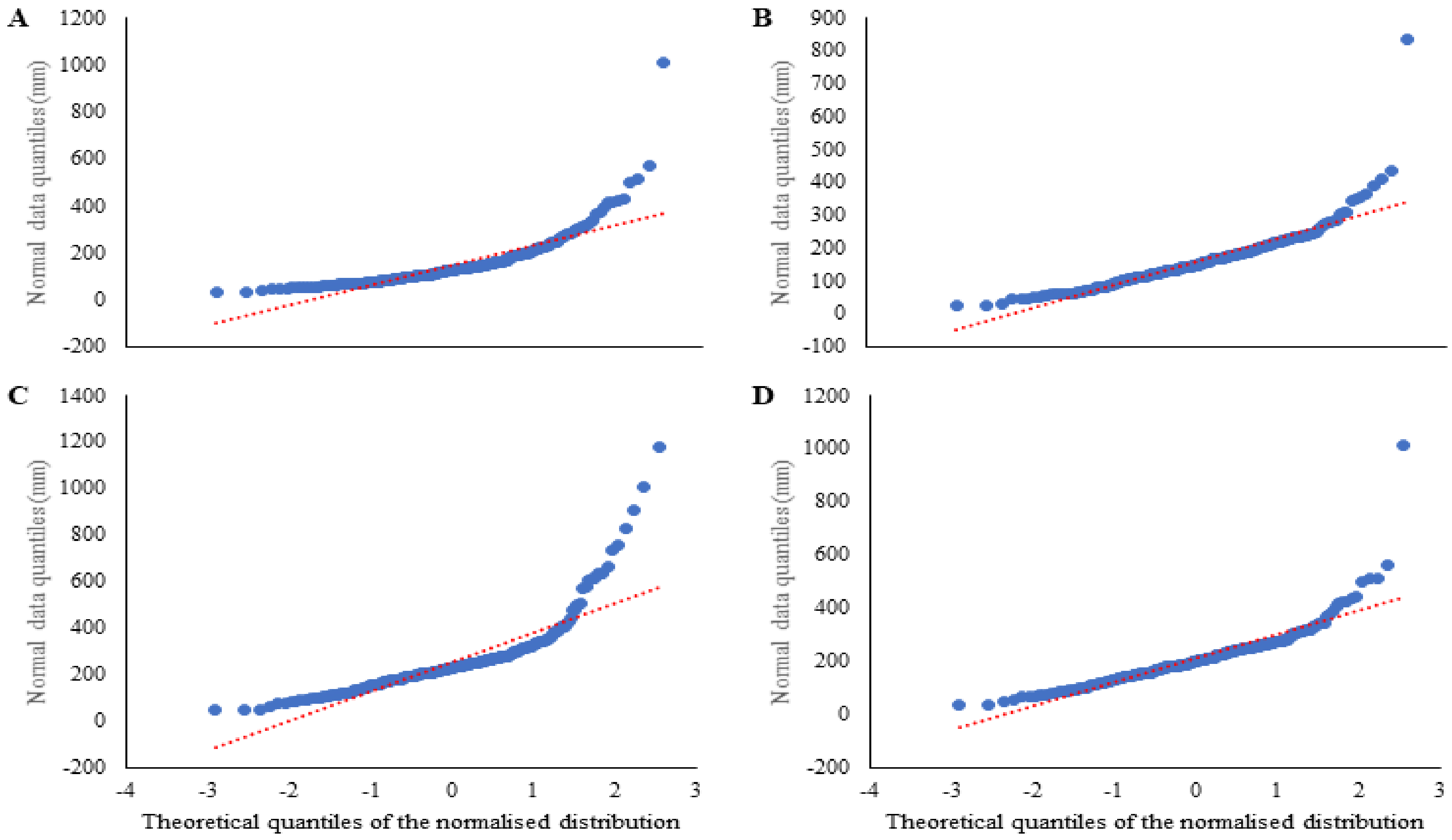

Figure A7.4. Q-Q plots of the body sway variables in the resultant axis where the frame of initial contact is the first frame. Figure A7.4A: Average distance raw. Figure A7.4B: Average distance SA. Figure A7.4C: Maximal distance raw. Figure A7.4D: Maximal distance SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

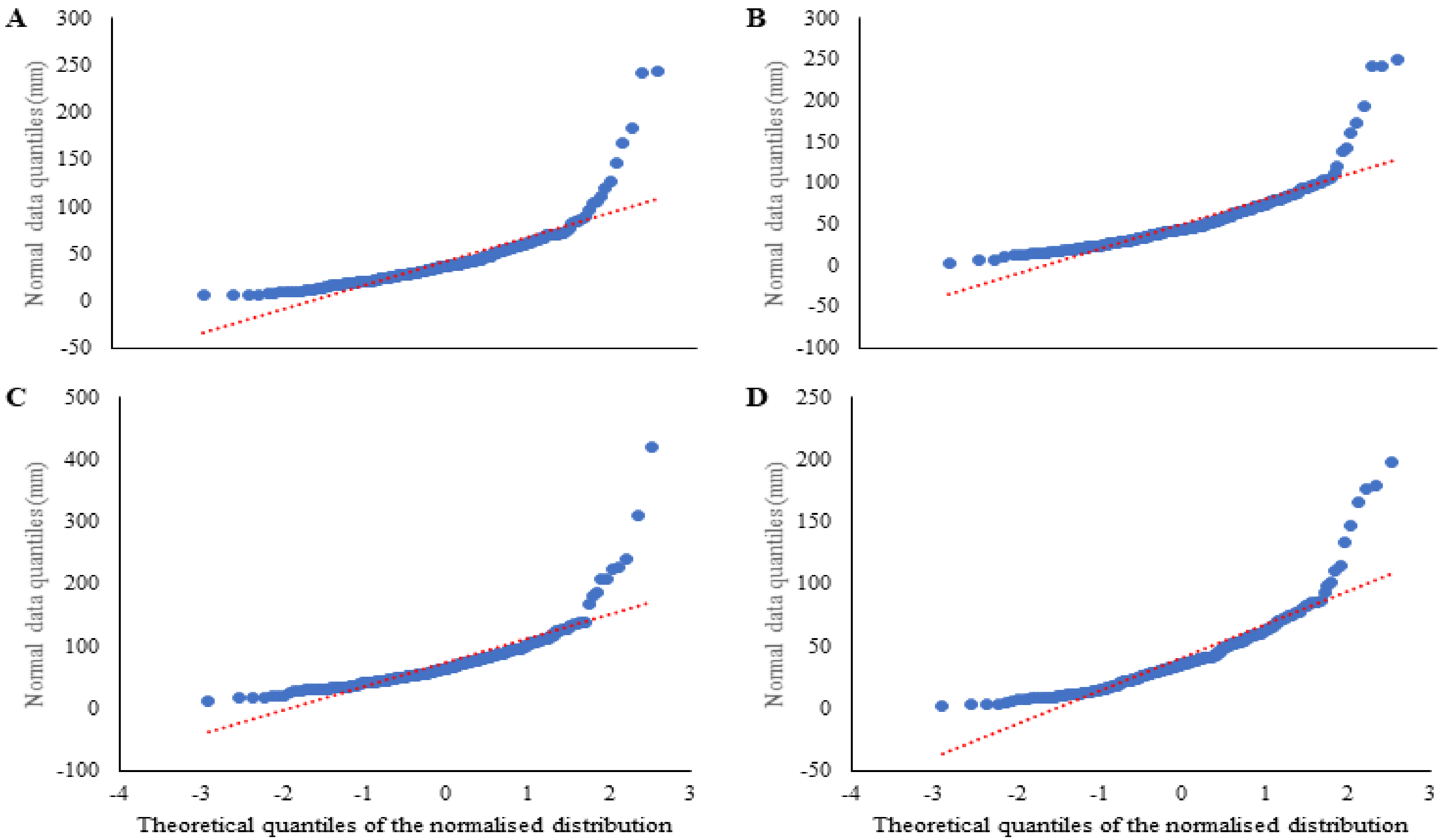

Figure A7.5. Q-Q plots of the body sway variables in the ML axis where $CoM_{PA} > BoS_{PA}$ is the first frame. Figure A7.5A: Average distance raw. Figure A7.5B: Average distance SA. Figure A7.5C: Maximal distance raw. Figure A7.5D: Maximal distance SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

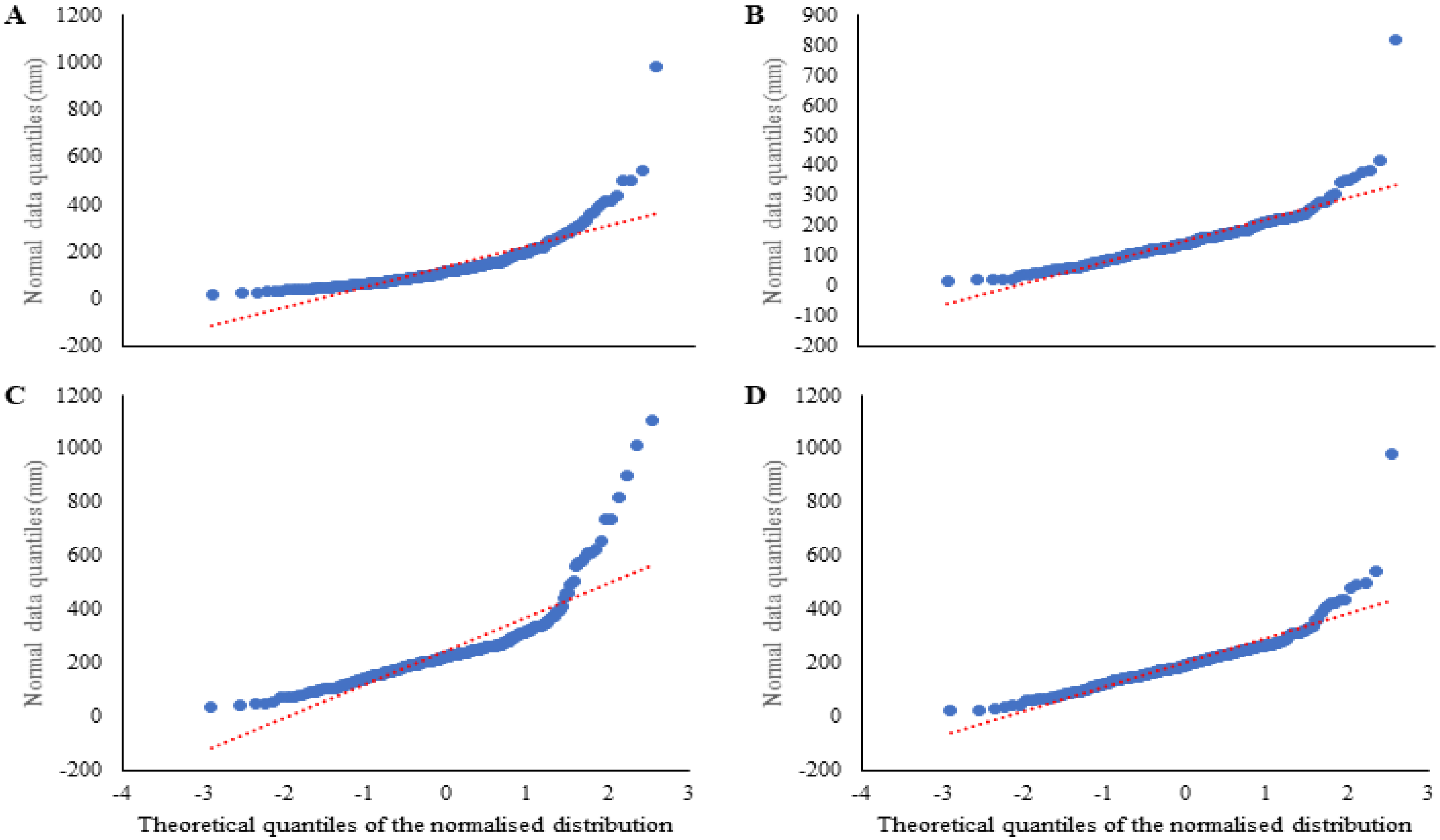

Figure A7.6. Q-Q plots of the body sway variables in the PA axis where $CoM_{PA} > BoS_{PA}$ is the first frame. Figure A7.6A: Average distance raw. Figure A7.6B: Average distance SA. Figure A7.6C: Maximal distance raw. Figure A7.6D: Maximal distance SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

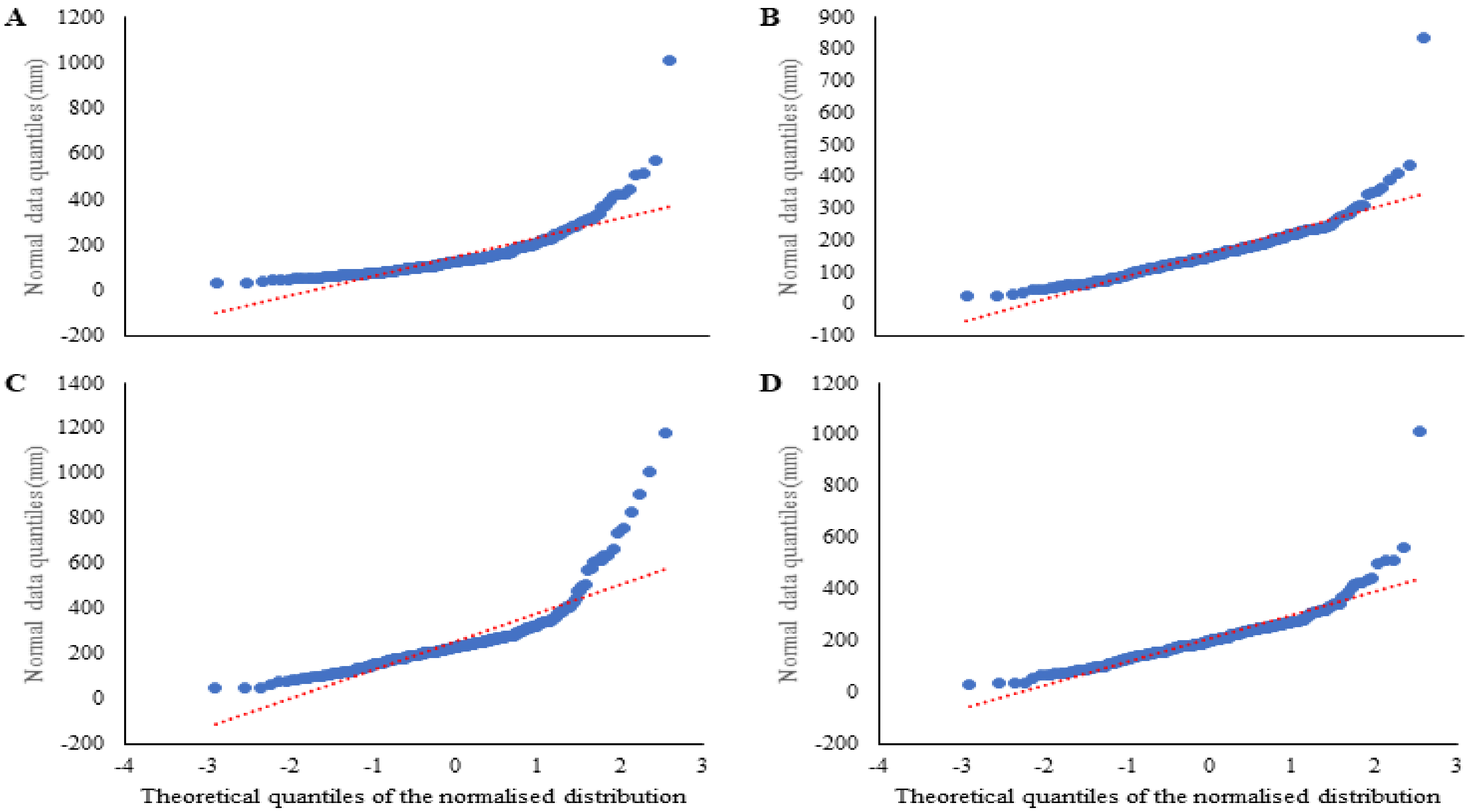

Figure A7.7. Q-Q plots of the body sway variables in the resultant axis where $CoM_{PA} > BoS_{PA}$ is the first frame. Figure A7.7A: Average distance raw. Figure A7.7B: Average distance SA. Figure A7.7C: Maximal distance raw. Figure A7.7D: Maximal distance SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

