Assessment of Postural Deviations Associated Errors in the Analysis of Kinematics Using Inertial and Magnetic Sensors and a Correction Technique Proposal

by

Monica Daniela Gomez Orozco

A thesis submitted in conformity with the requirements for the degree of Master of Applied Science
Institute of Biomaterials and Biomedical Engineering
University of Toronto

©Copyright Monica Daniela Gomez Orozco 2015
Assessment of Postural Deviations Associated Errors in the Analysis of Kinematics Using Inertial and Magnetic Sensors and a Correction Technique Proposal

Master of Applied Science
Institute of Biomaterials and Biomedical Engineering
University of Toronto
2015

Abstract
The MVN BIOMECH Awinda system has been used to analyze motion kinematics beyond laboratory conditions. However, it has a limitation in the rehabilitation field since it relies on a predefined posture to calibrate the sensors: the “N-Pose”, which is impossible to attain for some patient populations. The aims of this thesis are to assess the postural deviation error in gait kinematics measured with this system as well as two proposed correction approaches: the orientation correction (OC) and planar angle correction (PAC). After analyzing the crouch gait of four able-bodied participants it was found that the postural deviation error can be considered as a constant shift in kinematic values and that it can be corrected with both approaches. Digital images are explored as a means to capture the true body posture attained during the calibration.
Acknowledgments

I would like to thank my supervisor Dr. Jan Andrysek for giving me the opportunity to be part of his research team. It has being an honor to have him as a supervisor. I would like to thank his insightful comments and questions that led me to accomplish this research project. I also appreciate his unconditional support, patience and encouragement even during the busier times.

I owe much gratitude to my thesis committee, namely Dr. Elaine Biddiss, Dr. Karl Zabjek for their valuable feedback that along this two years motivated me to go beyond what is evident and keep finding better answers to my questions, and discover things about research that I could have not seen without them sharing their expertise.

I extend my gratitude to Dr. Jose Zariffa for accepting being part of the thesis committee and for his feedback in the final stage of the thesis that gave more insight into how this work can be improved.

I want to specially thank, Matt Leineweber and Alejandro Villaseñor who were of invaluable help with the hard work needed to put all the different parts of this project together. With all their help this work could have not been successful.

To Calvin Ngan, Jessica Tomasi, Arezoo Eshraghi, Francisco Morales, Emma Rogers, Kamil Pieaszschinki and Hank Yu Lee, I feel lucky to have shared the laboratory with you. It was always good to work in your company. Thank you for all your help and laughs in different occasions.

In general I want to thank all the members of the PROPEL lab, working with all of you always motivated me to do my best to contribute my part to this great team.

I also owe my appreciation to the BRI staff, they make this institute a wonderful environment to work in, not only because of the quality in work everybody does but also because of their warmth that make the work here being so enjoyable.
To my mom Monica, my dad Hector, my brother Hector and the rest of my family, because you were always supporting and loving me from home. You always trust in me and believe that I can achieve great things. That was the best source of motivation to complete this dream.

To my friends, the ones I knew before this period of my life and the ones I made throughout this process. Their friendship and words of encouragements made the hard work be lighter and fun. I specially thank Jahir Gutierrez for his continuous enthusiasm in learning that always inspires me to expand my creativity.

Finally, I am also thankful to CONACYT, SEP and NSERC, without their financial support, I could have not had the opportunity to learn so much during this past two years.
# Table of Contents

Abstract .............................................................................................................................................. ii  
Acknowledgments ................................................................................................................................ iii  
Table of Contents ................................................................................................................................... v  
List of Tables ........................................................................................................................................... viii  
List of Figures ......................................................................................................................................... x  
List of Appendices .................................................................................................................................... xiii  
List of Acronyms ....................................................................................................................................... xiv  

1. Introduction ........................................................................................................................................ 1  
   1.1 Overview ..................................................................................................................................... 1  
   1.2 Outline ....................................................................................................................................... 1  

2. Background and motivation .................................................................................................................. 3  
   2.1 Kinematics in Gait Analysis ........................................................................................................... 3  
   2.2 Joint angles ................................................................................................................................. 3  
   2.3 The gold standard in the assessment of kinematics ......................................................................... 4  
   2.4 Marker-less camera systems .......................................................................................................... 5  
   2.5 Inertial- and magnetic-based measurement units (IMMUs) in the assessment of kinematics .......... 6  
   2.6 Common problems of IMMUs in the assessment of kinematics .................................................... 7  
   2.7 The problem of posture during calibration ................................................................................... 7  
   2.8 Performance of MVN BIOMECH compared to the Gold Standard ............................................. 10
4.5 Assessing the effect of postural deviations in gait kinematics (Objective 1)........... 35

4.6 Applying OC and PAC towards improving the accuracy of gait kinematics (objective 2).................................................................................................................. 36

4.6.1 Orientation Correction (OC).............................................................................. 37

4.6.2 Planar Angle Correction (PAC).......................................................................... 45

5. Results...................................................................................................................... 48

5.1 Data ...................................................................................................................... 48

5.2 Baseline Error...................................................................................................... 48

5.3 Assessing the effect of postural deviations in gait kinematics (Objective 1)........... 53

5.4 Applying OC and PAC towards improving the accuracy of gait kinematics (objective 2).................................................................................................................. 63

6. Discussion............................................................................................................. 76

6.1 Key findings........................................................................................................ 76

6.2 Strengths and limitations.................................................................................... 77

6.3 Clinical relevance................................................................................................ 83

6.4 Future work......................................................................................................... 85

7. Conclusion............................................................................................................. 88

References.................................................................................................................... 90

Appendices.................................................................................................................. 104
List of Tables

Table 1: Description of reference frames and variables used in the estimation of kinematics by MVN system [77].................................................................................................................................................. 21

Table 2: Matrix representation of the MVN equations................................................................................................................. 23

Table 3: Experimental conditions............................................................................................................................................. 30

Table 4: CMC of all participants corresponding to the ideal calibration condition. Values ≥0.995 are written as 1. Shading indicates poor (lightest) to excellent (darker) correlation. CMC complex values are displayed as Not a Number (NaN).......................................................... 50

Table 5: Mean and standard deviation of DIFFD of the baseline error of all participants. Results are shown in degrees. The DIFFD of the FE angles are in bold and italics. Knee and ankle show DIFFD about zero and small standard deviation because of the excellent agreement between PiG and MVN. Hip DIFFD has high mean because PiG values are greater than MVN indicating a shift towards extensions values; as a result of different anatomical frame definition of the pelvis due to sensor location. However the standard deviation is low given the excellent agreement. .. 52

Table 6: CMC of all participants corresponding to the crouch gait experimental condition. Values ≥0.995 are written as 1. Shading indicates poor (lightest) to excellent (darker) correlation. CMC complex values are displayed as not a number (NaN).......................................................... 57

Table 7: Mean and standard deviation of the derivative of the difference between PiG and MVN corresponding to crouch gait experimental condition................................................................. 62

Table 8: CMC values corresponding to the crouch gait experimental condition before and after orientation and planar angle correction. Values ≥0.995 are represented as 1.00. Shading indicates poor (lightest) to excellent (darker) correlation. The different colors also correspond to the different conditions. ...................................................................................................................... 69

Table 9: Summary of DIFFD range across participants. The range of the mean of DIFFD across participants is shown for the baseline error, the CG condition as well as the two correction
approaches. Results are shown in degrees. Values in bold and italics represent the main improvements in FE angles of all joints. Underlined values represent improvement in other planes and values in italics represents errors added after the correction was applied. Notice how the range is smaller for the PAC correction where there was an improvement.

Table 10: Comparison of the joint angle of the static posture according to PiG, planar digital images, the difference between them and the percentage of error of the difference relative to PiG (DIFFS displayed as 0.00 had a value of 0.001 degrees and 0% error represents an error <0.002%). Results are shown in degrees. All percentages are shown as positive values.
List of Figures

Figure 1: a) N-pose used to calibrate MVN system. b) Assumed coordinate systems of segments during N-Pose [77] ......................................................................................................................................................... 10

Figure 2: Instrumentation of participant with sensors and markers. Front, back and right side views. .................................................................................................................................................................................. 18

Figure 3: Location of markers and known and unknown joint centers used in the "Chord function" used to find unknown joint centers with the Plug-in Gait model. ........................................... 19

Figure 4: Coordinate system of the Global Reference Frame (left) and coordinate system of Body Reference Frames (right). Sensor Reference Frame is drawn on the thigh sensor [77]. ............... 22

Figure 5: Segment definition with the MATLAB script used to measure planar joint angles .... 25

Figure 6: Experiment setup showing were the calibration spot and the beginning-of-trial spot are. .................................................................................................................................................................................. 28

Figure 7: Detection of gait cycles with the vertical acceleration of the shank sensor. .............. 32

Figure 8: a) Body segment represented as a vector in the Global Reference Frame. b) Segment deviation represented as a vector rotated by rotation matrix R. ................................................................. 42

Figure 9: Kinematics of the left limb of participant-1 according to PiG (blue) and MVN (red). The top right numbers are the mean of the CMC values of all the 9 gait cycles......................... 49

Figure 10: Each subplot shows the difference of kinematics of the left limb of participant-1, calculated according to PiG and MVN (DIFFD) as a function of time. The bars indicate the range within which DIFFD lies. On the right of each subplot the boxplot of DIFFD can be seen........... 51

Figure 11: Static difference in joint angle (DIFFS) measurement between PiG and MVN during calibration posture of the IC condition (green) and the CG condition (red)......................... 55
Figure 12: Kinematics of the crouch gait experimental condition recorded with PiG (blue) and MVN (orange). The numbers represent the CMC value.

Figure 13: Difference between PiG and MVN corresponding to crouch gait experimental condition (red) and baseline error (green) of left limb of participant-1. The bars represent the range within which this difference lies. The boxplots on the right of each graph represent the mean and dispersion of the difference throughout the gait cycle.

Figure 14: Mean and standard deviation of the difference deviation between PiG and MVN (DIFFD) corresponding to crouch gait experimental condition (red) and baseline error (green) for all participants; left limb (above) and right limb (below).

Figure 15: Derivative of DIFFD corresponding to the left limb of participant-1.

Figure 16: Comparison between the raw data, DIFFD and the derivative of DIFFD.

Figure 17: Kinematics of Crouch Gait according to PiG (blue) and MVN after the orientation correction was applied (MVN OC, orange).

Figure 18: Difference between PiG and MVN corrected corresponding to the baseline (green) and the crouch gait Kinematics corrected with the orientation correction (orange).

Figure 19: Kinematics of Crouch Gait according to PiG (blue) and MVN after the planar angle correction was applied (MVN PAC, purple).

Figure 20: Difference between PiG and MVN corrected corresponding to the baseline (green) and the crouch gait Kinematics corrected with the orientation correction (orange).

Figure 21: Mean and standard deviation of the difference deviation between PiG and MVN (DIFFD) according to the baseline error (green) and the crouch gait experimental condition before (red) and after orientation correction (orange) and planar angle correction (purple).

Figure 22: CMC values corresponding to crouch gait before correction and after correction with orientation and planar angle corrections. CMC that were complex numbers are not displayed. CMC of ankle AA are all complex numbers.
Figure 23: Dispersion of the difference between PiG and planar digital images (DIFFS) corresponding to each joint across all participants. ................................................................. 74

Figure 24: Kinematics of Crouch Gait according to PiG (blue) and MVN after the planar angle correction was applied (MVN PAC with planar digital images, black). .............................................. 74

Figure 25: Mean and standard deviation of the difference deviation between PiG and MVN (DIFFD) according to the baseline error (green) and the crouch gait experimental condition before (red) and after and planar angle correction using the angles measured from the planar digital images (black).............................................................................................................................................................................. 75
List of Appendices

Appendix A. Comparison between Plug-in Gait (PiG) and MVN BIOMECH coordinate systems ................................................................................................................................................................................................. 104

Appendix B. Procedure to find MVN equations to calculate kinematics ................................................................. 97

Appendix C. PiG segments orientation ................................................................................................................................................................................................................. 98

Appendix D. Comparison of PiG and MVN kinematics .......................................................................................... 100

Appendix E. Code use to calculate planar joint angles form digital images .......................................................... 103

Appendix F. Effect of environment on kinematics measured with the MVN system ................................................. 106

Appendix G. Gait Laboratory Mapping .................................................................................................................. 108

Appendix H. Posture reproducibility ...................................................................................................................... 112

Appendix I. Difference between PiG and MVN during the static trial (DIFFS) corresponding to the Ideal Calibration (IC) and Crouch Gait (CG) condition ................................................................................................................................. 113

Appendix J. Difference between PiG and MVN during the dynamic trial (DIFFD) corresponding to the Ideal Calibration (IC) and Crouch Gait (CG) condition ................................................................................................................................. 115

Appendix K. Magnetic declination of the gait laboratory ......................................................................................... 118
<table>
<thead>
<tr>
<th>Acronym</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>AA</td>
<td>Ab-Adduction</td>
</tr>
<tr>
<td>ASIS</td>
<td>Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>BRF</td>
<td>Body Reference Frame</td>
</tr>
<tr>
<td>CG</td>
<td>Crouch gait</td>
</tr>
<tr>
<td>CMC</td>
<td>Coefficient of Multiple Correlation</td>
</tr>
<tr>
<td>CMC-WD</td>
<td>Coefficient of Multiple Correlations within-day</td>
</tr>
<tr>
<td>DIFFD</td>
<td>Difference (dynamic trial)</td>
</tr>
<tr>
<td>DIFFS</td>
<td>Difference (static trial)</td>
</tr>
<tr>
<td>FE</td>
<td>Flexion-Extension</td>
</tr>
<tr>
<td>GRF</td>
<td>Global Reference Frame for MVN Biomech</td>
</tr>
<tr>
<td>IC</td>
<td>Ideal calibration</td>
</tr>
<tr>
<td>IE</td>
<td>Internal-External rotation</td>
</tr>
<tr>
<td>IMMU</td>
<td>Inertial and Magnetic Measurement Unit</td>
</tr>
<tr>
<td>ISB</td>
<td>International Society of Biomechanics</td>
</tr>
<tr>
<td>JRC</td>
<td>Joint Rotation Convention</td>
</tr>
<tr>
<td>KiC</td>
<td>Kinematic Coupling Algorithm</td>
</tr>
<tr>
<td>LBC</td>
<td>Lower body configuration</td>
</tr>
<tr>
<td>Acronym</td>
<td>Description</td>
</tr>
<tr>
<td>---------</td>
<td>--------------------------------------------------</td>
</tr>
<tr>
<td>MVN</td>
<td>MVN BIOMECH Awinda</td>
</tr>
<tr>
<td>N-Pose</td>
<td>Neutral Pose</td>
</tr>
<tr>
<td>OC</td>
<td>Orientation Correction</td>
</tr>
<tr>
<td>PAC</td>
<td>Planar Angle Correction</td>
</tr>
<tr>
<td>PiG</td>
<td>Plug-in Gait</td>
</tr>
<tr>
<td>PiGRF</td>
<td>Plug-in Gait Reference Frame</td>
</tr>
<tr>
<td>PSIS</td>
<td>Posterior Superior Iliac Spine</td>
</tr>
</tbody>
</table>
1. Introduction

1.1 Overview

The MVN BIOMECH Awinda system (MVN system) has been effectively used to analyze the kinematics of human motion in fields like sports science, ergonomics or rehabilitation[1]. The main advantage of this system is that it enables the study of kinematics beyond laboratory conditions, adding an advantage over the gold-standard camera-based system. The MVN system uses inertial and magnetic sensors to track the motion of the body segments, and then with the addition of a biomechanical model different kinematics parameters can be calculated [2] and translated into clinically meaningful information [3] that can be helpful for clinicians to evaluate treatments.

An important downfall of this system is that it relies on a predefined posture known as the “N-Pose” to calibrate the sensors. In the rehabilitation field, it is well known that this posture is sometimes impossible to be attained by some populations due to several factors (e.g. muscle weakness, bone deformity) [4]. Hence, the use of this technology is limited in this field since postures deviated from the N-Pose would produce wrong calibrations yielding inaccurate results.

The aim of this thesis is to assess the effect that these postural deviations have on the measurement of kinematics done with the MVN BIOMECH system. Additionally, two correction approaches based on the information of the “true body posture” assumed at the moment of calibration, the orientation correction (OC) and the planar angle correction (PAC), are proposed and evaluated. Planar digital images are explored as a means to provide the information of the true body posture.

1.2 Outline

This thesis is written in a single manuscript format where all its parts are related to the same objectives, meaning that there are no independent chapters each with its own objectives, results and discussion. Section 1 presents the “Introduction” where the reader can learn about the overall goal of the thesis together with the description of how it is addressed in the different sections of
the manuscript. In the “Background” contained in section 2, the reader will know about the context of the problem addressed in this thesis as well as a review of the literature concerning the related work. In this section a brief description of the solutions proposed to this problem, the OC and the PAC, are also introduced. Section 3 states the two objectives of the thesis each of them followed by particular questions that are sought to be answered with the methodology. Section 4 contains the “Methods” and it explains the instrumentation and protocol that were used during the experiments to collect the data as well as parameters used to analyze this data for each of the two objectives. Additionally, this section includes a detailed description of both correction approaches and how they are implemented in this study. Section 5 presents the “Results”, organized according to how objectives and questions were exposed in section 3. Each question is followed by the results and explanation as to how they answer the question. Following the results is the “Discussion” in section 6 that interprets the results in a more general perspective. This section also expands on the strengths and limitations as well as alternative routes that can be taken to address the problem alluded to in this work. As a part of this, the clinical relevance and future work suggest steps that can be taken towards the development of a technique that could correct for postural deviation in broader scenarios. Finally the “Conclusion” is presented in section 7 summarizing briefly the overall goal of the thesis and main findings to elaborate a “take-away” message.
2. Background and motivation

2.1 Kinematics in Gait Analysis

Gait analysis in the rehabilitation field is the study of human walking. It aids clinicians to diagnose abnormal gait patterns, decide about treatments and evaluate the patient’s progress. Techniques vary from basic visual assessment and surveys to complex equipped laboratories [4], [5].

According to a review done by Muro-de-la-Herran et al. (2014) [5], current technologies used to analyze gait can be either semi-subjective or objective. Semi-subjective techniques consist of the observations of specialists when the patient is performing a walking test. Objective techniques can be divided into floor sensors, image processing techniques and sensors placed on the body. Objective techniques, in contrast to semi-subjective techniques, provide accurate and quantifiable information that would not be obtainable by human observation alone. One of the characteristics that cannot be accurately measured using observational gait analysis is kinematics.

Generally, the term kinematics refers to the study of motion of an object from the perspective of displacement, velocity and acceleration. In gait kinematics these physical quantities are used to describe the translation and rotation of body segments, which serve to estimate joint angles. This thesis describes a technique that enhances the analysis of joint angles; therefore, particular attention is placed to this parameter.

2.2 Joint angles

Briefly, joint angles are the relative angles between two contiguous body segments, and they are used to describe the rotational motion of the distal segment relative to the proximal segment. Depending on how they are parametrized, these rotations can be understood as being performed about 3 different axes in a sequence of rotations (conventional definition of Euler) [6], [7]; about a single axis called the screw or helical axis (the screw or helical method) [8], [9]; and about 2 axes that are fixed to the body segments and one floating axis, result of the cross product of the two body-fixed axes (parametrization called joint rotation convention, JRC) [10].
Despite the parametrization approach, joint angles can be translated into the Clinical Reference System [3] as flexion-extension (FE), abduction-adduction (AA) or internal-external Rotation motion (IE) depending on whether the motion is in the sagittal plane of the segment, away or towards it in the frontal plane, or in the transverse plane of the segment respectively.

With this clinically meaningful description of joint angles, one can look at the changes in joint angles over the gait cycle to characterize the healthy as well as the pathological gait and as mentioned earlier this knowledge can be useful for clinicians. For instance, it was found that foot kinematics is related to the walking efficacy of adolescents with CP which can be used to decide and evaluate treatments [11]. It can also help to improve the design of assistive devices. Kinematic along with Kinetic data, for instance, are used to compare the gait of subjects wearing different prostheses [12] to determine which one aids the amputee to achieve a more efficient gait.

2.3 The gold standard in the assessment of kinematics

The gold standard technology used to assess kinematics is based on motion capture using camera recordings. Depending on the number of cameras used, kinematics can be estimated in 2 dimensions (2D) or 3 dimensions (3D). It can also be done with or without markers.

Systems with markers require the use of reflective (like Vicon [13]) or active markers (like Optotrac [14]) fixed to the subject’s body landmarks to provide reference points for the tracking algorithms. Different cameras detect the centroid of each marker from different perspectives and using the principle of triangulation their position in 3D space is estimated and tracked over time. Then with the addition of a model that relates each marker to a body segment and links each segment with a type of joint (e.g. hinge, pivot, ball-and-socket joints), joint angles are calculated.

The camera-based technology using markers is accurate and reliable, and it has been widely used for research purposes. However, given that tracking markers is essential for estimating kinematics, lighting conditions and location of all the cameras become essential making such systems only suitable for laboratory settings. This restriction is a disadvantage given that constraining the space might modify the subject’s normal walking [15]. Moreover, even when they are used in these conditions markers are sometimes occluded by objects and other body
parts increasing the time of data processing to estimate the lost data, in consequence adding the possibility of estimation error.

Additionally, the cameras require a lengthy set up and calibration process, making this technology unsuitable for day-to-day assessments in the clinical setting. Therefore, clinicians rely largely on the use of subjective techniques and miss relevant information about their patients’ mobility in real-life environments, preventing them from assessing their real performance.

Kinematics obtained in real-life environments are relevant not only to clinicians, but also to researchers that work towards developing technologies to improve independent ambulation. Assessment of kinematics of amputees wearing a prosthetic knee, for example, is done in the laboratory [16] using camera-based systems, whereas evaluation of the efficacy in daily life is done by means of questionnaires [17]. Ideally, accurate motion tracking systems that do not rely on cameras and can work in any environment would provide a means to objectively study and evaluate movements under a variety of life-relevant conditions.

2.4 Marker-less camera systems

Systems that do not require markers can be regular video recording cameras using [18]–[21] from which joint angles are extracted manually or using a commercial or customized software[22]–[27]; in such cases only the angle in the plane of capture can be measured[28], [29]. In the case where 3D angles are needed more than one camera can be used to apply stereo-triangulation techniques.

However, more modern systems such as the Kinect Sensor [30], a depth camera that functions under the principles of structured light [5], use some image processing combined with machine learning techniques to estimate the position and orientation [31] of joints. This information can then be used to estimate joint angles. The Kinect option is convenient because there is no problem with missing information due to marker occlusion and can be used in an inner space without the need of multiple cameras for stereo-triangulation. Recent studies have shown the potential of Kinect to be used as a tool to assess kinematics in the work place [32]. However, the
accuracy of Kinect to estimate the joint centers and therefore the joint angles, is still questionable [33]–[35].

In spite of the many advantages of marker-less camera-based technologies, their caveats including restricted utilization in outdoors environments, limited capture volumes, and impossible or inaccurate measurement due to occlusion under certain conditions, have prompted the pursuit of alternative motion tracking technologies.

2.5 Inertial- and magnetic-based measurement units (IMMUs) in the assessment of kinematics

To overcome the aforementioned limitations of camera-based assessment, an alternative technology has been emerging over the past two decades. This technology consists in the use of wearable inertial- and magnetic-based sensors [5], [36]. The goal of this technology is to be able to analyze human movement in real-life conditions.

The most commonly used sensor is the inertial measurement unit (IMU), which is comprised of one tri-axial accelerometer and one tri-axial gyroscope packed together [36]. The accelerometer enables the acquisition of sensor position by estimating the vertical virtual line traced by the component of gravity and double integrating the acceleration signal. When the angular velocity from the gyroscope is also integrated the orientation between the current instant and the previous initial position is obtained. Another usual combination includes also a tri-axial magnetometer: the inertial and magnetic measurement unit (IMMU). This additional magnetometer corrects the horizontal orientation by estimating the angle with respect to the Earth’s magnetic field, like a compass.

Human movement that has been studied using this technology varies from kinematics of only one joint (e.g. ankle, knee or hip) to whole body kinematics. Commercially available systems that allow for whole body kinematics assessments include the MVN BIOMECH system [37], FAB system [38], IGS-Cobra [39] and some others developed by research laboratories [40].

Whole body kinematics assessment is achieved when several sensors are placed on different body segments creating a network and, based on a biomechanical model, the position of each
sensor with respect to the others can be estimated. Then the position and orientation over time of
the segments to which the sensors are attached are known and the angle between segments can
be calculated. Because these systems do not require cameras and are not constrained to a
calibrated capture area, they enable the assessment of kinematics in essentially any environment.

2.6 Common problems of IMMUs in the assessment of
kinematics

As in every measurement system, inertial- and magnetic-based sensors, either used alone or in
networks, present some challenges. Although theoretically the position and orientation of the
sensors can be accurately estimated, in practice these systems are prone to errors caused by to
drift, especially with long data sets. Moreover, in the presence of ferromagnetic objects,
distortion can occur during the detection of Earth’s magnetic field by the magnetometer [41],
thus affecting the accurate estimation of the true heading (horizontal inclination from the
magnetic north).

Much research has been dedicated to reduce the aforementioned errors in wearable IMMUs.
With regards to drift error the most common solution is the use of Kalman filter to reduce the
noise of the estimation [42]–[48]. Other solutions have been resetting the estimation when
acceleration and angular velocity are known to be zero, during heel strike for instance[49]–[51];
having two sensors in tandem [52]; using de-drifted integration [53]; adding joint limit constrains
[54], [55]; fusing the sensor to cameras worn on feet [56], [57]; or even adding a magnetometer
[58], [59]. However, the magnetometer itself has its own issues that have being solved by adding
heading correction to the algorithm [60] or developing new types of Kalman filters[61], [62].
The research shows that the estimation of joint angles can be significantly improved when
removing drift, and using these techniques acceptable performance of IMUs can be achieved for
the purposes of human movement analysis.

2.7 The problem of posture during calibration

There is, however, another issue that has received less attention and is related to the estimation of
kinematics when using wearable IMMUs systems. As mentioned before, the orientation and
position of a sensor can be estimated based on the information from the sensor itself, however when estimating full body kinematics, the information of how sensors are linked to one another and to the body itself needs to be known. This information is provided by a biomechanical model.

Current biomechanical models used in whole body assessment of kinematics are proposed by the International Society of Biomechanics (ISB) [63]–[65]; the model proposed by Dempster et. al. (1967) [66] used by the FAB system; and the 23 segments 22 joints model used by the MVN BIOMECH (and its wireless version MVN BIOMECH Awinda) systems from Xsens Technologies (MVN systems) which is a modification of the ISB model [67].

