

**Concurrent Evaluation of Marker-Based and Markerless Motion Capture for Assessing
Sex-Based Differences in Lower Limb Kinematics and Anthropometrics**

Maxine Speers

Thesis submitted to the University of Ottawa
in partial Fulfillment of the requirements for the

Master of Applied Science in Biomedical Engineering

Ottawa-Carleton Institute of Biomedical Engineering (OCIBME)
Faculty of Engineering
University of Ottawa

© **Maxine Speers, Ottawa, Canada, 2026**

Abstract

Markerless (ML) motion capture has become increasingly popular in biomechanics because it addresses many practical limitations of marker-based (MB) systems by being less time consuming and more user friendly, and it enables movement data to be collected outside traditional laboratory environments. Systems such as Theia3D have been used to evaluate military and sports performance and in clinical assessments. However, while numerous validation studies have demonstrated promising accuracy, important limitations remain. These include limited transparency regarding kinematic model constraints and training datasets, as well as systematic differences between ML and traditional MB motion capture in joint centre definitions and allowable range of motion in each joint and anatomical plane. These differences make it challenging to distinguish true biomechanical differences from measurement-related differences. This consideration is particularly relevant when examining sex-based biomechanics, where known differences in movement patterns, injury mechanisms and musculoskeletal conditions exist. Understanding how ML and MB systems compare in capturing these differences is important for both research and clinical interpretation of results.

The purpose of this study was to compare MB and ML motion capture in their ability to detect established sex-based anthropometric and kinematic differences, and to determine whether the differences between motion capture systems are equivalent across sexes. Twenty-eight healthy adults (14 males, 14 females; 27 ± 4 years) completed self-selected-speed walking and drop vertical jumps (DVJ) while data were recorded simultaneously with both systems. To improve comparability, MB segment coordinate systems were aligned to the ML reference frame using a mean transformation matrix prior to kinematic analysis.

Both systems detected expected sex differences in hip width and shank length ($p < 0.001$). However, ML estimated notably shorter femur lengths compared to MB and did not identify a significant sex-based difference in thigh lengths, a well-established relationship in the literature. Comparison of joint centres revealed systematic offsets between the motion capture systems, with total mean offset of 3.37 ± 1.4 cm at the pelvis and 4.02 ± 1.3 cm at the hip. In general, ML pelvis and hip joint centres were positioned more distally compared to MB, across both static and dynamic trials.

For kinematics, both systems identified robust main effects of sex in frontal plane pelvis and hip motion. Sagittal plane waveforms showed very strong agreement between systems ($r = 0.98$). However, agreement was substantially weaker in the transverse plane for hip, knee, and ankle rotation ($r < 0.5$). Applying local coordinate system (LCS) alignment reduced overall root mean squared difference (RMSD) from $5.61 \pm 4.11^\circ$ to $3.12 \pm 1.88^\circ$ and significantly reduced systematic differences. For example, ankle rotation RMSD during walking decreased from 12.71° to 4.94° in males, and from 16.14° to 6.11° in females. Despite improvements, females generally exhibited larger RMSD between ML and MB at several joints, including pelvis rotation, ankle inversion and ankle rotation, suggesting a sex-specific effect in the comparison of MB and ML systems. Although LCS alignment substantially improved agreement, residual discrepancies remained.

Overall, both MB and ML systems identified key sex-based gait and landing characteristics. Both systems consistently identified greater pelvic list, hip adduction, and ankle flexion in females during walking, as well as greater anterior pelvic tilt, knee abduction, and ankle flexion in females during the drop vertical jump. However, agreement was generally higher in the sagittal plane, whereas lower agreement in non-sagittal plane kinematics – especially among females – highlights the need for careful consideration when choosing between ML and MB motion capture for research or clinical applications that rely on detecting subtle non-sagittal plane mechanics.

Table of Contents

List of Figures	vii
List of Tables.....	viii
Acronyms	xi
1. Introduction:.....	1
1.1 Background	1
1.2 Objectives and Hypotheses	3
1.2.1 Objective 1	3
1.2.2 Objective 2	5
1.3 Outline.....	6
2. Literature Review.....	6
2.1 Marker-Based Motion Capture	6
2.2 Markerless Motion Capture	8
2.3 Model Based Differences.....	10
2.4 Machine Learning Bias	10
2.5 Sex-Based Differences	12
2.5.1 Why sex matters in research	12
2.5.2 Anthropometrics Measurements	12
2.5.3 Kinematic Differences – Walking	14
2.5.4 Sex Based Differences – Drop Vertical Jump	17
2.5.5 Summary of sex-based differences	20
3. Methods.....	21
3.1 Participants.....	21
3.2 Ethics Approval.....	22
3.3 Experimental Protocol	22
3.4 Marker-Based Data Processing.....	26

3.5 Markerless Data Processing	28
3.6 Excluded Trials	28
3.7 Model Definition Sensitivity Analysis.....	29
3.8 Segment Alignment.....	30
3.9 Statistical Analysis	32
4. Results.....	34
4.1 Anthropometric Measurements and Joint Centre Comparison	35
4.2 Segment Alignment.....	39
4.3 Kinematic Comparison – Walking Trials.....	43
4.4 Kinematic Comparison – Drop Vertical Jump Trials	58
4.5 Kinematic RMSD and Bias.....	72
5. Discussion.....	75
5.1 Anthropometric measurements and joint centre comparison.....	76
5.2 Segment Alignment.....	78
5.3 Kinematic sex-based differences	80
5.3.1 Walking	80
5.3.2 DVJ	81
5.4 Kinematic method-based differences.....	82
5.4.1 Sex x Method interactions	84
5.4.2 RMSD differences.....	85
5.5 Implications.....	86
5.6 Limitations	86
6. Conclusion	88
6.1 Summary	88
6.2 Future Work	90

7. References.....	92
Appendix A – Participant anthropometrics and sizing.....	100
Appendix B – Certificate of ethics approval.....	101
Appendix C – Research consent form	102
Appendix D – Within-trial and between-trial comparison.....	106
Appendix E – Sensitivity Analysis	107
Appendix F – Mean Correction Angles	108
Appendix G – Q-Q plots	109
Appendix H – LMM Results of Similarly Sized Participants.....	117

List of Figures

Figure 1. Human Movement Biomechanics Laboratory cluster marker set [77]	24
Figure 2. A-Pose and Bikerpose.....	25
Figure 3. Harrington Hip Joint MB Model	27
Figure 4 - Local coordinate systems before alignment (left) and after alignment (right). MB and ML data are represented by red and blue.	31
Figure 5. Joint centre offset between raw MB and ML data in one female participant.....	38
Figure 6. Mean kinematic waveforms of left leg during the walk for raw MB and ML data across all participants	42
Figure 7. Mean kinematic waveforms of left leg during the walk for aligned MB and ML data across all participants	43
Figure 8. Mean pelvis angles during walk for male and female participants	46
Figure 9. Mean hip angle waveforms during walking	50
Figure 10. Mean knee angle waveforms during walking.....	54
Figure 11. Mean knee angle waveforms during walking.....	58
Figure 12. Mean pelvis angle waveforms during DVJ task.....	61
Figure 13. Mean hip angle waveforms during DVJ task	64
Figure 14. Mean knee angle waveforms during DVJ task.....	68
Figure 15. Mean ankle angle waveforms during DVJ task.....	72

List of Tables

Table 1. Mean Participant Anthropometrics	21
Table 2. Regression equations for hip joint centre estimation	29
Table 3. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from segment lengths. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size.	36
Table 4. Mean offset between MB and ML joint centres across all participants and trials	38
Table 5. p-values and Cohen's d effect sizes for comparisons of differences in joint centre offsets between sexes. Orange and yellow indicates that a very significant ($p < 0.001$) or a significant ($p < 0.05$) difference was found.	39
Table 6. RMSD ($^{\circ}$), bias ($^{\circ}$), and LOA ($^{\circ}$) between MB and ML kinematics in raw data and aligned MB data in males and females.	40
Table 7. RMSD ($^{\circ}$), bias($^{\circ}$), and LOA($^{\circ}$) between MB and ML kinematics in raw data and aligned MB data in males and females.	41
Table 8. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from pelvis kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	45
Table 9. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from hip kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	48
Table 10. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from knee kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate	

that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	52
Table 11. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from ankle kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	56
Table 12. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from pelvis DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	60
Table 13. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from hip DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	62
Table 14. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from knee DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.	66
Table 15. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from ankle DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes	

indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.....	70
Table 16. Mean, standard deviation, effect size, and significance of RMSD and Bias between MB and ML walking kinematics for male and female participants. Negative effect size indicates that mean of female data is greater than mean of male data. Yellow indicates that a significant difference ($p < 0.05$) was found.	73
Table 17. Mean, standard deviation, effect size and significance of RMSD and Bias between MB and ML DVJ kinematics for male and female participants. Negative effect size indicates that mean of female RMSD is greater than mean of male data. Yellow indicates that a significant difference ($p < 0.05$) was found.	74

Acronyms

2D – Two-dimensional

3D – Three-dimensional

CV – Coefficient of variation

ICC – Interclass correlation coefficient

HJC – Hip joint centre

LOA – Limits of agreement

MB – Marker-based

ML – Markerless

ROM – Range of motion

RMSD – Root means squared difference

1. Introduction:

1.1 Background

Biomechanical research is critical in developing advances in injury prevention, sports performance and clinical assessments. However, an increasing number of studies have highlighted that the empirical basis of biomechanical research is influenced by continuing sex-based disparities in participant recruitment, and data interpretation [1], [2]. It is well understood that women have historically been underrepresented as research participants, across medical fields and sport and exercise science. Women have often been treated as a special population rather than half of the worldwide population [3], [4]. These imbalances raise concerns regarding the generalisability, and validity of biomechanical models and technologies developed using predominantly male data.