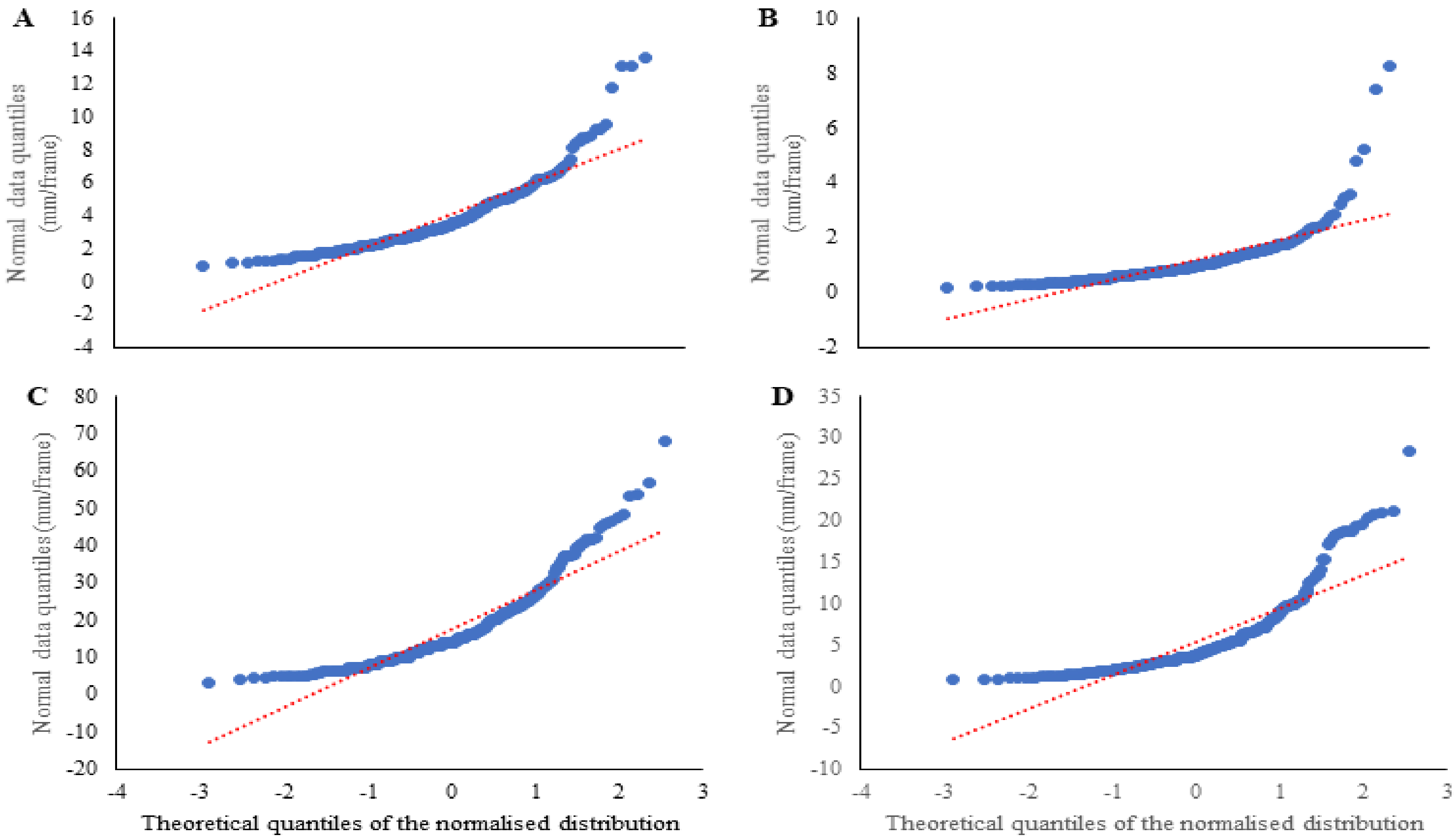

Figure A7.8. Q-Q plots of the body sway speed variables in the ML axis where the frame of initial contact is the first frame. Figure A7.8A: Average speed raw. Figure A7.8B: Average speed SA. Figure A7.8C: Maximal speed raw. Figure A7.8D: Maximal speed SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

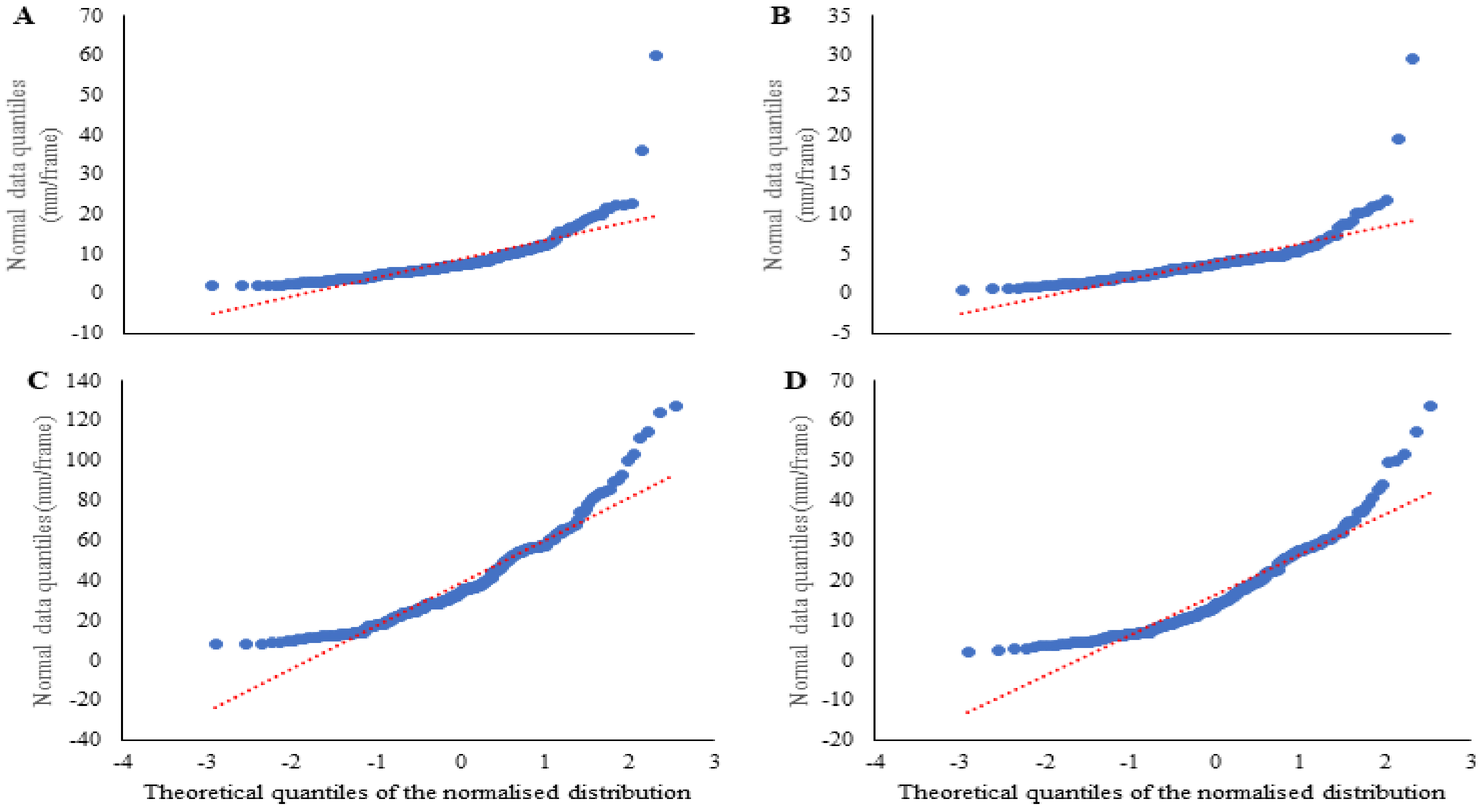

Figure A7.9. Q-Q plots of the body sway speed variables in the PA axis where the frame of initial contact is the first frame. Figure A7.9A: Average speed raw. Figure A7.9B: Average speed SA. Figure A7.9C: Maximal speed raw. Figure A7.9D: Maximal speed SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

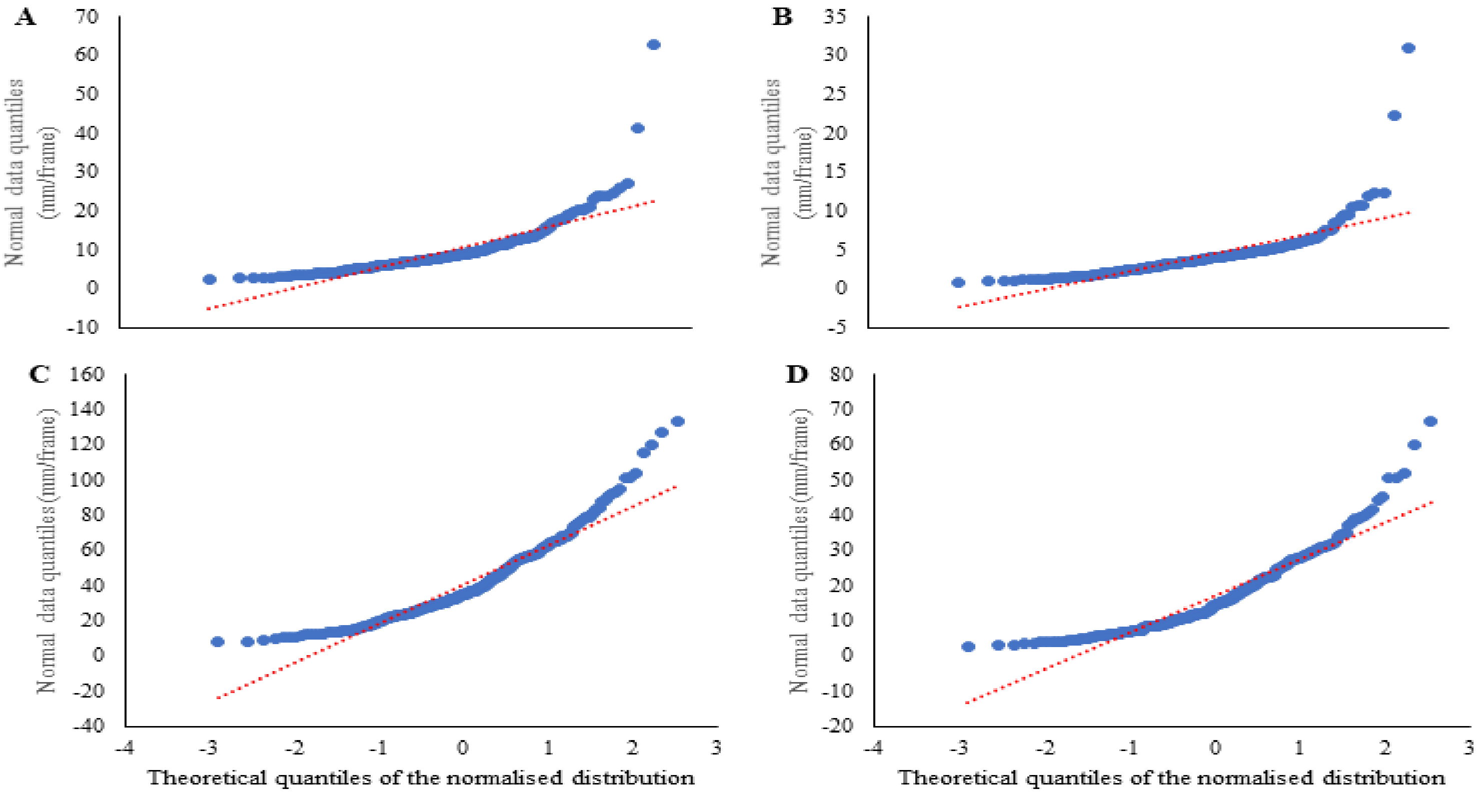

Figure A7.10. Q-Q plots of the body sway speed variables in the resultant axis where the frame of initial contact is the first frame. Figure A7.10A: Average speed raw. Figure A7.10B: Average speed SA. Figure A7.10C: Maximal speed raw. Figure A7.10D: Maximal speed SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

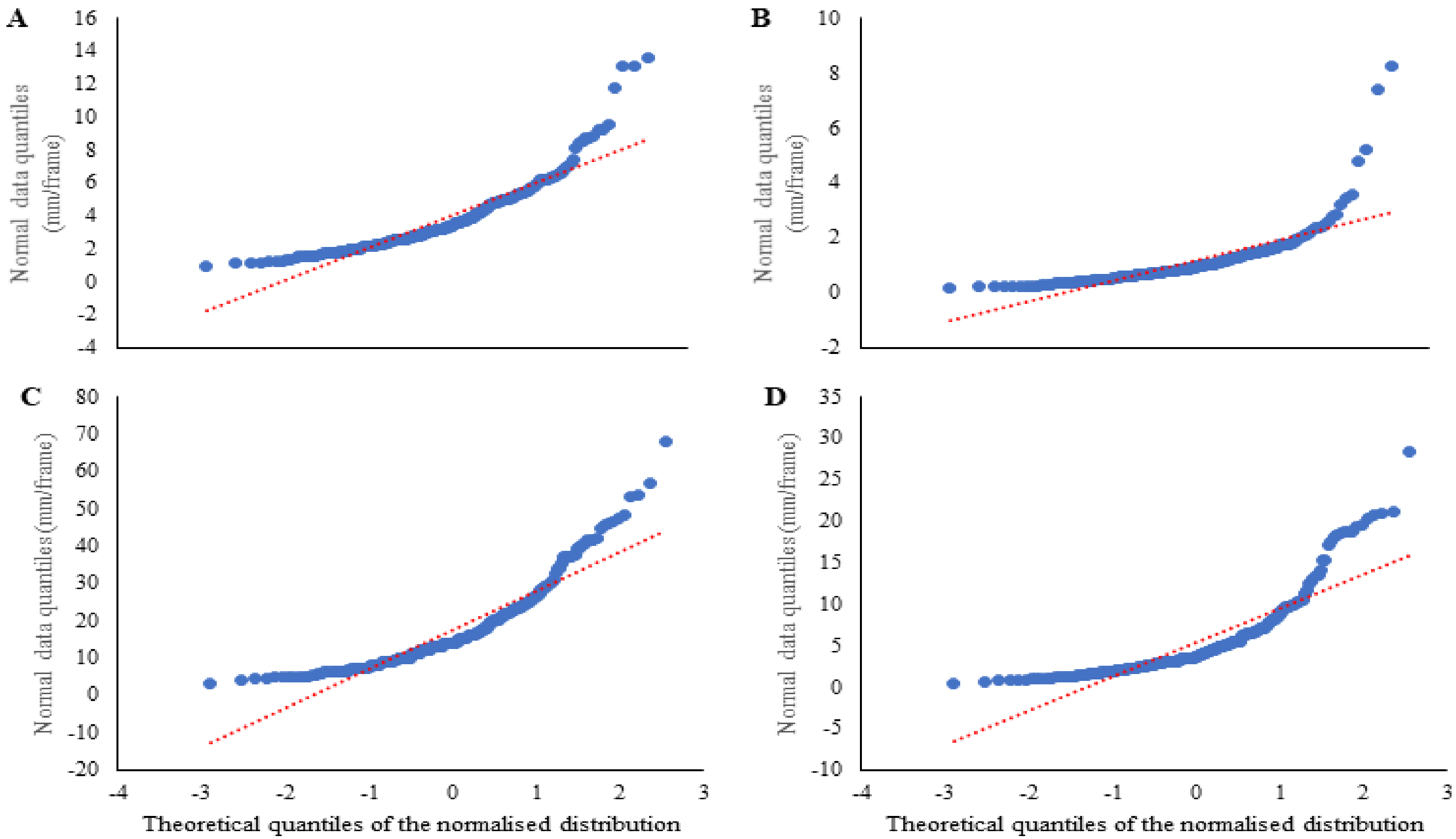

Figure A7.11. Q-Q plots of the body sway speed variables in the ML axis where $CoM_{PA} > BoS_{PA}$ is the first frame. Figure A7.11A: Average speed raw. Figure A7.11B: Average speed SA. Figure A7.11C: Maximal speed raw. Figure A7.11D: Maximal speed SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