The latter system (MVN BIOMECH) has become very popular and has been used to assess kinematics in different situations. Examples of these studies are the analysis of movement variability during skill refinement in Sport Science [68] or the assessment of joint loading in occupational settings [69]. In the rehabilitation field, MVN BIOMECH has been used to analyze the use of wheelchairs[70]; gait stability of powered above-knee prostheses[71] and visual feedback on gait retraining for patients with Trendelenburg gait [72] among other applications that can be found on the company’s website [1].

In most of the studies mentioned above, the participants were able to perform the required calibration prior the assessment of kinematics. The one group that did study patients with gait impairments such as acute pain or motor functionality, use of prostheses and leg length discrepancy of 2cm, did not look at the joint angles but at pelvis and trunk range of motion [72]; hence the patients’ calibration posture might not have affected the outcomes.

The problem of posture during calibration arises during this setup stage. During the calibration the subject has to stand in a calibration posture, the “Neutral-Pose” or “N-Pose” [73], in which the body segments are assumed to be all aligned. To assume the N-Pose the subject has to stand on a flat floor, feet parallel one foot apart, their back straight, arms straight alongside with thumbs forward and face forward; see Figure 1a. As it can be seen in Figure 1b, the coordinate systems of all the joints are assumed to be parallel to one another.
However, many patient populations presenting skeletal deformities, muscle weakness, sensory loss, pain or impaired control cannot hold such postures. As such, postural misalignments exist that deviate from the assumed N-Pose. Therefore, when the system is calibrated with individuals in these non-ideal postures, the true starting point of the segments will differ from the assumed orientations, yielding inaccurate further estimation of kinematics. This issue has affected the result of some studies, described below.

Van den Noort et al. (2013), studied kinematics of lower limbs of 6 children with CP using the Outwalk Protocol and 1 able-bodied child. Their objective was to compare between the IMMUs-based Outwalk Protocol and a standard camera-based protocol (CAST Protocol) [74]. They measured hip, knee and ankle angles in the three planes. They found the largest difference between protocols due to offsets. The average of root-mean-square error (RMSE) in the transversal plane was less than 17° and less than 10° in the sagittal and frontal planes. When they removed the offset RMSE was less than 4°. Also, the coefficients of multiple correlations (CMC) found in this study are lower in comparison to the ones found by Ferrari et. al. [75] using the same study protocol but with 4 healthy individuals. Some of the CMC of both studies (children with CP vs. healthy adults) for hip are flexion-extension 0.88, ab-adduction 0.71 and internal-external rotation 0.66 vs. flexion-extension/ ab-adduction/ and internal-external rotation all greater than 0.85. These results suggest that differences during the calibration might have affected the offset given that the camera-based accounted for real orientation and position of body segments but the IMMUs-based system failed to do it.

In her dissertation to get the degree of MASc, Laudanski [76] studied the feasibility of using wearable IMMUs (MVN BIOMECH system) to estimate kinematics of lower limbs during stair ambulation of healthy older adults and stroke survivors. When comparing CMC between kinematics of healthy individuals and stroke survivors obtained from the IMMUs-based system, the author found that values where lower among stroke survivors than among healthy individuals, again suggesting that the IMMUs system failed to estimate kinematics accurately in non-able-bodied population. The author concludes that new calibration procedures are necessary to increase the accuracy in estimation of kinematics.
2.8 Performance of MVN BIOMECH compared to the Gold Standard

The performance of the MVN BIOMECH system used under ideal calibration conditions has been studied. To understand the errors associated to postural deviations, it is important to first understand the errors inherent to the system when compared to the gold standard camera-based system.

Jun Tian Zhang et al. [78] evaluated the MVN BIOMECH system with 10 able-bodied individuals doing three different activities: level walking, stair ascent and descent. Flexion-extension motion was more accurately measured with MVN for all three activities (CMC>96) for all joints. CMC values of the other planes vary between 0.5 and 0.85. Other studies also compared other inertial-based systems to camera-based systems and found similar results [75], [79], [80].

The consistent disagreement in the frontal and transverse plane among the studies could be explained by a difference in the definition of the anatomical frames of the systems [81].
phenomenon is called ‘crosstalk’ and it exists because the anatomical frame is misaligned and motion that is supposed to be measured in one plane is measured in a different plane.

In addition to these crosstalk errors, the sensors rely on magnetometers that use the local magnetic north as a reference to define anatomical frames. It has been shown that ferromagnetic objects in laboratories where these type of studies are usually done present heterogeneous magnetic fields [82], introducing error to the estimation of the body segment that these systems are based on to measure kinematics [41]. Given that transverse and frontal plane have a smaller range of motion, magnetic disturbances as well as the known drift error present in inertial sensors used in motion tracking, might affect these planes more.

Literature presented here suggests that these IMMUs-based systems can be utilized to analyze flexion-extension motion and very similar results to the gold standard would be obtained. However, when analyzing other planes of motion, results must be carefully interpreted.

It would be ideal to understand errors due to deviations in all three planes, however due to the limitations of the sensors this study focuses primarily on deviations in the sagittal plane although deviations in the other planes are also explored in a supplementary way.

2.9 Crouch Gait

As mentioned in the previous section IMMUs-based systems tend to perform worse in measuring angles of the frontal and transverse plane, hence, system error would hinder the ability to recognize errors due to postural deviations in these planes. In practice there are situations where even if only the sagittal plane is analyzed useful information is found.

Crouch gait is a gait pattern that is commonly found in patients with cerebral palsy. Although it can include deviation in other planes, it is characterized mainly by excessive knee flexion during stance phase, consequently, affecting the hip and ankle flexion as well [83].

The expected outcome of crouch gait treatments is that hip and knee are fully extended and foot is level on ground to reach balance [84]. These are changes that occur mainly in the sagittal plane, therefore in some clinical studies gait kinematics of the sagittal plane are used as a primary outcome measurement of treatments [84].
In several studies, analysis of the sagittal plane for this condition have proved to be sufficient for assessing crouch gait. A study trying to classify gait patterns in cerebral palsy (a challenging task given the high-variability in gait patterns of this population) found that, out of sixteen different spatio-temporal and kinematic parameters, the most explicative gait patterns to do that were hip, knee and ankle maximal extension, flexion and dorsiflexion respectively during specific stages of the gait cycle [85]. Other studies have looked at flexion extension as means to analyze crouch gait and evaluate improvement in treatment [86], [87] or classify other gait patterns [88].

Crouch gait is also found in patients that have irregularities in spinal curvatures. To analyze improvement after surgery of these patients, the knee kinematics of the sagittal plane are also analyzed [89].

Since the sensors used for this study perform better in measuring sagittal plane kinematics, crouch gait is a clinical condition that would be explored as a postural deviation. This will allow the assessment of postural deviation aimed for in this study to be done in the plane where the sensors measure most reliably, while providing clinical relevance.

2.10 Overcoming the Calibration Posture Problem

Some work has been done to address the limitation of not knowing the real position and orientation of body segments during calibration.

Picerno et. al developed a calibration device to use anatomical landmarks to find the transformation rotation between the sensor and the anatomical frames[90]. Their device has an IMMU mounted on its body and two pointers directed towards to two body landmarks (i.e. femoral epicondyles to define horizontal axis and grater trochanter and lateral epicondyle to define vertical axes). With the known orientation of the sensor in the global reference frame, the line joining both pointers in the global reference frame can also be known. With this procedure they obtain orientation of 2 vectors of the actual anatomical frame. Given that another sensor, with its own reference frame, is placed on the body segment (e.g. the thigh) they can find the transformation matrix between sensor frame and the actual anatomical frame (the one found with the device).
Another alternative was proposed by Favre et al. [91], which is based on Picerno’s work and the JRC. In this study the authors aim to find the transformation matrix between sensor and anatomical frame of shank and thigh. Instead of using the calibration device to find the vectors that define the coordinate frame of the segments, they find these vectors by measuring the velocity of the shank in antero-posterior and medio-lateral directions. After that, the vectors are translated to the anatomical frame. Although they achieve this without the need of a calibration device, they still need the assumption of alignment between the two segments to estimate the transformation matrix of the thigh sensor, which again can be violated by certain patient populations.

A third solution, called the Outwalk Protocol®, is based on a dynamic calibration. This approach was proposed by a research group from The National Institute for Insurance Against Accidents at Work and Occupational Diseases (INAIL) in Italy[75], [80], and consists of three steps: “(1) positioning the sensing units (SUs) of the IMMS on the subjects’ thorax, pelvis, thighs, shanks and feet following simple rules; (2) computing the orientation of the mean flexion–extension axis of the knees; (3) measuring the SUs’ orientation while the subject’s body is oriented in a predefined posture, either upright or supine.”[75], [80] If the subject is unable to achieve an upright posture during step 2 and 3, a supine posture is assumed with knees bent and supported by a therapeutic foam cylinder and hip, knee and ankle has to be manually measured with a goniometer.

### 2.11 Image-based Correction Technique

Although all the mentioned calibration approaches have achieved good levels of reliability, repeatability and/or accuracy [75], [90]–[92], they can be time-consuming and exhaustive for the individual, especially if they are paediatric or geriatric patients. Furthermore, this technique only includes lower body assessments (some of them only one joint [90], [91]) and if this technology is expected to expand to whole body kinematics, the procedures might become impractical due to the addition of extra measurements and steps.

Besides the mentioned protocols other calibration procedures can be developed to overcome the need for a reference posture. However, if the true posture is known from image-based techniques, for example, other approaches can also be taken. Assuming that the wrong posture
during calibration can be considered as a rotation of the body segments from the N-Pose, if the true orientation of the body segments (true orientation) is known, then it can be used offline (after the recording was done) to correct the measured body segments orientation that do not account for the postural deviation. Another alternative is to consider the postural deviations as having joint angles offset from those of the N-Pose. These offset angles are the calibration-posture true joint angles (true joint angles) that can be added to the IMU-measured gait joint angles to account for the postural deviation. In this case it is assumed that the true joint angles are offset angles at the moment of calibration, and that this offset remains constant throughout the recording.

Therefore, this study aims to assess the effects that postural deviations introduced during the calibration have on gait kinematics measured with the MVN system. It was also envisioned to establish a basis for a further development of an image-based technique that would correct this calibration error. For the latter goal two correction approaches are proposed:

1. Orientation Correction (OC): that captures the body segments orientation at the moment of calibration then rotates all of the body segment orientations measured during the motion tracking and finally uses this corrected orientation to re-estimate the kinematics.

2. Planar Angle Correction (PAC): that measures the calibration-posture true joint angles and uses them directly to correct the gait joint angles.

In order to know the true body segments orientation and joint angles for both approaches an external source of measurements is needed. For the first approach the ‘true’ body segment orientation of the calibration posture was measured with a camera-based system (Vicon) system. For the second approach, Vicon was also used, but rather than segment orientations, joint angles during the calibration posture were calculated. Since the end use of the correction technique will not require Vicon to capture the calibration posture, photogrammetric techniques used on planar digital images taken with a standard camera were also explored to measure the true joint angles required for the PAC.

To assess the effect of postural deviation in kinematics measurements and to evaluate the two correction approaches, gait joint angles were collected from 4 able-bodied participants while
simulating crouch gait (walking with flexed knees). Kinematics were measured with the MVN BIOMECH and Vicon systems simultaneously.

This study focuses on establishing a basis for a further development of technique that will allow current technologies that require the N-Pose as a predefined posture such as MVN BIOMECH, to be used on a broader population and in particular individuals who do not exhibit normal postures. Creating a correction technique would be a time-effective alternative to systems like the Outwalk protocol since it doesn’t require calibration with extra postures (i.e. supine position) and hands-on patient maneuvers. It is hoped that such technique will be useful to analyze not only crouch gait and other postural deviations but also other types of movements since it only requires the capture of the initial posture during the calibration.
3. Objectives

As described in the background, the overall goal of the project is to establish a basis for the development of a technique that will correct gait kinematics acquired after a non-ideal calibration. This research is divided into two objectives.

Objective 1: Assess the effect of postural deviations during the calibration in gait joint angles measured with the MVN BIOMECH system.

Objective 2: Assess two potential correction approaches (OC and PAC) based on the measurements done on the orientation of the true body segments and joint angles. When the subject is unable to achieve an ideal calibration posture, these corrections would improve the accuracy of kinematics measured with MVN BIOMECH system.

The questions corresponding to each of these two objectives are summarized below:

Objective 1

I. In gait kinematics measured with the MVN BIOMECH system, what is the error associated with postural deviations (during the calibration)?

II. Is the error constant throughout the gait cycle?

Objective 2

I. If the true orientation of body segments is known, how well will the orientation correction (OC) reduce the error?

II. If the true joint angles are known, how well will the planar angle correction (PAC) reduce the error?

III. Are OC and PAC approaches equally effective in reducing the error?

If joint angles can be corrected using the PAC:

IV. Can angles measured from planar digital images, acquired using a standard optical digital camera, be used to implement the PAC?
4. Methods

Sections 4.1 to 4.5 include the description of the methods that were common in accomplishing both objectives. Specific methods are described in sections 4.6.

4.1 Instrumentation

4.4.1 Vicon system

The Vicon Nexus1.8.5 software connected to 7 MX13 motion-capture cameras (ViconPeak, Lake Forest, CA, USA) was used. 60Hz sampling rate was chosen to match the sampling rate of the MVN sensors. Sixteen optical markers were placed on lower half of each participant using the Plug-in Gait (PiG) model: both anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), lateral side of mid-thighs, lateral condyles of knees, lateral mid-tibias, lateral malleolus, posterior side of heels, and head of second metatarsals [93]. The two ASIS markers were placed on top of the belt strap used to attach the sensors to the body, rather than directly to the skin, and the inter-ASIS distance was measured and entered into the software. The markers on the thighs and shanks were placed on 5cm wide Styrofoam blocks, instead of the thigh and shank ‘wands’ proposed by the PiG model, and were fixated to the limb with transpore tape to remove motion artifact during heel strikes. Hip markers were also added and were later used for image processing only since they are not included in PiG. Figure 2 shows an example of how participants were instrumented with the sensors and markers.
Plug-in Gait Kinematics are calculated based on some modifications from [7], [94], [95]. To see these equations, please refer to Appendix D.

4.1.1.1 Plug-in Gait segment orientation

For this research the true body segments orientation is needed, however, the Plug-in Gait model used on Vicon does not directly provides this information. Hence, a MATLAB function that takes the marker data and retrieves the orientation of pelvis, thighs, shanks and feet in the Vicon coordinate frame was developed. The marker position data measured with Vicon and the PiG definitions of body segments coordinate frames [93], [96] was used to define the segment orientation. For a detail description on how the PiG segments orientation were found, please see appendix C.

To calculate the joint centers the “Chord function” described in the Plug-in Gait Manual was implemented [93]. It uses three points to define a plane. The first point is the pre-calculated or known joint center (KJC) and the second one is a real marker on the joint landmark (Joint Marker, JM) at a known perpendicular distance (Joint Centre Offset, JCO) from the required joint center (RJC), which is the third point. This function is called “Chord function” because it is assumed that the two joint centers and the marker lie on the periphery of a circle and the lines

Figure 2: Instrumentation of participant with sensors and markers. Front, back and right side views.
that link the points are chords of the circle. Thus, to calculate RJC the MATLAB function *lsqnonlin* [97] was used to find the point that will make JM be at a distance JCO from RJC as well as be perpendicular to the line between the KJC and RJC coplanar to this line and a second marker called Plane definition marker. The chord function is depicted in Figure 3.

**Figure 3: Location of markers and known and unknown joint centers used in the "Chord function" used to find unknown joint centers with the Plug-in Gait model.**

4.4.2 MVN BIOMECH AWINDA system (MVN system)

The motion tracking system MVN BIOMECH AWINDA (Xsens Technology B.V., Netherlands) consists of up to 17 wireless motion trackers, each of which contains one tri-axial accelerometer (± 160 m/s²), one tri-axial gyroscope (± 1200 °/s) and one tri-axial magnetometer (± 1.5 Gauss) with dimensions 34.5 x 57.8 x 14.5 mm and weight of 27g. For this research only the lower body configuration of the system was used. According to this configuration 7 sensors are placed on the body: sacrum at the level of L5S1, lateral side of mid-thighs, inner side of mid-shanks and middle of bridge of foot, by means of adjustable straps that come together with the system. Sensor data is transmitted to the AWINDA station (receiver that is connected to the computer) via AWINDA Protocol, developed by Xsens. Kinematics are obtained with the MVN Studio Pro software (MVN software). Figure 2 shows how the participants were instrumented with the sensors as well as the markers.
4.1.2.1 MVN standard calibration procedure

The standard calibration recommended by Xsens for use of the MVN system has 3 steps:

1) Body dimensions. These measurements are the calibration parameters needed to scale the model.

2) Data Fusion. Fusion engine refers to the algorithm used to filter the three types of sensor data. MVN system has 3 optional fusion engines: XKF-3, Kinematic Coupling Algorithm (KiC) and KiC without magnetometers. XKF-3 is a regular Kaman Filter that fuses acceleration, angular velocity and magnetic field. KiC and KiC without magnetometers are used when heterogeneous magnetic distortions are expected for long periods of time.

3) N-Pose. The subject has to stand on N-Pose or T-Pose and hold that posture for about 5 seconds. The N-Pose will be the reference calibration pose for further experiments, therefore is the one that will be described.

The N-Pose, as can be seen in Figure 1a, consists in the subject standing upright on a horizontal surface; feet parallel one foot apart; back straight; arms straight alongside with thumbs forward and face forward. During the calibration body segments are assumed to be aligned according to Figure 1b.

4.1.2.2 MVN calibration and kinematics principles

The model used by MVN system consists of 23 body segments (pelvis, L5, L3, T12, T8, neck, head, and right and left shoulder, upper arms, fore arms, hands, upper legs, lower legs, feet and toes) and 22 joints (L5S1, L4L3, L1T12, C1Head, right and left C7Shoulder, shoulders ZXY, shoulders XZY, elbows, wrists, hips, knees, ankles and ball-feet).

In order to understand how MVN estimates kinematics it is necessary to define the reference frames, as well as the variables involved. These descriptions can be seen in Table 1; Figure 4 shows a representation of the different reference frames.
Table 1: Description of reference frames and variables used in the estimation of kinematics by MVN system [77].

<table>
<thead>
<tr>
<th>Concept</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Global Reference Frame (GRF) (See figure 4)</strong></td>
<td>X pointing to local magnetic north; Y according to right-handed coordinates pointing west; Z positive when pointing up. Origin is set after calibration to be at right heel.</td>
</tr>
<tr>
<td><strong>Body Reference Frame (BRF) embedded in the body segment (Used for describing position only)</strong></td>
<td>X forward; Y up from joint to joint; Z pointing right. Each segment has its own origin which is located at the proximal center of rotation of the segment. See Figure 1b.</td>
</tr>
<tr>
<td>$\mathbf{q}_{\text{sen}}^{GS}$</td>
<td>The quaternion that describes the orientation of the sensor in the global frame.</td>
</tr>
<tr>
<td>$\mathbf{q}_{\text{seg}}^{GB}$</td>
<td>The quaternion vector that describes the orientation of the segment in the global frame.</td>
</tr>
<tr>
<td>$\mathbf{q}_{\text{sen}}^{BS}$</td>
<td>The quaternion that describes rotation from sensor to body segment.</td>
</tr>
<tr>
<td>$\mathbf{X}^{B}$</td>
<td>The vector that describes the positions of connecting joints and anatomical landmarks with respect to origin of that segment (in body frame B).</td>
</tr>
<tr>
<td>$\mathbf{P}_{\text{origin}}^{G}$</td>
<td>Vector that describes position of origin of anatomical landmarks in the global frame (during the calibration posture).</td>
</tr>
<tr>
<td>$\mathbf{P}_{\text{landmark}}^{G}$</td>
<td>Vector that describes position of anatomical landmarks in the global frame.</td>
</tr>
</tbody>
</table>

* BRF is only used to express position of body landmarks. **Segment orientation is expressed relative to the GRF.**

To estimate kinematics the orientation of the sensor with respect to the body segment and the distances between joints and segments are needed. Before calculating that information some calibration steps are necessary. These are: 1) Sensor to segment alignment; 2) Model scaling 3) Sensor to segment alignment and segment length re-estimation.

During the first step the orientation of the sensor in the global frame $\mathbf{q}_{\text{sen}}^{BS-\text{NPose}}$ is solved for based on the known orientation of the segment $\mathbf{q}_{\text{seg}}^{GB-\text{NPose}}$ to the global frame and the measure sensor orientation $\mathbf{q}_{\text{sen}}^{GS}$, following equation 1[67]:

$$
\mathbf{q}_{\text{sen}}^{BS-\text{NPose}} = \mathbf{q}_{\text{seg}}^{GB-\text{NPose}} \cdot \mathbf{q}_{\text{sen}}^{GS}
$$
\[ \mathbf{G}^B_q_{\text{seg-NPose}} = \mathbf{G}^S_q_{\text{sen}} \bigotimes \mathbf{B}^S_q_{\text{sen-NPose}}^* \quad \text{Eq. 1} \]

Where \( \mathbf{B}^S_q_{\text{sen-NPose}}^* \) is the conjugate of \( \mathbf{B}^S_q_{\text{sen-NPose}} \)

Here \( \mathbf{G}^B_q_{\text{seg-NPose}} \) has the information of the body segment in the GRF according to the assumed N-Pose.

During the model scaling step, body dimensions, previously measured, and regression equations based on anthropometric models are used to estimate joint and body landmark positions, which is used to posteriorly scale the model. Before that body segment orientations are calculated based on \( \mathbf{B}^S_q_{\text{sen-NPose}} \) found during the sensor to segment alignment as shown in equation 2.

\[ \mathbf{G}^B_q_{\text{seg}} = \mathbf{G}^S_q_{\text{sen}} \bigotimes \mathbf{B}^S_q_{\text{sen-NPose}}^* \quad \text{Eq. 2} \]

After calibration, the orientation and position of joints (origins of body segments) are known \( \mathbf{G}_P_{\text{origin}} \) and equation 2 can be used to obtain the position of the rest of body landmarks, \( \mathbf{G}_P_{\text{landmark}} \).

Figure 4: Coordinate system of the Global Reference Frame (left) and coordinate system of Body Reference Frames (right). Sensor Reference Frame is drawn on the thigh sensor [77].
Finally, joint angles $^{BA}_B q$ can be found by calculating the orientation of a distal segment $^{GB}_A q$ with respect to a proximal segment $^{GB}_B q$ using equation 4.

$$^{BA}_B q = ^{GB}_A q^* \otimes ^{GB}_B q$$  \hspace{1cm} \text{Eq. 4}$$

It is important to mention that since segment orientation is defined relative to the GRF, during the N-Pose all the segments are assumed to be aligned to one another regardless of the orientation of the person relative to the Earth’s frame. However, when the person is facing the magnetic north during the calibration, body segments are not only aligned to one another but also to the GRF.

### 4.1.2.3 Matrix representation of the MVN equations

#### Table 2: Matrix representation of the MVN equations.

<table>
<thead>
<tr>
<th>Quaternion operation</th>
<th>Rotation matrix equivalent</th>
</tr>
</thead>
<tbody>
<tr>
<td>$^{q1 \otimes q2} = (q1_0 \cdot q2_0 - v1 \cdot v2, q1_0 \cdot v2 + q2_0 \cdot v1 + v1 \times v2)$</td>
<td>$R = \begin{bmatrix} q_0^2 + q_1^2 - q_2^2 - q_3^2 &amp; 2q_1q_2 - 2q_0q_3 &amp; 2q_1q_3 + 2q_0q_2 \ 2q_1q_2 + 2q_0q_3 &amp; q_0^2 - q_1^2 + q_2^2 - q_3^2 &amp; 2q_2q_3 - 2q_0q_1 \ 2q_1q_3 - 2q_0q_2 &amp; 2q_2q_3 + 2q_0q_1 &amp; q_0^2 - q_1^2 - q_2^2 + q_3^2 \end{bmatrix}$ \hspace{1cm} \text{Eq. 5}</td>
</tr>
</tbody>
</table>

$$^{GB}_{q_{\text{seg}}} = ^{GS}_{q_{\text{sen}}} \otimes ^{BS}_{q_{\text{sen}}}^*$$  \hspace{1cm} \text{Eq. 6}$$

$$^{GB}_{R_{\text{seg}}} = ^{GS}_{R_{\text{sen}}} \otimes ^{BS}_{R_{\text{sen}}}^{-1}$$

$$^{GB}_{p_{\text{landmark}}} = ^{GP}_{\text{origin}} + ^{GB}_{q_{\text{seg}}} \otimes ^{BX} \otimes ^{GB}_{q_{\text{seg}}}^*$$  \hspace{1cm} \text{Eq. 7}$$

$$^{BA}_B q = ^{GB}_A q^* \otimes ^{GB}_B q$$  \hspace{1cm} \text{Eq. 8}$$

$$^{BA}_B R = ^{GB}_A R^{-1} \otimes ^{GB}_B R$$
The information that was explained above will be used later in this document to explain the orientation correction approach. Given that the PiG segment orientations are represented in rotation matrix format, the equations presented herein are used throughout the rest of this text in their rotation matrix equivalent as shown in Table 2.

4.4.3 Systems synchronization

Both Vicon and MVN have built-in synchronization features. The synchronization was configured according to the description presented in MVN User Manual [77] and the MVN-Vicon synchronization report that can be found online [98].

An additional setting that was required to make the synchronization successful, that was not included in neither of these manuals but was rather learned through discussions with the manufacturer, is the pulse width. When Vicon records at 60Hz pulse (16.7ms period) width has to be adjusted to 20ms, otherwise the synchronization would be sporadic, since the pulse width has to be greater or equal to the Vicon sampling rate.

4.4.4 Camera Setup

A Canon PowerShot SD1200 IS, Digital ELPH camera was used to acquire the images from the right side of the participant, from which joint angles were measured. The camera was placed 2.5m away from the participant. The height of the camera was adjusted for each participant, so that the lens was center at the knee joint when he or she assumed the N-Pose.

A MATLAB script was written to calculate the angles using right PSIS and ASIS, hip, knee, ankle, heel and toe markers. The script defines the segment in the sagittal plane similar to the Plug-in Gait definition of body segments, and then uses the dot product and the arc-cosine function to measure the angles between the segments. The MATLAB script prompts, the user to select the markers that form the segments, similar to what is done in Vicon Nexus when labeling the markers (e. g. Hip and knee markers define the femur). Figure 5 shows an image after it was processed with the customized MATLAB script. The equations to calculate the angles are shown in Appendix E.
Identifying and Reducing factors of error

The various equipment present in gait laboratories together with the metallic structures of the building create heterogeneous magnetic fields. This heterogeneity is known to affect the estimation of heading (estimation of the earth’s magnetic field) by the sensors, thus affecting all further calculations made based on it, such as orientation [41], [82] and since joint angles are measured from the body segments orientation, kinematics is also indirectly affected.

For that reason it was sought to identify the factors that could add errors to MVN joint angles calculation.