Large-scale reviews of sport and exercise science literature have consistently documented this disparity. Costello et al. reviewed 1382 articles published between 2011 and 2013 found that women were significantly underrepresented across major sport and exercise medicine journals, accounting for only 35% to 37% of participants [5]. In studies published between 2014 and 2020, Cowley et al. reported that females accounted for only 34% of participants across more than 5000 studies, with nearly one-third of studies recruiting exclusively males [1]. This pattern persists within biomechanics research specifically. A meta-analysis of all studies published in the *Journal of Biomechanics* in 2024 revealed that one-in-four studies had less than 30% female representation in their datasets, and found that male-only studies outnumbered female-only studies five-to-one [2]. Additionally, female-only studies typically provided clear reasoning for focusing on a single-sex, while male-only studies frequently omitted both justification and explicit sex-identification in titles or abstracts [1], [2]. Collectively, these reviews demonstrate that the underrepresentation of female participants has remained a persistent challenge in exercise science and biomechanics, despite the growing recognition of the importance of considering sex in research design. These practices implicitly position male participants as the default research population, despite well-established anatomical, physiological, and endocrinological differences that influence biomechanics, injury risk, and movement strategies. Although representation has improved over time, male dominant sampling remains common, particularly in foundational datasets used for model development and validation [6].

These concerns are particularly relevant in the context of markerless (ML) motion capture systems, which rely heavily on large datasets to train and validate machine-learning algorithms. Theia3D, developed by Theia3D Markerless Inc. (Kingston, ON, Canada), has become one of the most widely used commercial ML motion capture system due to its ability to estimate three-dimensional human kinematics without requiring physical markers [7]. As a result, the system offers several practical advantages, including reduced participant preparation time and the ability to capture movement outside traditional laboratory environments, while avoiding some sources of error associated with marker placement and soft tissue artefact. However, like all data-driven systems, its performance is inherently linked to the characteristics of the data on which it was developed [8], [9], [10].

Despite the widespread application of Theia3D in biomechanics research, the available literature evaluating the agreement between ML and MB systems across sexes is limited. One study evaluating the intersession reliability of MB and ML motion capture during a drop vertical jump (DVJ) found discrepancies in sex-based comparisons between methods at the hip and ankle [11], highlighting the need for further investigation. Given documented sex differences in pelvis morphology [12], lower limb alignment, and movement differences [13], investigations are needed to evaluate whether ML and MB systems capture these differences consistently. Moreover, limited knowledge of the ML system's training dataset, including participant sex and demographics, make it difficult for end users to assess whether the system is equally sensitive to various subject demographics [10].

In contrast, traditional marker-based (MB) motion capture systems are not dependent on machine learning algorithms or training datasets. Instead, MB approaches rely on predefined marker-placement and biomechanical models defined using functional or regression methods and constrained using physical assumptions that can be modified depending on the research requirements [14], [15], [16]. As such, model behaviour is determined by explicit definitions rather than learned patterns, and if the model assumptions are appropriate, the MB approaches are less likely to systematically favour one sex over the other due to historical representation alone. Still, MB motion capture is not without limitations. Errors can be caused by soft tissue artefacts, misplacement of markers, and assumptions within the musculoskeletal models, many of which were developed using male-dominant anatomical data, including the commonly applied Rajagopal

OpenSim model [17], [18], [19]. Additionally, differences in segment definitions, joint coordinate systems, and scaling procedures can influence derived kinematics and anthropometric measures.

Despite these limitations, MB motion capture remains a well-established and widely accepted reference method in biomechanics research. MB motion capture has frequently been employed to validate Theia3D across various applications, including cerebral-palsy research, osteoarthritis research, clinical settings, military contexts, and sports performance [20], [21], [22], [23]. In the context of investigating sex-based differences in ML motion capture, MB approaches provide a reliable comparative framework for evaluating systematic differences in anthropometric measurements and lower limb kinematics. Much of the existing literature on sex-based differences in kinematics is based on MB motion capture [13], [24], [25], [26], [27]. Accordingly, MB motion capture is not treated as ground truth, but rather as a well-established reference framework against which systematic differences and agreement with ML outputs can be evaluated.

1.2 Objectives and Hypotheses

Given the documented underrepresentation of females in biomechanics research and the increasing interest in machine-learning-based motion capture systems, there is a critical need to evaluate the agreement between MB and ML methods and determine whether both approaches capture similar sex-based differences in biomechanical outcomes. The primary aim of this project was to determine whether sex-based differences exist in joint kinematic and anthropometric outcomes derived from Theia3D ML motion capture. Improving understanding of potential sex-specific differences in ML motion capture is critical for the appropriate application of this technology in both clinical and research settings. Additionally, many studies note a limitation in the comparison of ML and MB data due to model definition differences. A secondary goal of this study was to evaluate whether differences between MB and ML motion capture systems attributable to model definition could be reduced through alignment of segment coordinate systems.

1.2.1 Objective 1

The first objective was to determine whether MB and ML motion capture systems identify comparable sex-based differences in lower limb anthropometric and kinematic measurements, and whether these differences are consistent with those previously reported in literature.

Based on previous literature, 3D Euclidian distances between joint centres estimated by MB and ML motion capture systems have previously demonstrated an offset, averaged over gait cycles, between 24 mm and 36 mm [17]. Agreement of the systems within individual sexes, however, has not been examined. Many anthropometric measurements, such as femoral or tibial length, scale proportionally with overall stature. Therefore, both methods are expected to estimate comparable lower-limb segment lengths that reflect established sex-based anthropometric differences [28], [29], [30]. In contrast, pelvic morphology does not scale linearly with height in the same manner as long bone lengths. Pelvic breadth, acetabular orientation, and mediolateral dimensions demonstrate sex-specific morphological characteristics that are not fully explained by stature alone [28]. Additionally, the exact method and equations used by Theia3D to estimate the hip joint centre (HJC) are not publicly documented, and multiple approaches exist. As a result, while similar lower-limb segment length estimates are anticipated between the MB and ML systems, greater variability and reduced agreement may be observed at the pelvis and HJC, which may make examining sex-based differences more challenging. Agreement between the kinematics estimated by MB and ML systems are well documented in the sagittal plane. Accordingly, it is projected that both motion capture methods will be capable of identifying sagittal plane kinematic differences between sexes, because these differences are typically larger in magnitude and less sensitive to modelling assumptions. However, differences between the systems have been shown to be greater in the frontal and transverse planes, with ML often estimating lower ROM relative to MB [11], [17], [22]. Importantly, the literature consistently reported that the most meaningful sex-based kinematic differences most strongly associated to injury risk occur in the frontal and transverse planes, particularly at the hip and knee. Given our participant pool of 14 males and 14 females, statistical power limits mean that significant findings are likely to emerge only when there are medium or large effect sizes [31]. Therefore, while sagittal plane differences may be detectable by both systems, it is anticipated that the ability to identify frontal and transverse plane sex-based differences will depend on the magnitude of those differences and the sensitivity of each motion capture method. Additionally, many studies comparing sex-based kinematics have used MB motion capture for data acquisition. Consequently, MB is expected to demonstrate a strong capability in detecting previously documented kinematic differences between sexes. It is for the same reasons that ML may demonstrate greater differences to the results previously reported by MB studies in the frontal and transverse planes.

1.2.2 Objective 2

The second objective is to evaluate the extent to which alignment of segment coordinate systems reduces differences between MB and ML motion capture outputs and assess whether the amount of variation differs between male and female participants.

Aligning the segment local coordinate systems (LCSs) derived from ML data to MB data has previously been shown to increase agreement between systems and reduce systematic differences. Antonini et al. demonstrated that optimizing a rotation matrix to minimize kinematic differences reduced knee joint angle RMSD from $3.9 \pm 1.5^\circ$, $6.1 \pm 1.3^\circ$, and $10.2 \pm 2.8^\circ$ in the sagittal, frontal, and transverse planes to $1.7 \pm 0.2^\circ$, $1.7 \pm 0.3^\circ$, and $2.5 \pm 0.5^\circ$ [32]. These findings indicate that a substantial portion of inter-system disagreement may be attributable to coordinate system misalignment rather than true differences in estimated joint motion. In the present study, similar reductions of differences are expected following the alignment of the MB LCSs to ML data using a mean rotation matrix derived across frames of a single trial. Because this alignment approach assumed a consistent rotational offset between systems, the greatest improvements in agreement are expected in planes where disagreement arises primarily from coordinate definition. However, the magnitude of improvement may differ across anatomical planes and joints. Given that the ML system has been shown to underestimate frontal and transverse plane ROM relative to MB, and that these planes are more sensitive to segment definition, alignment is expected to produce the greatest reduction in differences in the frontal and transverse planes. Sagittal plane agreement is anticipated to have modest improvements because the baseline agreement between systems is strong. Furthermore, differences in alignment effectiveness may emerge between sexes. Theia3D does not require the input of sex when processing data, and therefore the model is likely trained using the combined data from males and females. Given that the sex distribution of the training dataset is not disclosed, it remains unclear whether the model is equally representative of male and female movement patterns. Marker placement error is a documented limitation of MB systems; however, that error is generally random across a sample, whereas dataset bias in ML has the potential to be systematic and directionally biased toward the dominant training population. If so, baseline agreement between ML and MB outputs may be greater in males than in females prior to alignment.

1.3 Outline

This thesis is organised into five chapters. In Chapter 2, research on sex-based bias in biomechanical studies as well as the advantages and limitations of MB and ML motion capture systems is presented, followed by a summary of the current literature on sex-based differences during walking and landing tasks.

Chapter 3 presents the experimental method followed during the data collection process in the laboratory. It includes detail on the recruitment, ethics, instrumentation, data processing, and analysis methods.

Chapter 4 details the results of this project from the ML and MB data. It provides quantitative results on how well the methods agree with each other and whether sex-based differences were found.

Chapter 5 presents a discussion on the implications and limitations of this research.

Lastly, Chapter 6 summarises the thesis and discusses whether the objectives were met. It also provides suggestions on future directions for this research.

2. Literature Review

2.1 Marker-Based Motion Capture

Traditional optical motion capture systems (e.g., Vicon, Qualisys) that rely on reflective markers have been considered the “gold standard” for biomechanical analysis for decades [33]. These systems estimate human motion by tracking retro-reflective markers placed on anatomical landmarks. The markers are then captured by a calibrated array of infrared cameras, and the subsequent reconstruction of the marker locations in three-dimensional space. Segment coordinate systems and joint kinematics are subsequently derived using biomechanical models such as the Plug-in-Gait model [34].