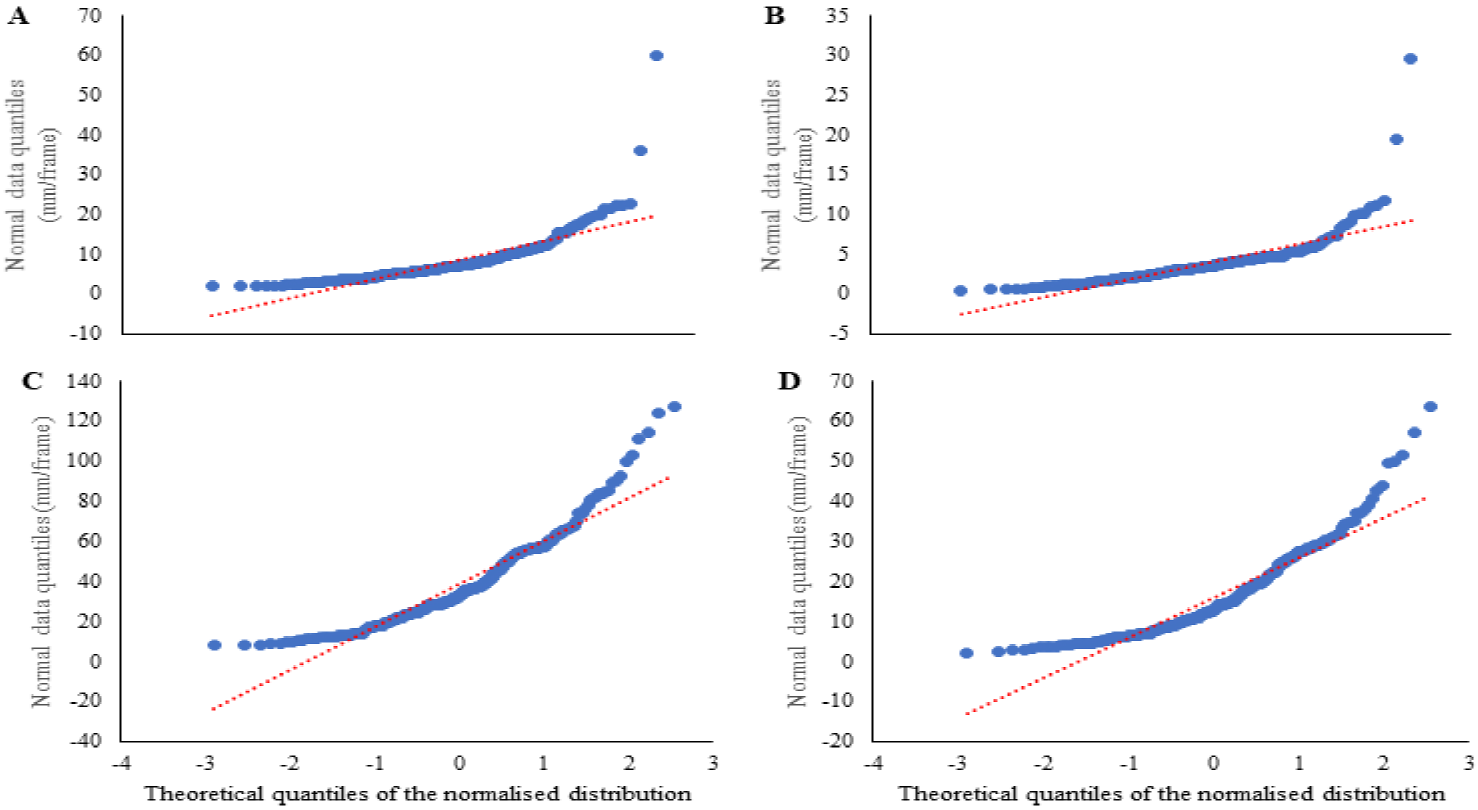

Figure A7.12. Q-Q plots of the body sway speed variables in the PA axis where $CoM_{PA} > BoS_{PA}$ is the first frame. Figure A7.12A: Average speed raw. Figure A7.12B: Average speed SA. Figure A7.12C: Maximal speed raw. Figure A7.12D: Maximal speed SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

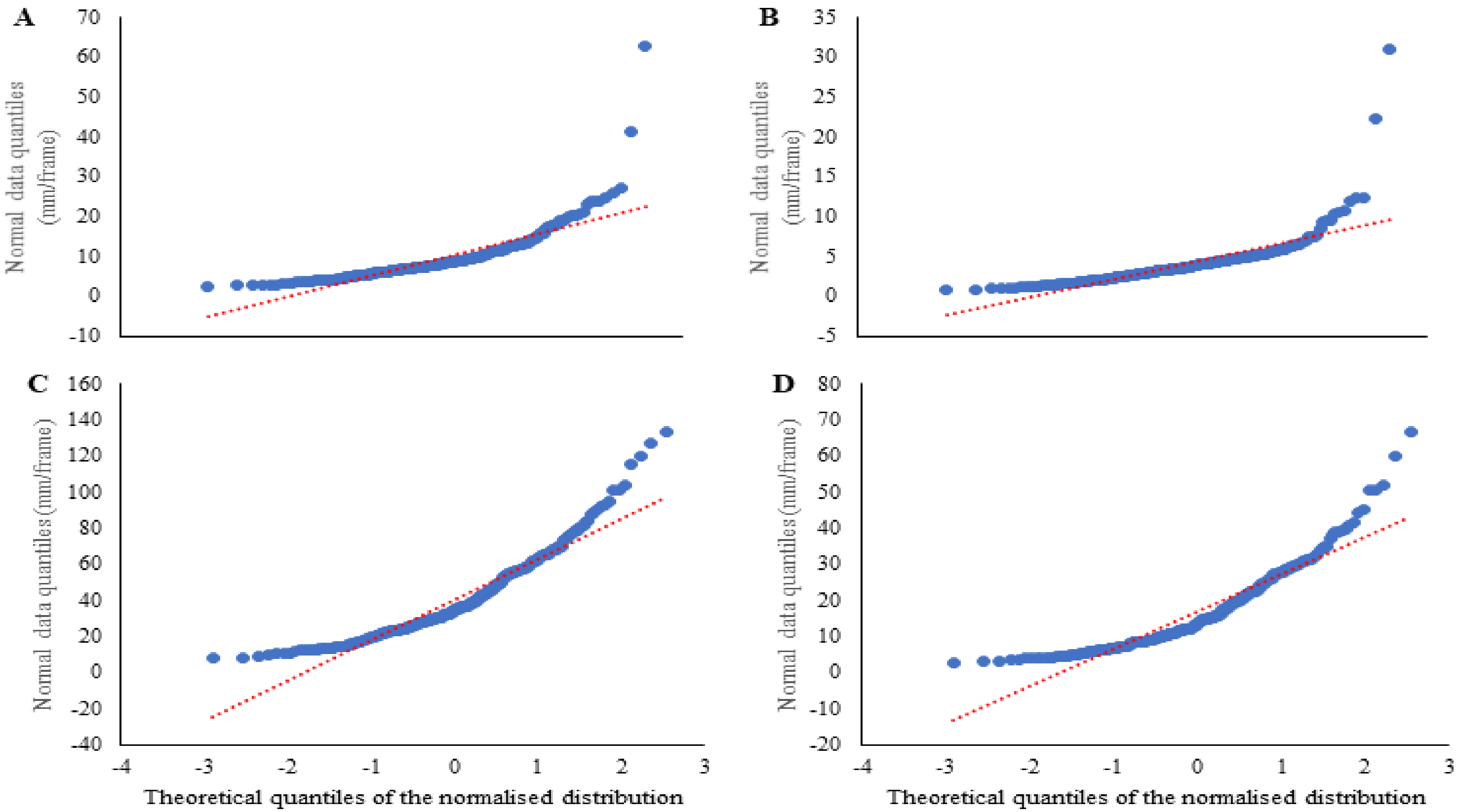

Figure A7.13. Q-Q plots of the body sway speed variables in the PA axis where CoM_{PA} > BoS_{PA} is the first frame. Figure A7.13A: Average speed raw. Figure A7.13B: Average speed SA. Figure A7.13C: Maximal speed raw. Figure A7.13D: Maximal speed SA. The blue lines represent the collected data, the red line represents the theoretical data of perfectly normal distributed data.

Appendix 8. Figures of within-subject variability of body sway variables.

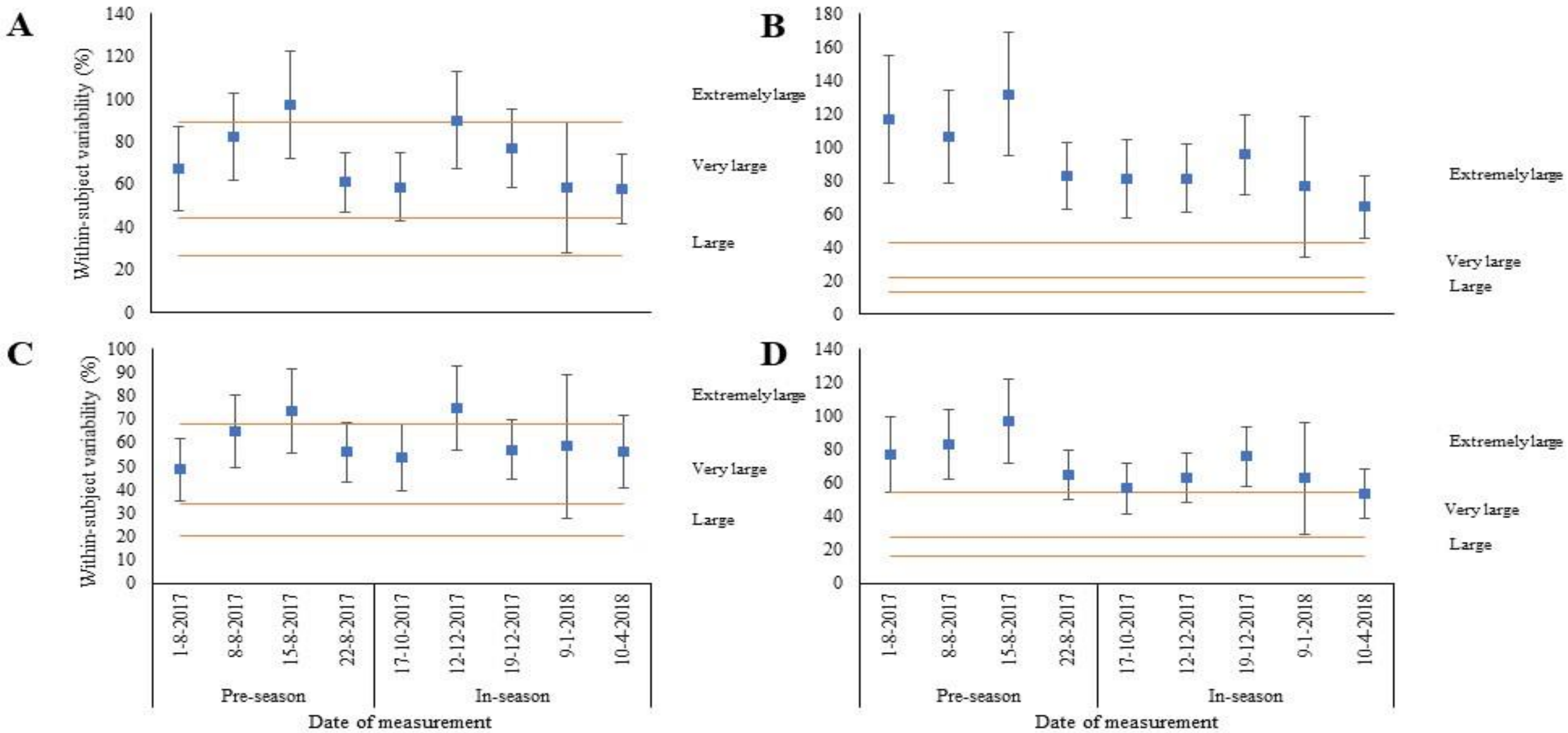

Figure A8.1. Within-subject variability of body sway variables in the ML axis where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A8.1A: Average distance raw. Figure A8.1B: Average distance SA. Figure A8.1C: Maximal distance raw. Figure A8.1D: Maximal distance SA.

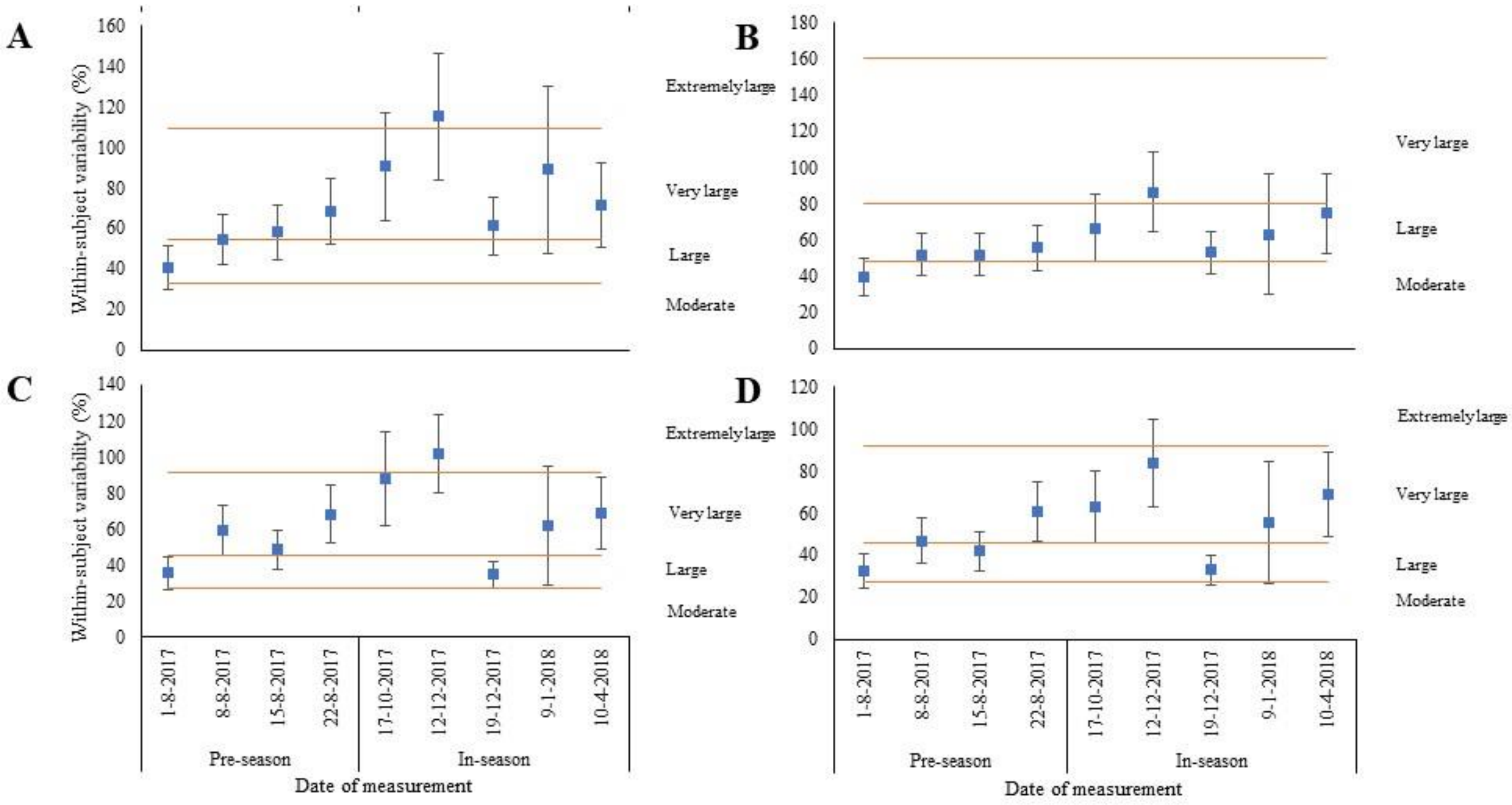

Figure A8.2. Within-subject variability of body sway variables in the PA axis where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A8.2A: Average distance raw. Figure A8.2B: Average distance SA. Figure A8.2C: Maximal distance raw. Figure A8.2D: Maximal distance SA.

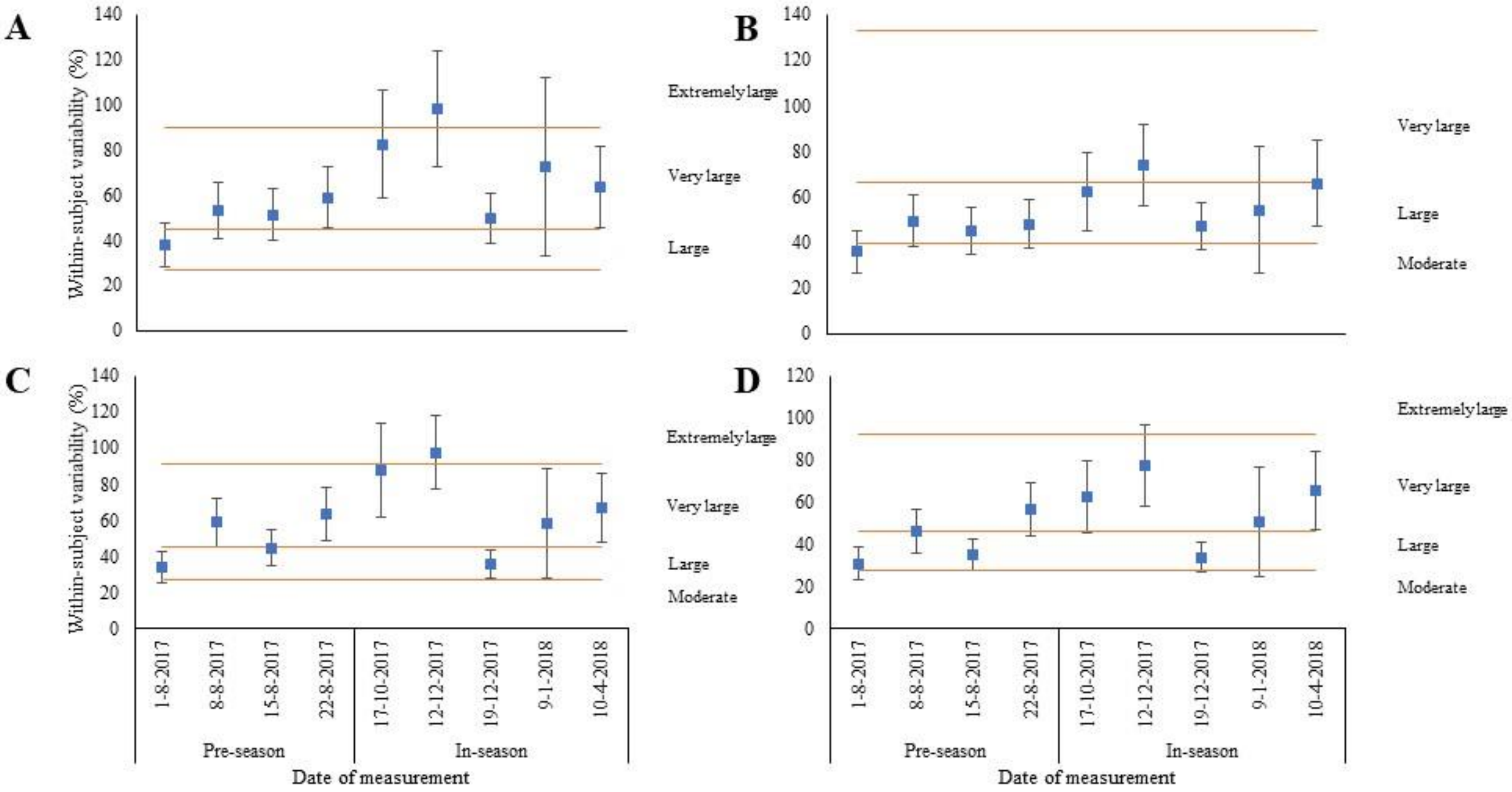

Figure A8.3. Within-subject variability of body sway variables in the resultant axis where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A8.3A: Average distance raw. Figure A8.3B: Average distance SA. Figure A8.3C: Maximal distance raw. Figure A8.3D: Maximal distance SA.