4.1.5.1 Fusion engine

As mentioned earlier MVN has 3 different algorithms to merge the 3 types of sensor data: XKF-3, KiC with magnetometer and KiC without magnetometers.

KiC without magnetometers was the first option to use because of expected high magnetic disturbances, however when testing the difference in kinematics using KiC and XKF-3, it was

Figure 5: Segment definition with the MATLAB script used to measure planar joint angles.
found that the error due to drift was affecting the kinematics more than the error due to the magnetic field, hence, XKF-3 was chosen.

4.1.5.2 **Effect of environment on kinematics measured with the MVN system**

The gait of one volunteer was analyzed inside the gait laboratory as well as outside (in the garden of the Hospital), where no magnetic distortions were found. In both cases the system was calibrated with the N-Pose and the subject walked normally at self-selected speed. The coefficient of multiple correlation within-day (CMC-WD) [94] was used to measure the similarity of kinematics recorded inside and outside the gait laboratory. For more details about this experiment refer to Appendix F.

The CMC-WD coefficients indicated that there is a very good to excellent correlation for joint angles of all joints of both sides (≥0.80; ranking of values was done following [99]). Despite of this good similarity, it was concluded that FE angles can be used safely whereas AA and IE could be affected by crosstalk error. This error is potentially introduced due to the perturbation of the antero-posterior axis of the body segments frame that relies on the magnetic field measurements. On the contrary, the CMC-WD is a metric that measures the similarity of waveforms of the same subject recorded in two different sessions, thus, the good to excellent CMC-WD values obtained could mean that the environment does not affect the similarity between waveforms more than what the within subject variability does. Hence, in this work all the planes of all the joints are measured while the crosstalk error is considered during the interpretation of the results.

4.1.5.3 **Calibration spot**

Xsens recommends calibrating the sensors in a magnetically safe environment and letting the sensors “warm up” for a minute; calibrating the sensors in a distortion-free area contributes to having acceptable measurement accuracy as long as data sets are less than 20 to 30 seconds in size, after which the effect of drift is prone to significantly increase measurement error [73].

In order to find this magnetically save environment the protocol described by De Vries et. al [82] was reproduced. In their work, they mapped their gait laboratory to understand how the equipment and ferromagnetic structures affect the heading estimation. They found that the closer
the sensor is to the floor the greater the distortions that effect it: the worst height was 5cm above ground with deviations of up to 30 degrees and the best height was 180 cm with distortions of about 3 degrees. They found that a good trade-off between height and accuracy in heading estimation is 40cm off the ground. They suggest to map the gait laboratory to find the area that has the lowest levels of magnetic distortion and calibrate the sensors there every time. Therefore, for this research the same approach was taken. To see the details on how the gait laboratory was mapped, please refer to Appendix G.

In this study, Vicon and MVN were used to record concurrently, hence, there was the need to find an area in the laboratory where the testing was conducted that was not only magnetically safe but also within the Vicon capture volume (the space that can be capture by Vicon cameras). However, no single area could be found to satisfy both requirements. An alternative was to use two areas, an area of low magnetic distortion for the MVN calibration and then have the participant transfer to the Vicon capture area, however this required that the participant walk through the magnetically unsafe area (highly distorted areas), which was found to adversely affect sensor calibration. To find a solution, the participant was asked to stand on top of a 50cm high wooden table for the MVN recording, which was placed within the Vicon capture area.

Figure 6 shows how the subject was calibrated on top of the wooden table near the magnetically safe area and inside the Vicon capture volume.
It was mentioned before that since the model PiG used on Vicon does not directly provide segments orientation they were calculated using marker data as well as PiG definition of segments coordinate system. To test whether the orientations calculated by the developed code were valid, the angles computed from these orientations using the equations described by Kadaba et al [7]. Details of this process can be found in Appendix C.

4.1.5.5 Comparison of Plug-in Gait and MVN kinematics

PiG and MVN BIOMECH joint angle calculations are based on Kadaba (1990) [7] and Grood and Suntay (1983) [10], respectively. Therefore, to show whether a difference in angle calculation would be a confounding factor to understand the effect of calibration postures, the
kinematics measured using equations in Appendix D, corresponding to both systems were compared when the same body segments orientation was given as input.

Based on the segment orientations that were calculated as explained in section 3.1.1.1 the joint angles according to the two aforementioned articles were found and plotted on the same graph. From those plots it was discovered that, apart from some differences in abduction-abduction sign, the outputs are identical. The only measurements that were totally different were ankle abduction-adduction angles. Therefore, higher disagreement is expected in this plane of motion for this joint when comparing kinematics measured with PiG and MVN. Appendix 0, contains a detailed description of the comparison between the coordinate system definitions and kinematics outputs of the 2 systems.

4.2 Experiment design

4.2.1 Participants

Four able-bodied adults age 23.5±1.73 years old (2 females and 2 males; 165±8.63cm height; 60.43±9.41Kg of body mass) were recruited for this study. This age range for recruitment (18 to 50) was decided to avoid variability in gait pattern due to difference in age [100]. None of the participants had gait (i.e. muscle weakness, bone deformities, sensory loss, pain or impaired control) or cognitive impairments. This research was approved by the Research Ethics Board at Holland Bloorview Kids Rehabilitation Hospital and at the University of Toronto. Before the experiments were done all participants provided written consent.

4.2.2 Simulated postural deviations

To assess the effect of postural deviations in the measurement of joint angles, the participants were asked to simulate crouch gait. This posture was selected to add deviations mainly in the sagittal plane due to limitations of the sensors to measure joint angles in other planes; as explained in section 1.8 and 1.9.

Crouch gait was simulated by bending their knees at an angle that would feel comfortable but that they could easily and reliably replicate. The participants were trained, prior to the actual experiment, by asking them to stand upright and then bend their knees a few times trying to
reach the same flexion each time. Posture repeatability measurements for each participants are presented in Appendix H.

4.2.3 Experiments

Each participant was asked to perform walking trials, where he/she had to walk a 5m distance under two different experimental conditions. An experimental condition consists in a calibration posture (the N-Pose or Crouch posture) and a type of walking either normal walking or crouch walking, both at self-selected speed. The experimental conditions are shown in Table 3.

<table>
<thead>
<tr>
<th>Experimental condition</th>
<th>Calibration Posture</th>
<th>Type of walking</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Ideal Calibration (IC)</td>
<td>N-Pose</td>
<td>Normal Walking</td>
</tr>
<tr>
<td>2. Crouch gait (GC)</td>
<td>Crouch Posture</td>
<td>Crouch Walking</td>
</tr>
</tbody>
</table>

Both conditions were recorded by MVN and Vicon systems simultaneously. The participants performed three walking trials per experimental condition; in total 6 trials were recorded with each system, per participant. Vicon can only record the walking of the participants while they remain inside the capture volume, therefore, from each trial only the gait cycles that were successfully recorded with Vicon were used. The sensors were calibrated at the beginning of each trial. During this calibration time the participant assumed the calibration posture and was asked to stay still while the static posture was recorded with both systems (Vicon and MVN) and the pictures were taken using the digital camera (i.e. there were six static trial per participant, three for each experimental condition).

Before the trials, the knee height of the participant standing on the table was measured to adjust the lens of the camera so that it was at the same height of the knee. Also, adhesive tape was put on the table around the feet of the participant, standing comfortably upright, to draw their shape for the participant to assume the same posture during subsequent trials. This was done to control for replicability.
For each walking trial the participants followed the steps presented below:

Step 1. Stand on a calibration posture: N-Pose or Crouch posture on top of the wooden table. Participant standing facing the geographic north of the gait laboratory (considered to be aligned to the magnetic north).

Step 2. Calibrate the sensors.

Step 3. Record the calibration posture for 5 seconds.

Step 4. Take the pictures from the right side.

Step 5. Down on ground level, the participant heard someone counting to three and at the order “go” he/she started walking with his/her right leg at normal comfortable speed and stopped a couple of steps after crossing the 5m mark.

Step 6. The recording was stopped and the participant went back to the starting point to repeat the same experimental condition 3 times.

Step 7. Steps 1 to 5 were repeated until all the trials were completed.

4.3 Data Processing

All required signal processing was done on MATLAB version 2013a (The MathWorks, Natick, MA, USA).

4.3.1 Signal alignment

Even after synchronizing Vicon and MVN BIOMECH a time lag was found in the signals. To improve aligning of the signals the peak of the knee flexion-extension angles in both systems were found and manually the frames at which these peaks occurred were identified and the frame lag was added to the code that detects the gait cycles in each recording.
4.3.2 Gait Cycle detection

Gait cycles were detected by manually identifying heel strike events from vertical acceleration signals of the shank MVN sensors as shown in Figure 7.

![Detecting gait cycles](image)

**Figure 7: Detection of gait cycles with the vertical acceleration of the shank sensor.**

4.3.3 Gait Cycle Normalization

All gait cycles were normalized to gait cycle percentage using a customized MATLAB Script that uses the function `interp1`.

4.4 Analysis

4.4.1 Analysis parameters

Analysis parameters were divided into static and dynamic parameters. Static parameters are used to measure the agreement in calculating static angles during the calibration posture whereas dynamic parameters represent the agreement in calculating angles during gait.

The static parameters refer to the difference in angles calculated between the Vicon and MVN systems during the static capture of the calibration posture. To calculate the static parameter first
the difference of joint angles measured with the two systems was found using the following equation:

\[
\text{Difference (t)} = \text{Static Angle}_{\text{PiG}}(t) - \text{Static Angle}_{\text{MVN}}(t)
\]

Eq. 9

Where \( t = 1, \ldots, T \) trials.

The mean (\( \text{DIFF}_S \)) and standard deviation (\( \sigma_{\text{DIFF}_S} \)) of this difference across trials were calculated, these are the static parameters.

\[
\text{DIFF}_S = \frac{\sum_{t=1}^{T} \text{Difference}(t)}{T}
\]

Eq. 10

\[
\sigma_{\text{DIFF}_S} = \sqrt{\frac{\sum_{t=1}^{T} (\text{DIFF}_S(t) - \mu_{\text{DIFF}_S})^2}{T}}
\]

Eq. 11

The dynamic parameters, on the contrary, were calculated from the waveforms acquired during the walking trials. These parameters are coefficient of multiple correlation (CMC) and difference (\( \text{DIFF}_D \)).

A variation of the coefficient of multiple correlation (CMC) [101] was used to compare the similarity between waveforms acquired from 2 different protocols (e.g. PiG and MVN). To ensure that CMC measures only the shape similarity, the signals were zeroed before calculating the CMC. This was done by subtracting the mean of the signal. Eq.1 shows the definition of the CMC.

\[
\text{CMC} = \sqrt{1 - \frac{\sum_{g=1}^{G} \left[ \sum_{f=1}^{P} (Y_{gpf} - \bar{Y}_{gf})^2 / GF_g (P - 1) \right]}{\sum_{g=1}^{G} \left[ \sum_{p=1}^{P} \sum_{f=1}^{P} (Y_{gpf} - \bar{Y}_{gf})^2 / G (PF_g - 1) \right]}}
\]

Eq.12

Where:

- \( P \) = number of protocols = number of waveforms for each gait cycle(one per recording system)
• \( G \) = number of gait cycles

• \( F_g \) = number of frames measured for the gth gait cycle

• \( Y_{grf} = \) ordinate at frame \( f \) of the average provided by recording system \( r \) at gait cycle \( g \)

• \( \bar{Y}_{gf} = \) ordinate at frame \( f \) of the waveform among the P waveforms for the gait – cycle \( g \)

• \( \bar{Y}_g = \) grand mean for the gait cycle \( g \) among its P waveforms

CMC close to 1 means that “…the variability of the P waveforms around their mean waveform is considerably smaller than the variance about their grand mean…” Otherwise CMC tends to be zero or a complex number [101]. The factors that make the CMC have complex values are various [102] (e.g. offset between signals and range of motion). Despite the possibility of having CMC complex values, from a clinical perspective CMC equal to zero or complex number could be interpreted as a total disagreement [103], even though mathematically zero and complex numbers have a different meaning. The following scale will be used to classify the CMC according to [99]: 0.65-0.74 moderate, 0.75-0.84 good, 0.85-0.94 very good, 0.95-1 excellent. Zero or complex values will be considered as disagreement in shape.

The parameter \( \text{DIFF}_S \) as well as \( \text{DIFF}_D \) reflect the difference in joint angle calculated by the two systems. The difference between dynamic \( \text{DIFF}_D \) and static \( \text{DIFF}_S \) is that the dynamic \( \text{DIFF} \) (\( \text{DIFF}_D \)) is the difference between systems at each point in time or frame (Eq.13) and the mean of these difference is found across all gait cycles (Eq.14) as opposed to trials:

\[
\text{Difference}(f) = \text{GaitAngle}_{\text{Pi}G}(f) - \text{GaitAngle}_{\text{MVN}}(f)
\]

Eq.13

Where \( f = 1, \ldots, F \) frames.

\[
\text{DIFF}_D(f) = \frac{\sum_{g_c=1}^{GC} \text{Difference}_{g_c}(f)}{GC}
\]

Eq.14

\[
\sigma_{\text{Difference}}(f) = \sqrt{\frac{\sum_{g_c=1}^{GC} (\text{Difference}_{g_c}(f) - \text{DIFF}_D(f))^2}{GC}}
\]

Eq.15

Where \( g_c = 1, \ldots, GC \) gait cycles
The great mean difference across all frames was also found according to Eq. 16

\[ \mu_{DIFF_D} = \frac{\sum_{f=1}^{F} DIFF_D(f)}{F} \]  

Eq. 16

\[ \sigma_{DIFF_D} = \sqrt{\frac{\sum_{f=1}^{F} (DIFF_D(f) - \mu_{DIFF_D})^2}{F}} \]  

Eq. 17

All these parameters were calculated for the three joints (hip, knee, ankle) and the joint angles in the three planes (abduction-adduction, AA; external-internal rotation, IE and flexion-extension, FE).

In the following sections, the specific methods used to achieve objective 1 and 2 are described.

4.5 Assessing the effect of postural deviations in gait kinematics  
(Objective 1)

For objective 1 the following questions were proposed.

I. In gait kinematics measured with the MVN BIOMECH system, what is the error associated with postural deviations (during the calibration)?

II. Is the error constant throughout the gait cycle?

To answer these questions the data from the Ideal Calibration condition (IC) and Crouch Gait condition (CG) were used to compare the static as well as the dynamic angles between MVN and PiG. The protocol to collect the data can be seen in section 4.2.3, however, the analysis is reported in this section.

Dynamic parameters of all the joints and planes were found for the IC. These parameters are considered as the baseline error given that is the best agreement that can be obtained between PiG and MVN under the ideal calibration condition.

The difference between PiG and MVN (DIFF_D) for both the IC and CG conditions, will reflect systemic error that contains instrumentation error as well as crosstalk error. The instrumentation
error is due to measurement errors by the sensor (e.g. error in acceleration calculation, drift, magnetic disturbances, etc.) or inherit error to Vicon (e.g. skin artifacts). The crosstalk error arises from differences in anatomical frame definition in joint angles [81], which could affect the shape of the gait kinematics. However, since the sensors and markers are not replaced on the person during the experiment, the IC as well as the CG are likely to experience the same potential crosstalk error, which would be considered part of the systemic error.

If all of these additional errors are included in the system error, when comparing the CG system error and IC system error (the baseline error), the error due to postural deviation, and its potential effects, will be evident.

To account for the variability of systemic error that might arise from one experimentation day to the next, the baseline error was calculated for each participant.

To analyze the postural deviation error during the static posture, which is right after the calibration and before the walking trial, $\text{DIFF}_S$ values of the CG condition were analyzed. If there is no error during the calibration then this difference should be zero. However, since MVN does not account for the postural deviation, large positive $\text{DIFF}_S$ values that are similar to the true joint angles are expected, especially for FE angles.

These comparisons were used to answer Question I. To answer Question II, the $\text{DIFF}_D(f)$ was differentiated with respect to time, and its mean was calculated across all frames. This derivative represents the change in $\text{DIFF}_D(f)$ from frame to frame. If this change is minimal from frame to frame, then it can be considered to be more or less constant. In this case, it would be acceptable to use the information of the static trial (an offset) to correct the dynamic trial frame by frame. The derivative of the difference could be considered constant if the mean is close to zero and the standard deviation is small.

### 4.6 Applying OC and PAC towards improving the accuracy of gait kinematics (objective 2)

Recalling from section 3, for objective 2 there are 4 different questions that are sought to be answered:
If the body segments orientation at the moment of calibration are known:

I. If the true orientation of body segments is known, will the orientation correction (OC) reduce the error?

II. If the true joint angles are known, will the planar angle correction (PAC) reduce the error?

III. Are OC and PAC approaches equally effective at reducing the error?

If joint angles can be corrected using the PAC:

IV. Can angles measured from planar digital images, acquired using a standard camera, be used to implement the PAC?

In the following sections the correction approaches as well as the methodology used to answer these questions will be described in detail.

4.6.1 Orientation Correction (OC)

Previously, the procedure for calibrating the MVN system was described in section 3.1.2.2. It was established that during the sensor to segment alignment step the orientation of each sensor in the global frame $^{GS}_{\text{q}_{\text{sen}}}$ is matched to the body segment $^{BS}_{\text{q}_{\text{sen}}}$ based on the known orientation of the segment $^{GB}_{\text{q}_{\text{seg}-\text{NPose}}}$ in the global frame. During this calibration step $^{GB}_{\text{q}_{\text{seg}-\text{NPose}}}$ is considered to be aligned to the GRF, however, if the individual is not assuming the N-Pose the subsequent calculations of $^{GB}_{\text{q}_{\text{seg}}}$ is going to be inaccurate. This is what is being referred to as the orientation error. The following section describes how this error is originated and how it can be corrected for.

4.6.1.1 Description

If E is the standard basis whose vectors span $\mathbb{R}^3$:

$$E = \{e_1, e_2, e_3\} = \{(1,0,0), (0,1,0), (0,0,1)\}$$  \hspace{1cm} \text{Eq.18}

The coordinate system of the GRF in terms of E can define as:
\[
\text{GRF} = \{X_G, Y_G, Z_G\} \quad \text{Eq. 19}
\]

Where:

\[X_G = e1; \quad Y_G = e2; \quad Z_G = e3\]

Also, it was explained in section 4.1.2.2, that the orientation of body segments is expressed in the GRF, differently from the position of body segments which is expressed in the body reference frame (BRF). Therefore, the BRF in the following description is expressed in the GRF and can be written as:

\[
\text{BRF} = \{X_B, Y_B, Z_B\} \quad \text{Eq. 20}
\]

Where:

\[
X_B = x_{B1}e1 + x_{B2}e2 + x_{B3}e3, \quad x_{B1}, x_{B2}, x_{B3} \in \mathbb{R} \\
Y_B = y_{B1}e1 + y_{B2}e2 + y_{B3}e3, \quad y_{B1}, y_{B2}, y_{B3} \in \mathbb{R} \\
Z_B = z_{B1}e1 + z_{B2}e2 + z_{B3}e3, \quad z_{B1}, z_{B2}, z_{B3} \in \mathbb{R}
\]

Recalling that during the calibration it is assumed that the all body segments are aligned, BRF can be expressed in terms of basis vector as:

\[
\text{BRF} = \text{GRF} \quad \text{Eq. 21}
\]

Thus, it is assumed that

\[
X_B = X_G; \quad Y_B = Y_G; \quad Z_B = Z_G
\]

In that case, \(G^B_{\text{seg-NPose}}\) which is the matrix expressing the change of basis from the BRF to the GRF, would be equal to the identity matrix.

\[
G^B_{\text{seg-NPose}} = T: \text{BRF} \rightarrow E = I \quad \text{where T means transformation} \quad \text{Eq. 22}
\]

Therefore, going back to equation 6 introduced in section 3.1.2.2 and rewriting it
Gratefully:

\[ GBR_{seg-NPose} = GSR_{sen-NPose} * BS_{sen-NPose}^{-1} \]  \text{Eq.23}

Leads to:

\[ I = GSR_{sen-NPose} * BS_{sen-NPose}^{-1} \]  \text{Eq.24}

Were \( BS_{sen-NPose}^{-1} \) can be solved for to find:

\[ BS_{sen-NPose}^{-1} = GSR_{sen-NPose}^{-1} \]  \text{Eq.25}

Allowing Eq.23 to be re-written as:

\[ GBR_{seg-NPose} = GSR_{sen-NPose} * GSR_{sen-NPose}^{-1} \]  \text{Eq.26}

Thus, to calculate the orientation of the segment in GRF frame by frame \( GBR_{seg}(f) \), that is during the recording time, the new sensor measurements \( GSR_{sen}(f) \) and the just calculated \( BS_{sen-NPose}^{-1} \) (equation 24), would be used according to Eq.27:

\[ GBR_{seg}(f) = GSR_{sen}(f) * BS_{sen-NPose}^{-1} \]  \text{Eq.27}

However, in the presence of postural deviations body segments would not be aligned to the GRF:

\[ BRF \neq GRF \]  \text{Eq.28}

and \( GBR_{seg-NPose} \) would not be equal to \( I \):

\[ GBR_{seg-NPose} \neq I \]  \text{Eq.29}

Instead it would be the result of rotating \( I \) by some rotation \( R_0 \):

\[ BRF = R_0 * I \]  \text{Eq.30}

This means that the BRF was moved from the N-Pose to another position:

And
\[ \text{GR}_{\text{True}} = \text{R}_0 \ast \text{GR}_{\text{seg-NPose}} = T: \text{R}_0 \ast \text{BRF} \rightarrow \text{E} = \text{R}_0 \ast \text{I} \quad \text{Eq. 31} \]

Therefore, the equation during the calibration step would be:

\[ \text{GR}_{\text{True}} = \text{R}_0 \ast \text{GR}_{\text{seg-NPose}} = \text{GSR}_{\text{sen-NPose}} \ast \text{BSR}_{\text{sen-NPose}}^{-1} \quad \text{Eq. 32} \]

Where \( \text{BSR}_{\text{sen-NPose}}^{-1} \) could be solved for as follows:

\[ \text{GSR}_{\text{sen-NPose}}^{-1} \ast \text{R}_0 \ast \text{GR}_{\text{seg-NPose}} = (\text{GSR}_{\text{sen-NPose}}^{-1} \ast \text{GSR}_{\text{sen-NPose}}) \ast \text{BSR}_{\text{sen-NPose}}^{-1} \]

\[ \text{GSR}_{\text{sen-NPose}}^{-1} \ast \text{R}_0 \ast \text{GR}_{\text{seg-NPose}} = \text{I} \ast \text{BSR}_{\text{sen-NPose}}^{-1} \quad \text{Eq. 33} \]

Now, when the recording starts \( \text{BSR}_{\text{sen-NPose}}^{-1} \) can be used to calculate \( \text{GR}_{\text{seg}}(f) \):

\[ \text{GR}_{\text{seg}}(1) = \text{GSR}_{\text{sen}}(1) \ast \text{BSR}_{\text{sen-NPose}}^{-1} \quad \text{Eq. 34} \]

Plugin-in Eq.33 into Eq.34:

\[ \text{GR}_{\text{seg}}(1) = \text{GSR}_{\text{sen}}(1) \ast \text{GSR}_{\text{sen-NPose}}^{-1} \ast \text{R}_0 \ast \text{GR}_{\text{seg-NPose}} \quad \text{Eq. 35} \]

However, in spite of the calibration posture it is assumed that equation 24 is true

\[ \text{BSR}_{\text{sen-NPose}}^{-1} = \text{GSR}_{\text{sen-NPose}}^{-1} \]

Then using this value of \( \text{BSR}_{\text{sen-NPose}}^{-1} \) to find \( \text{GR}_{\text{seg}}(f) \) as opposed to the value of \( \text{BSR}_{\text{sen-NPose}}^{-1} \) from Eq.33, yields an error that is connoted as the orientation error.

Knowing that \( \text{GSR}_{\text{sen}}(1) \), is the output of the sensor at frame \( f = 1 \), \( \text{GSR}_{\text{sen-NPose}}^{-1} \), is the inverse of the sensor output at the moment of calibration; \( \text{R}_0 \) is the true orientation of the body segment that can be measured with imaging techniques and that \( \text{GR}_{\text{seg-NPose}} \) is just the identity matrix, we could use Eq.35 to solve the orientation error problem and find the true \( \text{GR}_{\text{seg}}(1) \). However, MVN system does not provide the sensors output at the moment of calibration,
therefore there are more unknowns than equations and this equation cannot be solved. Because of what was explained here, a different alternative has to be sought.

As mentioned several times along this work, $^\text{GR} R_{\text{seg}}$ is the orientation of body segment in the GRF, in other words the change of basis from the BRF to the GRF. Nevertheless, it can also be interpreted as the matrix that rotates a 3-dimensional vector $v$ to a new position $v'$ in the GFR.

$$v' = ^{GB} R_{\text{seg}} \ast v$$

In analogy, the body segment is represented in Figure 8a as a vector $\vec{s}$ expressed in terms of the GRF. Following that analogy as shown in Figure 8b, the deviation from the N-Pose at the moment of calibration can be expressed as the matrix $^\text{GB} R_{\text{seg-NPose}}$ that rotates $\vec{s}_{\text{N-Pose}}$ to its “new” or true orientation $\vec{s}_{\text{True}}$:

$$\vec{s}_{\text{True}} = ^{\text{GB} R_{\text{seg-NPose}}} \ast \vec{s}_{\text{N-Pose}}$$ \hspace{1cm} \text{Eq.36}$$

During the calibration, regardless of the posture, the system will assume $^\text{GB} R_{\text{seg-NPose}}$ to be the identity matrix. Therefore, the equation to express the measured orientation of the body segment is:

$$\vec{s}_{\text{MVN}} = \vec{s}_{\text{N-Pose}} = ^{\text{GB} R_{\text{seg-NPose}}} \ast \vec{s}_{\text{N-Pose}} = \mathbf{I} \ast \vec{s}_{\text{N-Pose}}$$ \hspace{1cm} \text{Eq.37}$$

In reality, there is a rotation matrix $R_0$ different than the identity matrix that rotates vector $\vec{S}$ from the N-Pose to the true position:

$$\vec{s}_{\text{True}} = R_0 \ast \vec{s}_{\text{N-Pose}}$$ \hspace{1cm} \text{Eq.38}$$
Figure 8: a) Body segment represented as a vector in the Global Reference Frame. b) Segment deviation represented as a vector rotated by rotation matrix $R$.