The widespread adoption of MB systems has largely been driven by their high spatial resolution and compatibility with inverse dynamics approaches. Marker trajectories are typically localised with sub-millimetre accuracy, enabling precise estimation of joint centres, segment orientations, and joint moments when combined with force plate data [34], [22], [35]. Because MB motion

capture is well established in laboratories, the approach serves as the foundation for much of the biomechanical literature on walking, running, jumping, and other sport-specific tasks. Decades of use of the MB systems have also resulted in large datasets across wide ranges of populations and tasks. These datasets have established MB motion capture as the reference standard against which emerging technologies, including ML systems, are benchmarked [34].

Despite its strengths, MB motion capture is subject to several well-documented limitations. One of the most significant sources of errors is soft tissue artefact, which occurs when the skin and adipose tissue move relative to the bony landmark on which the marker was placed. As a result, the position of the marker relative to the anatomical landmark during a static position is not fixed. One study comparing MB HJC estimations to ground-truth hip joint centre data from dynamic biplane radiography have found error in MB HJC from 16.6 mm during gait up to 89.9 mm during step-up tasks, as well as hip flexion angle errors ranging from 7.2° to 23.8° [36]. Soft tissue artefacts can also introduce angular errors up to 7° in lower limb joint angles, with errors often being greater during high impact activities such as jumping or change-of-direction movements [18]. These errors disproportionately affect frontal and transverse plane kinematics which have a smaller range of motion.

Practical challenges also limit MB systems. Marker occlusion and loss are common during tasks involving rapid movements or additional equipment. Self-occlusions can also often lead to incomplete data [22]. In addition, MB motion capture requires lengthy preparation, staff trained to accurately place markers on anatomical landmarks, static calibration trials, and a tedious post-processing protocol to label markers and fill gaps in short periods of missing marker data. Even with knowledgeable personnel, human error is impossible to completely rule out and small errors of 10 mm in marker placement can produce joint angle errors of $>5^\circ$ [37], [38]. Additionally, if multiple technicians are responsible for placing the markers, inter-operator variability can produce inconsistencies in the marker sets [23]. Accuracy of the MB motion capture system also decreases when participants have a higher body-fat percentage as there is an increase in soft-tissue artefacts [35]. Finally, a MB system requires a controlled environment with a calibrated capture volume and consistent lighting, which makes field-based assessments challenging.

2.2 Markerless Motion Capture

ML motion capture emerged as a response to the methodological and practical limitations of the MB system. Instead of relying on physical markers, ML motion capture approaches estimate human pose data directly from video data, thereby reducing set up time and participant discomfort, enabling data collections outside of specialised laboratory environments, and removing soft-tissue artefacts and marker placement error by estimating joint centre and skeletal poses from visual information [33]. Commercial systems such as Theia3D (Theia Markerless, Kingston, ON), employ multiple cameras and deep-learning models to predict key anatomical landmarks, which are then fitted to a biomechanical skeleton using inverse kinematics [22].

Theia3D is one of the most widely used commercially available ML motion capture system. As a result, many studies have evaluated Theia3D's performance against MB motion capture technology for validation. Across literature comparing ML and MB approaches, strong agreement between the systems has been consistently reported in the sagittal plane for the hip, knee and ankle angles, with Pearson correlation coefficients often exceeding 0.9 for knee and ankle flexion [33], [22], [16], [7]. Sagittal plane kinematics are also the most widely reported across literature. All studies that evaluate lower limb kinematics discuss hip, knee, and ankle flexion; however, only a portion of the studies also discuss knee valgus, knee rotation, ankle eversion, or ankle rotation. In the frontal plane, agreement is more variable with moderate to good correlations ($r = 0.6 - 0.8$) reported for hip adduction/abduction and knee varus/valgus. The root mean square differences (RMSD) in the frontal plane typically range from 3-5° [33], [16]. Kinematics in the transverse plane result in the weakest agreement between the two motion capture approaches with correlation coefficient falling below 0.6 and RMSD frequently exceeding 8-15° for the hip and knee [17], [16], [39].

An important advantage of ML motion capture is its improved inter-session reliability. Because ML systems do not rely on the manual placement of markers, they avoid a major point of contention inherent to MB motion capture. It is very challenging to place markers in the same exact position over multiple sessions. In fact, data collection sessions are frequently restarted if a critical anatomical marker falls, or if a marker-cluster shifts. Studies evaluating inter-trial reliability of motion capture systems have shown Theia3D outperforming MB motion capture on reliability in both male and female datasets [11], [10].

Despite Theia3D's impressive sagittal-plane accuracy, Theia3D is still subject to limitations. Theia3D's internal model definitions and landmarks used in segment definitions are proprietary. As a result, the system acts as a "black box", outputting results but not providing details of how conclusions are drawn. Information on the segment coordinate systems is especially relevant when comparing ML and MB motion capture, as it is a necessary step to quantifying which differences between the approaches are due to measurement differences or inherent error. Additionally, not knowing how segments are defined makes it difficult to trace the source of systematic offsets [32]. Several studies have reported systematic angular offsets between the two approaches, highlighting the importance of understanding underlying model assumptions [17], [22], [34]. Furthermore, available information on the ML model's kinematic constraints provides little information other than the number of degrees of freedom (DOF). The default model consists of two kinematic chains (lower body: pelvis and legs, and upper body: torso, arms, and head), 6 DOF shoulder joints, 3 DOF hip joints, 2 or 3 DOF knee joints, and 6 DOF ankle joints with limited translation [40], [41]. Theia3D documentation states that improvements to the model and algorithm have allowed for the reduction of constraints on the joints; however, the specific limits applied within the model are not publicly described, making it challenging to determine how these constraints may influence the resulting joint kinematics [40].

Theia3D is a vision-based system, which requires a clean, unobstructed view of the participant to function to its fullest potential. Occlusions for sports or military equipment or bulky clothing can increase error in joint centre estimations. A study evaluating the effectiveness of Theia3D to evaluate lower-limb joint kinematics in participants wearing loose and tight-fitting clothing found greater differences between hip flexion angles in loose clothing [42]. Other studies have also noted that specific populations such as children, individuals with cerebral palsy, and high-body mass index (BMI greater than 30) participants show larger RMSD for hip rotation and poorer correlation for pelvic tilt [11], [21], [43], [20], [44].

Higher frequency noise is another concern, particularly in the transverse plane. Frame-by-frame pose estimation introduces greater high-frequency variability compared to MB systems, and while a low-pass filter can be used to reduce RMSD, it may also alter true signal characteristics [45].

2.3 Model Based Differences

Both MB and ML pipelines rely on biomechanical models to define segment coordinate systems and compute joint kinematics. Even when similar anatomical landmarks are used, differences in coordinate system orientations can introduce systematic angular offsets unrelated to measurement noise. Consequently, observed discrepancies between MB and ML outputs may reflect modelling differences as well as true differences in kinematic estimation accuracy. Many validation studies have implicitly accepted these offsets as unavoidable; however, this limits interpretability as it makes it impossible to separate measurement differences from local coordinate system differences from systematic differences. In 2024, Antonini et al. demonstrated that aligning the local coordinate systems using matrix rotation can significantly reduce RMSD across all planes and allow for a more accurate comparison of measurement differences [32].

Version dependency further complicates comparisons between Theia3D and MB systems. Continuous software updates have introduced changes to joint centre definitions and segment orientations that can alter joint angle outputs derived from identical video data [46], [47], [40]. One of the most consequential updates occurred during the 2024 Apollo release, which revised the pelvis definition. Earlier versions of Theia3D assumed a near-neutral pelvis in the sagittal plane, leading to systematic underestimation of hip flexion by approximately $6\text{-}13^\circ$ and of hip internal/external rotation by $8\text{-}12^\circ$ [46], [33]. Theia3D now employs a pelvis with moderate anterior tilt which better mimics the natural position of the pelvis while standing in a neutral position. Theia3D's updated pelvis definition also works better with Visual3D's composite pelvis, which is used in their skeletal model, and is designed to have the pelvis coordinate system tilted forward [46], [48].

2.4 Machine Learning Bias

Because ML motion capture relies on machine learning, additional sources of bias may influence its analyses. Theia3D employs a convolutional neural network to estimate human pose from a minimum of six synchronised video cameras. Theia3D states that its algorithm has been trained on more than 500,000 images of people performing varying movements, in various clothing and environments on which anatomical landmarks were manually marked by trained annotators [17]. After training, a testing dataset of new images was presented to the algorithm. When processing video data, the algorithm uses the knowledge from the training images to estimate the 2D positions

of 51 features on the participants one frame at a time for each camera [17]. The calibration of the camera array relative to the laboratory origin, allows for the 2D positions of the features to be combined into 3D positions. A multi-body model can then be scaled to the participant using these features. For optimal results, Theia3D recommends using at least eight cameras, arranged with different orientations to track the participant in the desired capture volume. Theia3D also recommends making sure all joints are visible by at least three cameras throughout the trials.

Even when best-practice guidelines are followed, machine learning-based ML motion capture systems remain subject to several sources of bias that can affect data processing and interpretation. Bias can arise at multiple stages of the machine learning lifecycle, most notably during data generation and model development. Commonly documented forms include representation bias and measurement bias, which occur during data collection, as well as learning bias, aggregation bias, evaluation bias, and deployment bias, which may emerge during model training, optimisation, and application [49].

Representation bias occurs when the dataset an algorithm is trained on is not sufficiently representative of the target population, potentially leading to reduced accuracy for underrepresented groups defined by sex, body-composition, age, ethnicity, or movement pathology [49], [44]. Although the general training framework used by Theia3D has been described, the demographic composition of the training dataset has not been disclosed. However, there is evidence of representation bias in the broader computer vision literature, as vision-based algorithms for facial classification have been shown to exhibit greater error when applied to darker-skinned females compared to lighter-skinned males. Similar forms of representation bias have also been identified in other ML pose estimation frameworks, including OpenPose, MoveNet, and PoseNet [9], [50].