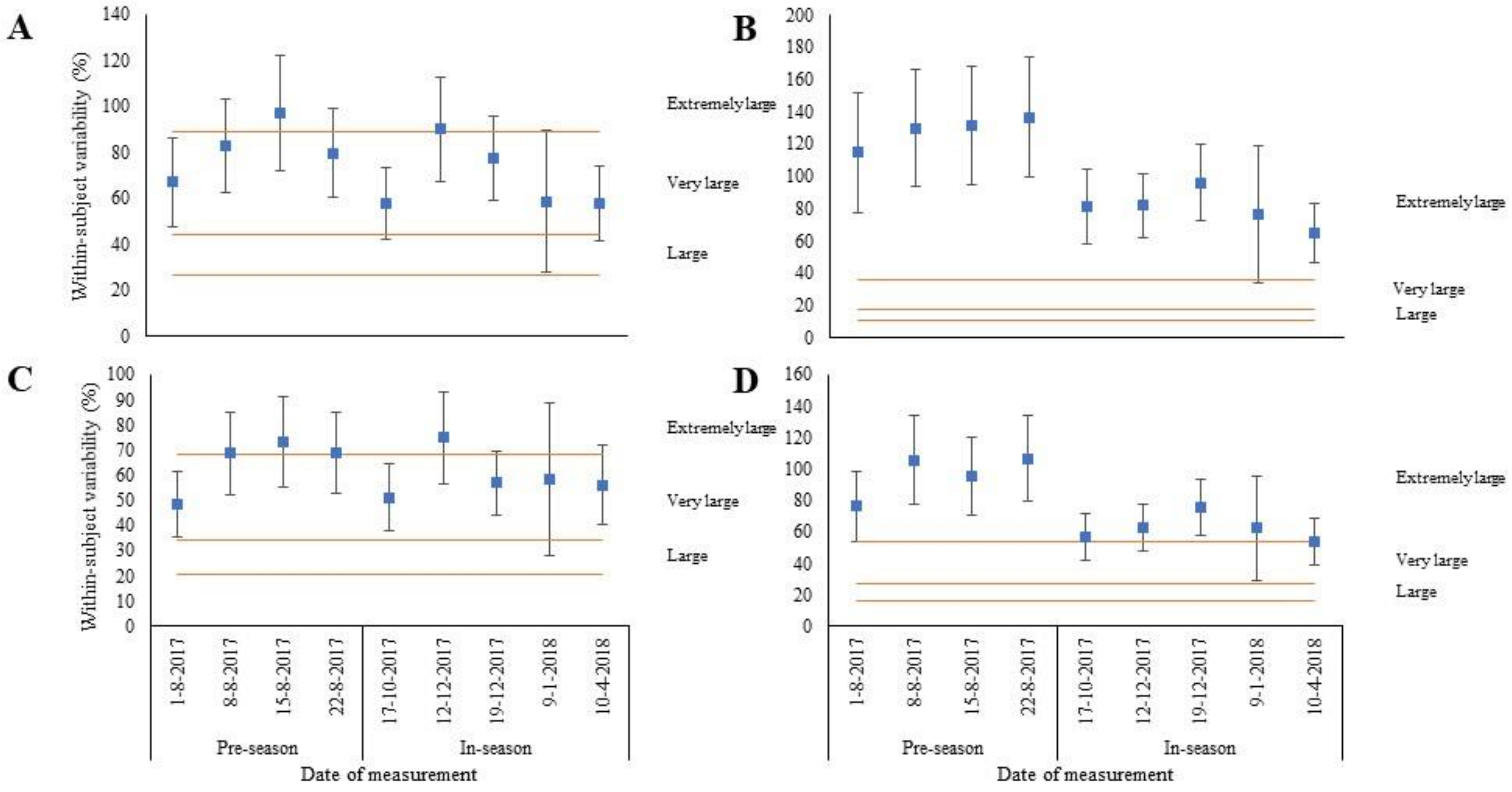

Figure A8.4. Within-subject variability of body sway variables in the ML axis where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A8.4A: Average distance raw. Figure A8.4B: Average distance SA. Figure A8.4C: Maximal distance raw. Figure A8.4D: Maximal distance SA.

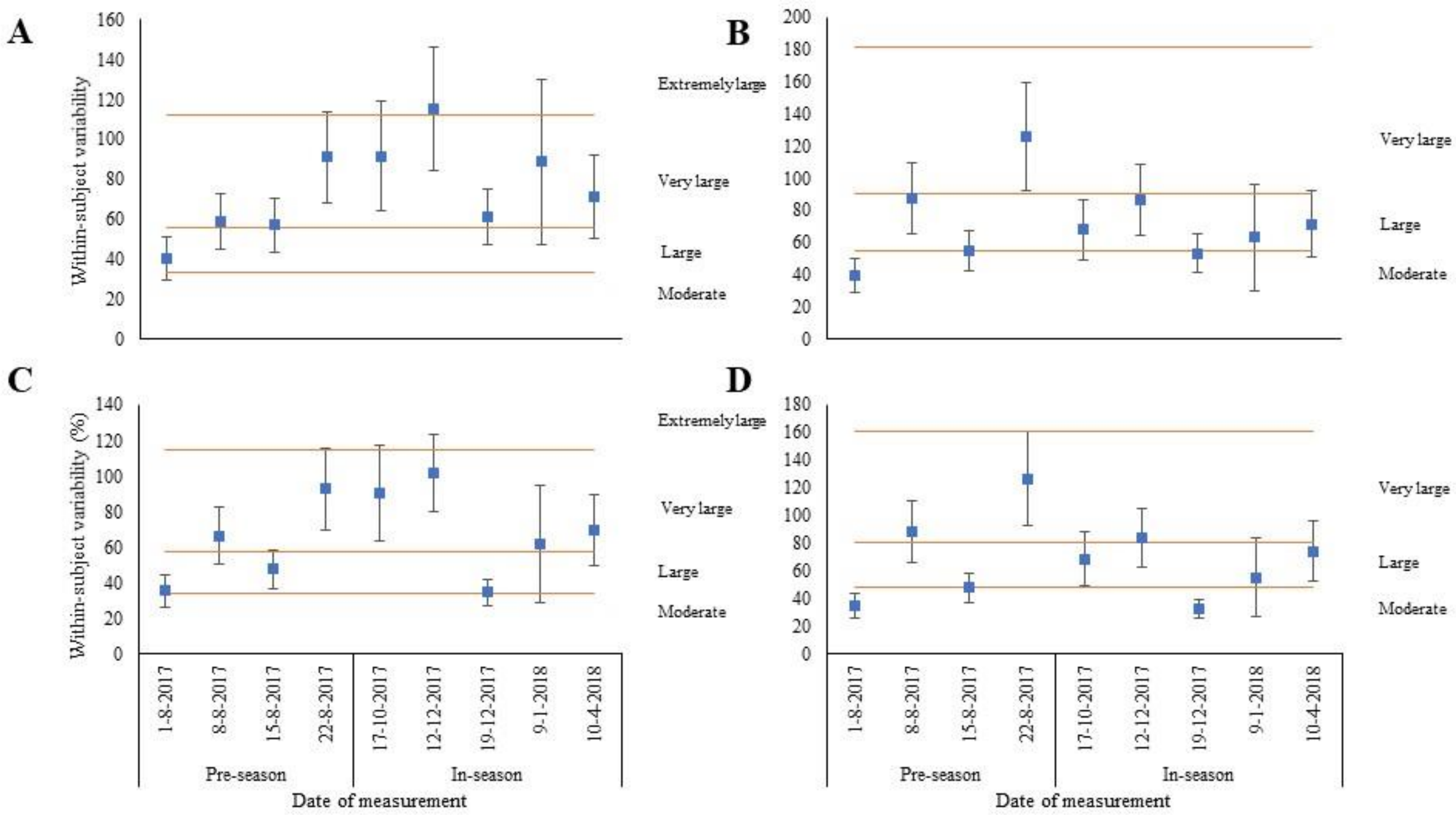

Figure A8.5. Within-subject variability of body sway variables in the PA axis where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A8.5A: Average distance raw. Figure A8.5B: Average distance SA. Figure A8.5C: Maximal distance raw. Figure A8.5D: Maximal distance SA.

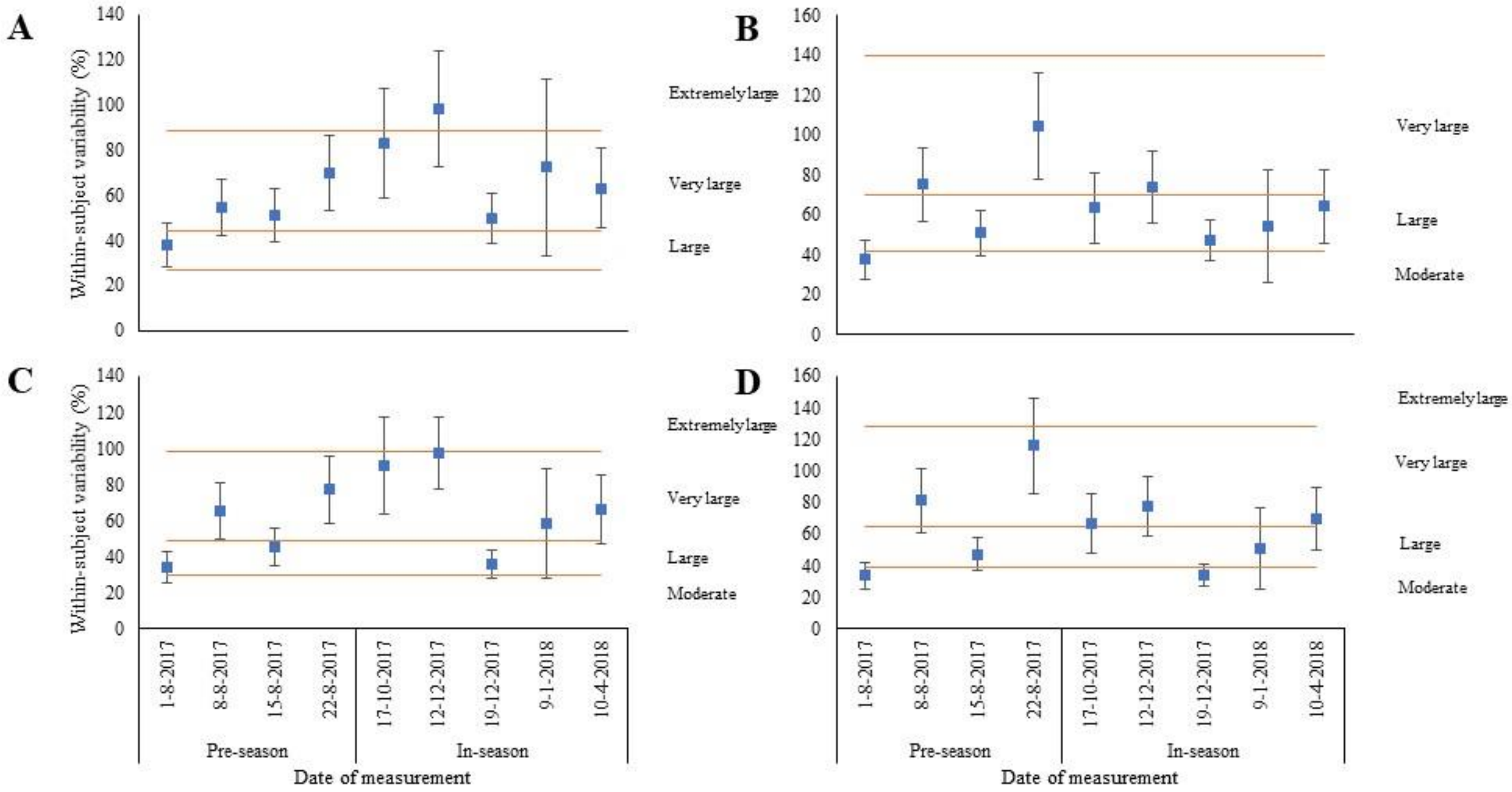

Figure A8.6. Within-subject variability of body sway variables in the resultant axis where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A8.6A: Average distance raw. Figure A8.6B: Average distance SA. Figure A8.6C: Maximal distance raw. Figure A8.6D: Maximal distance SA.

Appendix 9. Figures of within-subject variability of the body sway speed variables.

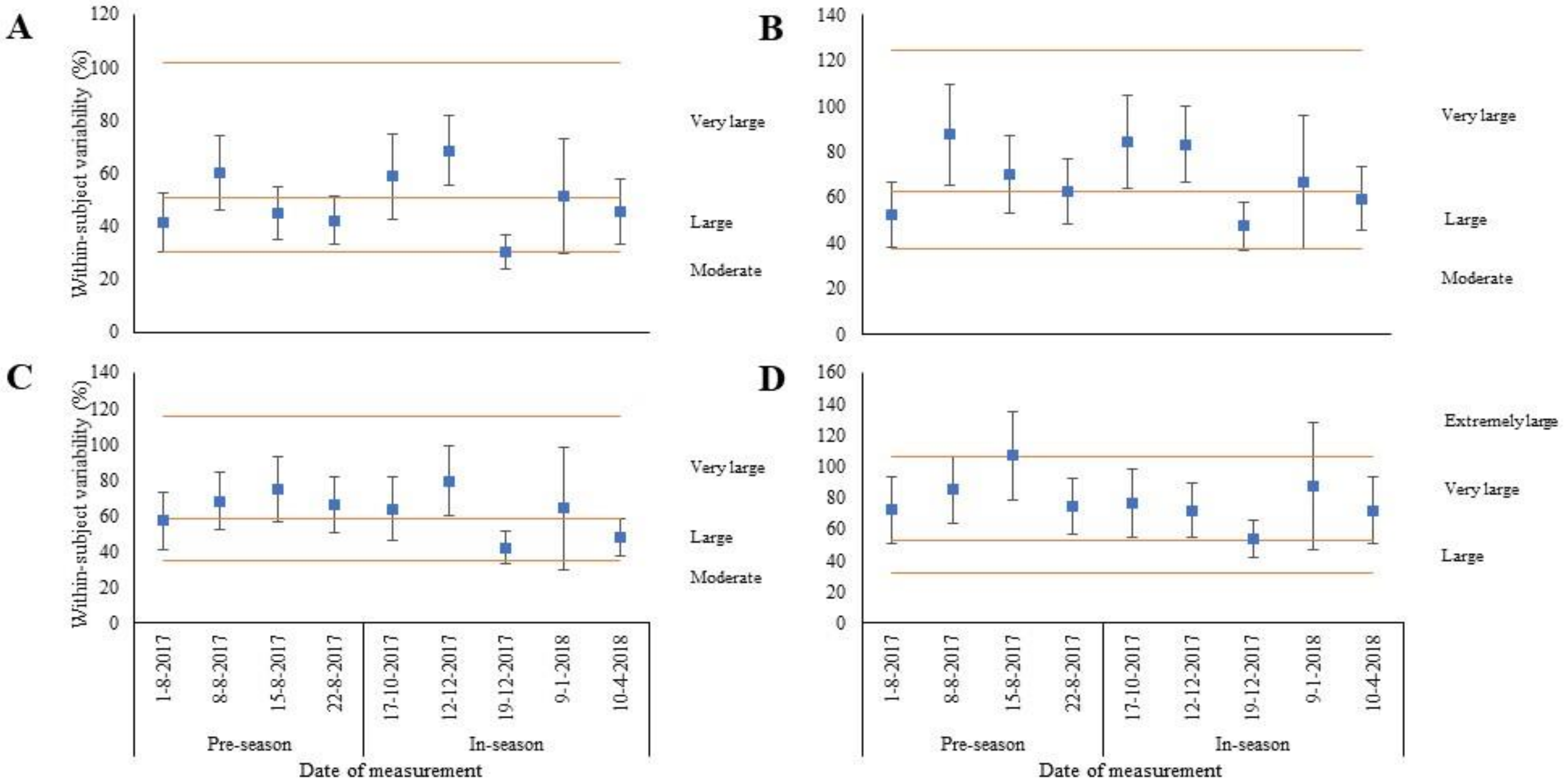

Figure A9.1. Within-subject variability of the body sway speed variables in the ML axis where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A9.1A: Average speed raw. Figure A9.1B: Average speed SA. Figure A9.1C: Maximal speed raw. Figure A9.1D: Maximal speed SA.