Therefore, 2 different cases can be identified:

<table>
<thead>
<tr>
<th>Moment</th>
<th>Measured with MVN case</th>
<th>True case</th>
</tr>
</thead>
<tbody>
<tr>
<td>N-Pose</td>
<td>$\vec{s}<em>{MVN} = \vec{I} \ast \vec{s}</em>{N-Pose}$</td>
<td>$\vec{s}<em>{True} = R_0 \ast \vec{s}</em>{N-Pose}$</td>
</tr>
</tbody>
</table>

When the recording starts at $f = 1$, $\vec{s}_{N-Pose}$ has been rotated due to the motion of the individual to its new position denoted by $\vec{s}_{MVN}'_1$ for the assumed case and $\vec{s}_{True}'_1$ for the true case. In the measured case the matrix responsible for the rotation is the MVN system output $^{GB}R_{seg}(1)$ whereas in the true case is a different matrix $M_1$ that takes into account the initial rotation from the N-Pose to the true calibration posture $R_0$ as well the new rotation $R_1$. In other words $M_1$ is a composite rotation of the two:

$$M_1 = R_1 \ast R_0$$

Eq.39

<table>
<thead>
<tr>
<th>Moment</th>
<th>Measured with MVN case</th>
<th>True case</th>
</tr>
</thead>
<tbody>
<tr>
<td>N-Pose</td>
<td>$\vec{s}<em>{assumed} = \vec{I} \ast \vec{s}</em>{N-Pose}$</td>
<td>$\vec{s}<em>{True} = R_0 \ast \vec{s}</em>{N-Pose}$</td>
</tr>
<tr>
<td>$t = t_1$</td>
<td>$\vec{s}<em>{assumed}'<em>1 = ^{GB}R</em>{seg}(1) \ast \vec{s}</em>{N-Pose}$</td>
<td>$\vec{s}_{True}'<em>1 = M_1 \ast \vec{s}</em>{N-Pose}$</td>
</tr>
</tbody>
</table>

Given that the rotation due to the motion of the individual is the same in both the assumed and the true cases, they can be considered to be equal:

$$^{GB}R_{seg}(1) = R_1$$

Eq.40
Thus:

\[ M(1) = R(1) * R_0 = \mathbf{GR}_{\text{seg}}(1) * R_0 \quad \text{Eq.41} \]

This equation corrects the orientations measured by MVN to account for the posture deviated from the N-Pose. Because this calculations are done frame by frame Eq.41, can be generalized to

\[ M(f) = R(f) * R_0 = \mathbf{GR}_{\text{seg}}(f) * R_0 \quad \text{Eq.42} \]

Where \( R_0 \) can be measured with imaging techniques.

Furthermore, to estimate the kinematics with the corrected orientations, Eq.8 can be used as follows:

\[ \mathbf{R}_{\text{BA}}(f) = \mathbf{GR}_{\text{BA}}^{-1}(f) * \mathbf{GR}_{\text{B}}(f) \quad \text{Eq.43} \]

In summary, equation 41 and 42 together constitute the orientation correction.

4.6.1.2 Implementation

The data recorded during the 5 seconds after the calibration of the sensors at the beginning of every trial was processed with the MATLAB script that was developed as described in section 4.4.4, in order to know the true orientation of body segments \( (\mathbf{R}_{\text{PiGRF}}) \) in the PiG reference frame (PiGRF). The mean orientation across the 5 seconds data was found and used to implement the correction technique. All the trials of all experimental conditions were corrected individually.

The orientation correction follows the steps mentioned below:

1) Find the reference segments orientations \( \mathbf{R}_{\text{PiGRF}} \).

2) Do a change of basis from the PiG reference frame to the MVN global reference frame.

MVN global reference frame was defined in Eq.19, as:

\[ \mathbf{GRF} = \{X_G, Y_G, Z_G\} \]
Where:

\[ X_G = e_1; \; Y_G = e_2; \; Z_G = e_3; \]

and \( e_1, e_2 \) and \( e_3 \) are the base vector that define the standard basis \( E = \{ e_1, e_2, e_3 \} = \{(1,0,0), (0,1,0), (0,0,1)\} \).

According to the geometry of the room and its geographic position it was assumed that the PiGRF in respect to the GRF is defined as follows:

\[
\text{PiGRF} = \{X_{\Pi G}, Y_{\Pi G}, Z_{\Pi G}\} \tag{Eq.44}
\]

Where:

\[ X_{\Pi G} = Y_G; \; Y_{\Pi G} = -X_G; \; Z_{\Pi G} = Z_G; \]

Therefore the change of basis matrix \( C \) is defined as:

\[
C = \begin{bmatrix}
0 & -1 & 0 \\
1 & 0 & 0 \\
0 & 0 & 1
\end{bmatrix}
\]

Hence, the reference segments orientations in the GRF were calculated according to equation 43:

\[
^{GB}R_{\text{True}} = C \times ^{PiGB}R_{\text{True}} \tag{Eq.45}
\]

For all segments but the feet, this equation was used to do the correction:

\[
M(f) = R(f) \ast R_0 = ^{GB}R_{\text{seg}(f)} \ast R_0 \tag{Eq.46}
\]

Equation 46, shows the same equation as above but with the experimental variables

\[
^{GB}R_{\text{orientCorr}}(f) = ^{GB}R_{\text{seg}(f)} \ast ^{GB}R_{\text{True}} \tag{Eq.47}
\]

For the feet orientations an extra step had to be taken before applying the correction.

The feet coordinate system (FCS) in PiG is defined as follows:
FCS = \{^{PiGB}X_{\text{foot}} = -Z_p; ^{PiGB}Y_{\text{foot}} = Y_p; ^{PiGB}Z_{\text{foot}} = X_p\}

As opposed to:

FCS = \{^{PiGB}X_{\text{foot}} = X_p; ^{PiGB}Y_{\text{foot}} = Y_p; ^{PiGB}Z_{\text{foot}} = Z_p\}

like the other segments. Therefore, $^GR_{\text{True(foot)}}$ was also multiplied by the following matrix:

$$
C_{\text{foot}} = \begin{bmatrix}
0 & 0 & -1 \\
0 & 1 & 0 \\
1 & 0 & 0
\end{bmatrix}
$$

After that the correction was implemented.

In the final step equations described in Appendix D were applied to re-calculate the MVN corrected joint angles (Ang$_{\text{MVN-OC}}$).

4.6.1.3 Analysis

To determine whether OC correction approach improves the kinematics accuracy as well as similarity to the gold standard, the true gait joint angles, Ang$_{\text{PiG}}$, were compared to the corrected MVN angles Ang$_{\text{MVN-OC}}$. The dynamic parameters calculated from the kinematics corrected with the OC approach were compared to the baseline error.

4.6.2 Planar Angle Correction (PAC)

4.6.2.1 Description

It was mentioned in the background that the error associated to postural deviation can also be interpreted as offset angles from the N-Pose.

According to equation 8,

$$
^{BA}B_R = ^G^{BA}R - ^G^{BB}R
$$

The kinematics are calculated based on the orientation of two contiguous body segments; if this orientation is wrong it follows that the kinematics will also be wrong. However, in the end the
outcome of interest is the kinematics, not the orientation; hence if the joint angles are thought of as angles in the plane of motion (planar angles), the difference between the measured and the true joint angles is going to be an offset. This error is mentioned as the planar joint angle error.

This error in angle can be expressed with the following equation:

\[
\text{True Joint Angles} = \text{Measured Joint Angles} + \text{Offset Joint Angles}
\] (1)

The Offset_Joint_Angles are not accounted for during the calibration and it is thought that they should remain constant throughout the gait cycle since the postural deviation of the subject being analyzed (patient with postural deviations) is maintained for the whole experiment. Hence, the gait joint angles should be corrected by adding the angle offset at each point in time. This procedure is referred to as the planar joint angle correction.

4.6.2.2 Implementation

To determine the true joint angles the mean of the Plug-in Gait angles across the 5 seconds of data recorded at the beginning of every trial were used. In the rest of these manuscript these angles are called PiG angles (Ang\text{PiG}).

To find the corrected angles Ang\text{MVN-PAC}, equation 47 was used as follows:

\[
\text{Ang}_{\text{MVN-PAC}} = \text{Ang}_{\text{PiG}} + \text{Ang}_{\text{MVN}}
\]

Where:

Ang\text{MVN-PAC} = the corrected angles (true angles)

Ang\text{MVN} = the angles measured by MVN

Ang\text{PiG} = angles measured by PiG (offset angles)

4.6.2.3 Analysis

Results obtained after applying the OC and the PAC were compared to evaluate whether both correction approaches are equally effective in reducing the postural deviation error.
To determine whether the PAC correction approach improves the kinematics accuracy as well as similarity to the gold standard, the true gait joint angles, $\text{Ang}_{\text{PiG}}$, were compared to the corrected MVN angles $\text{Ang}_{\text{MVN-PAC}}$. The dynamic parameters calculated from the kinematics corrected with the OC approach were compared to the baseline error; similarly to what was done for the OC approach.

Finally to determine whether planar angles measured from digital images taken with regular cameras are suitable for the implementation of PAC, the $\text{DIFF}_S$ measured between $\text{PiG}$ and the planar images was also analyzed.
5. Results

5.1 Data

Each of the two experimental conditions (IC, CG) that each participant performed was recorded three times with both systems (Vicon, MVN), resulting in six walking trials per participant (ICx3, CGx3) per system. In each trial, three gait cycles were detected for both limbs of each participant, therefore, in total each participant had nine gait cycles per limb per experimental condition per system (e.g. 9 right gait cycles for experimental condition IC recorded with Vicon and 9 recorded with MVN).

Also, because the static trials were acquired at the beginning of each experimental condition and they were repeated three times, there were three static trials per experimental condition (staticICx3, staticCGx3) per participant. Each of these static trials was acquired with Vicon, MVN and the digital camera.

5.2 Baseline Error

Before the results corresponding to each objective are introduced, the errors corresponding to the MVN system used under the ideal calibration are presented.

Figure 9 shows the kinematics of the left limb of participant-1 according to PiG in blue and MVN in red. This graph corresponds to the IC condition, which means that MVN was calibrated with the N-Pose, and the participant walked at self-selected speed. From left to right and top to bottom abduction-adduction, external-internal rotation and extension-flexion (AA, IE, and FE) of hip, knee and ankle are shown. The numbers represent the CMC values across the nine gait cycles of that limb. The CMC values of all participants can be seen in Table 4.
Figure 9: Kinematics of the left limb of participant-1 according to PiG (blue) and MVN (red). The top right numbers are the mean of the CMC values of all the 9 gait cycles.

As expected, FE angles of Figure 9 show very good to excellent agreement at the hip, knee and ankle. Even the hip FE has high CMC values regardless of the offset; recalling from section 4.4.1 the signals were zeroed before the correlation was calculated. The high values show the excellent agreement between PiG and MVN in calculating angles in the sagittal plane, as expected. The shift in value of the hip FE, can be explained by a difference in the anatomical frame definition of PiG and MVN, most likely due to the placement of the sensor. Since the sensor was placed lower than the markers, as can be seen in Figure 2, the anterior-posterior line that creates the pelvis frame was more horizontal to the floor as opposed to being pointing towards it, like the line joining the posterior and anterior pelvis markers does. This creates a virtual extension of the pelvis, hence the values of the Hip-FE on MVN were shifted to the extension side (below zero).

This situation was consistent in all participants. When looking at the CMC values shown in the hip, knee and ankle FE angles of all participants are excellent (>0.96). Furthermore when
looking at the DIFF₉ in Table 5, hip FE values are all positive (>16.47) meaning that PiG values where higher than MVN values, confirming the “virtual” extension due to anatomical frame definition of MVN. Given the nature of this shift, even if it happens during the IC condition, it is expected that the OC or the PAC could correct for it.

When looking at Figure 9, knee and ankle IE, the CMC values are moderate to excellent. However, only knee IE has consistently good to excellent correlation whereas ankle IE contains some moderate correlations as well.

Table 4: CMC of all participants corresponding to the ideal calibration condition. Values ≥0.995 are written as 1. Shading indicates poor (lightest) to excellent (darker) correlation. CMC complex values are displayed as Not a Number (NaN).

<table>
<thead>
<tr>
<th>Participant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Right</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>AA</td>
<td>0.91</td>
<td>0.95</td>
<td>0.96</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.49</td>
<td>0.65</td>
<td>0.88</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Knee</td>
<td>AA</td>
<td>0.48</td>
<td>NaN</td>
<td>0.31</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.90</td>
<td>0.95</td>
<td>0.79</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Ankle</td>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.84</td>
<td>0.69</td>
<td>0.74</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.98</td>
<td>0.98</td>
<td>0.96</td>
</tr>
<tr>
<td><strong>Left</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>AA</td>
<td>0.72</td>
<td>0.81</td>
<td>0.70</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.74</td>
<td>0.93</td>
<td>0.79</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>1.00</td>
<td>0.99</td>
</tr>
<tr>
<td>Knee</td>
<td>AA</td>
<td>0.81</td>
<td>0.64</td>
<td>NaN</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.93</td>
<td>0.92</td>
<td>0.78</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>1.00</td>
<td>0.99</td>
</tr>
<tr>
<td>Ankle</td>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.92</td>
<td>0.73</td>
<td>0.86</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.97</td>
<td>0.97</td>
<td>0.96</td>
</tr>
</tbody>
</table>
The only plane and joint that had a consistently poor agreement was ankle AA (all complex values). Moreover, when looking at the mean and standard deviation of ankle AA DIFF\textsubscript{D} in Table 5, the standard deviation is much higher than the mean in both limbs of all participant. This suggests that the error in this two cases is not likely just due to anatomical frame definitions, as can also be seen in Figure 9. No improvement is expected to be made in these joint angles with either the OC or the PAC. The other planes and joints hip and knee AA as well as hip IE are inconsistent across participants; as expected, since the MVN system is known to perform better in the sagittal plane.

Figure 10: Each subplot shows the difference of kinematics of the left limb of participant-1, calculated according to PiG and MVN (DIFF\textsubscript{D}) as a function of time. The bars indicate the range within which DIFF\textsubscript{D} lies. On the right of each subplot the boxplot of DIFF\textsubscript{D} can be seen.
Table 5: Mean and standard deviation of DIFF_D of the baseline error of all participants. Results are shown in degrees. The DIFF_D of the FE angles are in bold and italics. Knee and ankle show DIFF_D about zero and small standard deviation because of the excellent agreement between PiG and MVN. Hip DIFF_D has high mean because PiG values are greater than MVN indicating a shift towards extensions values; as a result of different anatomical frame definition of the pelvis due to sensor location. However the standard deviation is low given the excellent agreement.

<table>
<thead>
<tr>
<th>Participant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>μDIFF_D (σDIFF_D)</td>
<td>μDIFF_D (σDIFF_D)</td>
<td>μDIFF_D (σDIFF_D)</td>
<td>μDIFF_D (σDIFF_D)</td>
</tr>
<tr>
<td><strong>Right</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-5.67 (3.14)</td>
<td>-5.86 (2.63)</td>
<td>-0.38 (1.91)</td>
<td>-4.39 (1.93)</td>
</tr>
<tr>
<td>IE</td>
<td>-2.86 (4.09)</td>
<td>-8.44 (4.39)</td>
<td>-7.88 (2.26)</td>
<td>-1.56 (3.88)</td>
</tr>
<tr>
<td>FE</td>
<td><strong>28.80 (3.39)</strong></td>
<td><strong>18.58 (1.85)</strong></td>
<td><strong>22.88 (2.93)</strong></td>
<td><strong>24.98 (3.29)</strong></td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>2.35 (2.72)</td>
<td>-2.22 (7.30)</td>
<td>-2.14 (2.36)</td>
<td>5.26 (4.30)</td>
</tr>
<tr>
<td>IE</td>
<td>0.99 (2.01)</td>
<td>8.32 (1.75)</td>
<td>11.53 (2.18)</td>
<td>8.33 (2.69)</td>
</tr>
<tr>
<td>FE</td>
<td><strong>-2.22 (3.53)</strong></td>
<td><strong>1.91 (2.68)</strong></td>
<td><strong>-7.17 (3.89)</strong></td>
<td><strong>1.70 (3.96)</strong></td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-1.42 (8.53)</td>
<td>-6.04 (8.97)</td>
<td>-1.66 (8.90)</td>
<td>-1.06 (6.52)</td>
</tr>
<tr>
<td>IE</td>
<td>-7.20 (4.38)</td>
<td>-5.79 (4.96)</td>
<td>-23.17 (3.79)</td>
<td>-10.65 (3.41)</td>
</tr>
<tr>
<td>FE</td>
<td><strong>1.49 (2.45)</strong></td>
<td><strong>2.96 (1.93)</strong></td>
<td><strong>-0.51 (2.48)</strong></td>
<td><strong>1.18 (2.35)</strong></td>
</tr>
<tr>
<td><strong>Left</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>4.97 (7.17)</td>
<td>7.30 (7.00)</td>
<td>9.43 (6.98)</td>
<td>0.47 (5.04)</td>
</tr>
<tr>
<td>IE</td>
<td>3.18 (4.06)</td>
<td>4.66 (2.23)</td>
<td>15.00 (3.05)</td>
<td>5.45 (3.57)</td>
</tr>
<tr>
<td>FE</td>
<td><strong>28.12 (2.29)</strong></td>
<td><strong>16.47 (1.61)</strong></td>
<td><strong>25.49 (2.87)</strong></td>
<td><strong>24.37 (2.16)</strong></td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>0.99 (3.75)</td>
<td>0.91 (3.72)</td>
<td>-1.81 (6.54)</td>
<td>1.27 (1.50)</td>
</tr>
<tr>
<td>IE</td>
<td>2.19 (2.01)</td>
<td>2.61 (2.40)</td>
<td>-9.82 (1.66)</td>
<td>3.79 (1.17)</td>
</tr>
<tr>
<td>FE</td>
<td><strong>-4.30 (3.35)</strong></td>
<td><strong>-3.87 (2.42)</strong></td>
<td><strong>-1.92 (4.21)</strong></td>
<td><strong>-1.35 (3.88)</strong></td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-2.07 (6.85)</td>
<td>5.56 (6.25)</td>
<td>-3.25 (8.38)</td>
<td>-3.07 (7.60)</td>
</tr>
<tr>
<td>IE</td>
<td>-5.44 (3.43)</td>
<td>-32.07 (2.93)</td>
<td>-20.02 (3.64)</td>
<td>-9.27 (3.79)</td>
</tr>
<tr>
<td>FE</td>
<td><strong>3.02 (3.75)</strong></td>
<td><strong>5.84 (2.55)</strong></td>
<td><strong>2.79 (3.16)</strong></td>
<td><strong>-0.24 (2.33)</strong></td>
</tr>
</tbody>
</table>

To represent the error present in MVN when calibrated under the IC condition as compared to the gold standard, the **baseline error** was defined in section 4.5 as the dynamic parameters calculated for the IC condition. Figure 10 shows the dynamic difference DIFF_D as a function of time (frames) of the left limb of participant-1. As mentioned in section 4.5, the baseline error represents the disagreement in shape due to crosstalk or system error, as well as a difference in
shift due to difference in anatomical frame definition. Each subplot shows the DIFFD corresponding to each joint and plane; the bars represent the range of this difference. The boxplots of DIFFD is also shown on the right of each subplot. The means and standard deviations of DIFFD (μDIFFD(σDIFFD)) of the baseline error of all participants can be found in Table 5. The DIFFD of each participant represents his/her own baseline error, since it is the best agreement that can be obtained between PiG and MVN for that participant.

The shape of DIFFD in Figure 10 represents the difference between PiG and MVN at each point in time. This difference can be explained by instrumentation error or cross talk error. However, if both systems record about the same range of motion, then DIFFD should be about zero; as it is in most of the joint angles and planes. In the case of hip FE the mean of the error is above 20, this can be explained by the difference in anatomical frame definition between MVN and PiG due to sensor placement as mentioned above.

5.3 Assessing the effect of postural deviations in gait kinematics

(Objective 1)

5.3.1 Question 1

I. Is there an error introduced due to non-ideal calibration postures in Kinematics measurements?

Figure 11 shows the static difference in joint angle measurement between PiG and MVN during the calibration of the IC condition in green and the CG condition in red. The actual values of this table can be seen in Appendix I. The simulated posture was expected to add a deviation mainly in the sagittal plane. If the postural deviation had no effect on the MVN measurements, the difference between PiG and MVN should be close to zero, however this difference is high in that plane. Notice how the DIFFS values of knee and ankle FE during the IC condition are close to zero where as DIFFS of the CG condition are above 20, clearly indicating that MVN is not accounting for this flexion. In the hip case MVN values of IC condition are also about 20 due to the shift in hip FE angle towards the extension side as explained in the previous section. Although these results indicate crouch standing adds mainly a deviation in the sagittal plane,
achieving this posture might have required participants to rotate in other planes as well; that can be seen in ankle IE rotation.

When looking at the ankle IE DIFF_S of the IC condition, it can be seen that the values for participants 2 to 4 are close to -20, which means that MVN values are greater than PiG values. In addition to this, the DIFF_S of the CG condition for all participants were shifted to a more negative value. To understand why, the ankle IE angles of table in appendix 0, were inspected and it was found that all the PiG for both limbs of participants 1 to 3 are negative (between -21.57 and -7.38 degrees; -4.87 and 0.76 for right and left ankle of participant-1 respectively) indicating external rotation, whereas the MVN values are closer to zero (between -0.73 and 0.73, including all participants), which means that even during the IC condition, the MVN failed to measure the external rotation of the ankle. Furthermore, when inspecting the ankle IE DIFF_S of the CG condition of the appendix 0, it was found that the PiG values of each participant are even more negative (between -33.79 and -6.83 degrees, now including participant-1) than the PiG value of that participant during the IC condition, which might mean that the participants adjusted ankle rotation to achieve the crouch posture. In this case, however, the MVN values are still close to zero (between -0.47 and 1.14 degrees), suggesting that MVN failed to measure IE adjustment of the postural deviation. Why, the MVN failed to measure the ankle IE even during the N-Pose, needs to be explored.
These results show how the MVN fails to capture the simulated postural deviations. In a real situation, where the system is used with a real patient, a clinician or researcher using this MVN data could make incorrect conclusions about the patient’s posture. If the deviation was not detected even during the capture of the static posture right after the calibration, it is expected that the same error remains in the rest of the recording, during the walking trials.

Figure 12 shows the kinematics of the left limb of participant-1 according to PiG and MVN during the CG experimental condition. Table 6 shows the CMC values corresponding to the kinematics of this experimental condition of all participants.
According to Figure 12 a shift in FE joint angles for the hip, knee and ankle is evident. Comparing these plots to Figure 9, it can be seen that this shift is present in the FE angles of all joints. Even when the participant kept the hip, knees and ankle flexed at all times, as can be seen on the PiG kinematics (because he/she was walking with crouch gait), the MVN system measures some extension. Since for the calibration during crouch standing all FE values were considered to be those of a neutral standing, every time that the two contiguous body segments flex more than the starting flexed position, MVN measures them as flexion (degrees above zero). When the segments extend more than the starting flexed position, MVN will measure them as extension (degrees below zero).

As was mentioned above regarding Figure 11, a small shift in ankle IE can also be noticed. There is no apparent shift in any of the other planes.
Table 6: CMC of all participants corresponding to the crouch gait experimental condition. Values ≥0.995 are written as 1. Shading indicates poor (lightest) to excellent (darker) correlation. CMC complex values are displayed as not a number (NaN).

<table>
<thead>
<tr>
<th>Participant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Right</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>0.74</td>
<td>0.89</td>
<td>0.98</td>
<td>0.91</td>
</tr>
<tr>
<td>IE</td>
<td>0.74</td>
<td>0.77</td>
<td>0.94</td>
<td>NaN</td>
</tr>
<tr>
<td>FE</td>
<td>0.99</td>
<td>1.00</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>0.79</td>
<td>NaN</td>
</tr>
<tr>
<td>IE</td>
<td>0.88</td>
<td>0.93</td>
<td>0.91</td>
<td>0.89</td>
</tr>
<tr>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.98</td>
<td>0.99</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td>IE</td>
<td>0.85</td>
<td>0.83</td>
<td>0.87</td>
<td>0.81</td>
</tr>
<tr>
<td>FE</td>
<td>0.95</td>
<td>0.97</td>
<td>0.98</td>
<td>0.96</td>
</tr>
<tr>
<td><strong>Left</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>0.42</td>
<td>0.54</td>
<td>0.78</td>
<td>0.49</td>
</tr>
<tr>
<td>IE</td>
<td>0.79</td>
<td>0.84</td>
<td>0.89</td>
<td>0.51</td>
</tr>
<tr>
<td>FE</td>
<td>1.00</td>
<td>1.00</td>
<td>0.99</td>
<td>1.00</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>0.79</td>
<td>0.78</td>
<td>NaN</td>
<td>0.94</td>
</tr>
<tr>
<td>IE</td>
<td>0.92</td>
<td>0.94</td>
<td>0.95</td>
<td>0.94</td>
</tr>
<tr>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td>IE</td>
<td>0.91</td>
<td>0.85</td>
<td>0.91</td>
<td>0.90</td>
</tr>
<tr>
<td>FE</td>
<td>0.96</td>
<td>0.97</td>
<td>0.98</td>
<td>0.95</td>
</tr>
</tbody>
</table>

Because of the excellent agreement in hip, knee and ankle FE, and good agreement on knee and ankle IE, the CMC values did not change as compared to the IC condition, which supports the hypothesis that the postural deviation does not affect the shape but only the value. Hence, the CMC values in these planes and joints are not expected to change after applying the OC and PAC corrections.
Figure 13: Difference between PiG and MVN corresponding to crouch gait experimental condition (red) and baseline error (green) of left limb of participant-1. The bars represent the range within which this difference lies. The boxplots on the right of each graph represent the mean and dispersion of the difference throughout the gait cycle.

Figure 13 shows the difference between PiG and MVN corresponding to this experimental condition (red lines), as well as the baseline error (green lines). It can be seen that the difference between PiG and MVN of the CG condition is similar in shape and range of variation to the baseline error. This means that the system error is affecting both experimental conditions in the same way and it doesn’t change with the addition of the postural deviation error, instead, there is a shift on the mean of the system error. This shift is more evident on hip, knee and ankle FE since the postural deviations adds deviation on the sagittal plane. The shift can also be noticed, although at a lower degree, on ankle IE rotation.

What was explained for participant-1 can be generalized for participants across this study. Since it is already known that system error in CG condition is similar to the baseline error, a shift in the mean of this error can be seen as a difference in $\text{DIFF}_D$ between the CG condition and the baseline error shown in the bars of Figure 14. Values corresponding to Figure 14 can be seen in
the Appendix J. The mean $\text{DIFF}_D$ of CG condition of FE angles is higher than the baseline error mean $\text{DIFF}_D$, indicating that the PiG angles are larger than MVN angles in the CG condition. Again, this suggests that MVN is not measuring FE but PiG is. Smaller shifts can also be seen in ankle IE rotation as commented previously. The CMC values of the planes with apparent shift have very good to excellent agreement for all participants (Table 6). These results suggest that the error due to postural deviation in gait kinematics can be represented as a shift or offset in joint angles while the shape is not affected.