Measurement bias occurs when the features used to characterise a sample do not accurately capture the intended variables of interest. In ML motion capture, differences in camera configuration, lighting conditions, occlusions, and clothing can introduce uneven data quality across participants or experimental conditions, leading to systematic bias [51]. During model optimisation, learning bias may arise when modelling choices disproportionately amplify performance disparities across different subsets of the data, while labelling bias can be introduced if training annotations reflect subjective or systematic biases of human annotators. Aggregation bias occurs when a single,

generalised model is applied across data where there are sub-groups that should be considered differently, leading to a model that is sub-optimal for any specific group or a model that performs adequately when used on the dominant population. Finally, evaluation bias can arise if validation datasets do not reflect real-world user populations, and deployment bias may occur when a model is applied in contexts or tasks for which it was not designed [49].

2.5 Sex-Based Differences

2.5.1 Why sex matters in research

Many studies have investigated the biological influences of gait biomechanics, and that these differences can be linked to higher incidence of lower-extremity injuries in females, including patellofemoral pain, anterior cruciate ligament ruptures, iliotibial band syndrome, knee osteoarthritis, and stress fractures [52], [53], [54]. Anatomical differences, such as a wider pelvis and shorter femur length, combined with sex-related variations in muscle-mass distribution, are thought to derive systematic biomechanical differences during locomotion.

Several mechanical pathways have been proposed to explain why females might have a higher rate of injury occurrence. Greater hip adduction and internal rotation during the stance phase causes an increase in lateral knee loading. The greater load has been identified as a risk factor for patellofemoral pain [55]. During running, females exhibit larger peak knee-adduction angles (~7.5 vs 2.5 deg in males) which are strongly associated with injuries to the anterior cruciate ligament [56]. Load carriage, walking, or jumping activities, amplify these patterns; a systematic review of load carriage studies reported that female are more susceptible to lower-limb overuse injuries when carrying absolute loads, a finding attributed to the combination of higher relative effort and altered frontal-plane mechanics [52].

2.5.2 Anthropometrics Measurements

Biological sex is a major determinant of human anthropometry, influencing skeletal dimensions, pelvis morphology, and joint geometry. These differences are central to applications such as forensic identification, stature estimation, biomechanical modelling, clinical gait analysis, and orthopedic implant design. One measure that is particularly interesting is how sex-based differences in pelvis and lower limb dimensions affect the estimation of the HJC, a critical input for inverse kinematics.

Across geographically and ethnically diverse adult populations, males consistently exhibit longer and wider lower limb segments than females. Studies evaluating Sudanese, Egyptian, Indian, and Malaysian samples report significantly greater tibial length, femoral length, foot length, and foot widths in males, with these studies typically reporting very significant differences between sexes ($p < 0.001$) [57], [58], [59], [60]. For example, femoral measurements from Indian skeletal samples indicate a mean femur length of approximately 437 mm in males compared to 402 mm in females, alongside larger trochanteric and femoral head dimensions in males [59]. Width measurements, such as bimalleolar breadth and foot width, often display greater sexual dimorphism indices than segment lengths, which makes them particularly important discriminators in forensic sex classification. For example, bimalleolar breadth alone classified sex with nearly 90% accuracy in a Sudanese cohort, with marginal improvement when incorporated into multivariate models [57]. Lower limb dimensions also demonstrate strong correlation to stature. Across studies, tibial length consistently is another strong predictor, explaining approximately 60-76% of the variance in stature across populations [58], [60].

In contrast to males exhibiting larger segment dimensions, sex-differences in pelvis-related measures diverge. In a sample of professional soccer athletes, females exhibited a greater pelvic width below the femoral head (approximately 16.8 cm in females versus 15.3 cm in males), despite males being taller and having longer femurs and tibias [29]. These findings indicate that pelvis geometry does not scale uniformly with the overall body size. The pelvis has evolved with trade-offs allowing for bipedal locomotion, while also needing to meet obstetric demands. As a result, the pelvis is often the most reliable skeletal segment for identifying sex [61]. Female pelvises typically present with a wider pelvic inlet and outlet, a broader sciatic notch, and a sub-pubic angle that can often measure up to 30° more than that of males, while male pelvises present with thicker bones and a narrower pelvic architecture [62]. Quantitative imaging studies further reinforce these morphological distinctions. Analyses based on computed tomographic images confirm sex-differences in pelvic width, depth, and angular measures [63]. Radiographic analyses have demonstrated that even simple linear measurements such as the distance from the femoral head to the pelvic midline differ significantly between sexes. For example, the average distance from the femoral head to the pelvic midline has been reported as 90.6 mm in males compared to 87.8 mm in females [12]. The femoral head diameter in males has also been reported to be an average of 6 mm larger than female counterparts [12]. These findings are directly relevant to biomechanical

modelling because pelvis landmarks and dimensions form the basis for many HJC estimation methods.

Accurate localisation of the HJC is essential for reliable gait kinematics. As stated in the section ‘2.1 Marker-Based Motion Capture’, changing the position of the anatomical landmark markers – and as a result the joint centres – can cause significant changes in kinematics. There are two types of methods for estimating the HJC: predictive methods which rely on regression equations derived from external marker positions or segment lengths, and functional methods that estimate the HJC from motion of the femur relative to the pelvis. Systematic evaluations indicate that predictive methods achieve mean errors of 7-17 mm, which is acceptable for most clinical and research applications [64], [65], [66]. Among predictive approaches, regression equations that consider leg length, have shown greater robustness. These methods have achieved mean absolute errors below 6 mm [30]. However, in populations with hip dysplasia, the Harrington pelvis-only method was shown to be the most accurate and had the lowest directional bias to HJC estimated with magnetic resonance imaging compared to Harrington pelvis-and-leg, Bell, and Hara methods [61]. It is important to consider that, in many cases, marker placement error and soft-tissue artefacts has greater effects on the results than the choice of HJC equation, meaning that that sex-specific adjustments may have limited practical impact [67]. Comparative gait studies report that different predictive methods produce modest differences in hip kinematics, with typical differences of less than 1° in the sagittal plane [68].

Functional methods for HJC have been shown to perform better than predictive equations when participants can achieve a normal hip range of motion, but their accuracy deteriorates with populations with constrained movement [64]. Quality-control approaches, such as SCoRE residual analysis combined with optimal common shape techniques, provide a means of monitoring HJC estimation reliability and can reduce errors to below 4 mm [69].

2.5.3 Kinematic Differences – Walking

Walking is one of the most frequently studied tasks in biomechanics, and a substantial body of literature has examined whether males and females employ different kinematic strategies during gait. Across a wide range of experimental conditions, women tend to walk with a higher cadence and a shorter stride length while achieving comparable self-selected walking speeds. In a population of healthy adults, women average 113 steps per minute versus 108 steps per minute for

men, and men produced a slightly longer stride of 1.49m versus 1.43m for women [13], [70], [28]. However, when speed and stride are normalised to leg length the differences disappear, indicating that stature rather than sex explains most of the spatiotemporal differences [13]. Data from older adults between the ages of 50 and 96 from the Baltimore Longitudinal Study of Aging confirm that women adopt a higher cadence (115 steps per minute) and a narrower step width to maintain a speed of approximately 1.1 metre per second [25].

Although walking is often considered a low-exertion task, consistent sex-based differences have been reported, particularly in the frontal and transverse planes of motion at the pelvis and hip. In contrast, sagittal plane kinematics at the hip, knee, and ankle tend to show smaller, less consistent sex-based differences. Sex-based differences in pelvic motion are among the most consistent in the literature. Multiple studies have reported greater frontal plane pelvic range of motion (ROM) in females compared to males [13], [28], [71]. Smith et al. reported pelvic obliquity ROM values of $9.4 \pm 3.5^\circ$ in females versus $7.4 \pm 3.1^\circ$ in males ($p < 0.01$) [28]. Similar significant differences in ROM magnitudes have also been reported across the literature by Bruening, Chocklingam, Cho and Chumanov [13]. Greater transverse-plane motion in females has also been reported, although results are somewhat more variable. Bruening et al. observed pelvic transverse ROM of $14.1 \pm 4.6^\circ$ in females compared to $12.0 \pm 3.0^\circ$ in males ($p < 0.01$) [13]. Smith et al. reported slightly smaller pelvis rotation values in males ($11.4 \pm 4.2^\circ$ and $10.4 \pm 4.1^\circ$ for females and males respectively), with no significance noted [28], [71]. However, other studies have reported smaller or non-significant differences or even slightly larger pelvic rotation ROM in males [13]. Larger transverse plane motion in males could be because of the faster walking speed and longer stride lengths [71]. These discrepancies suggest that the magnitude of this difference is sensitive to population characteristics and methodological factors. Sagittal-plane pelvic motion shows minimal sex-based differences. Bruening et al. reported nearly identical sagittal-plane ROM between females ($4.4 \pm 1.3^\circ$) and males ($4.5 \pm 1.2^\circ$), similar non-significant differences have been reported in many studies [13], [24], [28], [71]. Collectively, these results indicate that sex-based pelvic differences during walking are primarily expressed in the frontal plane, with more modest and inconsistent effects in the sagittal and transverse plane.

The motion of the hip joint exhibits the most pronounced and consistent sex-based kinematic differences of the joints during walking. In the frontal plane, females consistently achieve greater

hip adduction ROM and peak hip adduction angles compared to males. Bruening et al. reported hip adduction ROM of $16.2 \pm 3.4^\circ$ in females versus $14.2 \pm 3.1^\circ$ in males ($p < 0.01$). Similar significant differences in ROM, peak hip adduction and hip adduction angle at initial contact have also been described in other young adult populations by Mullerpatan et al., Phinyomark et al., and Takano et al., Ko et al. also reported that elderly women continue to demonstrate greater hip adduction than elderly men [24], [25], [53], [54]. Greater hip adduction in females has been attributed to sex-based differences in pelvic morphology, including a wider pelvis, as well as lower relative hip adductor strength. These factors may contribute to greater frontal-plane motion during stance to maintain dynamic stability [71].

Transverse-plane hip rotation also differs between sexes, with females generally exhibiting greater internal rotation ROM. Bruening et al. reported hip transverse-plane ROM values of $5.9 \pm 3.2^\circ$ in females compared to $3.7 \pm 2.2^\circ$ in males ($p < 0.01$) [13]. Chen et al. similarly reported greater peak hip internal rotation in females than males with angles of $6.3 \pm 2.5^\circ$ and $3.3 \pm 1.3^\circ$ [72]. Significant differences in hip rotation between sexes were also reported by Mullerpatan et al. for peak angles, Chumanov et al. for ROM during the walk, and Takano et al. for peak angles, ROM and initial contact angle [24], [53], [55].