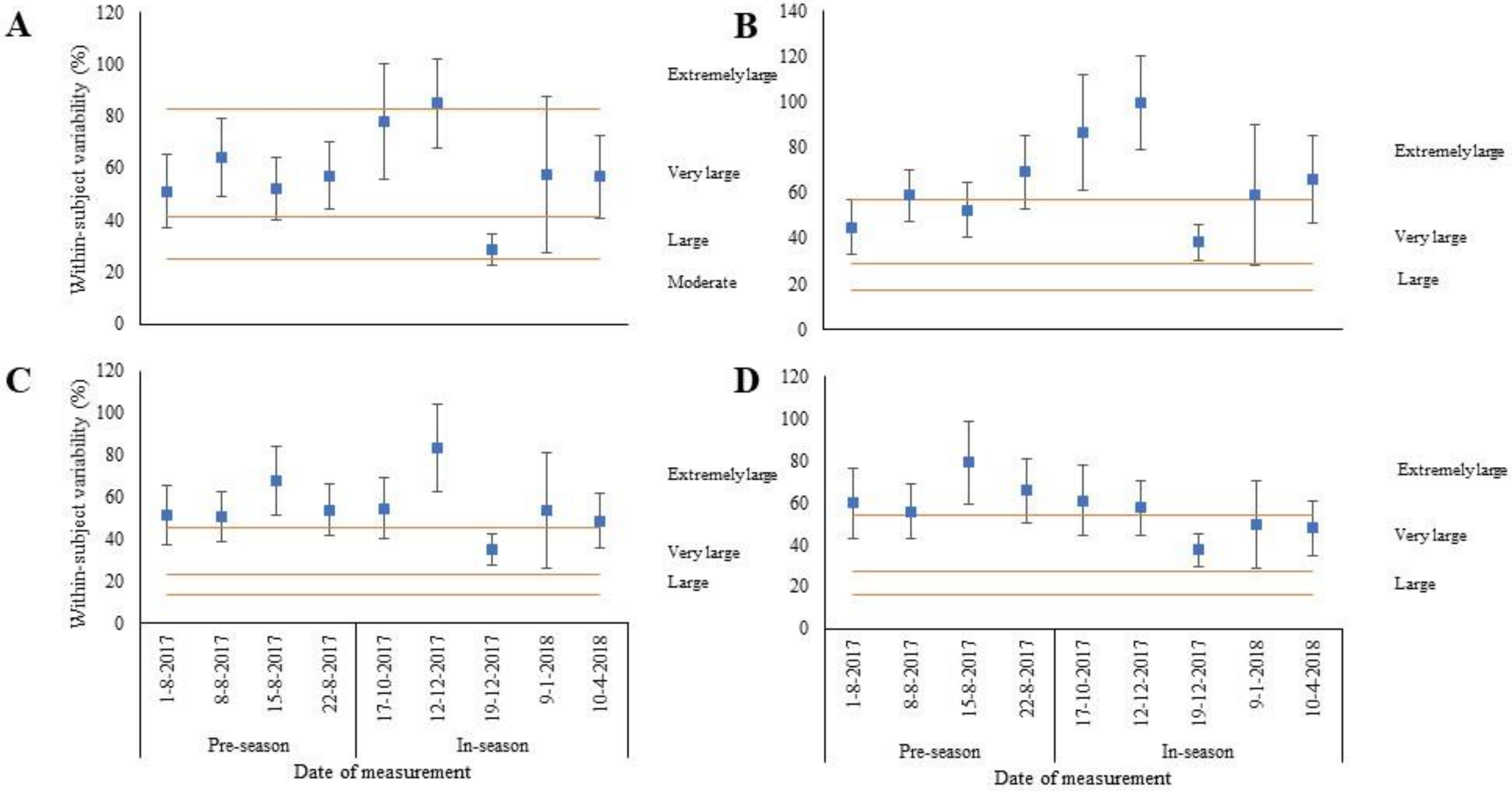

Figure A9.2. Within-subject variability of the body sway speed variables in the PA axis where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A9.2A: Average speed raw. Figure A9.2B: Average speed SA. Figure A9.2C: Maximal speed raw. Figure A9.2D: Maximal speed SA.

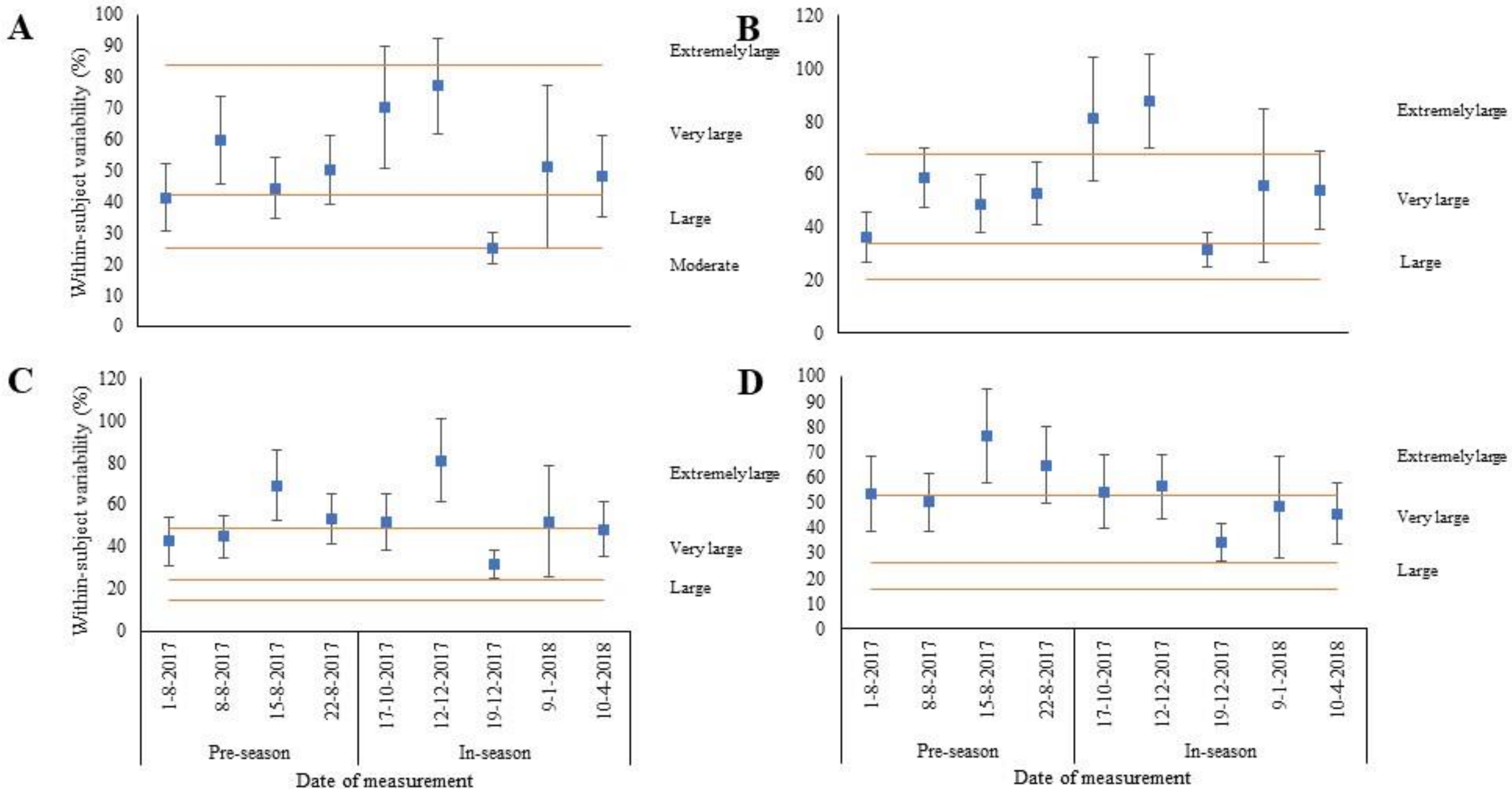

Figure A9.3. Within-subject variability of the body sway speed variables in the resultant axis where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A9.3A: Average speed raw. Figure A9.3B: Average speed SA. Figure A9.3C: Maximal speed raw. Figure A9.3D: Maximal speed SA.

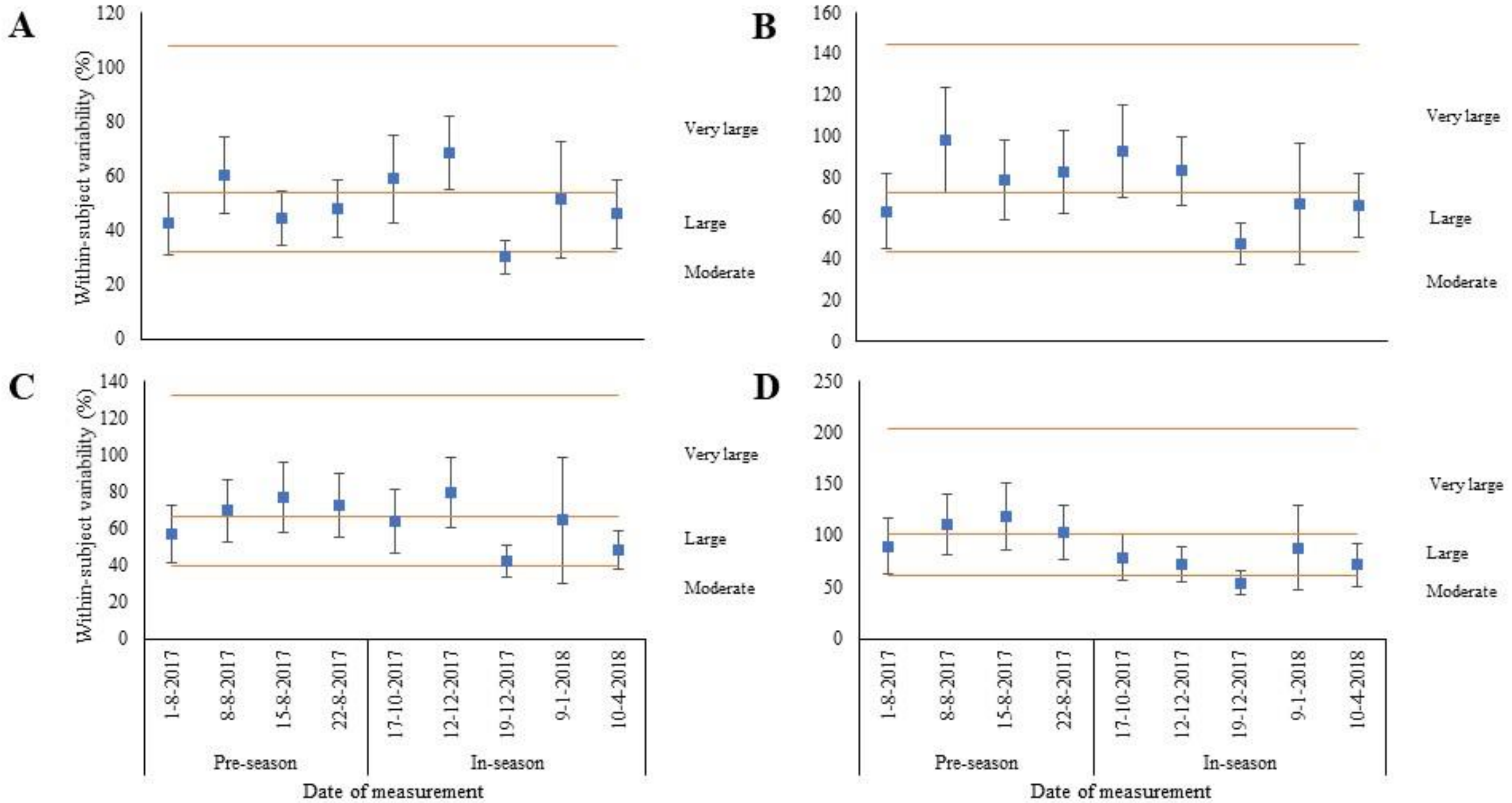

Figure A9.4. Within-subject variability of the body sway speed variables in the ML axis where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A9.4A: Average speed raw. Figure A9.4B: Average speed SA. Figure A9.4C: Maximal speed raw. Figure A9.4D: Maximal speed SA.

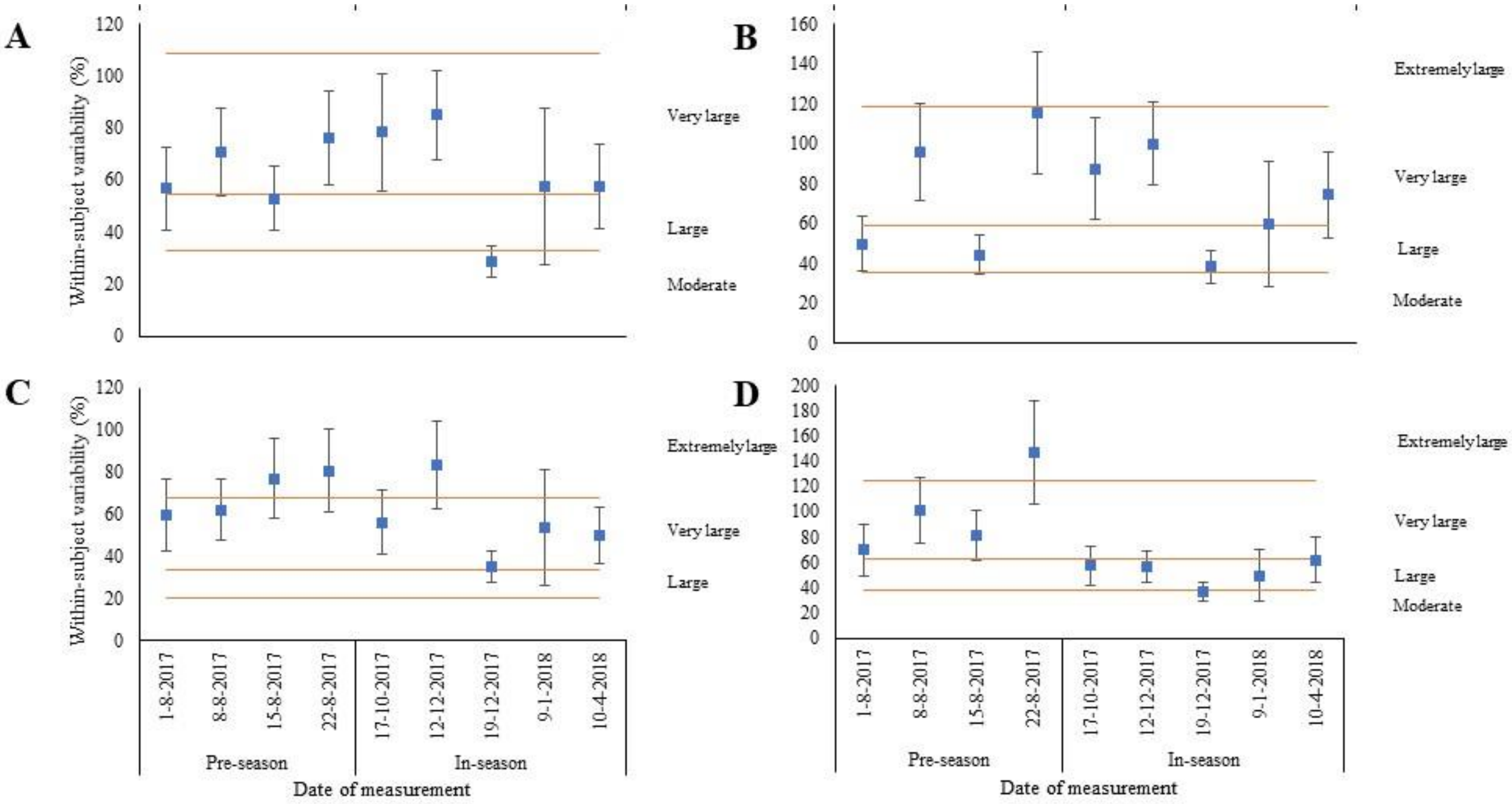

Figure A9.5. Within-subject variability of the body sway speed variables in the PA axis where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A9.5A: Average speed raw. Figure A9.5B: Average speed SA. Figure A9.5C: Maximal speed raw. Figure A9.5D: Maximal speed SA.