There are some other planes that also show a shift in the system error mean, for instance, knee IE. If this shifts are due to the postural deviation, then after applying the correction they should be reduced to about the same mean of the baseline error.

Measuring the offset angles during the walking trial in a scenario where the principal measuring tool is MVN could be hard. Therefore, one of the aims of this thesis was to answer whether the error found in dynamic trials can be corrected using the information of static capture of the calibration posture alone. To determine this it is important to know whether the error frame to frame is more or less constant. This is explored in the following section.
Figure 14: Mean and standard deviation of the difference deviation between PiG and MVN (DIFF<sub>D</sub>) corresponding to crouch gait experimental condition (red) and baseline error (green) for all participants; left limb (above) and right limb (below).
5.3.2 Question 2

II. Is the error constant throughout the gait cycle?

An additional check was done on the difference of PiG and MVN of the CG to see whether the difference from frame to frame was constant.

Figure 15 shows the derivative of DIFF\(_D\) in black, corresponding to the left limb of participant-1 during the CG condition. This derivative represents the change in difference between two consecutive time points. Table 7 shows the mean and standard deviation across frames of the derivative of DIFF\(_D\) of all participants. The mean of all of them is close to zero, and the standard deviation is small, indicating that the difference between PiG and MVN is nearly constant. However, it can be seen in Figure 15 that this change from frame to frame, although constant in most of the gait cycle, has some spikes at some points.

![Derivative of the difference between PiG and MVN (DIFF\(_D\)) (Left)](image)

**Figure 15:** Derivative of DIFF\(_D\) corresponding to the left limb of participant-1.
Table 7: Mean and standard deviation of the derivative of the difference between PiG and MVN corresponding to crouch gait experimental condition.

<table>
<thead>
<tr>
<th>Participant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip AA</td>
<td>-0.001 (0.31)</td>
<td>0.006 (0.31)</td>
<td>-0.010 (0.16)</td>
<td>0.00 (0.17)</td>
</tr>
<tr>
<td>IE</td>
<td>-0.002 (0.39)</td>
<td>-0.003 (0.41)</td>
<td>0.002 (0.22)</td>
<td>0.00 (0.31)</td>
</tr>
<tr>
<td>FE</td>
<td>0.006 (0.28)</td>
<td>0.001 (0.25)</td>
<td>0.008 (0.22)</td>
<td>0.00 (0.24)</td>
</tr>
<tr>
<td>Knee AA</td>
<td>-0.003 (0.37)</td>
<td>0.008 (0.58)</td>
<td>0.003 (0.17)</td>
<td>0.00 (0.33)</td>
</tr>
<tr>
<td>IE</td>
<td>0.002 (0.44)</td>
<td>-0.004 (0.41)</td>
<td>-0.004 (0.30)</td>
<td>-0.01 (0.34)</td>
</tr>
<tr>
<td>FE</td>
<td>0.008 (0.45)</td>
<td>0.001 (0.38)</td>
<td>0.008 (0.43)</td>
<td>0.01 (0.33)</td>
</tr>
<tr>
<td>Ankle AA</td>
<td>0.016 (1.10)</td>
<td>0.004 (1.05)</td>
<td>0.007 (0.99)</td>
<td>0.00 (0.55)</td>
</tr>
<tr>
<td>IE</td>
<td>0.007 (0.54)</td>
<td>0.006 (0.61)</td>
<td>0.006 (0.47)</td>
<td>0.01 (0.29)</td>
</tr>
<tr>
<td>FE</td>
<td>0.005 (0.56)</td>
<td>0.001 (0.54)</td>
<td>0.003 (0.47)</td>
<td>0.00 (0.52)</td>
</tr>
<tr>
<td>Left</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip AA</td>
<td>0.002 (0.46)</td>
<td>-0.005 (0.51)</td>
<td>-0.003 (0.41)</td>
<td>0.00 (0.38)</td>
</tr>
<tr>
<td>IE</td>
<td>0.011 (0.45)</td>
<td>-0.005 (0.39)</td>
<td>-0.001 (0.33)</td>
<td>0.00 (0.28)</td>
</tr>
<tr>
<td>FE</td>
<td>-0.002 (0.30)</td>
<td>0.001 (0.25)</td>
<td>0.000 (0.23)</td>
<td>0.00 (0.20)</td>
</tr>
<tr>
<td>Knee AA</td>
<td>-0.003 (0.26)</td>
<td>0.000 (0.33)</td>
<td>-0.005 (0.46)</td>
<td>0.00 (0.16)</td>
</tr>
<tr>
<td>IE</td>
<td>-0.007 (0.27)</td>
<td>-0.008 (0.37)</td>
<td>-0.008 (0.26)</td>
<td>-0.01 (0.21)</td>
</tr>
<tr>
<td>FE</td>
<td>-0.003 (0.35)</td>
<td>-0.002 (0.38)</td>
<td>-0.002 (0.44)</td>
<td>0.00 (0.32)</td>
</tr>
<tr>
<td>Ankle AA</td>
<td>0.002 (0.87)</td>
<td>0.000 (1.03)</td>
<td>0.000 (1.08)</td>
<td>0.00 (0.70)</td>
</tr>
<tr>
<td>IE</td>
<td>0.003 (0.49)</td>
<td>0.000 (0.58)</td>
<td>0.007 (0.42)</td>
<td>0.01 (0.35)</td>
</tr>
<tr>
<td>FE</td>
<td>-0.002 (0.70)</td>
<td>0.001 (0.68)</td>
<td>-0.006 (0.44)</td>
<td>0.00 (0.54)</td>
</tr>
</tbody>
</table>

When comparing DIFFD and the derivative of DIFFD overtime in Figure 16, although the overall shape of the waveforms is similar as shown by the CMC values, there are some spikes in the MVN angles, shown in the black circles, which are not seen in the PiG angles. These spikes were found in all joints and all participants, especially in the beginning and the middle of the gait cycle. The reason of this spikes at these specific points needs to be investigated. It can be seen that these spikes contribute to increasing the standard deviation of the derivative even as the mean remains relatively small (e.g. left ankle AA or participant-1 has a mean DIFFD of 0.016±1.10 degrees). The points of the derivative with highest variation are only at those spike locations; however, the rest of the signal is maintained around zero. Therefore, it is possible to assume the difference between PiG and MVN as constant and use the information of the static trial to correct frame by frame the kinematics of the dynamic trial.
5.4 Applying OC and PAC towards improving the accuracy of gait kinematics (objective 2)

The previous section established that error due to postural deviations can be considered as a shift in angle. It was also shown that frame-by-frame correction of the dynamic kinematics is possible since the error can be considered to be constant. This section shows the results after applying both correction approaches.
5.4.1 Orientation Correction (OC)

I. If the initial body segments orientation is known, how well will the orientation correction (OC) reduce the error?

Figure 17: Kinematics of Crouch Gait according to PiG (blue) and MVN after the orientation correction was applied (MVN OC, orange).

Figure 17 shows the kinematics measured with PiG (blue) and MVN after the OC was applied (orange).

Figure 18 shows the difference between these two measurements in orange and the baseline error in green. The data is also an example of the left limb of participant-1. The OC corrected for the postural error as can be seen in the hip, knee and ankle FE as well as ankle IE. The shift was removed, (i.e. the mean of the system error of the CG is about the same as the mean of the baseline error), while the CMC were kept the same.
Figure 18: Difference between PiG and MVN corrected corresponding to the baseline (green) and the crouch gait Kinematics corrected with the orientation correction (orange).

The improvement for all participants can be seen in the mean of $\text{DIFF}_D$ shown in Figure 21. The mean of $\text{DIFF}_D$ of the CG condition was brought down to the level of the $\text{DIFF}_D$ of the baseline error in the FE angles of all joints of all participants. Recalling that the hip FE baseline error had a high mean $\text{DIFF}_D$ (ranging between 16.47 and 28.80 across both limbs of all participants as shown in Table 5) due to the sensor placement that affected the hip FE angle, the $\text{DIFF}_D$ after the OC was applied improved to the range between 6.70 and 14.72. This suggests that the OC approach could also correct for the initial difference in anatomical frame. Surprisingly, the shift was also decreased on hip IE to be smaller than the baseline error, potentially due to the same reason as for the hip FE. Regarding the ankle IE angles that was not even detected during IC condition, the mean $\text{DIFF}_D$ after applying the OC was smaller (between -7.07 and -0.28 degrees) than the mean $\text{DIFF}_D$ of the baseline error of that joint (between -32.07 and -5.44 degrees). On the contrary, it can be noticed that the OC tends to add a shift on the mean $\text{DIFF}_D$ of the AA angles and knee IE angles.
The CMC of the OC of all participants are summarized in Figure 22 and Table 8. Some of the FE and ankle IE CMC values among all joints and all participants were decreased by up to 0.03 (e.g. right ankle FE of participant 3) after applying the OC, however, they still have excellent correlation. Similarly, other planes had worse correlation after applying the correction. This can be explained because the OC requires matrix multiplication, potentially making changes to the shape of the waveform as can be seen in the knee AA of Figure 17.

5.4.2 Planar Angle Correction (PAC)

II. If the initial offset angles are known, how well will the planar angle correction (PAC) reduce the error?

Figure 19 shows the Kinematics corresponding to PiG and MVN after the PAC was applied, and Figure 20 shows the difference between these two measurements in purple and the baseline error in green. The example of the left limb of participant-1 is presented.

The results of the PAC are similar to the OC for the planes where major improvements were expected. The FE of all participants improved, which are evidenced by the mean DIFFD of the FE angles of the CG condition that was brought to the level of the near-zero baseline error DIFFD (see Figure 21. The values corresponding to this figure can be found in Appendix J). The mean DIFFD of hip was not reduced to zero but was reduced to less than the DIFFD range according the baseline error (ranging between 16.47 and 28.80 as shown in Table 5), between 6.87 and 9.29, taking right as well as left hip into account. The same happened to the ankle IE angles; the range of mean DIFFD decreased from the range between -32.07 and -5.44 degrees (baseline error) to the range between -2.2 and 2.10 degrees (CG condition). Similar to the OC, it also reduced the shift of hip IE and introduced a shift on AA angles. Furthermore, a correction can be noticed on the knee IE, especially on the left limb of all participants. Since this correction approach was just implemented as the addition of planar angles, the CMC values remain the same as the CMC values before the correction; that can be seen in Figure 22 and Table 8.
Figure 19: Kinematics of Crouch Gait according to PiG (blue) and MVN after the planar angle correction was applied (MVN PAC, purple).

Figure 20: Difference between PiG and MVN corrected corresponding to the baseline (green) and the crouch gait Kinematics corrected with the orientation correction (orange).
Figure 21: Mean and standard deviation of the difference deviation between PiG and MVN (DIFF$_D$) according to the baseline error (green) and the crouch gait experimental condition before (red) and after orientation correction (orange) and planar angle correction (purple).
Table 8: CMC values corresponding to the crouch gait experimental condition before and after orientation and planar angle correction. Values $\geq 0.995$ are represented as 1.00. Shading indicates poor (lightest) to excellent (darker) correlation. The different colors also correspond to the different conditions.

<table>
<thead>
<tr>
<th>Participant</th>
<th>Crouch Gait Uncorrected</th>
<th>Crouch Gait Corrected</th>
<th>MVN OC</th>
<th>MVN PAC</th>
<th>Crouch Gait Uncorrected</th>
<th>Crouch Gait Corrected</th>
<th>MVN OC</th>
<th>MVN PAC</th>
<th>Crouch Gait Uncorrected</th>
<th>Crouch Gait Corrected</th>
<th>MVN OC</th>
<th>MVN PAC</th>
<th>Crouch Gait Uncorrected</th>
<th>Crouch Gait Corrected</th>
<th>MVN OC</th>
<th>MVN PAC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>AA</td>
<td>0.74</td>
<td>0.69</td>
<td>0.74</td>
<td>0.89</td>
<td>0.79</td>
<td>0.89</td>
<td>0.98</td>
<td>0.97</td>
<td>0.98</td>
<td>0.91</td>
<td>0.95</td>
<td>0.96</td>
<td>0.95</td>
<td>0.91</td>
<td>0.95</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.74</td>
<td>0.61</td>
<td>0.74</td>
<td>0.77</td>
<td>0.83</td>
<td>0.77</td>
<td>0.94</td>
<td>0.95</td>
<td>0.94</td>
<td>0.91</td>
<td>0.95</td>
<td>0.96</td>
<td>0.95</td>
<td>0.91</td>
<td>0.95</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>1.00</td>
<td>0.98</td>
<td>1.00</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Knee</td>
<td>AA</td>
<td>NaN</td>
<td>0.83</td>
<td>NaN</td>
<td>NaN</td>
<td>0.61</td>
<td>NaN</td>
<td>0.79</td>
<td>0.91</td>
<td>0.79</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.88</td>
<td>0.92</td>
<td>0.88</td>
<td>0.93</td>
<td>0.92</td>
<td>0.93</td>
<td>0.91</td>
<td>0.85</td>
<td>0.91</td>
<td>0.91</td>
<td>0.94</td>
<td>0.93</td>
<td>0.94</td>
<td>0.91</td>
<td>0.94</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.98</td>
<td>0.98</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Ankle</td>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.85</td>
<td>0.85</td>
<td>0.85</td>
<td>0.83</td>
<td>0.80</td>
<td>0.83</td>
<td>0.87</td>
<td>0.83</td>
<td>0.87</td>
<td>0.81</td>
<td>0.69</td>
<td>0.81</td>
<td>0.69</td>
<td>0.81</td>
<td>0.69</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.95</td>
<td>0.94</td>
<td>0.95</td>
<td>0.97</td>
<td>0.97</td>
<td>0.97</td>
<td>0.98</td>
<td>0.95</td>
<td>0.95</td>
<td>0.96</td>
<td>0.95</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
<td>0.95</td>
</tr>
<tr>
<td>Left</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>AA</td>
<td>0.42</td>
<td>0.59</td>
<td>0.42</td>
<td>0.54</td>
<td>0.59</td>
<td>0.54</td>
<td>0.78</td>
<td>0.79</td>
<td>0.78</td>
<td>0.49</td>
<td>0.54</td>
<td>0.49</td>
<td>0.49</td>
<td>0.49</td>
<td>0.49</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.79</td>
<td>0.72</td>
<td>0.79</td>
<td>0.84</td>
<td>0.89</td>
<td>0.84</td>
<td>0.89</td>
<td>0.79</td>
<td>0.89</td>
<td>0.51</td>
<td>0.56</td>
<td>0.51</td>
<td>0.51</td>
<td>0.51</td>
<td>0.51</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>1.00</td>
<td>0.98</td>
<td>1.00</td>
<td>1.00</td>
<td>0.98</td>
<td>1.00</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>1.00</td>
<td>1.00</td>
<td>1.00</td>
<td>1.00</td>
<td>1.00</td>
<td>1.00</td>
</tr>
<tr>
<td>Knee</td>
<td>AA</td>
<td>0.79</td>
<td>0.56</td>
<td>0.75</td>
<td>0.78</td>
<td>0.34</td>
<td>0.78</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>0.94</td>
<td>0.94</td>
<td>0.94</td>
<td>0.94</td>
<td>0.94</td>
<td>0.94</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.92</td>
<td>0.75</td>
<td>0.92</td>
<td>0.94</td>
<td>0.89</td>
<td>0.94</td>
<td>0.95</td>
<td>0.88</td>
<td>0.95</td>
<td>0.94</td>
<td>0.90</td>
<td>0.94</td>
<td>0.90</td>
<td>0.94</td>
<td>0.90</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.98</td>
<td>0.98</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Ankle</td>
<td>AA</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td></td>
<td>IE</td>
<td>0.91</td>
<td>0.84</td>
<td>0.91</td>
<td>0.85</td>
<td>0.61</td>
<td>0.85</td>
<td>0.91</td>
<td>0.84</td>
<td>0.91</td>
<td>0.90</td>
<td>0.80</td>
<td>0.90</td>
<td>0.80</td>
<td>0.90</td>
<td>0.80</td>
</tr>
<tr>
<td></td>
<td>FE</td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
<td>0.97</td>
<td>0.96</td>
<td>0.97</td>
<td>0.98</td>
<td>0.98</td>
<td>0.98</td>
<td>0.95</td>
<td>0.92</td>
<td>0.95</td>
<td>0.92</td>
<td>0.95</td>
<td>0.92</td>
</tr>
</tbody>
</table>

The summary of DIFF range across participants is shown in Table 9. Results presented in this section suggest that the OC, as well as the PAC, can correct the gait kinematics for the postural deviation; especially for the FE angles of all joints. It was also noticed that small deviations on other planes like ankle IE that were not detected even during the IC condition, could also be corrected. In addition to it, both correction approaches reduced the DIFF to be smaller than the baseline error of the hip FE and IE angles. This decrease suggests that the correction approaches could have also corrected for errors that existed even during the IC condition as a result of different anatomical frame definition.
Figure 22: CMC values corresponding to crouch gait before correction and after correction with orientation and planar angle corrections. CMC that were complex numbers are not displayed. CMC of ankle AA are all complex numbers.

5.4.3 Comparison between Orientation Correction (OC) and Planar Angle Correction (PAC)

III. Are OC and PAC approaches equally effective in reducing the error?

When comparing the two correction approaches, PAC seems to outperform OC since the mean across participants of the error in FE angles (the principal target errors to correct) after applying PAC is smaller (range from -3.93 to 9.18 degrees) than the mean of the error after the OC correction (range from -5.42 to 14.72 degrees). Specifically, PAC reduced the error at the hip (between 6.87 and 9.29 degrees) more than what the OC did (between 6.70 and 14.72 degrees) as compared to the range between 16.47 and 28.80 degrees corresponding to the baseline error, however, both showed a good improvement. In general, the range of the mean DIFFD after the PAC correction is smaller than after the OC correction. OC introduced error in planes where
there was no error related to postural deviation: AA angles. Furthermore, the OC had an effect upon the shape of the waveforms, which was reflected on the CMC values that were decreased. Although PAC also introduced an error on AA angles, it does not change the shape of the waveform, leaving all the CMC values unchanged. The reason of this introduced shift on AA angles, needs more investigation.

Table 9: Summary of DIFFD range across participants. The range of the mean of DIFFD across participants is shown for the baseline error, the CG condition as well as the two correction approaches. Results are shown in degrees. Values in bold and italics represents the main improvements in FE angles of all joints. Underlined values represent improvement in other planes and values in italics represents errors added after the correction was applied. Notice how the range is smaller for the PAC correction where there was an improvement.

<table>
<thead>
<tr>
<th></th>
<th>Baseline error</th>
<th>Crouch gait</th>
<th>MVN OC</th>
<th>MVN PAC</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>min</td>
<td>max</td>
<td>min</td>
<td>max</td>
</tr>
<tr>
<td><strong>Right</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-5.86</td>
<td>-0.38</td>
<td>-7.35</td>
<td>0.69</td>
</tr>
<tr>
<td>IE</td>
<td>-8.44</td>
<td>-1.56</td>
<td>-12.18</td>
<td>-0.53</td>
</tr>
<tr>
<td>FE</td>
<td>18.58</td>
<td>28.80</td>
<td>34.62</td>
<td>67.10</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-2.22</td>
<td>5.26</td>
<td>-1.87</td>
<td>11.96</td>
</tr>
<tr>
<td>IE</td>
<td>0.99</td>
<td>11.53</td>
<td>6.15</td>
<td>17.55</td>
</tr>
<tr>
<td>FE</td>
<td>-7.17</td>
<td>1.91</td>
<td>33.79</td>
<td>53.54</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-6.04</td>
<td>-1.06</td>
<td>-1.90</td>
<td>0.00</td>
</tr>
<tr>
<td>IE</td>
<td>-23.17</td>
<td>-5.79</td>
<td>-33.22</td>
<td>-10.77</td>
</tr>
<tr>
<td>FE</td>
<td>-0.51</td>
<td>2.96</td>
<td>17.02</td>
<td>25.12</td>
</tr>
<tr>
<td><strong>Left</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>0.47</td>
<td>9.43</td>
<td>-2.65</td>
<td>6.09</td>
</tr>
<tr>
<td>IE</td>
<td>3.18</td>
<td>15.00</td>
<td>-2.50</td>
<td>12.88</td>
</tr>
<tr>
<td>FE</td>
<td>16.47</td>
<td>28.12</td>
<td>36.23</td>
<td>66.71</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-1.81</td>
<td>1.27</td>
<td>-2.19</td>
<td>6.69</td>
</tr>
<tr>
<td>IE</td>
<td>-9.82</td>
<td>3.79</td>
<td>-5.23</td>
<td>17.50</td>
</tr>
<tr>
<td>FE</td>
<td>-4.30</td>
<td>-1.35</td>
<td>31.70</td>
<td>51.98</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-3.25</td>
<td>5.56</td>
<td>-2.36</td>
<td>4.58</td>
</tr>
<tr>
<td>IE</td>
<td>-32.07</td>
<td>-5.44</td>
<td>-26.16</td>
<td>-8.46</td>
</tr>
<tr>
<td>FE</td>
<td>-0.24</td>
<td>5.84</td>
<td>18.21</td>
<td>25.77</td>
</tr>
</tbody>
</table>
In practice, the Vicon system would not be used as a reference to know the true orientation of body segments or joint angles. Since the PAC is easier to implement, the use of planar digital images was proposed as an alternative to capture the true body posture, from which the joint angles can be measured. Therefore, the comparison between joint angles measured according to PiG and planar images is shown in next section.

Originally, it was expected that the OC would outperform the PAC since it would directly correct the orientation of the body segments that are measured wrongly. However, it has been shown here that the error can be considered as a shift in the values of kinematics, and that the PAC alone seems to be enough to correct for the postural deviation error. Therefore, the idea of using images to measure planar joint angles appears supported by results in this study.

5.4.4 **Use of planar digital images to implement PAC**

IV. Can angles measured from planar digital images taken with a standard camera be used to implement the PAC?

Table 10 shows the joint angles according to PiG and the planar digital images, the difference between them and the percentage of error of this difference relative to PiG. Only the right FE angles were measured and those are presented here. The angles measured from the planar images with less percentage error are the hip FE angles, which have a $\text{DIFF}_S$ range across participants of -4.54 to 7.31 (%Error ranges from 6% to 13%). The knee FE has a range from -1.69 to 8.41 (%Error ranges 0 to 17%). Finally, the joint angle with highest percentage of error was ankle FE that can be measured expecting a $\text{DIFF}_S$ range between 1.36 and 14.04 (%Error ranges 6% to 44%). The dispersion of these ranges can be seen in Figure 21. The ankle has a smaller range of FE values as compared to the other two joints, therefore the percentage of error is higher, however, notice how the $\text{DIFF}_S$ range is within the same range as the other two joints, as exception of the outlier value (14.04 degrees of trial 2, participant-4). That indicates the overall performance of the measurement of planar angles from the images. The overall range and percentage of error of the $\text{DIFF}_S$ across all joints and all participants is -4.54 to 14.04 with percentage of error from 0% to 44% (removing the outlier range is -4.54 to 8.60 and %Error 0 to 32%). It was mentioned in the previous section, the participants showed external rotation, therefore, their feet could have also rotated from the plane of the image (parallel to the
participant’s sagittal plane) influencing how the ankle FE angles are calculated, potentially showing smaller values than the true ones.

Table 10: Comparison of the joint angle of the static posture according to PiG, planar digital images, the difference between them and the percentage of error of the difference relative to PiG (DIFF$_S$ displayed as 0.00 had a value of 0.001 degrees and 0% error represents an error <0.002%). Results are shown in degrees. All percentages are shown as positive values.

<table>
<thead>
<tr>
<th></th>
<th>Vicon Planar image</th>
<th>DIFF$_S$ % Error</th>
<th>Vicon Planar image</th>
<th>DIFF$_S$ % Error</th>
<th>Vicon Planar image</th>
<th>DIFF$_S$ % Error</th>
<th>Vicon Planar image</th>
<th>DIFF$_S$ % Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>52.20</td>
<td>46.88</td>
<td>5.32</td>
<td>10</td>
<td>34.97</td>
<td>39.52</td>
<td>-4.54</td>
<td>-13</td>
</tr>
<tr>
<td>Knee</td>
<td>47.57</td>
<td>43.44</td>
<td>4.13</td>
<td>9</td>
<td>40.69</td>
<td>42.38</td>
<td>-1.69</td>
<td>-4</td>
</tr>
<tr>
<td>Ankle</td>
<td>56.98</td>
<td>52.70</td>
<td>4.29</td>
<td>8</td>
<td>38.51</td>
<td>38.51</td>
<td>0.00</td>
<td>0</td>
</tr>
</tbody>
</table>

When measuring the angles from the images, each of the body segments was defined with the two markers that where on that segment (e.g. foot segment is the line joining heel and toe markers, see Figure 5). The difference in angle measurement between PiG and the planar digital images might also be due to differences in the definitions of the body segments.

In this study the smallest value of FE DIFF$_D$ that was measured is 18.2 corresponding to left ankle FE of participant 3 (see Figure 14) and even then the higher DIFF$_S$ found from the joint angles measured with the images (8.60, without considering the outlier) would still offer a significant correction.

Figure 24 shows an example of the right limb of participant-1 after correcting the kinematics with the PAC using the planar joint angles measured from the digital images. Figure 25 shows the DIFF$_D$ mean and the standard deviation of all participants.
Dispersion of the difference between PiG and planar digital images ($\text{DIFF}_S$) across all participants

Figure 23: Dispersion of the difference between PiG and planar digital images ($\text{DIFF}_S$) corresponding to each joint across all participants.

Figure 24: Kinematics of Crouch Gait according to PiG (blue) and MVN after the planar angle correction was applied (MVN PAC with planar digital images, black).
Figure 25: Mean and standard deviation of the difference deviation between PiG and MVN (DIFF\textsubscript{D}) according to the baseline error (green) and the crouch gait experimental condition before (red) and after and planar angle correction using the angles measured from the planar digital images (black).

Similar to the results of the PAC, the PAC using the planar joint angles from the images also corrects for the postural deviation error in all joints as well as the anatomical frame definition on the pelvis that causes the virtual extension on the hip FE angles. Even the system error of the ankle FE angles were decreased to about the same level of the baseline error. Only the system error at the ankle of participant-4 was not completely reduced, which was expected given the larger error found in the planar ankle angles of this participant. The correction done on hip and knee FE angles of all participants as well as the ankle angles of 3 out of 4 participants suggests that this technique is a good approach to correct for the FE kinematics.
6. Discussion

6.1 Key findings

The overall goal of this study was to assess the effect that postural deviations have on gait kinematics when measured with the MVN BIOMECH system. Two correction approaches, the orientation correction, OC, and the planar angle correction, PAC were explored as a means to account for these errors and to establish a basis for the further development of a correction technique. To achieve these goals, the gait kinematics of four able-bodied individuals simulating crouch gait was analyzed before and after applying both correction approaches. So that the effect of postural deviations in gait kinematics was understood, the system error of the crouch gait condition (CG) was compared against the system error of the ideal calibration condition (IC), the baseline error. To implement the correction approaches, the true body posture was captured with the Plug-in Gait model using a camera-based system (Vicon). Both the orientation of body segments and the joint angles were considered as the true orientation and joint angles used to correct for the CG kinematics. In the OC approach the orientation of body segments measured with MVN was corrected with a right matrix multiplication by the true orientation, to add the rotation of body segment from the N-Pose to the true posture. After that the kinematics were re-calculated. In the PAC approach, the true joint angles were seen as an offset from the measured joint angles. In this case the true joint angles measured according PiG were added to the measured gait joint angles. As a means to assessing the performance of the correction approaches, the system error of the corrected CG condition was compared against the baseline error.