In the sagittal plane, discussion on sex-based hip flexion differences is mixed. Bruening et al. reported non-significant difference between hip flexion ROM values of $45.7 \pm 4.5^\circ$ in females and $44.1 \pm 3.9^\circ$ in males [13]. Mullerpatan et al., Chen et al. and Chumanov et al. described similar results [24], [55], [72]. Ko et al. reported that in adults between the ages of 50-59, males had significantly greater ROM [25]. Meanwhile, other studies have reported significantly greater peak hip flexion and ROM in women, including Kerrigan et al, Araki et al., and Wills et al. [44], [73], [74].

Sex-based differences at the knee during walking are less pronounced and more variable than those observed at the pelvis and hip. In the sagittal plane, studies generally reported small or non-significant differences in knee ROM and peak angles. Kerrigan et al. found that women achieved significantly greater knee flexion at initial contact with angles of $7.1 \pm 4.8^\circ$ compared to $3.7 \pm 5.1^\circ$ in males ($p < 0.001$), significant greater peak knee flexion was also measured with angles of $61.5 \pm 6.2^\circ$ in females and $59.4 \pm 5.0^\circ$ in males ($p = 0.03$) [73]. However, in terms of ROM, many studies did not find significant sex-based differences [13], [25], [72].

Frontal plane knee kinematics also show mixed results. Some studies report greater knee valgus angles in females, while others report no significant differences. Phinyomark et al. reported greater peak frontal-plane knee angles in females ($-5.20 \pm 3.07^\circ$) compared to males ($-1.77 \pm 2.86^\circ$, $p < 0.05$, $d > 0.8$), suggesting a meaningful sex-based difference [54]. Chen et al. reported that, on average, females had greater peak knee adduction angles; however, significance was not reported between sexes [72]. Ko et al. found no significant sex-based differences in knee adduction ROM [25]. Transverse plane knee motion is not as commonly reported in kinematic studies; however, Ko et al. found that both the mean knee rotation ROM and peak angles were greater in women [25].

Sex-based differences at the ankle during walking are generally small and primarily confined to the sagittal plane. Bruening et al. reported slightly greater ankle sagittal-plane ROM in females ($34.6 \pm 3.5^\circ$) compared to males ($32.4 \pm 4.5^\circ$, $p < 0.02$) [13]. Kerrigan et al. similarly reported greater, minimum flexion angle or peak plantarflexion in females ($-22.2 \pm 7.5^\circ$) than males ($-19.3 \pm 6.3^\circ$, $p < 0.05$). Araki et al. found similar significances in minimum and maximum ankle flexion angles [74]. Frontal- and transverse-plane ankle kinematics show inconsistent and generally small sex-based differences. Al-Kharaz et al. reported event-specific differences in ankle inversion and rotation at heel strike, mid-stance, and heel-off; however, variability was high, particularly in the transverse plane, and significances were not reported [75]. Ko et al. reported no significant sex-based differences in ankle frontal-plane ROM in older adults, despite observing greater sagittal-plane ROM in females [25]. Studies using biplane radiography further suggest that after normalising for bone size, sex-based differences in ankle rotational and translational motion are minimal.

2.5.4 Sex Based Differences – Drop Vertical Jump

In addition to walking, landing tasks such as the drop vertical jump (DVJ) have been widely used to investigate sex-based differences in lower-limb kinematics due to their potential role in the elevated risk of anterior cruciate ligament injury observed in female athletes [76]. Across the literature, consistent differences between male and female participants have been reported at the pelvis, hip, knee, and ankle, with differences most evident in the frontal and transverse planes at key events during landing, including initial contact, peak knee flexion, and maximum knee flexion.

However, the magnitude and direction of these differences depend on the task demands, landing strategies, and the motion capture method used to collect the data.

Pelvis and trunk differences have consistent findings across landing kinematics. Static comparison of lower extremity alignment found that females had greater pelvic tilt, femoral anteversion and Q-angle compared to males, which could influence the kinematics [76]. During dynamic landing tasks, several studies have reported that females consistently demonstrate a more upright trunk posture, as well as larger anterior pelvic tilt [77], [78]. Frontal and transverse plane pelvis kinematics are rarely reported in landing task studies, with research focusing primarily on the hip and knee kinematics.

Sex differences in hip flexion during landing are frequently reported, though the magnitude and direction of findings are somewhat event dependent. Chun et al. reported that females achieved significantly greater peak hip flexion angles than males, with values of $78.6 \pm 15.3^\circ$ and $75.5 \pm 15.1^\circ$, respectively [76]. Similar results were also reported by Sudo et al. [79]. However, some studies report that females land with smaller hip flexion and achieve peak hip flexion faster than male participants, while others have found no significant difference between sexes in initial contact or peak hip flexion during DVJ tasks [80], [81], [82].

In the frontal plane, females have shown lower hip adduction at initial contact compared to males with angles of -7.6° and -10.8° [76]. Chun et al. continued to observe smaller hip adduction in females through the peak with angles of -6.8° and -3.1° for males and females. Earl et al. also reported similar results with mean angles of $1.9 \pm 4.9^\circ$ for females compared to $-1.9 \pm 4.9^\circ$ for males [26], indicating a clear directional difference in frontal-plane hip posture. However, a study comparing Theia3D to MB motion capture during DVJ found that the ML system observed greater hip adduction over 17-89% of the DVJ task in females, where no differences were observed between sexes in MB [11].

Sex-based differences are also evident in the transverse plane at the hip with female participants achieving greater internal hip rotation compared to males during DVJ landing. Earl et al reported significant difference in peak hip internal rotation values of $1.6 \pm 5.1^\circ$ in females compared to $5.2 \pm 5.1^\circ$ in males [26]. Chun et al. also reported greater hip internal rotation at initial contact ($-9.8 \pm 10.2^\circ$ in females vs. $-6.2 \pm 9.8^\circ$ in males) and greater peak hip internal rotation ($5.9 \pm 9.6^\circ$ vs. $6.1 \pm 10.2^\circ$), although not all transverse-plane differences reached statistical significance [76].

Wills et al. further supported these findings reporting that significant difference in hip internal rotation between sexes was found from 70-72% of the DVJ phase in ML, which corresponds to the frames with the maximum hip rotation; however, the same observation was not made by MB [11].

Sagittal plane knee kinematics consistently indicate that males land with greater knee flexion compared to women, with Chun et al. reporting significantly lower knee flexion in females ($23.9 \pm 6.0^\circ$) at initial contact compared to males ($25.3 \pm 7.1^\circ$) [76]. Decker et al, also reported similar results with knee flexion angles of $22.8 \pm 8.0^\circ$ in females versus $30.0 \pm 7.7^\circ$ in males during initial contact [82]. Another study comparing MB and ML during DVJ reported that both methods found that women had significantly less hip flexion between 1-5% of the DVJ [11]. Sex-based differences continue to be observed through the peak loading phase where Chun et al. reported peak knee flexion of $97.8 \pm 12.9^\circ$ in females compared to $100.4 \pm 15.7^\circ$ in males ($p < 0.05$) [76]. Kawaguchi et al. also reported peak knee flexion at maximum knee flexion of $77.5 \pm 12.3^\circ$ in females versus $87.5 \pm 17.3^\circ$ in males ($p < 0.01$). However, significant difference in peak knee flexion angle or ROM is not reported in all studies. Both Peebles et al. and Earl et al. reported that no significant difference was observed in knee ROM during DVJ [27], [26].

In the frontal plane of the knee, females demonstrate significantly greater knee adduction and abduction angles. Chun et al. reported peak knee adduction of $9.8 \pm 9.0^\circ$ in females compared to $2.3 \pm 7.9^\circ$ in males, along with greater knee abduction throughout the landing phase [76]. Earl et al. reported knee adduction values of $6.0 \pm 4.7^\circ$ in females versus $1.0 \pm 3.2^\circ$ in males, reinforcing the presence of greater frontal plane collapse in females [26]. Kawaguchi et al. expanded on these findings, reporting significantly greater knee valgus angles in females at initial contact, maximum knee flexion, and toe-off. At maximum knee flexion, females exhibited valgus angles of $6.2 \pm 11.6^\circ$, whereas males demonstrated varus-biased angles of $-9.5 \pm 17.5^\circ$ [80]. Peebles et al. similarly reported greater peak knee abduction angles in females, with values of $-8.22 \pm 7.36^\circ$ compared to $-2.15 \pm 4.62^\circ$ in males [27]. Collectively, these findings indicate that increased frontal-plane knee valgus in females is a consistent characteristic across landing tasks.

Transverse plane differences at the knee also further distinguish the landing mechanics of the sexes, with females often exhibiting less rotation compared to males. Chun et al. reported that males had significantly greater internal knee rotation at landing with rotation of $4.4 \pm 1.1^\circ$

compared to female's $1.1 \pm 8.1^\circ$ rotation [76]. This observation continued during peak rotation where males were reported to have internal knee rotation of $23.9 \pm 5.0^\circ$, whereas women only achieved $18.7 \pm 9.3^\circ$ [76]. Earl et al. similarly reported a smaller peak internal rotation angle of $0.3 \pm 3.2^\circ$ in females versus $3.6 \pm 6.9^\circ$ in males [26].

Although the ankle kinematics are less frequently discussed compared to the knee or the hip, they are nonetheless important as sex-based differences have been reported during the DVJ. Females have demonstrated significantly greater ankle dorsiflexion with Decker et al. reporting ROM values of $58.0 \pm 7.9^\circ$ in females compared to $41.6 \pm 6.9^\circ$ in males [82]. Decker also found that the dorsiflexion at initial contact was significantly greater in females with values of 21.3 ± 9.6 compared to 11.3 ± 5.1 in males [82]. These findings are in line with the results presented by Chun et al. which reported initial contact angles of 14.8 ± 11.3 and 10.6 ± 11.6 in females and males, respectively [76]. Both Chun et al. and Sudo et al. also discussed that females had greater peak dorsiflexion angles during initial, however a significant main effect of sex was not identified [76], [79].