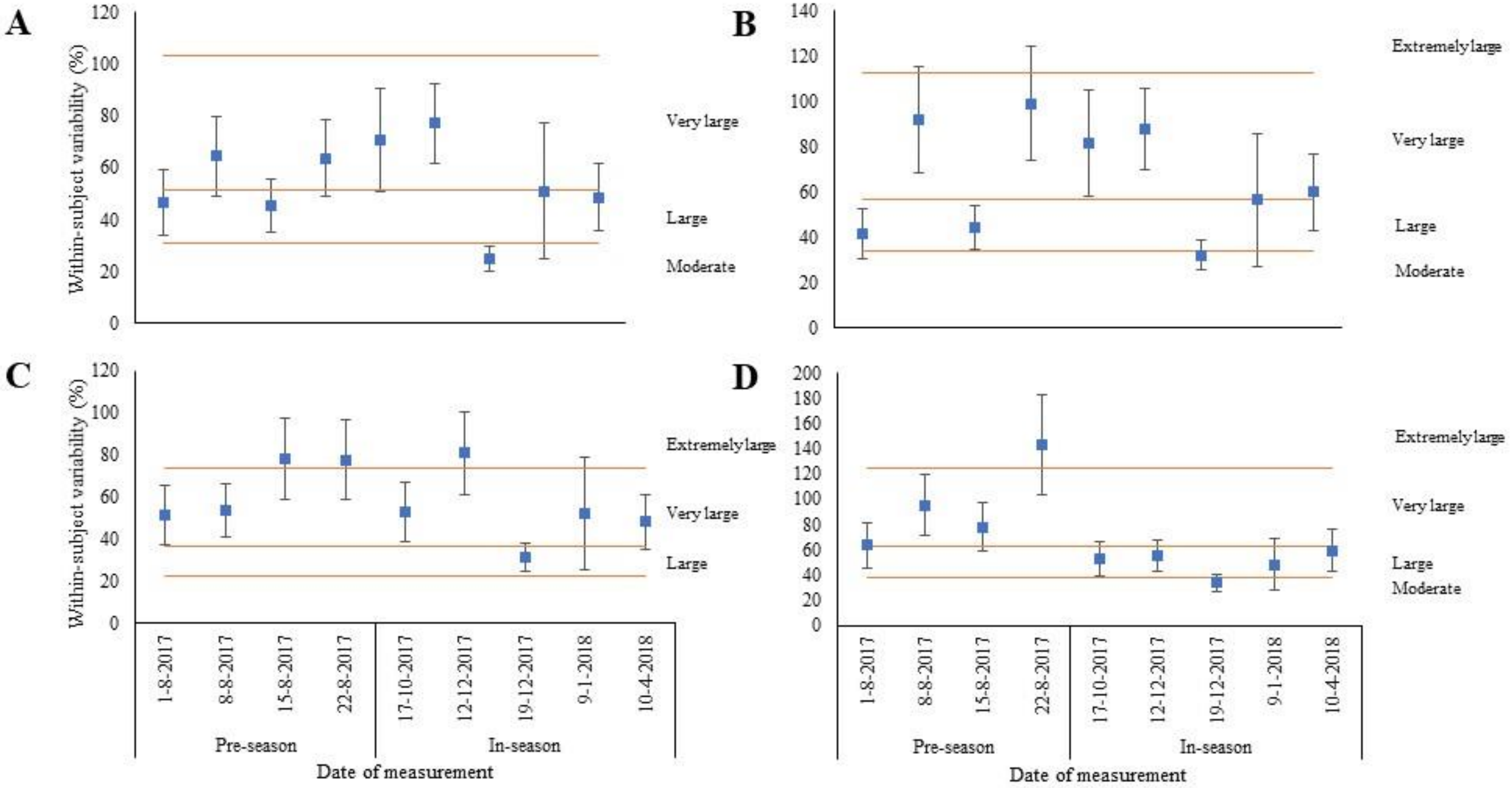

Figure A9.6. Within-subject variability of the body sway speed variables in the resultant axis where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. The red lines display the thresholds for meaningful changes. The qualitative inferences are displayed on the right side of the graphs. Figure A9.6A: Average speed raw. Figure A9.6B: Average speed SA. Figure A9.6C: Maximal speed raw. Figure A9.6D: Maximal speed SA.

Appendix 10. Figures of seasonal variability of the body sway variables.

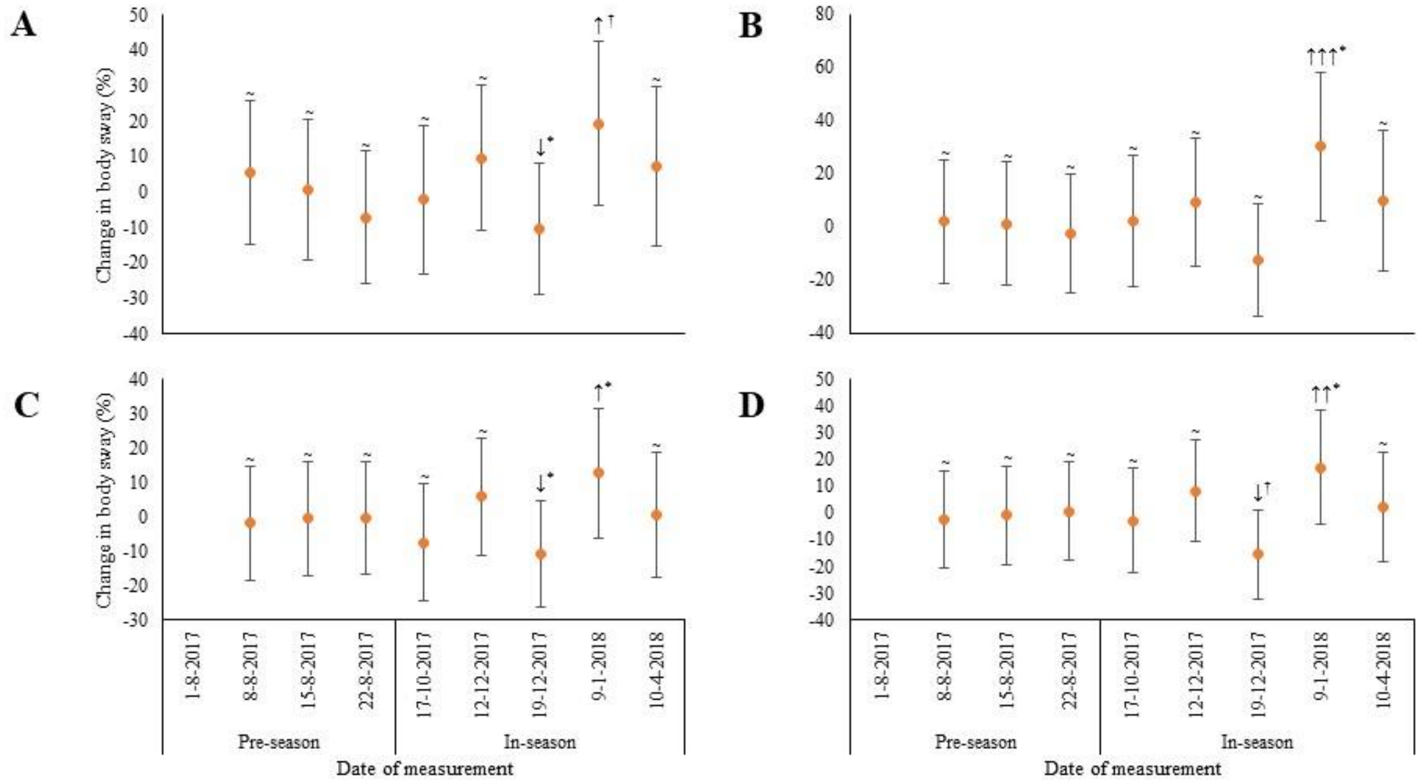

Figure A10.1. Changes in body sway in the ML axis during all sessions compared to the baseline measurement, where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. Figure A10.1A: Average distance raw. Figure A10.1B: Average distance SA. Figure A10.1C: Maximal distance raw. Figure A10.1D: Maximal distance SA. ↑, ↑↑↑ represent small and large increases in body sway compared to baseline and ↓ represents a small decrease in body sway compared to baseline. ~: unclear, *: possibly, †: likely.

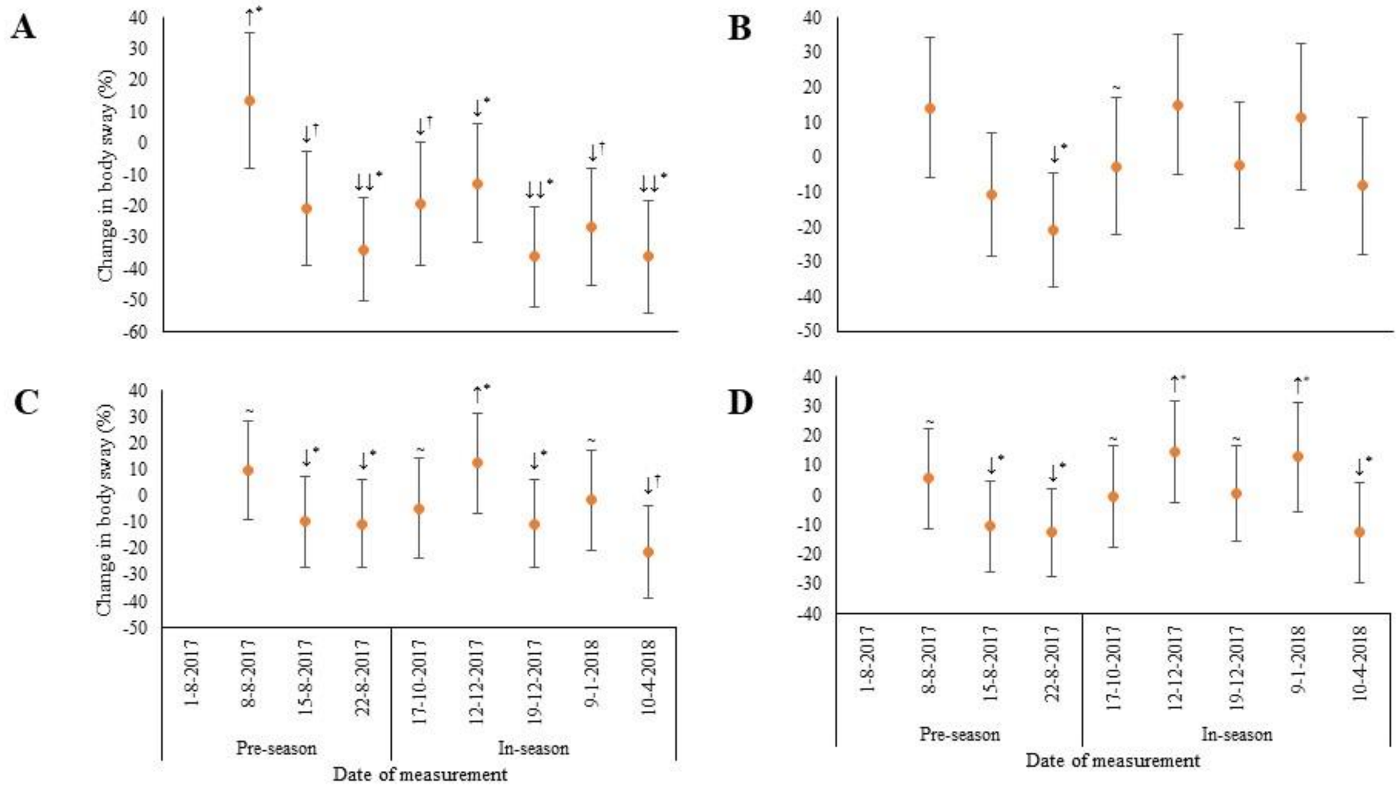

Figure A10.2. Changes in body sway in the PA axis during all sessions compared to the baseline measurement, where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. Figure A10.2A: Average distance raw. Figure A10.2B: Average distance SA. Figure A10.2C: Maximal distance raw. Figure A10.2D: Maximal distance SA. ↑ represents a small increase in body sway compared to baseline, ↓ and ↓↓ represent a small and moderate decrease in body sway compared to baseline. ~: unclear, *: possibly, †: likely. All sessions without a symbol represent a trivial change compared to baseline.

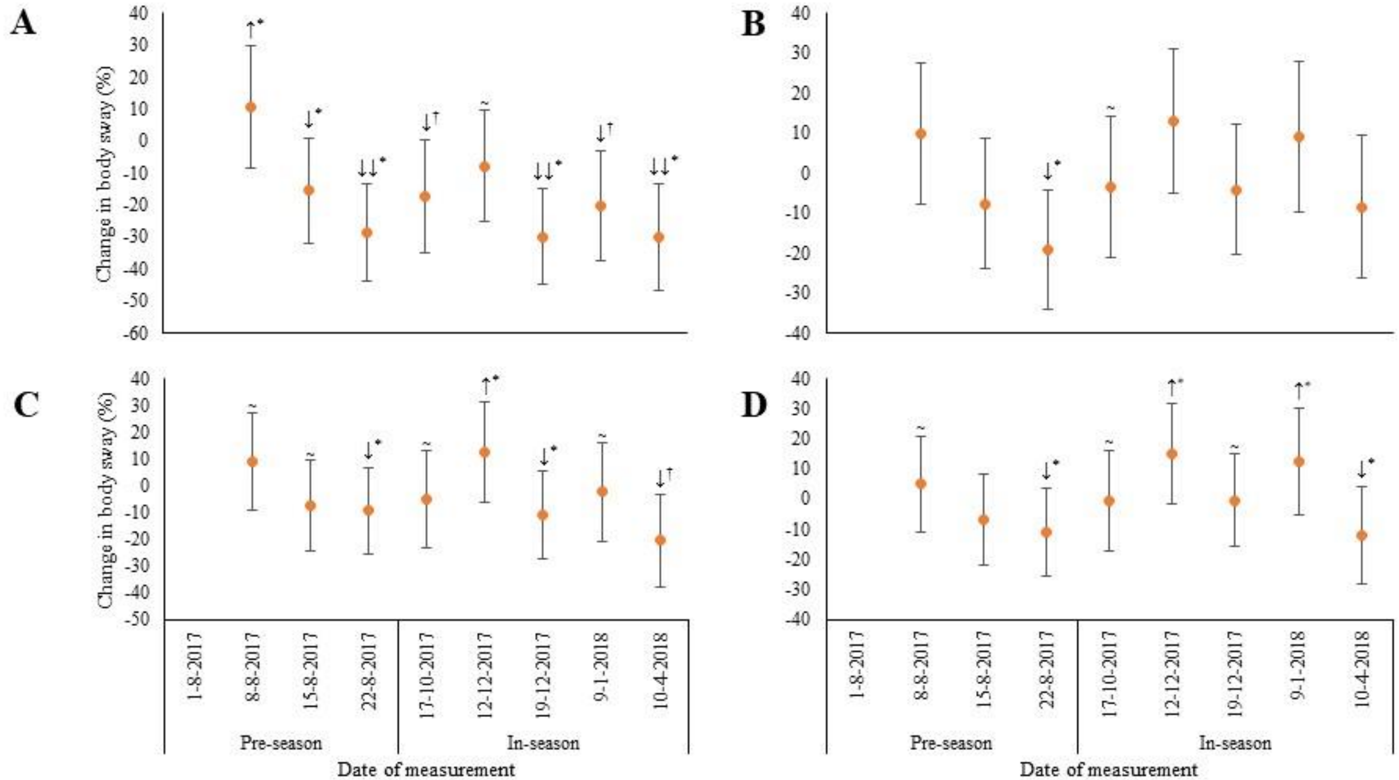

Figure A10.3. Changes in body sway in the resultant axis during all sessions compared to the baseline measurement, where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. Figure A10.3A: Average distance raw. Figure A10.3B: Average distance SA. Figure A10.3C: Maximal distance raw. Figure A10.3D: Maximal distance SA. ↑ represents a small increase in body sway compared to baseline, ↓ and ↓↓ represent a small and moderate decrease in body sway compared to baseline. ~: unclear, *: possibly, †: likely. All sessions without a symbol represent a trivial change compared to baseline.