The most important findings, regarding the error of postural deviation on joint angles is that postural deviation can be considered as a shift in value of the system error. Additionally, since the change in difference between the PiG and MVN kinematics is relatively constant, the information of the true body posture captured during the calibration may be sufficient to correct the gait kinematics frame by frame. That conclusion was supported by the findings that the mean of the derivative of the difference between PiG and MVN was close to zero with small standard deviation. It was also found that the OC as well as the PAC can correct for the postural deviation in the FE angles of all joints since they can reduce the difference in gait kinematics between PiG
and MVN to about zero. Additionally, when trying to explore other planes it was also found that, although the error of the postural deviation is smaller than the system error, the PAC tends to outperform since overall it reduced the postural deviation error more than the OC did.

Finally, it was also shown that since PAC is easier to implement, the joint angles needed to implement this approach can be measured from planar digital images with an expected range of error between -5.54 and 14.04 degrees.

6.2 Strengths and limitations

To the best of the author’s knowledge, this study was the first attempt to assess and correct experimentally for postural deviation of an IMMUs-based system such as the MVN BIOMECH system that relies on predefined postures (N-Pose) to calibrate the system. Other studies in the past have also addressed the issue of knowing the true body postures with sensor calibration approaches [15], [80], [99], [104], [105]. In this study two correction approaches (OC and PAC) were presented as a different way to account for true body postures during the calibration. These correction approaches provide a basis for developing an easy to use and time-effective alternative using basic digital photographic images; this has the potential to eliminate the need for extra measurement steps and calibration devices that are part of other techniques.

The main idea described here was that it is possible to correct the gait kinematics by applying a constant offset value, if that value is known. It has been commented on the literature that the difference in coordinate systems adds an offset between the inertial- and the camera-based systems [78], [79], [99]. In these studies it was shown that removing the offset reduces the error between the kinematics measured with the two systems while keeping the shape similarity. This literature supports the basis of the correction technique presented here.

Although it was shown in this work that the postural deviation could be seen as a shift in value, since the true orientation of body segments are deviated from the assumed orientation of the N-Pose, it could also be considered as perturbations of the anatomical frame definition, like the ones presented in [81]. Therefore, the effect of postural deviations could also add crosstalk errors. In this study, the crosstalk error was considered as part of the system error, hence this
error as an effect of postural deviation was neglected. Implementing the analysis applied in [81] to postural deviations would make clear the effect of more complex postural deviations.

These results suggest that both the OC and PAC can correct the gait kinematics for the postural deviation, especially on the FE angles of all joints, and the ankle IE. It was noticed that it is challenging to ensure that sensors are aligned to body segments even for able-bodied people during the N-Pose calibration. This was noticed in the hip FE angles and ankle IE rotation. The most interesting finding regarding the correction was that PAC as well as OC seem to correct for this anatomical frame definition errors present even during the IC condition. The mean \( \text{DIFF}_D \) of hip FE were reduced even more than the baseline error. That was also evident when looking at unexpected improvements on hip IE. This finding suggest that the best place to attach the sacrum sensor is at the same height of the PSIS, this would create a pelvic reference frame similar to the PiG reference frame. This recommendation, should be implemented in future studies. Additionally, the effect of changing the sensor location on other body segments, together with the correction approaches to correct for potential shifts in kinematic values due to this sensor misplacement should be explored.

The CMC values found in this work are similar to the ones found in studies that evaluated the IMMUs-based system against the camera-based system [78], [79], [99]. In literature it has being described how the performance of kinematic measurements by this systems is better for the sagittal plane. It was found that even during the IC condition (the system was calibrated with the N-Pose), MVN failed to measure the ankle external rotation. In addition to the anatomical definition difference that has been suggested in the literature as a reason for this performance, another potential reason could be that even natural standing, some people deviate from the N-Pose assumption. However, it was shown that the OC as well as the PAC corrected for that initial deviation. Hence, a possibility to improve kinematics measurements even with the IC conditions could also be explored for both correction approaches.

Even though both approaches could correct well the kinematics of the target sagittal plane of all joints and some other additional planes (hip and ankle IE), PAC seems to outperform OC since in general the mean of the error across participants after applying PAC was smaller than the
mean of the error after the OC correction. Furthermore, PAC does not affect the shape of the waveforms.

Although the study sample size was small and only one postural deviation was simulated, thus potentially limiting the generalizability of the findings, the repeatable results across participants on FE angles (IE and AA were not expected to be repeatable due to the known disagreement between systems) support the internal validity of the study.

Although care was taken to minimize errors when applying the study protocols there are some known errors in both camera-based as and IMMU-based systems that might have hindered the effect of the postural deviations. Plug-in Gait was considered as the gold standard, however it is highly dependent on accurate marker placement [7] and the correct identification of anatomical landmarks [106]. Incorrect marker placement can lead to wrong definition of anatomical frame, hence, introducing the above discussed crosstalk errors that can potentially affect the measurements as well.

Another known source of error is the skin movement artifact [107]. In this study movement artifact errors were minimized on the shank and thigh makers by wrapping with transpore tape the Styrofoam blocks (where they were placed) around the limbs. The PSIS and ASIS markers could have been affected by the motion of the strap where the sensor was placed. Similar to marker misplacement and skin artifacts, there is an error associated with the movement of the straps where the sensors are attached. One study compared a system similar to MVN and put the marker on a lycra suit. They suggest that attachment of the sensors on the suit adds an error similar to the known skin artifact [79].

In addition to the motion artifacts, the MVN sensors also suffer from measuring errors: drift error and magnetic disturbances. The drift error on the MVN system can be corrected for with the addition of the magnetometers that define the heading of the sensor (declination from the magnetic north). However, given the magnetic disturbances inside the laboratory, MVN system also provided the KiC algorithm with no magnetometer to be used in such situations. Removing the magnetometer, however, increases the probability of experiencing drift error. Preliminary experiments were run in the laboratory to determine which fusion engine to use, and it was found that the regular filter (XKF-3) that uses magnetometer performed best. Additionally, when
looking at the results of a different experiment to compare kinematics measured outside and inside the gait laboratory (shown in Appendix F) the results are similar to the ones found in [81]. That suggests that, when measuring inside the laboratory, the anterior-posterior axis (the one estimated by the magnetometer) being affected by the magnetic distortions can be considered as a perturbation in this axis, consequently introducing the crosstalk effect. However, results from a study that validated the MVN system against a camera-based system [108] used the KiC algorithm and found similar results to the ones found in this study, in terms of kinematics similarity (better for FE and not as good for IE and AA). Therefore, since the end use of the sensors is to be used outside, using the XKF-3, could still be valid for this study, although exploring the use of the other fusion algorithm should be done as part of future work.

Besides the magnetic distortion, IMMUs-based systems are also prone to drift, especially during longer periods of recording. The trials recorded during the experiments were short as participants only walked a 6m distance. In the results, it was shown that the system error can be considered constant, however, during longer recording periods this might not be the case. In the future, longer trials should be explored to study the effect of drift as a function of time.

The metric $\text{DIFF}_D$ was used to represent the system error including all the types of error mentioned above, that are inherent in both the IMMUs-based and the camera-based systems. This was done to isolate the postural deviation error from all the other sources of error. However, there is another error that was not mentioned before and was evident when analyzing the data. It was noticed that the FE baseline error had a high range of variation even when they showed an excellent shape similarity (CMC values $\geq 0.95$). It can be seen that the baseline error has a better agreement during the first 50% of the gait cycle and then there is a drastic change for the other half. This could be because of a time lag caused during the manual alignment of the signals and could be corrected by aligning the signals automatically using methods such as the autocorrelation function. However, when looking separately at the stance phase and swing phase of the FE angles, it can be seen that the agreement is better on the stance phase part. Potential factors to this dependency on gait cycle phase could be the range of motion of the gait cycle phase; in other words, the accuracy of the two systems can be different at different gait cycle phases. For instance, it is known that the skin artifact causes the marker move over the actual bony landmark (e.g. greater trochanter) when the skin is stretched as the joint extends or flexes.
This will cause the joint center calculated by the camera-based system to be shifted back and forth along the different phases of the gait cycles, affecting the calculation of the true joint motion. In contrast, the IMMUs-based system relies on the magnetic field calculation potentially adding crosstalk error even to flexion-extension angles, although at a lower degree. No study has looked at the validation of the IMMUs system against camera-based systems to analyze gait-cycle-phase-dependent error. Therefore, more research needs to be done to understand this dependency. This would provide a more thorough evaluation of the system, which would help the community of clinicians and researchers decide whether these IMMUs-based systems could be used as clinical tools and under what conditions.

Regarding the correction approaches, although it was found that OC as well as PAC can correct for the FE angles, it was determined that the OC approach increased the errors in knee IE and AA angles. The orientation of body segments that was considered as true orientations of body segments was calculated from the marker position recorded by Vicon, following the coordinate systems described in the PiG model. Moreover, the “Chord function” used to estimate joint centers was implemented using the Isqnonlin MATLAB function. According to Appendix C, the angles measured from the calculated orientations were not totally identical to PiG joint angles. An offset was seen in most of the angles. This might be due to a different anatomical frame definition, likely stemming from the way that the function was implemented. Furthermore, the PiG calculates some angles during the calibration of the subject such as tibia offset [109]. This tibia offset is used to create a second tibia reference frame (untorsioned tibia) used solely to calculate knee angles. Differences in calculating the untorsioned tibia will create a slightly different reference frame. That could explain why the OC affected the knee angles. A similar situation happens when calculating foot offsets.

Additionally, it was mentioned that as PiG does not directly output orientations, they were calculated based on the marker position and PiG equations (PiG orientations, see Appendix C). The reason why OC could have changed the shape of the waveforms is because it directly changes the orientation of the MVN measured body segments. As noted in the appendix, angles calculated based on the PiG equations and the calculated PiG orientations have a perfect agreement for FE angles, however for AA as well as IE some offsets and change of shape can be seen. That suggests that the calculated PiG orientations introduced some round-off error while
performing the quaternion-to-matrix transformation or the matrix multiplications during the OC. This situation could be specially noticed in the change of shape of knee AA and IE of the results presented in this study.

It was pointed out in section 4.1.2.2 that body segments orientation measured with the MVN are expressed in the global reference frame of the sensors and PiG orientation are expressed in the PiGRF. The X axis of the PiGRF faces the north of the gait laboratory. Participants were asked to face the geographic north of the room as an intent to align both reference frames (assuming that the magnetic north and the geographic north of the room are equivalent). However, the geographic north of the room might not be the same as the magnetic north of the room potentially affecting the orientation correction when applying the change of basis from one reference frame to the other. To add simplicity in calculus the declination was neglected, however, results of the OC could improve if the change of basis from the PiGRF to the GRF accounts for the actual magnetic declination. The calculation of the magnetic declination can be seen in appendix 0.

Finally, regarding the use of planar digital images to measure the joint angles other studies, have measured the knee FE angle and they found that the range of the error of true vs measured angle was (0 to 5 degrees with standard deviation from 1.1 to 2.4 degrees [29]) similar to the ones found in this study including all angles (-5.54 to 8.60, with one outlier or 14.04). Even the ankle joint angles that had the highest percentage of error (6% to 32%, with the 44% outlier) had a range of DIFFS that was similar to literature. In a study that assessed the reliability of standing posture to be implemented as a diagnostic approach, found that the angles of the cervical, thoracic and lumbar spines from planar images ranged from -38.23 to 35.63 degrees (similar to the results found in this thesis for knee and hip angles) with a standard deviation ranging from 5.58 to 14.05 degrees [27]. The results from literature suggest that the errors found in this study are usual for this photogrammetric technique. In addition to that, it was also shown that the PAC using the planar joint angles measured from the images were able to correct for the FE angles of all participants, with exception of the ankle of participant-4. These results support the potential of planar angles measured from digital images to apply the PAC.
Although we found our errors of measuring angles from planar digital images to be acceptable and similar to that of other published literature, in the presence of more complex deviations this technique might not work as well. Two studies have shown that rotating the limb from the plane of the camera of 15 degrees can produce errors of 5 degrees or less, and rotating it about 30 degrees can produce errors of 4 degrees [22], [29]. This could have also introduced errors in ankle angles, since it was shown that participants had external rotation even during normal standing. Adjusting the camera to be parallel to the sagittal plane of each joint as opposed to body sagittal plane might improve ankle angles. In that case, the accuracy of the calculation of ankle angle from planar images would increase, however, it would take time to adjust the camera to each plane. It has also been shown that angles measured close to the edge of the image tend to have higher errors [22], since the ankle joint was closer to the edge than knee, the ankle was affected more by this error. Again, acquiring the image for individual joints would be ideal, however this would add time to the evaluation. Hence, a more robust method to measure the true body posture should be used.

The calibration of any motion analysis system such as the camera-based as well as the IMMUs-based system is required prior to its use. The correction technique will not be an exception. It would be expected that the capture of the body posture should be acquired every time the system is calibrated, since the person would not be able to assume the exact same posture every time, especially if they are patients. In addition to the normal calibration needed between sessions, the current stage of the IMMUs-based system requires frequent calibration given the drift and magnetic distortions, hence, increasing the need to capture the body posture each time. Although the option of using a single capture of the body posture within-session to do the correction of each recording of motion analysis could be explored, it is expected that when the sensors are used in magnetically save environments the sensors will not need to be recalibrate so often.

6.3 Clinical relevance

Only one postural deviation was simulated in this study to add deviations on the sagittal plane. Other studies have evaluated the gait of able-bodied adults simulating crouch gait to learn about the difference between voluntary crouch gait and pathological crouch gait [110] and looked only
at the sagittal kinematics of hip, knee and ankle. In one study, the researchers simulated crouch gait just by training the participant, as was done in the experiments presented in this thesis. Results presented in the cited study are similar in range and shape to the kinematics presented in this work. In addition to that, our results are similar when comparing the kinematics of the other planes to the work done in [111] where children (mean age of 10.6 years old) simulated crouch gait (with a belt that had strings tied to straps on their shanks to constrain their motion; It has been shown that kids over 7 years old do not have significantly different hip, knee or ankle kinematics to the adult population [112], therefore it is valid to make this comparison. This suggests that the approach followed in this thesis to simulate crouch gait would have the same effect if participants had a physical apparatus to constrain their motion.

Although it is true that more complex postural deviations can be found in clinical practice, the findings of this study based on the examination of crouch gait can still be of clinical relevance.

When different surgical methods were evaluated [84] to treat spastic diplegia (a musculoskeletal disorder that causes crouch gait) of 24 patients, an improvement in 20 degrees was found in knee flexion during the gait cycle. Other studies consider crouch gait improvement if a reduction in knee FE of 10 degrees is reached [86]. In this thesis, knee FE error of up to 50 degrees was corrected with both techniques (OC and PAC) to a level where the MVN kinematics could be used to reliably detect this range of flexion improvement of 10 to 20 degrees.

In the results shown here it was also found that the FE values can be represented as knee extensions when they are in fact flexions. In clinical practice, crouch gait is characterized mainly by excessive knee flexion during stance phase, but without the proper correction applied to the MVN data, the knee flexion data would not indicate excessive knee flexion [84], [86]. Therefore, seeing the measurements from MVN presented here without proper adjustment to reflect the true kinematics, could lead to the wrong clinical treatment.

Since crouch gait is a common pathological gait pattern not only in patients with cerebral palsy but also with other impairments [89], literature suggest that the MVN could be a valuable evaluation tool if it is able to provide accurate results. In this thesis it was shown that such improvements are possible.
In addition to this, since it was shown that the correction approaches could correct for postural deviation on the sagittal plane, the study of other conditions can benefit from such correction. In the population with cerebral palsy alone, deviations in ankle like equines foot are also common [11], [85]. Furthermore, this work was implemented on lower limbs, however, analyzing the postural deviation on MVN measuring the upper body could potentially add relevance to the study of conditions affecting spine and pelvic alignment such as spine deformities [113]–[116].

6.4 Future work

This thesis had as a goal to establish a basis for the further development of a correction technique. In the following paragraphs some recommendations and steps that need to be taken towards that goal are presented.

This study assessed the effect of postural deviation on gait kinematics in an experimental way. Analyzing the problem from an analytical point of view as done in [81], would help to gain a more thorough understanding of how the wrong measurement of body segment orientations affect the measurement of kinematics. Such study could result in the proposal of a more complete correction technique. Once the error is understood, the correction approaches proposed here should be evaluated. It was demonstrated in [81] that perturbing the alignment of alignment of the axes of the anatomical frame produces crosstalk errors. In future studies, it could be analyzed whether the postural deviation produce crosstalk errors. Then, given the nature of the OC approach it should be assessed analytically to see if this technique is less prone to introduce this type of error, or alternatively if PAC might be sufficient to correct for them.

Regardless of the details on how the correction technique is applied, the most important information to correct for kinematics is the true body posture. Better techniques to measure body posture in a single capture should be developed. The planar images technique using a standard camera was proposed since it is a tool that has been commonly used in literature to analyze posture [23]–[27] and a number of commercially available systems exist [23]–[25]. Although, in this study only the sagittal was analyzed, other studies have also looked at the frontal plane [23], [24], [27], [117]. The transverse plane is more difficult to analyze, however one study looked at the rotation in three-dimensional space of the rib cage measured from planar images [27],
suggesting that it is possible with proper image processing. For that, it might be needed to acquire images from the frontal as well as the sagittal plane.

Another alternative is the use of the depth sensor Kinect. Due to the capability of Kinect to track objects in three dimensional space, it could be potentially used to capture the true posture of the person [31], [118], [119]. However, to date the performance of Kinect in finding joint centers and measuring joint angles is not suitable to measure the true body posture as accurately as needed for the correction technique [33]–[35]. Nevertheless, when the Kinect is used to track markers rather than unmarked body segments, the accuracy has been shown to improve [120], [121]. Although the application of markers would add extra time, they would help to capture the true posture based on the position of the markers (as done in the gold standard) tracked with the Kinect, as supposed to rely on biomechanical models [122] that can be violated. Sometimes the addition of markers is a good alternative where the extra few minutes to add them is a worth trade-off for accuracy. Even the commercial systems mentioned above are based on markers [24], [25]. Adding markers to Kinect would allow a more robust control over the algorithm to measure not only joint angles but also body segments orientation, which would provide the ability to improve the OC correction approach.

The joint motion is described not only by the three-dimensional angles but also three-dimensional displacement [123] of the segment. MVN not only outputs joint angle but also joint position which means that postural deviation could also affect this measurement. This study considered postural deviation to affect only the joint angles, hence, future work should focus on assessing joint displacement as a deviation from the N-Pose.

A necessary step in the development of a correction technique is to test its validity and repeatability. First, other postural deviations should be analyzed to see if a) they can be detected over and above the errors that were found between MVN and the gold standard in frontal and transverse planes and b) to see if improvement in gait kinematics is consistently obtained. Second, a larger sample size should be considered to evaluate repeatability. Finally, the technique should be tried on real patients to validate it.

The correction approaches proposed and analyzed here have been developed specifically for the MVN BIOMECH Awinda system. Although there are some specific parameters that are needed
to implement the OC (e.g. change of basis matrix between the different reference frames, equations to calculate joint angles), if a different IMMUs-based system also uses the same principles to measure kinematics (i.e. calculates the orientation of body segments as a first step to measure kinematics and if it assumes a certain alignment among the body segments and these assumptions are known), the overall idea of the OC can be adjusted (i.e. by finding a specific change of basis matrix and equations to calculate angles) and implemented according to the specific characteristic of the other system. In contrast, given that the PAC is easier to implement, it will likely be easier to implement with any other IMMUs-system. However, the applicability of the correction approaches presented here to other IMMUs-based systems needs further investigation.
7. Conclusion

This thesis aimed to understand the effect that postural deviations have on gait kinematics measured with the MVN BIOMECH Awinda system when it is calibrated with postures that deviate from the N-Pose. Additionally, it was also envisioned to assess two correction approaches (OC and PAC) in order to establish a basis for the further development of a correction technique to account for the true body posture dismissed during the calibration.

The results of this thesis demonstrated how this error can be represented in a simple way and thus, can be corrected. The conclusions are summarized below.

- The error introduced at the moment of calibration due to postural deviations can be considered as a shift in joint angle values while the shape is not affected. The CMC values as well as the system error (DIFF\(_D\)) of CG conditions were similar in shape and range of variation to the system error (the difference in kinematics measured with PiG and MVN that includes instrumentation, crosstalk and anatomical frame definition errors) of the IC condition (the baseline error), in contrast there was a shift in the mean of this system error whose magnitude is similar to the deviation that the MVN failed to measure.

- It was found that this error can be corrected with the information of true body posture captured during the calibration procedure, since the change of system error frame to frame can be considered constant (the mean of the derivative of DIFF\(_D\) is about zero and the standard deviation is small).

- Therefore, if the information of the true body posture (measured as the true orientation of body segments or joint angles) is known, the postural deviation in the primarily affected planes of motion (hip, knee and ankle FE) can be corrected with either approach.

- These correction approaches not only correct for major postural deviations but also small deviations not measured even during the IC condition. It was shown how challenging it is to ensure that the sensors are actually aligned to the body segments even during the IC condition when the participant assumes the N-Pose. This was seen in the ankle IE angles and hip FE and IE angles.
• When comparing OC and PAC, results suggested that PAC outperforms the OC approach, since in general, the range of the mean $\text{DIFF}_D$ after the PAC correction is smaller than after the OC correction. Additionally, since PAC does not require the change of segments orientation it does not change the shape of the waveforms (as was seen with CMC that decreased after the OC was applied).

• Finally, the results suggest that PAC is enough to correct for the postural deviations with the removal of the known offsets (without having to change the orientation of the body segments), hence, the use of images to measure the planar angles needed for the implementation of this correction approach, appears supported. However, it is suggested that more robust methods are explored to capture this true posture, since the one used here would not perform as well in the presence of more complex postural deviations.

In the discussion, several suggestions are given to expand the understanding of the postural deviations and find more robust methods to capture the true body posture, a key step in this endeavor, since the technique relies on this information.

Overall, it was shown that the correction approaches represent a potential alternative solution to use the MVN BIOMECH Awinda with populations that cannot attain the N-Pose. This will ultimately make the study of human motion in real-life environments, the goal of IMMUs-based systems, be possible even with different patient populations.
References


Appendices

Appendix A. Comparison between Plug-in Gait (PiG) and MVN BIOMECH coordinate systems

A.1 PiG Marker placements

<table>
<thead>
<tr>
<th>Marker</th>
<th>Placement</th>
<th>Note</th>
</tr>
</thead>
<tbody>
<tr>
<td>RASI/LASI</td>
<td>Both Anterior Superior Iliac Spine</td>
<td>The height is critical. They have to be at the same height of the ASIS. Medio-Lateral placement can vary.</td>
</tr>
<tr>
<td>RPSI/LPSI</td>
<td>Both Posterior Superior Iliac Spine</td>
<td>The height is critical. They have to be at the same height of the ASIS. Medio-Lateral placement can vary.</td>
</tr>
<tr>
<td>RTHI/LTHI</td>
<td>Mid-Thigh, on a short stick on the lateral 1/3 surface of the thigh, just below the swing of the hand</td>
<td>Antero-posterior placement of the marker is critical. The marker must be placed so that it is aligned in the plane that contains the hip, knee joint centers and the knee flexion/extension axis. Height is not important.</td>
</tr>
<tr>
<td>RKNE/LKNE</td>
<td>Most lateral aspect of the femur lateral condyles</td>
<td></td>
</tr>
<tr>
<td>RTIB/LTIB</td>
<td>Mid-Thigh, on a short stick on the lateral 1/3 surface of the shank</td>
<td>Antero-posterior placement of the marker is critical. The marker must be placed so that it is aligned in the plane that contains the knee and ankle joint centers and the ankle flexion/extension axis. Height is not important.</td>
</tr>
<tr>
<td>RANK/LANK</td>
<td>On the most lateral aspect of the lateral malleolus</td>
<td></td>
</tr>
<tr>
<td>RHEE/LHEE</td>
<td>On the Calcaneus</td>
<td>Height is critical. Same height as RTOE</td>
</tr>
<tr>
<td>RTOE/LTOE</td>
<td>Second metatarsal head</td>
<td></td>
</tr>
</tbody>
</table>

Table A. 1: PiG marker placement.

A.2 PiG and MVN Coordinate System Definitions

Vicon Plug-in Gait is the commercial version of the Convention Gait Model developed at the Helen Hayes Hospital by Kadaba et al. and Davis et al. [7], [95]. MVN BIOMECH has a biomechanical model described in [67] that is based on the recommendations by the International Society of Biomechanics [10], [63]–[65]. Below there is a table that compares the coordinate systems of PiG [93], [124] and MVN[77].
<table>
<thead>
<tr>
<th>Pelvis</th>
<th>Pelvis</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>MVN BIOMECH (joint coordinate system)</strong></td>
<td><strong>MVN BIOMECH (body segment coordinate system)</strong> *</td>
</tr>
<tr>
<td><strong>O</strong></td>
<td>Midpoint between right and left hip center of rotation</td>
</tr>
<tr>
<td><strong>X</strong></td>
<td>Perpendicular to Y and Z Pointing forward</td>
</tr>
<tr>
<td><strong>Y</strong></td>
<td>jhipOrigin to jL5S1 Pointing up</td>
</tr>
<tr>
<td><strong>Z</strong></td>
<td>jLeftHip to jRight Hip Pointing right</td>
</tr>
</tbody>
</table>

*This coordinate system is not explicitly described on the manual.
**First the coordinate system is defined with the origin at the Mid-Point between RASI and LASI, after the HJC is defined, the coordinate system is shifted to the hip with its origin at the HJC.