In the frontal plane, Chun et al. found significant difference between sexes for the peak angles during the DVJ cycle with values of -1.1 ± 1.9 for males and -2.0 ± 2.0 for females, significant difference was not found between the initial contact angles [76]. Transverse plane ankle kinematics show limited sex-based differences. Chun et al. reported a significant sex difference during IC, with females demonstrating ankle internal rotation angles of -0.6 ± 1.8 compared to 0.3 ± 1.9 for males. However, no significant sex difference was observed in peak transverse plane ankle rotation during the DVJ cycle [76].

2.5.5 Summary of sex-based differences

Overall, the literature indicates that kinematic sex-based differences are not distributed uniformly across lower-limb joint or movement tasks. During walking, the most consistent differences occur at the pelvis and hip, particularly in the frontal and transverse planes. Females consistently demonstrate greater pelvis obliquity, increased hip adduction, and greater hip internal rotation, whereas sagittal plane differences at the hip, knee, and ankle are generally smaller and less consistent. Knee kinematics during the walk show mixed findings across planes, while ankle differences are typically limited to greater dorsiflexion in females. In contrast, the DVJ produces larger and more pronounced sex-based differences. During dynamic landing tasks, females

consistently demonstrate greater frontal-plane knee valgus, reduced knee flexion during landing, and increased ankle dorsiflexion compared to males. Sex-based differences at the hip are primarily expressed in the frontal and transverse planes, where females typically exhibit greater hip adduction and internal rotation, whereas differences in sagittal-plane hip flexion remain inconsistent. These findings suggest that the frontal and transverse planes of the hip and knee may provide the most sensitive measures for evaluating sex-specific differences in movement analysis.

3. Methods

3.1 Participants

Twenty-eight participants took part in this project – fourteen males and fourteen females. The average participant height was 1.73 ± 0.09 m, the average participant mass was 73.1 ± 13.5 kg, and the average participant age was 27 ± 4 years old. Table 1 presents the mean anthropometric data of each sex. Appendix A presents a more detailed table of the participants' height, mass, and age.

In 2024, Statistics Canada reported the mean height, weight and BMI for males between the ages of 20 and 39 to be 177.9 cm, 89.13 kg and 28.1 kg/m^2 , whereas the females had measures of 163.7 cm, 74.5 kg and 27.6 kg/m^2 [83]. Both the males and the females in this study are taller and lighter compared to the mean Canadian adult population; the difference in height between groups remained comparable, while the difference in weight is slightly larger in this study's participants.

Table 1. Mean Participant Anthropometrics

	Male	Female	P-Value
Age	27.0 ± 4.3	26.9 ± 3.8	0.965
Height (m)	1.81 ± 0.1	1.66 ± 0.1	< 0.001
Weight (kg)	83.0 ± 9.2	63.3 ± 9.3	< 0.001
BMI (kg/m^2)	25.5 ± 2.7	22.9 ± 3.3	0.035

The participants were recruited among healthy adults over the age of 18. To be included in the study, participants had to be able to achieve their full joint range of motion safely. Specifically, participants who suffered from a broken bone, sprained joint, muscle tear, or any injury that would

limit their ability to perform walking or jumping tasks at the time of recruitment were excluded from participating.

A priori power analysis was conducted using G*Power 3.1 to determine the minimum number of participants required to detect a medium-sized interaction effect between sex and motion capture method [84]. The choice of effect size is supported by literature reporting moderate kinematic differences between MB and ML systems through gait [16] and moderate-to-large kinematic differences between sexes during walking and landing tasks [13], [80], [82]. A repeated measures ANOVA was selected with two groups (male and female) and two measurements (MB and ML). The analysis used the following parameters: effect size $f = 0.30$, $\alpha = 0.05$, power $(1 - \beta) = 0.80$, correlation among repeated measures = 0.5, and nonsphericity correction $\epsilon = 1$. The analysis indicated that a minimum requirement of 24 participants (12 per sex) would be required. To account for potential data loss due to technical issues such as marker occlusion, tracking errors, or processing issues, additional participants were recruited. As a result, 14 participants of each sex were included in the study.

These participants were recruited on a first-come, first-served basis. Participants were informed of the study via word-of-mouth, email, or Microsoft Teams.

3.2 Ethics Approval

This research project was approved by the University of Ottawa Ethics Review Board (ethics file number: H-06-25-11691). Appendix B includes a copy of the approved ethics application that was submitted for this project.

Before beginning a data collection, every participant gave informed consent by signing the consent form. A copy of the consent form is presented in Appendix C. Participants also consented to having video of their static poses and dynamic trials recorded.

3.3 Experimental Protocol

To perform the data collection, participants were invited to the University of Ottawa Movement Biomechanics and Analytics Research Laboratory (200 Lees Avenue, 231) for an approximately one-hour data collection session. The study protocol involved three static poses, functional joint movements, and two dynamics movements: walking over a 5-meter distance and drop-vertical

jumps. Both dynamic trials were practiced a minimum of three times, or until the participant felt comfortable with the movement, before recording the trials.

For every trial collected, data were recorded simultaneously using three different systems. The first system was the MB motion capture system including ten infrared optical motion capture cameras recording at 80 Hz (Vantage V5, Vicon, Oxford, UK). The MB system recorded the positions of 37 individual reflective markers placed on anatomical landmarks on the participant's upper and lower limbs, trunk, pelvis, and head. Ten marker clusters, containing four markers each, were also placed on the participant (Appendix D). Data were also collected using a ML motion capture system through seven red-green-blue video cameras also recording at 80 Hz (Vue, Vicon, Oxford, UK). Dynamic body scans were also simultaneously captured using the Move4D high-speed 4D body scanner (Move4D, Instituto de Biomechanica de Valencia, Spain); however, the results of the Move4D data will not be discussed in this paper. Although many previous studies have recorded data at a frequency between 100 – 120 Hz [11], [13], [24], [25], [54], [55], 80 Hz was selected as it is the maximum frequency that allowed for the synchronous data collection of the markers, Theia3D, and Move4D. Four in-ground force plates (AMTI Force and Motion, MA, USA) were aligned in a T-shape to allow for the collection of both a full left-leg walking stride (left heel strike to left heel strike), and a vertical jump where participants land with feet on two different force plates. The force plates recorded at 1200 Hz. The marker-based cameras, ML cameras and force plates were run synchronously through Vicon's Nexus software (Nexus, Vicon, Oxford, UK).

Before participants arrived at the laboratory, both the ML and MB motion capture systems were turned on and allowed to warm up for thirty minutes prior to calibration. For the MB system, some pixels were masked to avoid reflective objects in the laboratory that may cause errors in the data. The masked pixels are consistent across all participants. Next, the MB and ML systems were calibrated, and the volume origin was set to the corner of the centre force plate on the ground for all trials. A new session and participant were created in Nexus following a standardized naming convention. The force plates were zeroed prior to beginning the static poses, and once again after placing the plyometric box used for the drop-vertical jump trials on the first force plate.

When participants arrived at their session, the experimental protocol and the instrumentation used were explained to the participant. Participants were given time to ask questions about the protocol, the systems, or any concerns they might have had. The participants were given time to read through

and sign the consent form. Afterwards, height and mass were measured. It is to be noted that when first informing participants about this study via word-of-mouth, email, or Microsoft Teams, participants were informed that they were to wear athletic shorts and shoes to the data collection. Participants also had the option to wear a sport bra, a tight-fitting shirt, or no shirt. Once in the laboratory, flexible Velcro bands were applied around the participants shanks, thighs, pelvis, torso, head, upper arms, and lower arms. Finally, participants were equipped with 37 individual reflective markers and ten marker clusters (Figure 1).

Figure 1. Human Movement Biomechanics Laboratory cluster marker set [77]

Once participants were equipped with all the reflective markers and understood the experimental protocol, the data collection could begin. Participants completed two one-second static poses: an A-pose and a biker-pose (Figure 2). Next, participants performed functional movement tasks which tested the range of motion of their lower limb joints. To capture the centre of rotation of the hip, participants performed a five-pointed star with their leg straight and torso upright, as well as circumduction of the hip. For the knee and ankle, participants held their balance while holding out the working leg and performed three flexion and extensions of the knee, followed by plantarflexion and dorsiflexion of the ankle. Participants were instructed to take their time and to try their best to hold their balance but were permitted to put their foot down as needed. Finally, a squat was

performed to evaluate range of motion across all joints at once. For the first dynamic task, participants practiced walking over the force plates, until they felt confident walking over the force plates and striking the force plates near with their heel and ensuring their feet only hit a single force plate at a time. Participants walked at a self-selected pace and repeated the walking trials three times. After, a drop-vertical jump off a 30 cm tall plyometric box was demonstrated to the participant. Participants were instructed to keep their arms in front of them as they step off the box with their dominant foot, land with their feet on separate force plates, and then immediately perform a maximum vertical jump. Participants practiced the drop jump a minimum of three times before the three jumping trials were recorded.