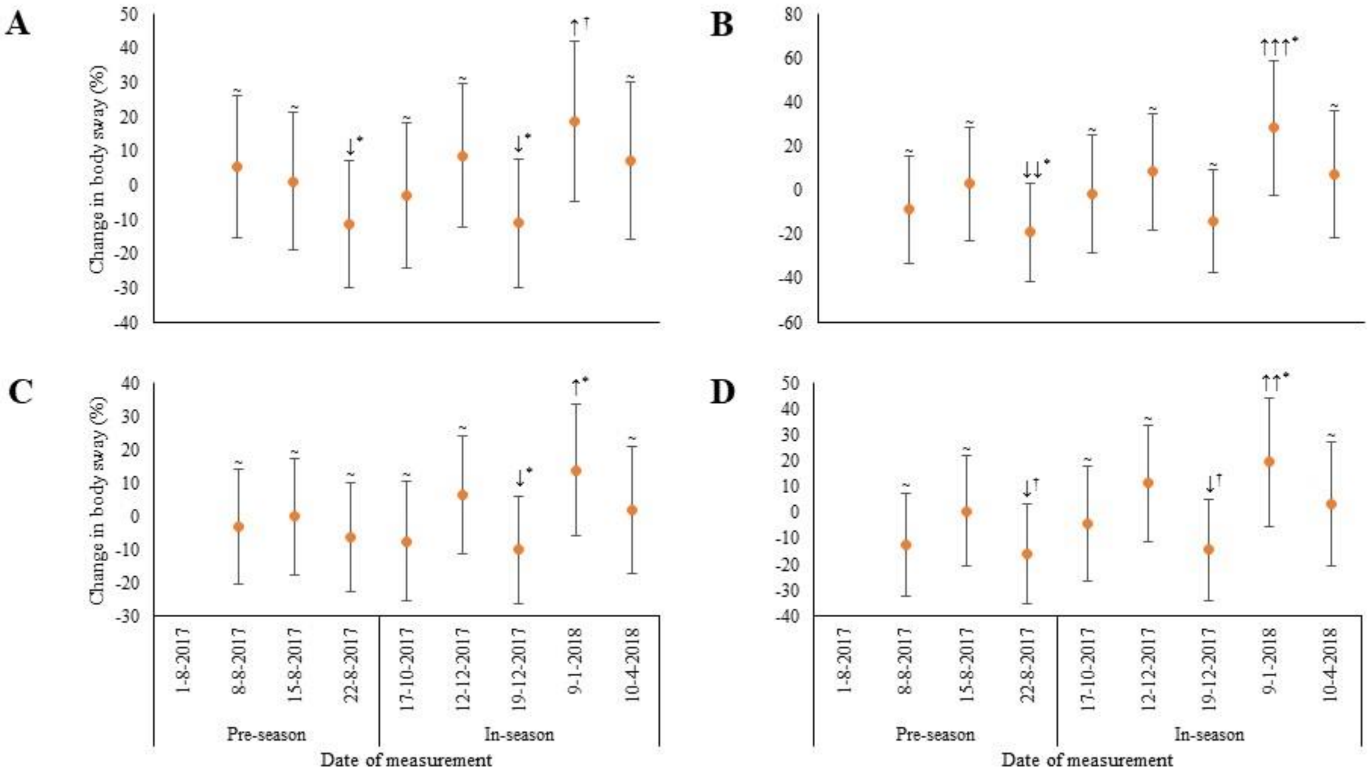

Figure A10.4. Changes in body sway in the ML axis during all sessions compared to the baseline measurement, where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. Figure A10.4A: Average distance raw. Figure A10.4B: Average distance SA. Figure A10.4C: Maximal distance raw. Figure A10.4D: Maximal distance SA. ↑, ↑↑ and ↑↑↑ represent a small, moderate and large increase in body sway compared to baseline, ↓ and ↓↓ represent a small and moderate decrease in body sway compared to baseline. ~: unclear, *: possibly, †: likely.

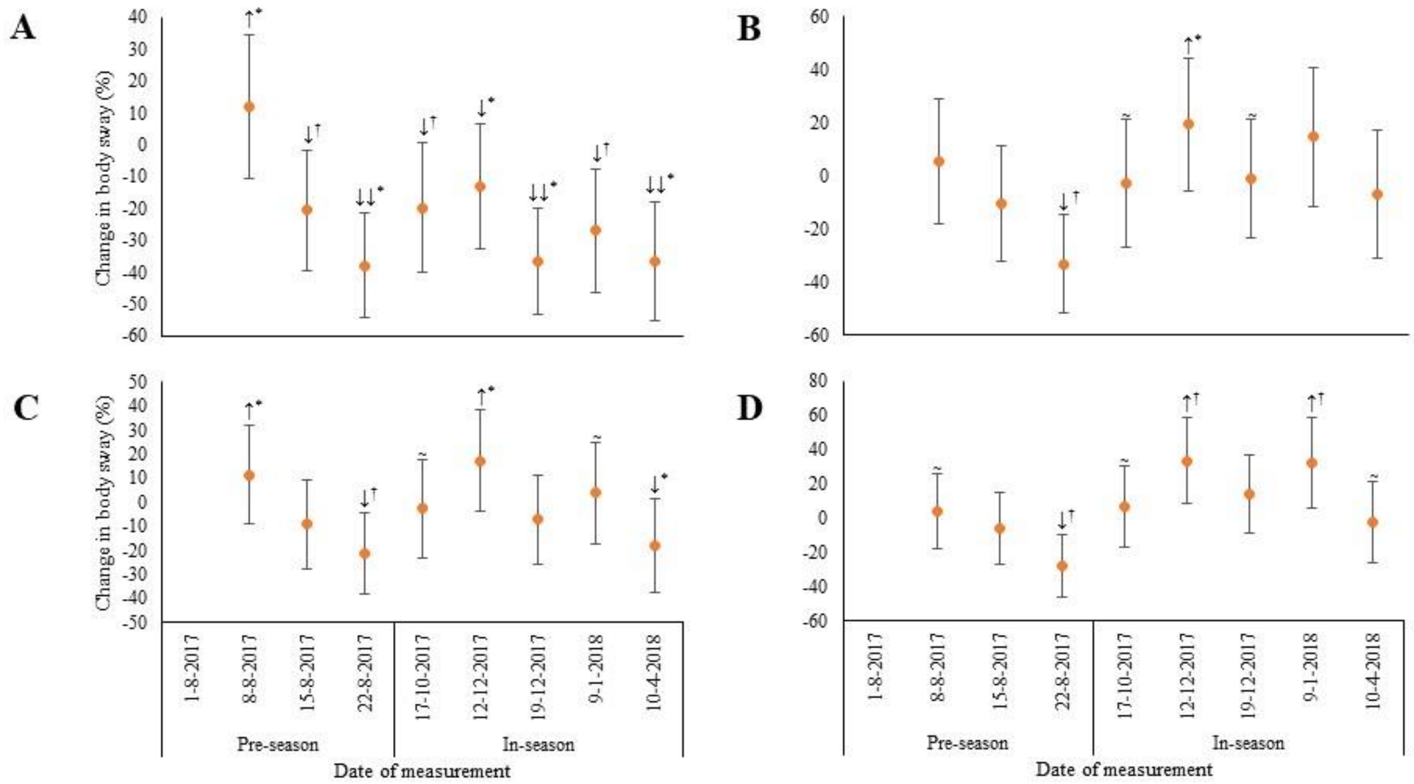

Figure A10.5. Changes in body sway in the PA axis during all sessions compared to the baseline measurement, where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. Figure A10.5A: Average distance raw. Figure A10.5B: Average distance SA. Figure A10.5C: Maximal distance raw. Figure A10.5D: Maximal distance SA. ↑ represents a small increase in body sway compared to baseline, ↓ and ↓↓ represent a small and moderate decrease in body sway compared to baseline. ~: unclear, *: possibly, †: likely. All sessions without a symbol represent a trivial change compared to baseline.

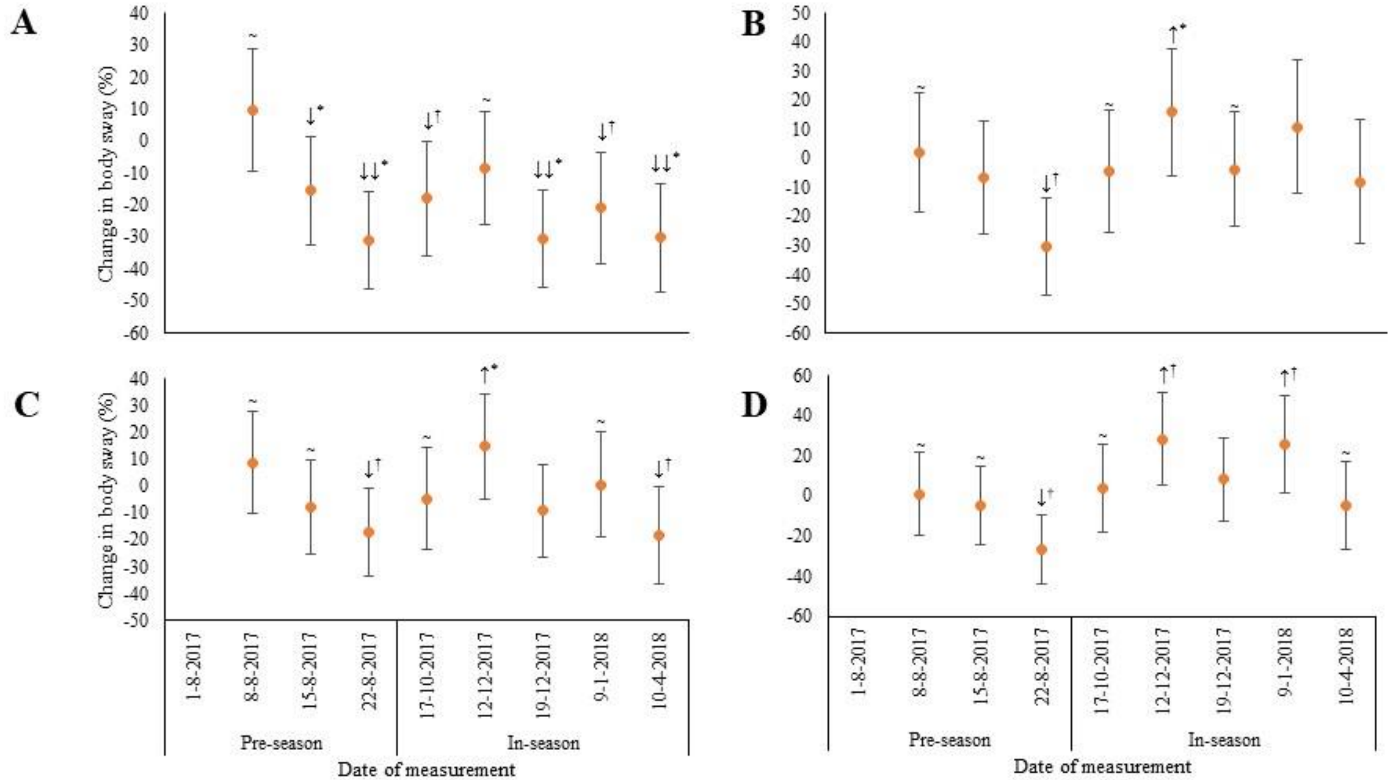

Figure A10.6. Changes in body sway in the resultant axis during all sessions compared to the baseline measurement, where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. Figure A10.6A: Average distance raw. Figure A10.6B: Average distance SA. Figure A10.6C: Maximal distance raw. Figure A10.6D: Maximal distance SA. ↑ represents a small increase in body sway compared to baseline, ↓ and ↓↓ represent a small and moderate decrease in body sway compared to baseline. ~: unclear, *: possibly, †: likely. All sessions without a symbol represent a trivial change compared to baseline.

Appendix 11. Figures of seasonal variability of the body sway speed variables

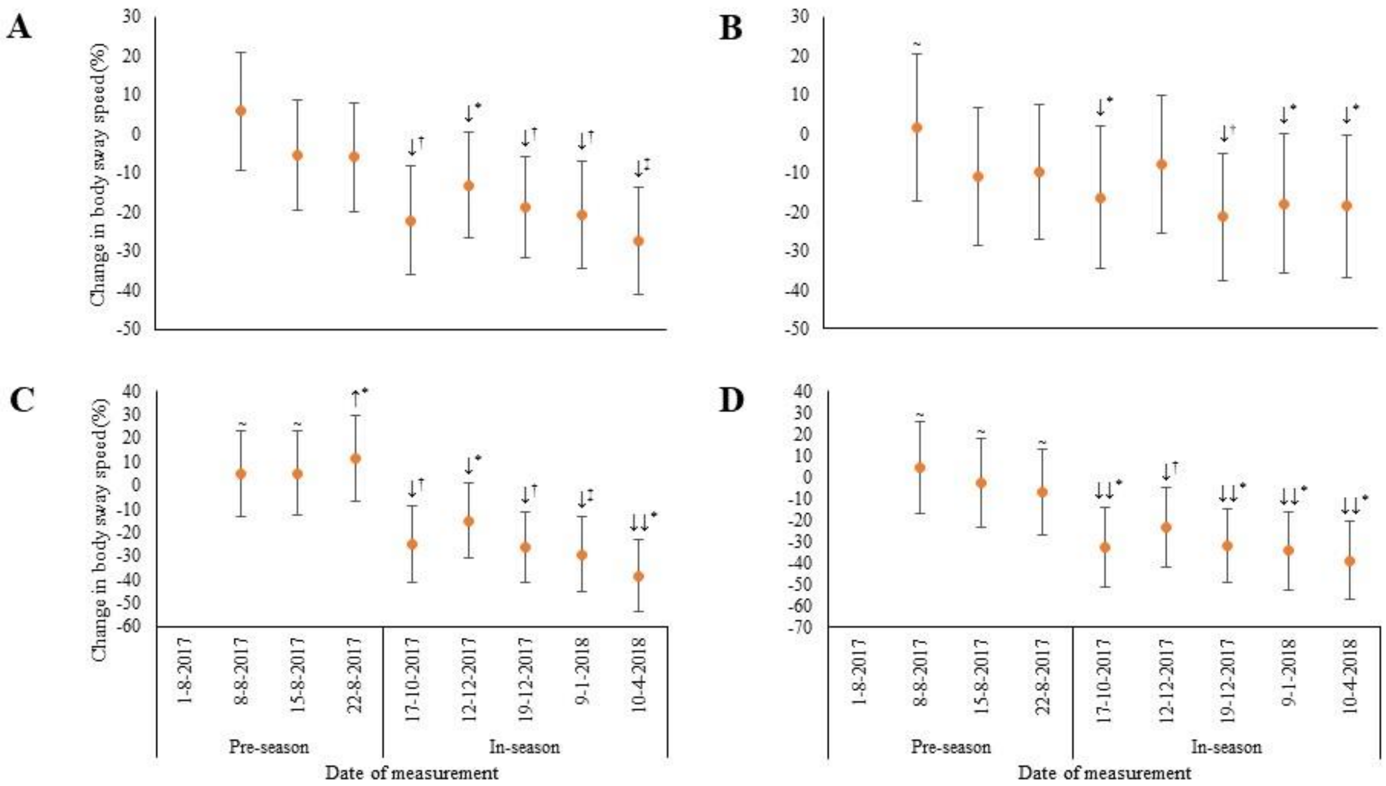

Figure A11.1. Changes in body sway speed in the ML axis during all sessions compared to the baseline measurement, where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. Figure A11.1A: Average speed raw. Figure A11.1B: Average speed SA. Figure A11.1C: Maximal speed raw. Figure A11.1D: Maximal speed SA. ↑ represents a small increase in body sway speed compared to baseline, ↓ and ↓↓ represent a small and moderate decrease in body sway speed compared to baseline. ~: unclear, *: possibly, †: likely, ‡: very likely. All sessions without a symbol represent a trivial change compared to baseline.

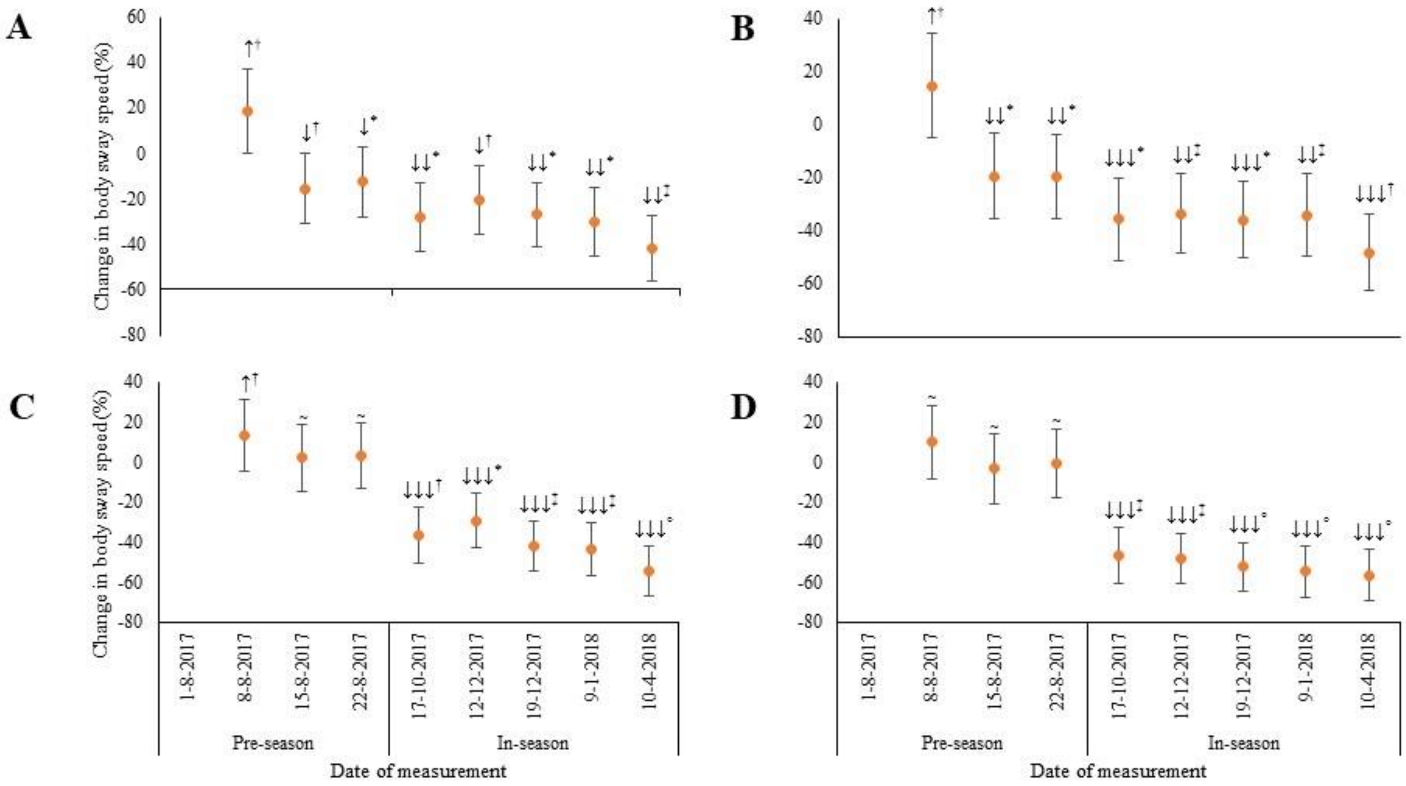

Figure A11.2. Changes in body sway speed in the PA axis during all sessions compared to the baseline measurement, where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. Figure A11.2A: Average speed raw. Figure A11.2B: Average speed SA. Figure A11.2C: Maximal speed raw. Figure A11.2D: Maximal speed SA. ↑ represents a small increase in body sway speed compared to baseline, ↓, ↓↓ and ↓↓↓ represent a small, moderate and large decrease in body sway speed compared to baseline. ~: unclear, *: possibly, †: likely, ‡: very likely, °: most likely.