### Upper Leg

<table>
<thead>
<tr>
<th>Upper Leg</th>
<th>Thigh</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>MVN BIOMECH (joint coordinate system)</strong></td>
<td><strong>MVN BIOMECH (body segment coordinate system)</strong></td>
</tr>
<tr>
<td><strong>O</strong></td>
<td>jRightHip/jLeftHip</td>
</tr>
<tr>
<td><strong>X</strong></td>
<td>Perpendicular to Y and Z Pointing forward in sagittal plane</td>
</tr>
<tr>
<td><strong>Y</strong></td>
<td>Right: jRightKnee to jRightHip Left: jLeftKnee to jLeftHip</td>
</tr>
<tr>
<td><strong>Z</strong></td>
<td>Right: Medial to Lateral Left: Lateral to Medial Pointing right</td>
</tr>
</tbody>
</table>

### Lower Leg

<table>
<thead>
<tr>
<th>Lower Leg</th>
<th>Tibia</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>MVN BIOMECH (joint coordinate system)</strong></td>
<td><strong>MVN BIOMECH (body segment coordinate system)</strong></td>
</tr>
<tr>
<td><strong>O</strong></td>
<td>jRightKnee jLeftKnee IM</td>
</tr>
<tr>
<td><strong>X</strong></td>
<td>Perp. Y and Z</td>
</tr>
</tbody>
</table>

*Table continues*
<table>
<thead>
<tr>
<th>Orientation</th>
<th>Definition</th>
<th>Coordinate System</th>
<th>Plane Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Y</strong></td>
<td>Right: jRightAnkle to jRightKnee</td>
<td>Right-handed co-ordinates (West or Left when participant is facing the Magnetic North).</td>
<td>Cross product between Z and X unit vectors Pointing Left</td>
</tr>
<tr>
<td></td>
<td>Left: jLeftAnkle to jLeftKnee</td>
<td>Right-handed co-ordinates (West or Left when participant is facing the Magnetic North).</td>
<td>Cross product between Z and X unit vectors Not necessarily pointing Left</td>
</tr>
<tr>
<td><strong>Z</strong></td>
<td>Right: Medial to Lateral</td>
<td>Positive when pointing up.</td>
<td>ACJ → KJC</td>
</tr>
<tr>
<td></td>
<td>Left: Lateral to Medial</td>
<td>Pointing right</td>
<td>ACJ → KJC</td>
</tr>
</tbody>
</table>

*To obtain the untorsioned tibia Plug-in Gait rotates the Torsioned tibia around its Z axis by **Tibial Torsion** degrees (calculated from the subject measurement)

**Untorsioned Tibia X unit vector is the same as the X unit vector of the Thigh segment

**Table A. 2: Comparison of Coordinate System Definitions between PiG and MVN. The terminology used to describe each model is the terminology used according to that model.
Appendix B. Procedure to find MVN equations to calculate kinematics

In section 4.6.1.2 it was said that the orientation correction approach uses the “true” body segments orientation to correct the orientation measured with the MVN system and after that kinematics are re-calculated.

To recalculate the kinematics, first the equations that the MVN system uses had to be found. MVN system calculates kinematics based on the work done by Grood and Suntay [10]. Therefore the equations from this paper were modified (changed in the signs in AA equations, due to differences in the definition of the direction of the Y axis of the coordinate system) and applied to the MVN segments orientation (in their matrix format) to prove whether they would reproduce the actual MVN kinematics. These equations can be seen in Table D. 2. As can be seen in Figure B. 1, it was proved that the equations would output the same kinematic values as the MVN system. This figure shows actual MVN angles displayed in the MVN Studio Pro software and the calculated angles in a superimposed plot.

![Figure B. 1: Example of how the joint angles were reproduced (above) to match the kinematics of the MVN Studio Pro software (below). This was done using the equations from Grood and Suntay [10] and the MVN segments orientation. This graphs correspond to the right knee angles.](image-url)
Appendix C. PiG segments orientation

PiG does not directly outputs segments orientation, therefore a MATLAB script was created. This script takes the marker position data and the information of the PiG coordinate system definition (Table A. 2) and outputs the orientation of the body segments in rotation matrix format in the PiG global reference frame (PiG orientations).

In order to check whether the orientations as calculated according to this script were correct, a similar procedure to appendix 0 was followed. Based on the Kadaba et al. [7] the equations to reproduce PiG angles were found. Besides some changes in sing an additional modification was the removal of the AA angle in the denominator of the argument of the arcsine. These equations are shown in Table D.2. Then, the actual PiG angles and the angles calculated based on the PiG orientation were plotted. That can be seen in Figure C. 1, Figure C. 2 and Figure C. 3. The figures correspond to the left limb of participant-1.

Figure C. 1 Flexion – extension angles of left hip, knee and ankle of participant-1. The black solid lines are the original PiG angles, the dotted lines are the angles calculated based on the PiG orientations and the green line represents the difference between the two (DIFFD).
Figure C. 2 External – internal rotation angles of left hip, knee and ankle of participant-1. The black solid lines are the original PiG angles, the dotted lines are the angles calculated based on the PiG orientations and the green line represents the difference between the two (DIFF).</figure>

Figure C. 3 Abduction -adduction angles of left hip, knee and ankle of participant-1. The black solid lines are the original PiG angles, the dotted lines are the angles calculated based on the PiG orientations and the green line represents the difference between the two (DIFF).
Appendix D. Comparison of PiG and MVN kinematics

To prove whether PiG and MVN kinematics are equivalent, kinematics were calculated based on the PiG orientations with both MVN and PiG equations (Table D.2). It was hypothesized that if the equations calculate the joint angles in an equivalent way, then, when plotting the joint angles calculated with both groups of equations (PiG and MVN), they would overlap. Figure D. 1, show that the kinematics are the same. It was found that the AA angles that MVN calculates are the negative of the PiG hip and knee AA angles, however ankle AA are not. The plots correspond to the left hip of participant-1.

![Figure D. 1 Joint angles of left hip of participant-1. Black and blue lines represent the angles calculated using MVN and PiG angles respectively (Table D.2). The green line represents the difference between the two (DIFF\(_D\)).](image-url)

**Figure D. 1** Joint angles of left hip of participant-1. Black and blue lines represent the angles calculated using MVN and PiG angles respectively (Table D.2). The green line represents the difference between the two (DIFF\(_D\)).
The MVN and PiG equations are expressed in terms of the basis vectors of the coordinate system of the proximal and distal segments. The basis vectors of both systems (PiG and MVN) as well as the two reference papers (Grood and Suntay [10] and Kadaba [7] respectively) can be seen in Table D. 1. The summary of the equations to calculate kinematics according to both systems as well as the two reference papers can be found in Table D. 2.

<table>
<thead>
<tr>
<th></th>
<th>MVN BIOMECH basis vectors of the coordinate system*</th>
<th>MVN basis vector according to Grood and Suntay [10] terminology (Joint Rotation Convention)**</th>
<th>PiG basis vectors of the coordinate system</th>
<th>Equivalent of PiG basis vector according to Kadaba [7]</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Proximal segment</strong></td>
<td>X</td>
<td>X, I</td>
<td>X, I</td>
<td>X, I</td>
</tr>
<tr>
<td></td>
<td>Y (pointing left)</td>
<td>Y, J, e₁ (pointing right)</td>
<td>Y (pointing left)</td>
<td>Y, J (pointing left)</td>
</tr>
<tr>
<td></td>
<td>Z</td>
<td>Z, K</td>
<td>Z</td>
<td>Z, K</td>
</tr>
<tr>
<td><strong>Distal segment</strong></td>
<td>X</td>
<td>x, i</td>
<td>X</td>
<td>X₃, I₃</td>
</tr>
<tr>
<td></td>
<td>Y (pointing left)</td>
<td>y, j (pointing right)</td>
<td>Y (pointing left)</td>
<td>Y₃, J₃ (pointing left)</td>
</tr>
<tr>
<td></td>
<td>Z</td>
<td>z, k, e₃</td>
<td>Z</td>
<td>Z₃, K₃</td>
</tr>
</tbody>
</table>

*Recall from Table A. 1A.2. That MVN BIOMECH has the body reference frames for joints and the body segment reference frame. The latter is the one presented here.

**e₂ = e₁ X X₃**

Table D. 1: Basis vectors of MVN and PiG and their equivalent terminology used in the reference papers.
<table>
<thead>
<tr>
<th>Internal-External Rotation</th>
<th>Hip</th>
<th>Knee</th>
<th>Ankle</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Flexion-Extension</strong></td>
<td>Hip</td>
<td>Knee</td>
<td>Ankle</td>
</tr>
<tr>
<td>α = arcsin(−e₂ · K)</td>
<td>β = arccos(J · k)</td>
<td>β = arccos(J · k)</td>
<td>β = arccos(J · k)</td>
</tr>
<tr>
<td>Flexion when positive</td>
<td>Adduction right hip = β − (\frac{\pi}{2})</td>
<td>Adduction right knee = β − (\frac{\pi}{2})</td>
<td>Inversion right ankle = β − (\frac{\pi}{2})</td>
</tr>
<tr>
<td></td>
<td>Adduction left hip = (\frac{\pi}{2}) − β</td>
<td>Adduction left knee = (\frac{\pi}{2}) − β</td>
<td>Inversion left ankle = (\frac{\pi}{2}) − β</td>
</tr>
<tr>
<td></td>
<td>Abduction when positive</td>
<td>Abduction when positive</td>
<td>Abduction when positive</td>
</tr>
<tr>
<td></td>
<td>α = −arcsin(−e₂ · K)</td>
<td>α = arccos(J · k)</td>
<td>α = arccos(J · k)</td>
</tr>
<tr>
<td></td>
<td>Flexion when positive</td>
<td>Adduction when positive</td>
<td>Inversion when positive</td>
</tr>
<tr>
<td><strong>Ab-Adduction</strong></td>
<td>Hip</td>
<td>Knee</td>
<td>Ankle</td>
</tr>
<tr>
<td>β = arccos(J · k)</td>
<td>β = arccos(J · k)</td>
<td>β = arccos(J · k)</td>
<td>β = arccos(J · k)</td>
</tr>
<tr>
<td>Adduction right hip = β − (\frac{\pi}{2})</td>
<td>Adduction right knee = β − (\frac{\pi}{2})</td>
<td>Inversion right ankle = β − (\frac{\pi}{2})</td>
<td>Inversion left ankle = (\frac{\pi}{2}) − β</td>
</tr>
<tr>
<td>Adduction left hip = (\frac{\pi}{2}) − β</td>
<td>Adduction left knee = (\frac{\pi}{2}) − β</td>
<td>Abduction when positive</td>
<td>Abduction when positive</td>
</tr>
<tr>
<td>Abduction when positive</td>
<td>β = arcsin(−K₃ · θ)</td>
<td>β = arcsin(−K₃ · θ)</td>
<td>β = arcsin(−K₃ · θ)</td>
</tr>
<tr>
<td></td>
<td>Kneee varus when positive</td>
<td>Knee varus when positive</td>
<td>Knee varus when positive</td>
</tr>
<tr>
<td><strong>Internal-External Rotation</strong></td>
<td>Hip</td>
<td>Knee</td>
<td>Ankle</td>
</tr>
<tr>
<td>γ = arcsin(−e₂ · J)</td>
<td>γ = arccos(e₂ · J)</td>
<td>γ = arccos(e₂ · J)</td>
<td>γ = arcsin(−e₂ · J)</td>
</tr>
<tr>
<td>right</td>
<td>right</td>
<td>left</td>
<td>right</td>
</tr>
<tr>
<td>γ = arcsin(e₂ · J)</td>
<td>Internal rotation when positive</td>
<td>Internal rotation when positive</td>
<td>Internal rotation when positive</td>
</tr>
<tr>
<td>left</td>
<td>γ = arcsin(e₂ · J)</td>
<td>left</td>
<td>γ = arcsin(−e₂ · J)</td>
</tr>
<tr>
<td></td>
<td>Internal rotation when positive</td>
<td>Internal rotation when positive</td>
<td>Internal rotation when positive</td>
</tr>
<tr>
<td><strong>Plug-in Gait modification of [7]</strong></td>
<td>Hip</td>
<td>Knee</td>
<td>Ankle</td>
</tr>
<tr>
<td>θ₁ = arcsin[(K₂ · (\frac{1}{cos(θ_j))</td>
<td>θ₁ = arcsin[(K₂ · (\frac{1}{cos(θ_j))</td>
<td>θ₁ = arcsin[(K₂ · (\frac{1}{cos(θ_j))</td>
<td></td>
</tr>
<tr>
<td>Flexion when positive</td>
<td>Flexion when positive</td>
<td>Flexion when positive</td>
<td>Flexion when positive</td>
</tr>
<tr>
<td>θ₁ = arcsin[(K₃ · (\frac{1}{cos(θ_j))</td>
<td>θ₁ = arcsin[(K₃ · (\frac{1}{cos(θ_j))</td>
<td>θ₁ = arcsin[(K₃ · (\frac{1}{cos(θ_j))</td>
<td></td>
</tr>
<tr>
<td>Flexion when positive</td>
<td>Flexion when positive</td>
<td>Flexion when positive</td>
<td>Flexion when positive</td>
</tr>
<tr>
<td>θ₂ = arcsin(−K₃ · J)</td>
<td>θ₂ = arcsin(−K₃ · J)</td>
<td>θ₂ = arcsin(−K₃ · J)</td>
<td>Equations to calculate ankle AA was not found, hence, not sure what equation PiG uses to calculate ankle AA</td>
</tr>
<tr>
<td>Right</td>
<td>Left</td>
<td>Left</td>
<td>Not included</td>
</tr>
<tr>
<td>θ₂ = arcsin(−K₃ · J)</td>
<td>Left</td>
<td>Left</td>
<td>Not included</td>
</tr>
<tr>
<td>Left</td>
<td>Left</td>
<td>Left</td>
<td>Not included</td>
</tr>
</tbody>
</table>

*The negative sign modifying β − \(\frac{\pi}{2}\) and \(\frac{\pi}{2}\) − β is there because in MVN coordinate system J points left, whereas in [10] points right. It is the same for hip and ankle flexion.*

Note: Angles would be expressed in radians.

Table D. 2: Comparison of the equations to calculate kinematics as presented in both reference papers, and both systems. The equations from column 1 and 4 (MVN BIOMECH and Plug-in Gait) are written so that when used provide the same kinematic values (i.e. by default the MVN system calculates hip and knee AA angles that are the negative of the AA PiG angles).
Appendix E. Code use to calculate planar joint angles form digital images

```matlab
function [hip, knee, ankle] = measureAnglesFromPhoto(plane)
% Call the function:
% [hip, knee, ankle] = measureAnglesFromPhoto(plane)
% where plane refers to the plane where angle are going to be measured: Sagittal

% 1. Load photo
% The software will prompt you to select the photo to analyze
% 2. Get marker position
% Get markers in the following order when side=sagittal
% 1. ASI
% 2. PSI
% 3. HIP
% 4. KNE
% 5. ANK
% 6. HEE
% 7. TOE
% To select the maker click on it:
% -If you are happy with the selection hit enter
% -If you want to reselect the marker just left click with the mouse and try to select again
% Each of the following variables contain the 2D coordinates of the marker in the photos

% 3. Define segments
% The function defines the coordinate system of the segment in the sagittal plane according to the PiG model
% z = vertical axis
% x = antero-posterior axis
% Then defines the segments

% 4. Draw segments
% Draws the segments on the image

% 5. Measure angles
% Measure the angle between the two vectors used to calculate the clinical FE angles

function [hip, knee, ankle] = measureAnglesFromPhoto(plane)
% 1. Load photo
[fileName, pathName] = uigetfile('.jpg');
cd(pathName)
image = imread(strcat(pathName,fileName));
imshow(image)
hip=0;
knee=0;	ankle=0;

% 2. Get marker position
switch plane
    case 'Sagittal'
        ASI=getMarker(1);
```
PSI=getMarker(1);
HIP=getMarker(1);
KNE=getMarker(1);
ANK=getMarker(1);
HEE=getMarker(1);
TOE=getMarker(1);

%% 3. Define segments
%X axis of pelvis reference frame
pelvis_x=ASI-PSI;
%Slope of the pelvis_x axis
m=(ASI(2)-PSI(2))/(ASI(1)-PSI(1));
minv=-1/m;
%Find the line equation y-y1=m*(x-x1)
%endPoint(2)-PSI(1)=minv*(endPoint(1)-PSI(2));
%to find the x value for endpoint
d endPoint(1)=PSI(1)+(abs(PSI(1)-ASI(1))/5);
d endPoint(2)=minv*(endPoint(1)-PSI(1))+PSI(2);

%Z axis of pelvis reference frame
pelvis_z=(endPoint-PSI);

%Z axis of femur reference frame
femur=HIP-KNE;

%Z axis of femur reference frame
shank_z=KNE-ANK;
mshank=(KNE(2)-ANK(2))/(KNE(1)-ANK(1));
mshankinv=-1/mshank;
endPointShank(2)=ANK(2)+(abs(KNE(2)-ANK(2))/5);
endPointShank(1)=(endPointShank(2)-KNE(2)+mshankinv*KNE(1))/mshankinv;

%X axis of femur reference frame
shank_x=(endPointShank-ANK);

%X axis of foot reference frame
foot=TOE-HEE;

%% Draw segments
%Pelvis
line([PSI(1) ASI(1)],[PSI(2) ASI(2)],'LineWidth',2)
line([PSI(1) endPoint(1)],[PSI(2) endPoint(2)],'LineWidth',2)

%Femur
line([HIP(1) KNE(1)],[HIP(2) KNE(2)],'LineWidth',2)

%Shank
line([KNE(1) ANK(1)],[KNE(2) ANK(2)],'LineWidth',2)

%Foot
line([HEE(1) TOE(1)],[HEE(2) TOE(2)],'LineWidth',2)

%% Measure angles
hip=calcAng(pelvis_x,femur)-90;
```matlab
knee = calcAng(femur, shank_z);
ankle = 90 - calcAng(shank_z, foot);

end
end
% Sub functions
function angle = calcAng(v1, v2)
    angle = acos(dot(v1, v2) / (norm(v1) * norm(v2))) * 180 / pi;
end

% Accept of reject selected marker
function marker = getMarker(varargin)
    switch nargin
        case 0
            h = gca;
        case 1
            h = gca;
            measure = varargin{1};
            axes(h);
        otherwise
            disp('Too many input arguments.');
    end
    cudata = get(gcf, 'UserData'); % current UserData
    hold on;
    while measure == 1
        k = waitforbuttonpress;
        marker = get(h, 'Currentpoint'); % get point
        marker = marker(1, 1:2); % extract x and y
        lh = plot(marker(1), marker(2), '+:');
        key = waitforbuttonpress;
        if key == 1
            set(gcf, 'UserData', cudata, 'WindowButtonMotionFcn', '', 'DoubleBuffer', 'off');
            % reset UserData, etc..
            measure = 0;
            else
            delete(lh);
            measure = 1;
    end
end
end
```
Appendix F. Effect of environment on kinematics measured with the MVN system

The following results are obtained from recording kinematics of one subject inside the gait laboratory and a second time outside the gait laboratory, in a magnetically safe environment. In both cases the system was calibrated with the N-Pose and the subject walked normally.

Figure F. 1 shows the kinematics recorded inside and outside as well as the difference between them for all the planes of all the joints of the left limb of the subject. The orange line represents kinematics recorded outside the laboratory and the brown lines inside. The green line represents the difference between them at each point in time. The thick lines represent the mean of the waveforms and the thin lines represent the minimum and maximum values of the measured kinematics across the six recorded gait cycles.

Figure F. 1 MVN kinematics of the left limb of the subject recorded inside (orange) and outside (Brown) the gait lab. The difference between the two recordings are shown with the green lines.

Figure F. 2 shows only the difference in kinematic estimation between the inside and outside environments. The numbers represent the range of error; that is the maximum difference minus the minimum difference along the gait cycle. Notice how they are constant for most part of the gait cycle and the high range of error is due to troughs present in small portions of the gait cycle where they are different.
Figure F. 2 Difference in kinematic estimation between the inside and outside environments. The numbers represent the range of error along the gait cycle.

The coefficient of multiple correlation within-day (CMC-WD) [103] was used to compare the similarity of the waveforms recorded in both environments.

CMC-WD values in Table F. 1 show good to excellent agreement (0.80 to 0.99). It cannot be known for sure what percentage of the error is due to subject variability or environment error. Given that this metric measures the similarity of waveforms of the same subject recorded in two different sessions, and all the CMC-WD values are good (considering that one environment is affected by magnetic distortions), it could be said that the shape of the kinematics measured inside the laboratory is not affected by the environment more than what it would be affected by the within-day variability of the subject. However, looking at the graphs that were affected by cross-talk shown in [81], it could be possible that the magnetic distortions affected the waveforms in a similar way to anatomical frame perturbations, since the definition of anteroposterior axis of the body segments frame rely on the magnetic field measurement. This is more evident on AA and IE angles. Hence it is safe to use the FE angles in the gait laboratory, while the possibility of cross-talk errors in IE and AA angles should be considered.

<table>
<thead>
<tr>
<th></th>
<th>Hip AA</th>
<th>Hip IE</th>
<th>Hip FE</th>
<th>Knee AA</th>
<th>Knee IE</th>
<th>Knee FE</th>
<th>Ankle AA</th>
<th>Ankle IE</th>
<th>Ankle FE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right</td>
<td>0.95</td>
<td>0.87</td>
<td>0.97</td>
<td>0.95</td>
<td>0.80</td>
<td>0.98</td>
<td>0.89</td>
<td>0.80</td>
<td>0.96</td>
</tr>
<tr>
<td>Left</td>
<td>0.99</td>
<td>0.93</td>
<td>0.99</td>
<td>0.98</td>
<td>0.96</td>
<td>0.99</td>
<td>0.93</td>
<td>0.89</td>
<td>0.99</td>
</tr>
</tbody>
</table>

Table F. 1: CMC-WD values
Appendix G. Gait Laboratory Mapping

To find the area within the gait laboratory that would have less magnetic distortions to perform the calibrations, a 2.35m wooden stick was instrumented according to the protocol found in [82] was followed. Figure G. 1 shows how the stick was instrumented. Instead of using five sensors as the protocol describes, 8 sensors were used and placed on a wooden stick at distances from the floor that would represent the height of the different sensors of the full body configuration of MVN, to see how they would be affected when worn by the participant. The distance of the sensors as well as the markers used can be seen in Table G.1.

![Instrumented wooden stick used to map the gait lab.](image)

The magnetic field of the sensors at height 5, 60 and 100 cm are shown in figures C.2 to C.4. During the experiments the sensors were calibrated while the person was standing of a 50cm high wooden table, hence the magnetic field that influenced the sensors during the calibration is shown in Figure G. 3.
<table>
<thead>
<tr>
<th>Sensor height</th>
<th>Marker height</th>
</tr>
</thead>
<tbody>
<tr>
<td>5cm – foot height</td>
<td>2 markers at 60cm</td>
</tr>
<tr>
<td>20cm –</td>
<td>1 marker at 100cm</td>
</tr>
<tr>
<td>40cm – shank height</td>
<td>2 markers at 120cm</td>
</tr>
<tr>
<td>60cm – thigh height</td>
<td></td>
</tr>
<tr>
<td>100cm – sacrum height</td>
<td></td>
</tr>
<tr>
<td>120cm – upper arm height</td>
<td></td>
</tr>
<tr>
<td>140cm – chest height</td>
<td></td>
</tr>
<tr>
<td>180cm – head height</td>
<td></td>
</tr>
</tbody>
</table>

Table G. 1: Sensors and markers height.

Figure G. 2: Magnetic field at 5cm height. The red rectangle represents the path were the participants walked during the experiments. Hence the feet sensor were affected by this magnetic field during the walking trials.
Figure G. 3: Magnetic field at 60cm high. The red rectangle represents the area of the "calibration spot". Since the participant were standing on the wooden table 50cm height, the magnetic field within the area represented by the red rectangle was the magnetic field affecting the feet sensors during the calibration.
Figure G. 4 Magnetic field at 180cm.
Appendix H. Posture reproducibility

Figure H. 1 PiG joint angles of the left limb of all participants during the three static trials of the Crouch Gait condition.

<table>
<thead>
<tr>
<th>Participant</th>
<th>1 min</th>
<th>1 max</th>
<th>2 min</th>
<th>2 max</th>
<th>3 min</th>
<th>3 max</th>
<th>4 min</th>
<th>4 max</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Right</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>IE</td>
<td>0.352</td>
<td>1.931</td>
<td>-7.68</td>
<td>-6.72</td>
<td>-5.08</td>
<td>-2.97</td>
<td>3.566</td>
<td>3.886</td>
</tr>
<tr>
<td>FE</td>
<td>52.2</td>
<td>61.24</td>
<td>34.97</td>
<td>37.02</td>
<td>32.71</td>
<td>34.67</td>
<td>24.3</td>
<td>30.17</td>
</tr>
<tr>
<td>Knee AA</td>
<td>6.164</td>
<td>8.522</td>
<td>2.74</td>
<td>3.653</td>
<td>-2.52</td>
<td>-0.55</td>
<td>11.25</td>
<td>12.19</td>
</tr>
<tr>
<td>FE</td>
<td>47.57</td>
<td>56.98</td>
<td>38.51</td>
<td>41.84</td>
<td>34.28</td>
<td>35.33</td>
<td>45.13</td>
<td>49.91</td>
</tr>
<tr>
<td>Ankle AA</td>
<td>1.988</td>
<td>2.425</td>
<td>2.37</td>
<td>3.11</td>
<td>6.09</td>
<td>6.767</td>
<td>4.244</td>
<td>4.949</td>
</tr>
<tr>
<td>IE</td>
<td>-11.9</td>
<td>-9.65</td>
<td>-13.3</td>
<td>-9.78</td>
<td>-35.4</td>
<td>-32.7</td>
<td>-25.8</td>
<td>-22.3</td>
</tr>
<tr>
<td>FE</td>
<td>21.57</td>
<td>24.31</td>
<td>18.59</td>
<td>20.47</td>
<td>22.67</td>
<td>24.02</td>
<td>26.61</td>
<td>31.8</td>
</tr>
<tr>
<td><strong>Left</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip AA</td>
<td>2.307</td>
<td>3.357</td>
<td>-1.31</td>
<td>-0.41</td>
<td>4.599</td>
<td>5.214</td>
<td>-4.98</td>
<td>-3.3</td>
</tr>
<tr>
<td>IE</td>
<td>-5.47</td>
<td>-5.21</td>
<td>-3.69</td>
<td>-2.99</td>
<td>8.424</td>
<td>9.758</td>
<td>0.268</td>
<td>3.084</td>
</tr>
<tr>
<td>FE</td>
<td>51.16</td>
<td>60.4</td>
<td>34.24</td>
<td>35.74</td>
<td>32.76</td>
<td>35.57</td>
<td>24.99</td>
<td>30.61</td>
</tr>
<tr>
<td>Knee AA</td>
<td>-3.01</td>
<td>-1.98</td>
<td>5.207</td>
<td>5.354</td>
<td>-0.93</td>
<td>0.835</td>
<td>6.123</td>
<td>8.108</td>
</tr>
<tr>
<td>IE</td>
<td>2.711</td>
<td>5.32</td>
<td>13.11</td>
<td>14.57</td>
<td>-3.66</td>
<td>-0.93</td>
<td>20.19</td>
<td>22.24</td>
</tr>
<tr>
<td>FE</td>
<td>46.2</td>
<td>54.55</td>
<td>34.88</td>
<td>39.75</td>
<td>33.5</td>
<td>35.46</td>
<td>45.14</td>
<td>50.19</td>
</tr>
<tr>
<td>IE</td>
<td>-8.79</td>
<td>-5.55</td>
<td>-25.3</td>
<td>-22.1</td>
<td>-26.7</td>
<td>-25.5</td>
<td>-25.4</td>
<td>-23.2</td>
</tr>
<tr>
<td>FE</td>
<td>23.34</td>
<td>24.65</td>
<td>19.68</td>
<td>22.89</td>
<td>20.88</td>
<td>21.96</td>
<td>27.2</td>
<td>33.53</td>
</tr>
</tbody>
</table>

Table H. 1 Range of PiG joint angles of the left limb of all participants during the three static trials
Appendix I. Difference between PiG and MVN during the static trial (\texttt{DIFF}_3) corresponding to the Ideal Calibration (IC) and Crouch Gait (CG) condition.