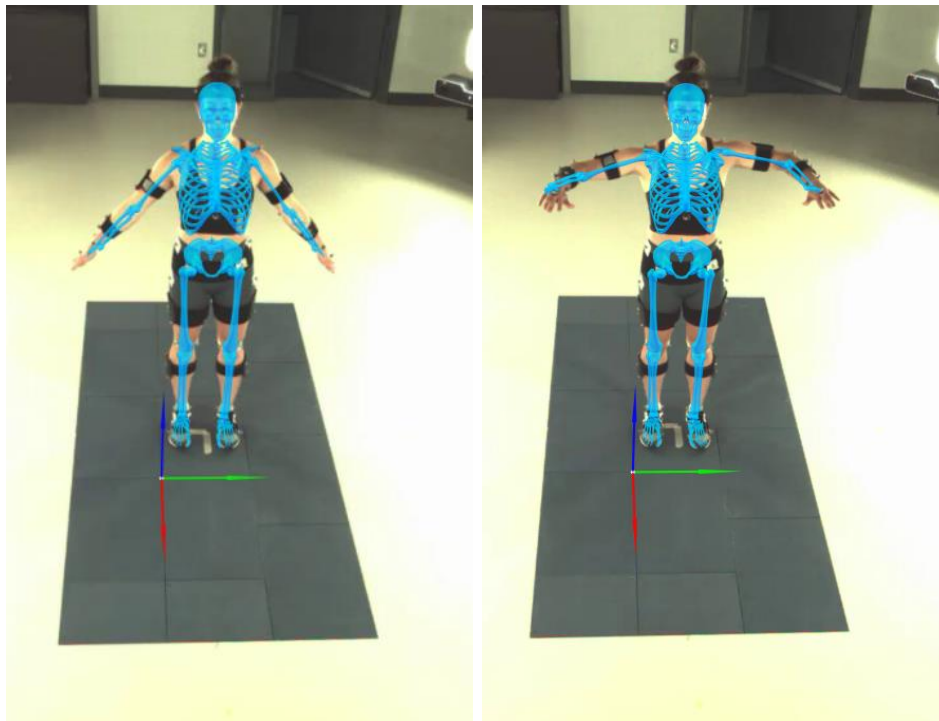


Figure 2. A-Pose and Bikerpose

After all the tasks were completed, the markers were removed from the participants, and they were free to leave the laboratory. All Velcro straps used during the trial were sanitised after every trial. In Nexus, the trials were batch processed using the Theia3D v2025.1.6 (Theia Markerless, Kingston, ON, Canada) software plug-in to obtain the ML human pose estimations. In this study, the Theia3D model that allows for three DOF at the knee was selected. The MB trials were methodically labelled and gap-filled. Gap filling allowed for the estimation of marker positions

during short windows – up to 60 frames – where the marker was out of sight from the cameras. After ensuring no markers were missing or mislabelled, a functional joint estimation pipeline was used to estimate the HJC of each participant based on their functional range of motion trial. Appendix D can be referenced for the naming conventions of the markers used in this study.

3.4 Marker-Based Data Processing

The dynamic walking and jumping trials were trimmed according to events detected by the force plates. The force plate data were down sampled from 1200 Hz to 80 Hz to match the cameras. For walking, only the frames between the first left heel strike on the force plate and the second left heel strike were analysed. For DVJ, the trials were trimmed between the frame the first foot contacted a force plate after dropping from the box to the takeoff frame of the maximum vertical jump. After trimming the data, the MB data were imported to Visual3D (HAS Motion, Kingston, Ontario), where a pipeline batch processed the trials that extracted to a MATLAB file (.mat) the location of the markers, joint angles, and the 4x4 segment transformation matrices which provide information on the position of the joint centre and the segment orientations. ML and MB data share the same global coordinate system, defined as the corner of the second force plate, and the local coordinate systems (LCS) of each segment were rotated to match the general direction of the LCS defined by Theia3D where the x-axis points to the direction of movement, the y-axis points upward, and the z-axis points to the right side of the participant.

The trial data were processed using four near identical pipelines, which differed only in the MB model applied. The four models created were all seven-segment models of the lower limbs that varied only in the definition of the HJC. Previous studies comparing sex-based kinematics using MB motion capture either employed a full-body 14- or 15-segment model like Bruening et al. [13], or a 7- or 8-segment lower-limb model like Chen et al. [72]. The joint centre definition for the knees and ankles was very consistent across the literature, with almost all studies using the midpoint between lateral and medial joint markers. However, HJC definition varied. Bruening et al. employed the Harrington et al. regression equations [13], whereas other studies estimated the hip using functional methods [55], or the Bell method [53]. The present study focuses on lower limb kinematics. As a result, a 7-segment model composed of the feet, shanks, thighs and pelvis was deemed sufficient. The section 3.7 Model Definition Sensitivity Analysis contains further details on the hip definitions for each model. Figure 3 illustrates the MB model used in the analysis. Each

model also included a virtual foot segment that corrected the orientation of the foot segment to reflect the actual ankle angle. In Visual3D, the foot of the MB skeletal model was first defined using the ankle joint centre, lateral ankle marker, and markers placed on the first and fifth metatarsals. As a result, the foot was positioned to have 12-20 degrees of plantarflexion during the static trials on a flat surface. This was corrected by defining virtual markers for the toe markers that have the same coordinates in the x- and z-planes, but the y-plane coordinates would be equal to the height of the ankle joint centre. This resulted in a neutral angle of the ankle during static position and rotated the ankle angle during dynamic trials to be within acceptable ranges.

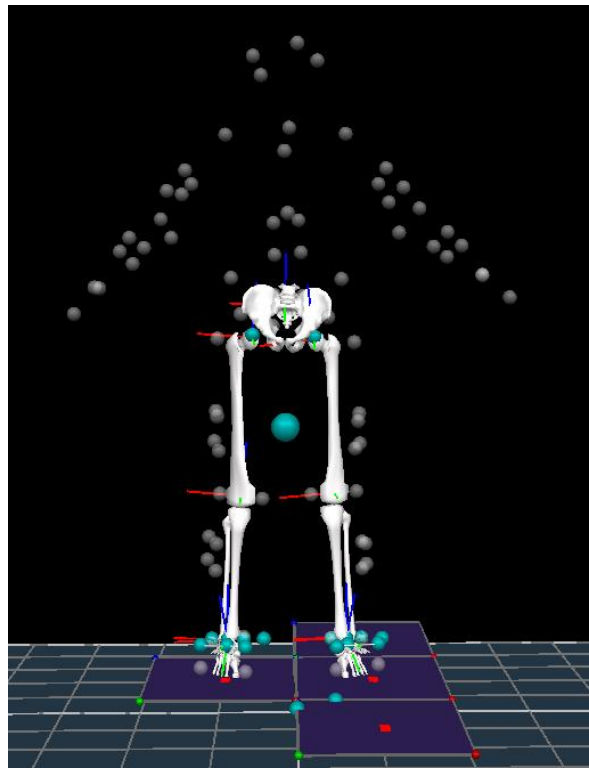


Figure 3. Harrington Hip Joint MB Model

After extracting the relevant information from Visual3D, the MB 4x4 segment transformation matrices were imported to a MATLAB code that calculated the joint angles. Each transformation matrix contained information on the orientation and position of a given segment relative to the laboratory coordinate system. Pelvis kinematics were computed by expressing the pelvis transformation matrix relative to the global reference frame. A fixed rotation was first applied to account for differences between the global reference frame and the LCS, after which pelvis orientation was decomposed into Euler angles using an XYZ rotation sequence corresponding to pelvic tilt, list, and axial rotation. The remaining joint angles were then calculated by determining

the relative angle between the segment in question and its parent segment (pelvis-thigh, thigh-shank, shank-foot). These relative rotation matrices were subsequently decomposed into joint angles using an XYZ Euler rotation sequence, with sign conventions applied to ensure consistency with anatomical definitions and left-right symmetry.

Finally, the joint angles were filtered using a low-pass second-order Butterworth filter with a cutoff frequency of 6 Hz, implemented using zero-phase forward-backward (i.e., dual-pass) filtering to avoid phase distortion. These filter parameters are common for filtering kinematic walking data and reflects similar filtering performed on the data by Bruening et al., Mullerpatan et al., and Takano et al. [13], [24], [53], [85]. Joint angles extracted directly from Visual3D were not used in the primary analysis, as recalculation from the transformation matrices was necessary to allow for subsequent rotation and alignment of local coordinate systems between MB and ML data. However, Visual3D-derived joint angles were used to verify the accuracy of the MATLAB-based joint angle calculations.

3.5 Markerless Data Processing

ML motion capture data were processed using the same procedure as the MB data. Both MB and ML data were collected using the same frame rate of 80 Hz, as a result the ML data were trimmed to match the frames used in each MB trial. ML trials were then batch processed in Visual3D. During this process, a Theia3D skeletal model was automatically applied to each trial, and segment-level kinematics were computed by Visual3D.

Following processing, the 4×4 homogeneous transformation matrices describing the local coordinate systems of each segment, as well as the associated joint angles, were exported from Visual3D to MATLAB. To ensure consistency between MB and ML analyses, joint angles were recalculated in MATLAB directly from the exported 4×4 transformation matrices using the same custom code and rotation sequences applied to the MB data. This approach ensured that both datasets were subject to identical joint angle computation, filtering, and sign conventions.

3.6 Excluded Trials

Of the 168 dynamic trials collected (28 participants x 3 walking trials x 3 jumping trials), two walking trials and three drop jump trials were excluded from the dynamic trials due to prematurely stopped recordings: p01_walk01 and p08_walk01. Although some trials were found to have

significantly greater differences between the motion capture systems, all trials that capture the full frames of the dynamic movements were included in the analysis.

3.7 Model Definition Sensitivity Analysis

A sensitivity analysis was conducted to systematically determine the best MB skeletal model to use in comparisons with the Theia3D model. The MB skeletal models compared shared the same pelvis, knee, and ankle definitions. The pelvis was defined using the left and right anterior superior iliac spine and posterior superior iliac spine markers and used the Visual3D composite pelvis, the knee was defined as the midpoint between the lateral and medial knee markers, whereas the ankle was defined as the midpoint between the lateral and medial ankle markers. First, four different MB models with different hip definitions were defined and applied to all static and dynamic trials to determine which method had the best agreement with the Theia3D model in joint centre estimations and kinematics. The hip definitions compared included functional HJC, Bell, Harrington pelvis-only, and Harrington pelvis-and-leg. Table 2 contains the equations used for each regression method.

Table 2. Regression equations for hip joint centre estimation

Method	ML	AP	Axial
Bell [48], [86]	$0.36 * \text{ASIS_Distance}$	$-0.19 * \text{ASIS_Distance} + (0.5 * \text{RPV_Depth} - \text{Target_Radius_ASIS})$	$-0.3 * \text{ASIS_Distance}$
Harrington pelvis-only [48], [65]	$0.33 * \text{ASIS_Distance} + 0.0073$	$-0.24 * \text{RPV_Depth} - 0.0099 + (0.5 * \text{RPV_Depth} - \text{Target_Radius_ASIS})$	$-0.30 * \text{ASIS_Distance} - 0.0109$
Harrington pelvis-and-leg [48], [65]	$0.28 * \text{RPV_Depth} + 0.16 * \text{ASIS_Distance} + 0.0079$	$-0.24 * \text{RPV_Depth} - 0.0099 + (0.5 * \text{RPV_Depth} - \text{Target_Radius_ASIS})$	$-0.16 * \text{ASIS_Distance} - 0.04 * \text{Leg_Length} - 0.0071$

After comparing HJC estimation methods, additional models were created to compare the effects of kinematic constraints. These included Harrington pelvis-only no kinematic constraints,

Harrington pelvis-only with kinematic constraints (6 DoF pelvis; 3 DoF hip; 3 DoF knee; 6 DoF Ankle), and Harrington pelvis-only with kinematic constraints (6 DoF pelvis; 3 DoF hip, 6 DoF knee; 6 DoF Ankle). For the kinematic constraints only the translational DoF were constrained; the knee was constrained to ± 7 mm in antero-posterior direction and ± 3 mm in both the medio-lateral and axial directions [87], [88], [89]. The ankle was constrained to ± 4 mm in the antero-posterior direction and ± 2 mm in the medio-lateral and axial directions [90]. After processing all the data using each model, the HJC were compared to see which had the best agreement to the ML HJC, and which model had the best fit to the ML kinematic data. The performance of the models in both metrics determined the model used for the remaining analyses.