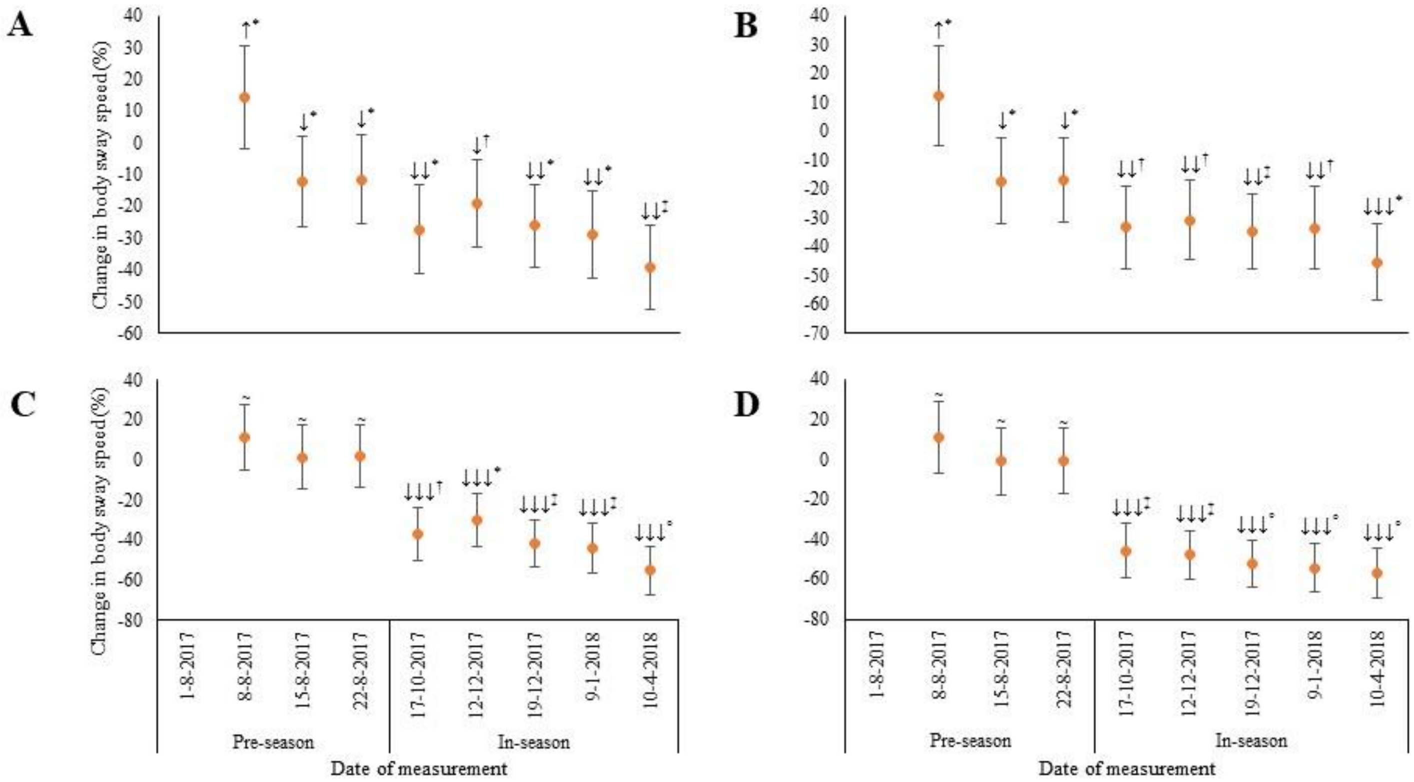

Figure A11.3. Changes in body sway speed in the resultant axis during all sessions compared to the baseline measurement, where the frame of initial contact is the first frame. The error bars display the 90% confidence limits. Figure A11.3A: Average speed raw. Figure A11.3B: Average speed SA. Figure A11.3C: Maximal speed raw. Figure A11.3D: Maximal speed SA. ↑ represents a small increase in body sway speed compared to baseline, ↓, ↓↓ and ↓↓↓ represent a small, moderate and large decrease in body sway speed compared to baseline. ~: unclear, *: possibly, †: likely, ‡: very likely, °: most likely.

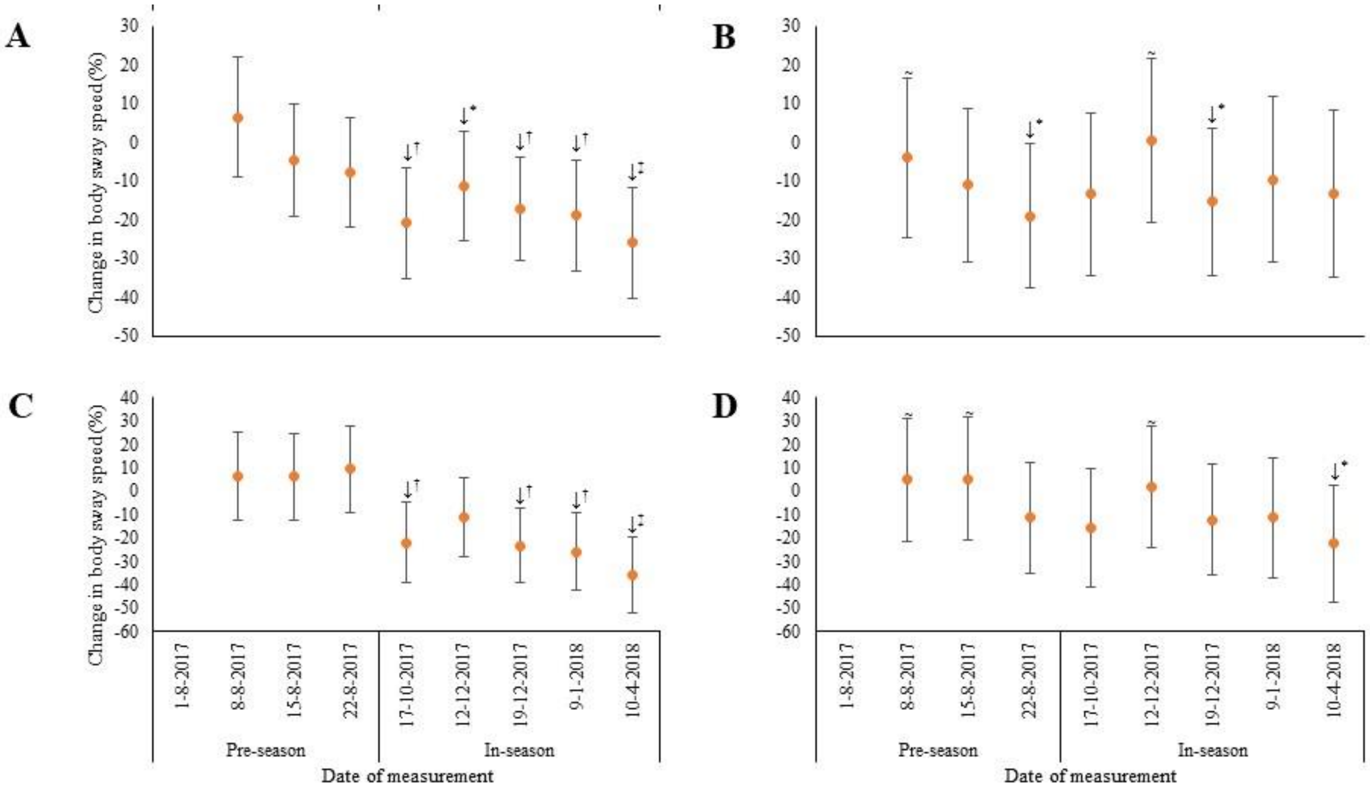

Figure A11.4. Changes in body sway speed in the ML axis during all sessions compared to the baseline measurement, where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. Figure A11.4A: Average speed raw. Figure A11.4B: Average speed SA. Figure A11.4C: Maximal speed raw. Figure A11.4D: Maximal speed SA. ↓ represents a small decrease in body sway speed compared to baseline. ~: unclear, *: possibly, †: likely, ‡: very likely. All sessions without a symbol represent a trivial change compared to baseline.

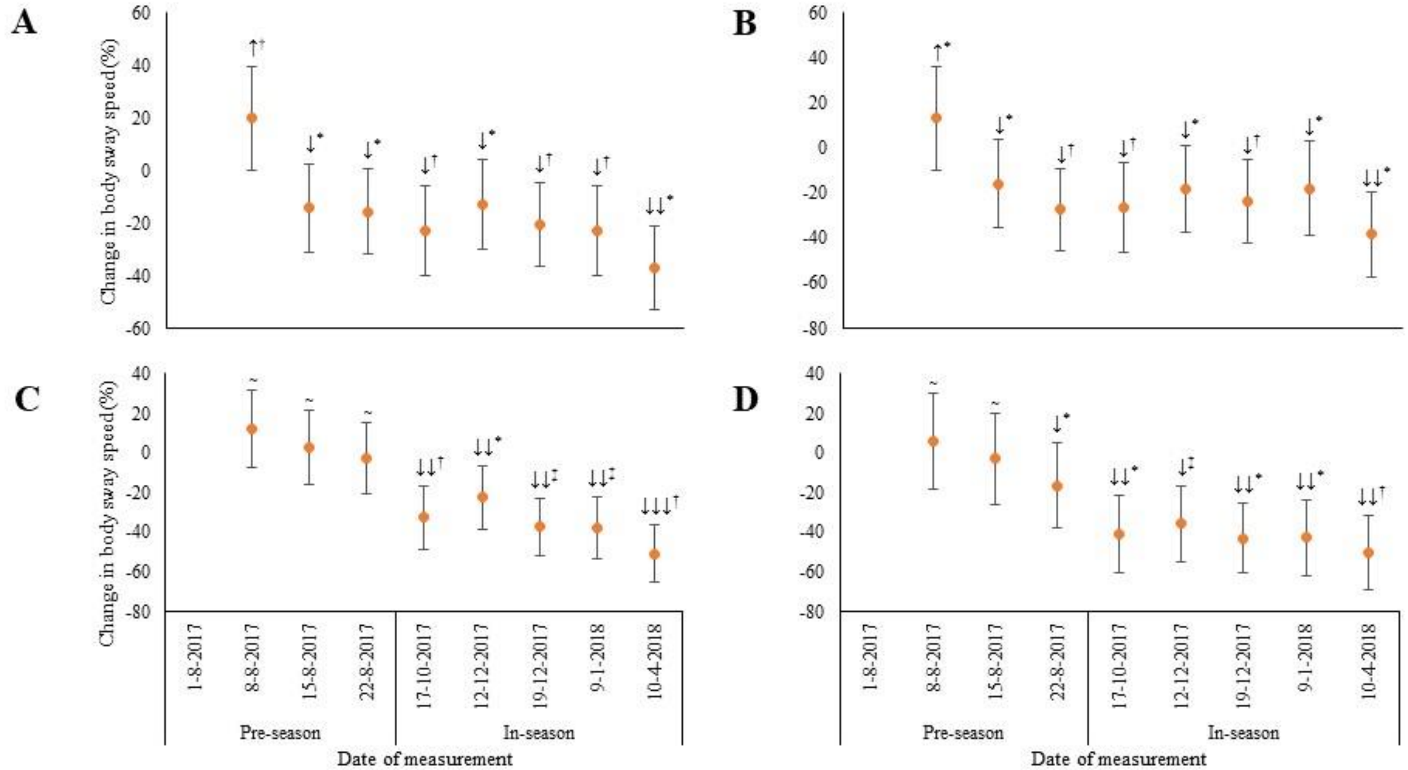

Figure A11.5. Changes in body sway speed in the PA axis during all sessions compared to the baseline measurement, where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. Figure A11.5A: Average speed raw. Figure A11.5B: Average speed SA. Figure A11.5C: Maximal speed raw. Figure A11.5D: Maximal speed SA. ↑ represents a small increase in body sway speed compared to baseline, ↓, ↓↓ and ↓↓↓ represent a small, moderate and large decrease in body sway speed compared to baseline. ~: unclear, *: possibly, †: likely, ‡: very likely.

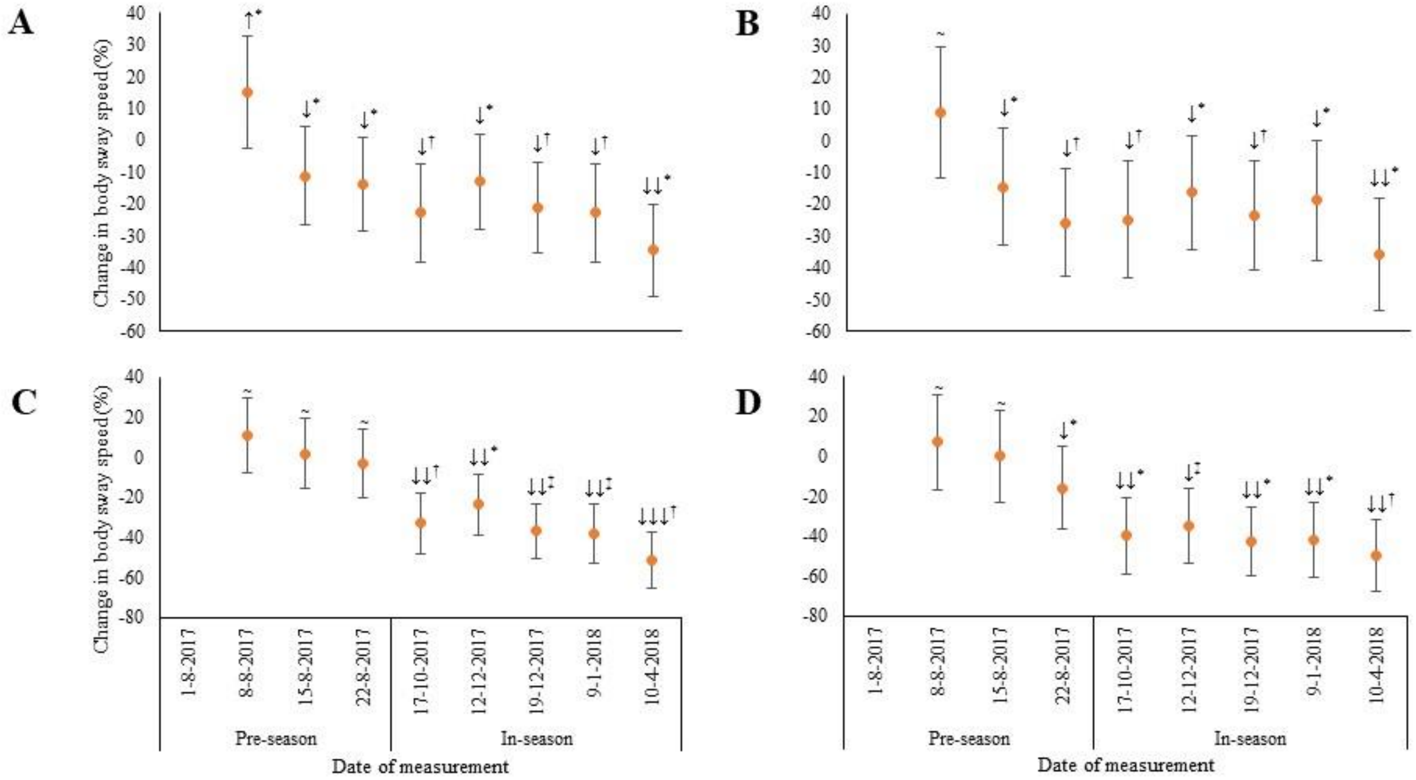

Figure A11.6. Changes in body sway speed in the resultant axis during all sessions compared to the baseline measurement, where the frame where $CoM_{PA} > BoS_{PA}$ is the first frame. The error bars display the 90% confidence limits. Figure A11.6A: Average speed raw. Figure A11.6B: Average speed SA. Figure A11.6C: Maximal speed raw. Figure A11.6D: Maximal speed SA. ↑ represents a small increase in body sway speed compared to baseline, ↓, ↓↓ and ↓↓↓ represent a small, moderate and large decrease in body sway speed compared to baseline. ~: unclear, *: possibly, †: likely, ‡: very likely.