<table>
<thead>
<tr>
<th></th>
<th>\texttt{PIG}</th>
<th>\texttt{MVN}</th>
<th>\texttt{DIFF}</th>
<th>\texttt{PIG}</th>
<th>\texttt{MVN}</th>
<th>\texttt{DIFF}</th>
<th>\texttt{PIG}</th>
<th>\texttt{MVN}</th>
<th>\texttt{DIFF}</th>
<th>\texttt{PIG}</th>
<th>\texttt{MVN}</th>
<th>\texttt{DIFF}</th>
</tr>
</thead>
<tbody>
<tr>
<td>\texttt{Right}</td>
<td>\texttt{Hip} AA</td>
<td>-5.21</td>
<td>-0.07</td>
<td>-5.14</td>
<td>-7.30</td>
<td>0.35</td>
<td>-7.65</td>
<td>0.77</td>
<td>0.15</td>
<td>0.62</td>
<td>-4.63</td>
<td>0.23</td>
</tr>
<tr>
<td></td>
<td>\texttt{IE}</td>
<td>0.08</td>
<td>0.88</td>
<td>-0.80</td>
<td>-7.20</td>
<td>4.24</td>
<td>-11.44</td>
<td>-4.97</td>
<td>2.27</td>
<td>-7.24</td>
<td>-0.94</td>
<td>1.94</td>
</tr>
<tr>
<td>\texttt{Ankle} AA</td>
<td>2.93</td>
<td>-0.11</td>
<td>3.04</td>
<td>0.86</td>
<td>-0.15</td>
<td>1.01</td>
<td>-2.28</td>
<td>-0.32</td>
<td>-1.96</td>
<td>5.82</td>
<td>-0.23</td>
<td>6.05</td>
</tr>
<tr>
<td>\texttt{IE}</td>
<td>-2.10</td>
<td>-1.58</td>
<td>-0.51</td>
<td>0.78</td>
<td>-1.31</td>
<td>2.10</td>
<td>8.84</td>
<td>-1.42</td>
<td>10.26</td>
<td>3.97</td>
<td>-1.45</td>
<td>5.42</td>
</tr>
<tr>
<td>\texttt{FE}</td>
<td>2.43</td>
<td>1.90</td>
<td>0.53</td>
<td>6.31</td>
<td>2.54</td>
<td>3.77</td>
<td>-2.83</td>
<td>1.79</td>
<td>-4.62</td>
<td>7.47</td>
<td>2.60</td>
<td>4.87</td>
</tr>
<tr>
<td>\texttt{Left}</td>
<td>\texttt{Hip} AA</td>
<td>1.57</td>
<td>0.12</td>
<td>1.45</td>
<td>4.45</td>
<td>-0.18</td>
<td>4.64</td>
<td>5.90</td>
<td>-0.03</td>
<td>5.93</td>
<td>-3.37</td>
<td>-0.08</td>
</tr>
<tr>
<td></td>
<td>\texttt{IE}</td>
<td>-7.52</td>
<td>-0.35</td>
<td>-7.17</td>
<td>1.79</td>
<td>-2.69</td>
<td>4.48</td>
<td>10.50</td>
<td>-0.69</td>
<td>11.19</td>
<td>-0.19</td>
<td>-0.30</td>
</tr>
<tr>
<td></td>
<td>\texttt{FE}</td>
<td>21.22</td>
<td>-8.40</td>
<td>29.62</td>
<td>8.76</td>
<td>-8.80</td>
<td>17.56</td>
<td>17.90</td>
<td>-8.11</td>
<td>26.01</td>
<td>15.34</td>
<td>-8.29</td>
</tr>
<tr>
<td>\texttt{Ankle} AA</td>
<td>0.09</td>
<td>0.03</td>
<td>0.06</td>
<td>1.02</td>
<td>0.12</td>
<td>0.89</td>
<td>-5.30</td>
<td>0.04</td>
<td>2.96</td>
<td>2.96</td>
<td>0.06</td>
<td>2.89</td>
</tr>
<tr>
<td>\texttt{IE}</td>
<td>7.19</td>
<td>-0.36</td>
<td>7.55</td>
<td>5.84</td>
<td>0.10</td>
<td>5.74</td>
<td>-6.28</td>
<td>0.29</td>
<td>8.26</td>
<td>8.26</td>
<td>0.17</td>
<td>8.09</td>
</tr>
<tr>
<td>\texttt{FE}</td>
<td>2.03</td>
<td>1.98</td>
<td>0.06</td>
<td>1.03</td>
<td>3.14</td>
<td>-2.12</td>
<td>3.24</td>
<td>2.88</td>
<td>2.26</td>
<td>2.26</td>
<td>1.96</td>
<td>0.31</td>
</tr>
<tr>
<td>\texttt{Ankle} AA</td>
<td>-0.01</td>
<td>-0.03</td>
<td>0.02</td>
<td>4.44</td>
<td>0.27</td>
<td>4.17</td>
<td>4.24</td>
<td>0.34</td>
<td>3.27</td>
<td>3.27</td>
<td>-0.16</td>
<td>3.43</td>
</tr>
<tr>
<td>\texttt{IE}</td>
<td>0.76</td>
<td>0.62</td>
<td>0.14</td>
<td>-20.58</td>
<td>-0.73</td>
<td>-19.85</td>
<td>-17.94</td>
<td>-0.56</td>
<td>-16.73</td>
<td>-16.73</td>
<td>-0.45</td>
<td>-16.28</td>
</tr>
<tr>
<td>\texttt{FE}</td>
<td>4.99</td>
<td>4.02</td>
<td>0.97</td>
<td>7.73</td>
<td>5.33</td>
<td>2.40</td>
<td>6.53</td>
<td>5.02</td>
<td>4.56</td>
<td>4.56</td>
<td>4.39</td>
<td>0.17</td>
</tr>
</tbody>
</table>

Table I. 1: Mean and standard deviation of joint angles measured with PiG, MVN as well as the difference between them during the static calibration posture of the IC condition. Results are shown in degrees.
<table>
<thead>
<tr>
<th></th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th></th>
<th>2</th>
<th>3</th>
<th>4</th>
<th></th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>mean</td>
<td>std</td>
<td>mean</td>
<td>std</td>
<td>mean</td>
<td>std</td>
<td>mean</td>
<td>std</td>
<td>mean</td>
<td>std</td>
<td>mean</td>
<td>std</td>
</tr>
<tr>
<td>Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>-8.76</td>
<td>0.28</td>
<td>-9.05</td>
<td>0.28</td>
<td>0.11</td>
<td>0.22</td>
<td>0.20</td>
<td>0.20</td>
<td>0.15</td>
<td>-9.06</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(1.46)</td>
<td>(0.11)</td>
<td>(1.54)</td>
<td>(0.28)</td>
<td>(0.74)</td>
<td>(0.50)</td>
<td>(0.83)</td>
<td>(0.83)</td>
<td>(0.29)</td>
<td>(0.13)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>IE</td>
<td>1.15</td>
<td>1.02</td>
<td>0.13</td>
<td>-7.30</td>
<td>1.41</td>
<td>-8.71</td>
<td>-3.68</td>
<td>0.52</td>
<td>-4.20</td>
<td>3.70</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.79)</td>
<td>(1.34)</td>
<td>(0.55)</td>
<td>(0.51)</td>
<td>(0.79)</td>
<td>(0.66)</td>
<td>(1.21)</td>
<td>(2.31)</td>
<td>(1.80)</td>
<td>(0.16)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>FE</td>
<td>57.93</td>
<td>-9.30</td>
<td>67.23</td>
<td>35.80</td>
<td>-7.75</td>
<td>43.55</td>
<td>33.37</td>
<td>-9.02</td>
<td>42.38</td>
<td>27.04</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(4.98)</td>
<td>(0.91)</td>
<td>(5.88)</td>
<td>(1.08)</td>
<td>(0.54)</td>
<td>(0.54)</td>
<td>(1.13)</td>
<td>(0.87)</td>
<td>(0.46)</td>
<td>(2.95)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee</td>
<td>7.32</td>
<td>-0.15</td>
<td>7.47</td>
<td>3.08</td>
<td>0.17</td>
<td>2.92</td>
<td>-1.78</td>
<td>-0.23</td>
<td>-1.55</td>
<td>11.67</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(1.18)</td>
<td>(0.70)</td>
<td>(1.58)</td>
<td>(0.50)</td>
<td>(0.46)</td>
<td>(0.77)</td>
<td>(1.07)</td>
<td>(0.05)</td>
<td>(1.11)</td>
<td>(0.48)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td>2.25</td>
<td>-0.96</td>
<td>3.21</td>
<td>2.77</td>
<td>-0.82</td>
<td>3.59</td>
<td>6.36</td>
<td>0.46</td>
<td>5.90</td>
<td>4.53</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.23)</td>
<td>(1.05)</td>
<td>(0.99)</td>
<td>(1.71)</td>
<td>(0.37)</td>
<td>(1.34)</td>
<td>(0.36)</td>
<td>(0.15)</td>
<td>(0.26)</td>
<td>(0.24)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Left</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>2.93</td>
<td>0.09</td>
<td>2.84</td>
<td>-0.99</td>
<td>0.06</td>
<td>-1.05</td>
<td>4.86</td>
<td>-0.06</td>
<td>4.92</td>
<td>-4.10</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.55)</td>
<td>(0.45)</td>
<td>(0.21)</td>
<td>(0.51)</td>
<td>(1.02)</td>
<td>(1.44)</td>
<td>(0.32)</td>
<td>(0.11)</td>
<td>(0.42)</td>
<td>(0.84)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>IE</td>
<td>-5.33</td>
<td>-0.60</td>
<td>-4.73</td>
<td>-3.36</td>
<td>0.31</td>
<td>-3.67</td>
<td>9.24</td>
<td>0.96</td>
<td>8.28</td>
<td>1.96</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.13)</td>
<td>(0.27)</td>
<td>(0.24)</td>
<td>(0.35)</td>
<td>(0.85)</td>
<td>(1.09)</td>
<td>(0.71)</td>
<td>(0.55)</td>
<td>(0.48)</td>
<td>(1.49)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>FE</td>
<td>57.42</td>
<td>-9.38</td>
<td>66.80</td>
<td>34.87</td>
<td>-7.38</td>
<td>42.25</td>
<td>33.76</td>
<td>-8.46</td>
<td>42.22</td>
<td>27.36</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(5.42)</td>
<td>(0.72)</td>
<td>(6.06)</td>
<td>(0.78)</td>
<td>(0.46)</td>
<td>(0.45)</td>
<td>(1.57)</td>
<td>(0.81)</td>
<td>(0.86)</td>
<td>(2.91)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee</td>
<td>-2.46</td>
<td>-0.23</td>
<td>-2.23</td>
<td>5.30</td>
<td>0.21</td>
<td>5.09</td>
<td>-0.10</td>
<td>0.22</td>
<td>-0.32</td>
<td>7.26</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.52)</td>
<td>(0.46)</td>
<td>(0.25)</td>
<td>(0.08)</td>
<td>(0.55)</td>
<td>(0.59)</td>
<td>(0.89)</td>
<td>(0.47)</td>
<td>(1.05)</td>
<td>(0.50)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>IE</td>
<td>4.44</td>
<td>-0.97</td>
<td>5.41</td>
<td>13.97</td>
<td>-0.98</td>
<td>14.96</td>
<td>-1.88</td>
<td>-0.27</td>
<td>-1.61</td>
<td>21.07</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(1.50)</td>
<td>(0.42)</td>
<td>(1.07)</td>
<td>(0.77)</td>
<td>(0.88)</td>
<td>(0.52)</td>
<td>(1.55)</td>
<td>(0.78)</td>
<td>(2.33)</td>
<td>(1.06)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>FE</td>
<td>51.69</td>
<td>0.81</td>
<td>50.88</td>
<td>37.41</td>
<td>2.10</td>
<td>35.31</td>
<td>34.68</td>
<td>1.59</td>
<td>33.09</td>
<td>46.98</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(4.75)</td>
<td>(0.25)</td>
<td>(4.84)</td>
<td>(2.44)</td>
<td>(0.89)</td>
<td>(1.84)</td>
<td>(1.04)</td>
<td>(0.65)</td>
<td>(1.22)</td>
<td>(2.78)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td>1.49</td>
<td>-0.23</td>
<td>1.72</td>
<td>5.35</td>
<td>0.13</td>
<td>5.22</td>
<td>6.53</td>
<td>-0.12</td>
<td>6.65</td>
<td>4.79</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.34)</td>
<td>(1.62)</td>
<td>(1.81)</td>
<td>(1.09)</td>
<td>(0.63)</td>
<td>(0.18)</td>
<td>(0.54)</td>
<td>(0.40)</td>
<td>(0.24)</td>
<td>(0.47)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>IE</td>
<td>-6.83</td>
<td>0.20</td>
<td>-7.02</td>
<td>-23.94</td>
<td>-0.42</td>
<td>-23.52</td>
<td>-26.04</td>
<td>0.07</td>
<td>-26.11</td>
<td>-24.51</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(1.73)</td>
<td>(1.43)</td>
<td>(2.28)</td>
<td>(1.67)</td>
<td>(0.82)</td>
<td>(1.77)</td>
<td>(0.58)</td>
<td>(0.51)</td>
<td>(0.78)</td>
<td>(1.16)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>FE</td>
<td>23.84</td>
<td>3.78</td>
<td>20.06</td>
<td>21.38</td>
<td>4.48</td>
<td>16.90</td>
<td>21.28</td>
<td>4.09</td>
<td>17.19</td>
<td>29.70</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(0.71)</td>
<td>(0.70)</td>
<td>(1.18)</td>
<td>(1.61)</td>
<td>(0.66)</td>
<td>(1.97)</td>
<td>(0.59)</td>
<td>(0.43)</td>
<td>(0.18)</td>
<td>(3.37)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table I. 2: Mean and standard deviation of joint angles measured with PiG, MVN as well as the difference between them during the static calibration posture of the CG condition. Results are shown in degrees.
Appendix J. Difference between PiG and MVN during the dynamic trial (DIFF) corresponding to the Ideal Calibration (IC) and Crouch Gait (CG) condition.

<table>
<thead>
<tr>
<th>Participant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>Baseline error</td>
<td>Crouch gait</td>
<td>Baseline error</td>
<td>Crouch gait</td>
</tr>
<tr>
<td>AA</td>
<td>$-5.67$ (3.14)</td>
<td>$-7.26$ (2.39)</td>
<td>$-5.86$ (2.63)</td>
<td>$-6.73$ (1.89)</td>
</tr>
<tr>
<td>IE</td>
<td>$-2.86$ (4.09)</td>
<td>$-5.21$ (2.08)</td>
<td>$-8.44$ (4.39)</td>
<td>$-12.18$ (3.57)</td>
</tr>
<tr>
<td>FE</td>
<td>$28.80$ (3.39)</td>
<td>$67.10$ (1.91)</td>
<td>$18.58$ (1.85)</td>
<td>$42.68$ (1.18)</td>
</tr>
<tr>
<td>Knee</td>
<td>Baseline error</td>
<td>Crouch gait</td>
<td>Baseline error</td>
<td>Crouch gait</td>
</tr>
<tr>
<td>AA</td>
<td>$2.35$ (2.72)</td>
<td>$8.13$ (3.89)</td>
<td>$-2.22$ (7.30)</td>
<td>$1.64$ (6.22)</td>
</tr>
<tr>
<td>IE</td>
<td>$0.99$ (2.01)</td>
<td>$6.15$ (2.41)</td>
<td>$8.32$ (1.75)</td>
<td>$9.45$ (2.33)</td>
</tr>
<tr>
<td>FE</td>
<td>$-2.22$ (3.53)</td>
<td>$53.54$ (2.98)</td>
<td>$1.91$ (2.68)</td>
<td>$38.14$ (2.27)</td>
</tr>
<tr>
<td>Ankle</td>
<td>Baseline error</td>
<td>Crouch gait</td>
<td>Baseline error</td>
<td>Crouch gait</td>
</tr>
<tr>
<td>AA</td>
<td>$-1.42$ (8.53)</td>
<td>$-0.52$ (9.82)</td>
<td>$-6.04$ (8.97)</td>
<td>$-1.90$ (9.78)</td>
</tr>
<tr>
<td>IE</td>
<td>$-7.20$ (4.38)</td>
<td>$-10.84$ (3.10)</td>
<td>$-5.79$ (4.96)</td>
<td>$-10.77$ (4.12)</td>
</tr>
<tr>
<td>FE</td>
<td>$1.49$ (2.45)</td>
<td>$23.17$ (4.23)</td>
<td>$2.96$ (1.93)</td>
<td>$17.02$ (3.45)</td>
</tr>
<tr>
<td>Left</td>
<td>Hip</td>
<td>Baseline error</td>
<td>Crouch gait</td>
<td>Baseline error</td>
</tr>
<tr>
<td>AA</td>
<td>$4.97$ (7.17)</td>
<td>$2.56$ (4.64)</td>
<td>$7.30$ (7.00)</td>
<td>$1.45$ (5.92)</td>
</tr>
<tr>
<td>IE</td>
<td>$3.18$ (4.06)</td>
<td>$-0.49$ (3.64)</td>
<td>$4.66$ (2.23)</td>
<td>$-2.50$ (3.20)</td>
</tr>
<tr>
<td>FE</td>
<td>$28.12$ (2.29)</td>
<td>$66.71$ (1.62)</td>
<td>$16.47$ (1.61)</td>
<td>$42.23$ (1.03)</td>
</tr>
<tr>
<td>Knee</td>
<td>Baseline error</td>
<td>Crouch gait</td>
<td>Baseline error</td>
<td>Crouch gait</td>
</tr>
<tr>
<td>AA</td>
<td>$0.99$ (3.75)</td>
<td>$-2.19$ (2.94)</td>
<td>$0.91$ (3.72)</td>
<td>$4.38$ (2.32)</td>
</tr>
<tr>
<td>IE</td>
<td>$2.19$ (2.01)</td>
<td>$2.19$ (1.32)</td>
<td>$2.61$ (2.40)</td>
<td>$14.71$ (1.86)</td>
</tr>
<tr>
<td>FE</td>
<td>$-4.30$ (3.35)</td>
<td>$51.98$ (2.91)</td>
<td>$-3.87$ (2.42)</td>
<td>$35.76$ (2.43)</td>
</tr>
<tr>
<td>Ankle</td>
<td>Baseline error</td>
<td>Crouch gait</td>
<td>Baseline error</td>
<td>Crouch gait</td>
</tr>
<tr>
<td>AA</td>
<td>$-2.07$ (6.85)</td>
<td>$-1.77$ (8.49)</td>
<td>$5.56$ (6.25)</td>
<td>$4.58$ (9.34)</td>
</tr>
<tr>
<td>IE</td>
<td>$-5.44$ (3.43)</td>
<td>$-8.46$ (3.17)</td>
<td>$-32.07$ (2.93)</td>
<td>$-26.16$ (4.00)</td>
</tr>
<tr>
<td>FE</td>
<td>$3.02$ (3.75)</td>
<td>$25.19$ (4.67)</td>
<td>$5.84$ (2.55)</td>
<td>$20.31$ (3.88)</td>
</tr>
</tbody>
</table>

Table J.1: Mean and standard deviation of the CG condition and the baseline error of all participants. Results are shown in degrees. Notice how the values for the IC condition are
smaller than the values of the CG condition, indicating that MVN fails to measure the deviation in the sagittal plane. Values in other planes are similar between IC and CG since the deviation was mainly in the sagittal plane. Ankle IE also show a difference between conditions probably because the participants adjusted ankle rotation to adapt the crouch position, as external rotation is higher than during the natural standing. These are shown in italics.

<table>
<thead>
<tr>
<th>Participant</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-16.02</td>
<td>15.59</td>
<td>-16.21</td>
<td>-0.60</td>
</tr>
<tr>
<td>(3.38)</td>
<td>(3.48)</td>
<td>(1.89)</td>
<td>(1.27)</td>
<td>(0.88)</td>
</tr>
<tr>
<td>IE</td>
<td>-6.71</td>
<td>-4.88</td>
<td>-4.07</td>
<td>-2.76</td>
</tr>
<tr>
<td>(1.91)</td>
<td>(2.07)</td>
<td>(1.61)</td>
<td>(1.87)</td>
<td>(2.67)</td>
</tr>
<tr>
<td>FE</td>
<td>7.63</td>
<td>6.87</td>
<td>8.36</td>
<td>9.09</td>
</tr>
<tr>
<td>(3.05)</td>
<td>(1.36)</td>
<td>(1.18)</td>
<td>(2.21)</td>
<td>(2.27)</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-13.20</td>
<td>4.72</td>
<td>-2.13</td>
<td>-3.65</td>
</tr>
<tr>
<td>(2.76)</td>
<td>(6.22)</td>
<td>(1.45)</td>
<td>(1.90)</td>
<td>(2.98)</td>
</tr>
<tr>
<td>IE</td>
<td>-6.09</td>
<td>2.06</td>
<td>1.95</td>
<td>1.95</td>
</tr>
<tr>
<td>(3.69)</td>
<td>(2.33)</td>
<td>(1.74)</td>
<td>(1.55)</td>
<td>(2.15)</td>
</tr>
<tr>
<td>FE</td>
<td>-4.41</td>
<td>-2.21</td>
<td>-1.55</td>
<td>-1.06</td>
</tr>
<tr>
<td>(3.08)</td>
<td>(2.37)</td>
<td>(2.27)</td>
<td>(3.37)</td>
<td>(3.41)</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>2.20</td>
<td>0.87</td>
<td>9.07</td>
<td>4.98</td>
</tr>
<tr>
<td>(9.46)</td>
<td>(9.78)</td>
<td>(11.70)</td>
<td>(10.02)</td>
<td>(7.07)</td>
</tr>
<tr>
<td>IE</td>
<td>-0.69</td>
<td>0.89</td>
<td>-3.41</td>
<td>0.56</td>
</tr>
<tr>
<td>(4.32)</td>
<td>(4.12)</td>
<td>(4.23)</td>
<td>(3.82)</td>
<td>(2.31)</td>
</tr>
<tr>
<td>FE</td>
<td>-5.42</td>
<td>-2.81</td>
<td>-0.16</td>
<td>-3.13</td>
</tr>
<tr>
<td>(4.31)</td>
<td>(3.41)</td>
<td>(3.45)</td>
<td>(4.19)</td>
<td>(2.98)</td>
</tr>
<tr>
<td>Left</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>26.30</td>
<td>0.46</td>
<td>11.46</td>
<td>10.95</td>
</tr>
<tr>
<td>(5.98)</td>
<td>(5.92)</td>
<td>(5.42)</td>
<td>(5.32)</td>
<td>(5.14)</td>
</tr>
<tr>
<td>IE</td>
<td>-0.97</td>
<td>0.86</td>
<td>2.43</td>
<td>3.65</td>
</tr>
<tr>
<td>(2.32)</td>
<td>(3.20)</td>
<td>(3.33)</td>
<td>(2.94)</td>
<td>(1.94)</td>
</tr>
<tr>
<td>FE</td>
<td>10.26</td>
<td>7.36</td>
<td>6.70</td>
<td>8.28</td>
</tr>
<tr>
<td>(5.06)</td>
<td>(1.03)</td>
<td>(2.78)</td>
<td>(2.29)</td>
<td>(1.26)</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>-25.48</td>
<td>9.68</td>
<td>-14.34</td>
<td>1.40</td>
</tr>
<tr>
<td>(6.63)</td>
<td>(2.32)</td>
<td>(4.70)</td>
<td>(5.26)</td>
<td>(1.58)</td>
</tr>
<tr>
<td>IE</td>
<td>-16.38</td>
<td>0.74</td>
<td>-7.86</td>
<td>-3.35</td>
</tr>
<tr>
<td>(4.16)</td>
<td>(1.86)</td>
<td>(2.12)</td>
<td>(0.94)</td>
<td>(1.57)</td>
</tr>
<tr>
<td>FE</td>
<td>-0.69</td>
<td>-1.65</td>
<td>-2.87</td>
<td>-2.98</td>
</tr>
<tr>
<td>(2.92)</td>
<td>(3.78)</td>
<td>(3.35)</td>
<td>(2.26)</td>
<td>(1.93)</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AA</td>
<td>11.98</td>
<td>9.93</td>
<td>5.20</td>
<td>4.17</td>
</tr>
<tr>
<td>(10.50)</td>
<td>(9.34)</td>
<td>(10.41)</td>
<td>(10.12)</td>
<td>(7.26)</td>
</tr>
<tr>
<td>IE</td>
<td>-3.83</td>
<td>-2.22</td>
<td>-7.07</td>
<td>0.32</td>
</tr>
<tr>
<td>(8.49)</td>
<td>(8.83)</td>
<td>(8.83)</td>
<td>(8.83)</td>
<td>(8.83)</td>
</tr>
</tbody>
</table>
Table J. 2: Mean and standard deviation of the CG condition after it was corrected with the OC and the PAC. Results are shown in degrees. The FE angles are shown in bold and italics, where the mayor improvement happened. Notice how they are close to zero. Underlined values only represent the other planes that also show improvement after the correction. The hip FE are not close to zero but it smaller than what it was before correction. In general ankle IE was corrected with both approaches. The values in italics of the knee AA show how an error was introduced.
Appendix K. Magnetic declination of the gait laboratory

Magnetic declination was calculated using the magnetic declination calculator from the National Centers for Environmental Information (NOAA). Please see Figure K.1. First, the latitude and longitude of the room where calculated using Google Maps and then introduced on the NOAA website. The calculated declination on August 21st 2015 is 10.51°W ± 0.38°.

Figure K. 1 Magnetic declination of the Gait laboratory.