3.8 Segment Alignment

Once the optimal model was identified, different alignment methods were compared: translating the MB joint centres to the position of the ML joint centres, rotating the MB LCSs to match the ML LCSs during the static pose (the same rotation matrix applied to all trials), and rotating the MB LCSs to match the ML LCSs according to the average rotation matrix to align the LCSs across the frames of a single trial (different rotation matrix applied to each trial). The kinematics from each alignment method were compared to determine which reduced the most amount of bias to ML data.

Once the skeletal model was corrected, it was then possible to extract the 4x4 transformation matrices of each joint. These 4x4 transformation matrices were then used to calculate the joint angles at each frame. The pelvis angle was calculated by determining the relative angles between the pelvis LCS and the laboratory's global coordinate system, whereas the hip and subsequent joints were calculated by determining the relative angle between the hip LCS and its parent segment's (i.e., pelvis) LCS. After determining the angles for each joint, root mean squared difference (RMSD) and bias (i.e., bias = MB – ML) of the raw kinematic data were calculated. Finally, the Bland-Altman limits of agreement (LOA) were computed to assess the agreement between the two approaches.

In addition to the raw kinematic data, the 4x4 matrices from both methods were used to calculate a correction matrix that would align the MB LCS to the ML LCS. Figure 4 presents the LCS before and after alignment during the static biker pose. Perfect alignment of the systems can be achieved in the static trial because the relative angle between the MB and ML coordinate systems do not

change. However, during the dynamic trials, the orientation of the LCS is changing, as a result the position of the MB LCS relative to the ML LCS is not constant because of the system measurement differences. Only a single correction matrix can be applied per joint and per trial, which was calculated by taking the average correction matrix from each frame. Once all the correction matrices were applied to the MB 4x4 transformation matrices and they were rotated to be more aligned with the ML data, the joint angles were determined using the same method of measuring the relative angles between LCSs. RMSD, bias, Pearson's correlation coefficients and Bland-Altman LOAs were then calculated between the aligned MB data and the raw ML data. For the aligned data, the average kurtosis results were 2.18, 2.39, 3.41, and 2.65 for the pelvis, hip, knee, and ankle respectively. These values were considered sufficiently close to a mesokurtic distribution to assume normality.

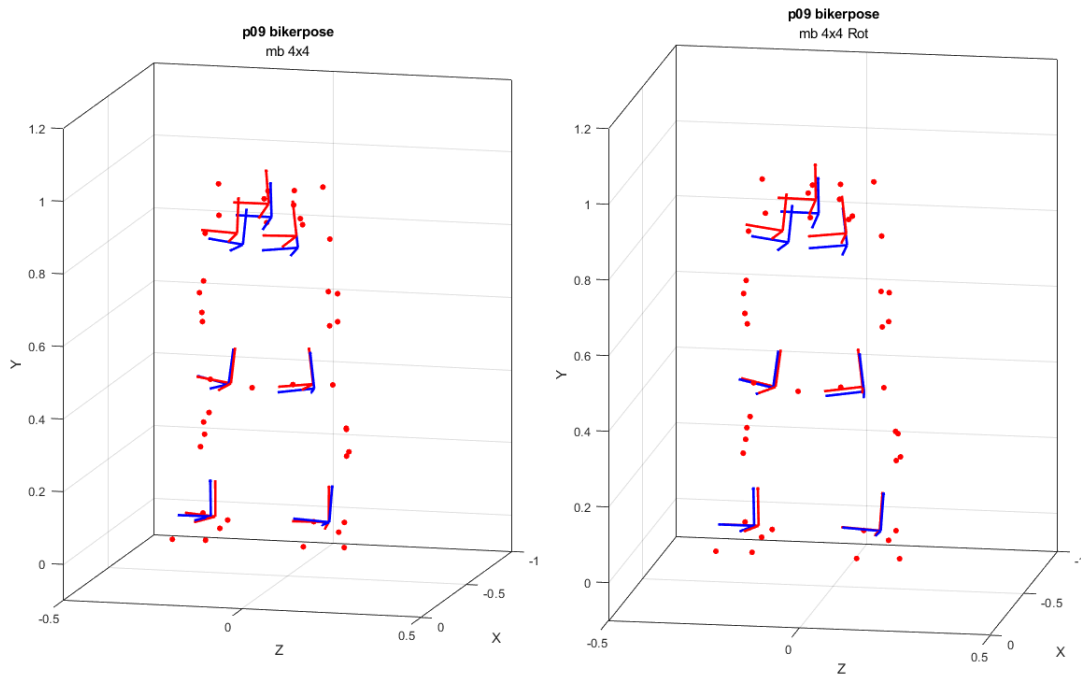


Figure 4 - Local coordinate systems before alignment (left) and after alignment (right). MB and ML data are represented by red and blue.

3.9 Statistical Analysis

Statistical analysis of the data collected in this study was separated into three parts. First, several metrics were computed from the joint centre data. Anthropometric measurements were calculated between joint centres to obtain pelvis-to-hip distance, inter-hip width, thigh length, and shank length. These values were first compared to previously reported values in the literature to determine whether the present data identified similar sex-based differences between male and female participants. Thus, the comparison of anthropometric measurements will aid in answering the first objective as listed in Section 1.2.1 Objective 1. In addition, RMSD and bias between the MB and ML joint centres were computed to quantify spatial agreement between systems, which serves to answer the second objective listed in Section 1.2.2 Objective 2.

The anthropometric data were then compared within each trial and within each participant to evaluate the consistency and repeatability of each motion capture method during data collection. For within-trial comparisons, the coefficient of variation (CV) and standard deviation (SD) were calculated for each anthropometric measurement to assess frame-to-frame stability, with a CV of less than 5% interpreted as excellent stability [91]. For within-participant comparisons, the mean anthropometric measurement was calculated for each trial. Intraclass correlation coefficients (ICC), CVs, and SDs were then calculated for each participant across trials to assess relative and absolute reliability, subsequently results were averaged across all 28 participants to provide an overall estimate of method consistency and repeatability [91], [92].

Next, linear mixed models (LMMs) were used to assess the effects of sex, method, and their interaction on anthropometric and kinematic outcomes while accounting for repeated measurements within participants. Separate LMMs were constructed for each dependent variable, including RMSD, bias, pelvis-to-hip distance, inter-hip width, thigh length, and shank length.

For RMSD and bias, the following model structure was applied:

$$(DV \sim Sex + (I|Participant)) \quad \text{Equation 1}$$

For all other anthropometric and kinematic variables, the following model structure was used:

$$(DV \sim Sex + Method + Sex * Method + (I|Participant)) \quad \text{Equation 2}$$

In these models, sex (male, female) and method (MB, ML) were treated as fixed effects, while participant was included as a random intercept to account for repeated measured and inter-

individual variability. The inclusion of the Sex x Method interaction allowed assessment of whether the effect of the measurement method differed between male and female participants.

The LMMs provided estimates of fixed effects, associated confidence intervals, and statistical significance, allowing for the determination of whether systematic differences existed between sexes, between motion capture methods, or whether sex-specific differences were method dependent. When significant main effects or interactions were identified, post-hoc pairwise comparisons were performed using a simplified LMM that evaluated fixed effects of method or sex to assist in interpreting statistically significant differences between subgroups. Cohen's d effect sizes were calculated to quantify the magnitude and direction of these differences, where effect size of 0.2 is small, 0.5 is medium, and greater than 0.8 is large [93]. An effect size of 0.5 indicates that the difference between groups equals half of a standard deviation [93]. The LMMs contributed to answering the first objective as listed in Section 1.2 Objectives and Hypotheses.

The model residuals were examined to assess the normality of the data. Shapiro-Wilk tests were applied to the residuals of each LMM [94]. Variables that failed the normality test were further evaluated using Q-Q plots. Visual inspection of the Q-Q plots indicated only minor deviations from normality, primarily at the distribution tails, while the central portion of the residual distributions closely followed the theoretical normal line [95]. Based on these observations, the residuals were considered sufficiently normal to proceed with parametric analyses. The Q-Q plots created to evaluate normality are presented in Appendix G.

Using the anthropometric measurements, a linear mixed model (LMM) was also applied to determine the effects of sex, method, and their interaction. Additionally, the RMSD between the ML and MB joint centres during each trial were calculated. These values help evaluate the overall agreement between the systems and allow RMSD to be compared with values found in the literature. A LMM was also applied to the RMSD to determine if significant differences exist between the sexes (Objective 1). In addition to comparing the anthropometrics, offset between the joint centres was also measured. The offset was calculated on the average distance between the ML and MB joint centres in the three planes separately to gain an understanding of which planes have better agreement. The total offset between the joint centres was also calculated by determining the Euclidian distance between the joint coordinates.

For the kinematic data, the RMSD and bias were computed across all dynamic trials for each joint and plane of motion. Mean RMSD values across all participants were used to facilitate comparison with values reported in prior validation studies of ML motion capture systems.

RMSD values were additionally segregated by sex and analysed using LMMs to determine whether agreement between the systems differed between males and females. RMSD and bias data were then analysed using LMMs to evaluate the main effect of sex, following the model structure described in Equation 1. Discrete kinematic metrics including ROM, minimum angle, maximum angle, and initial contact angle were extracted for each joint and plane of motion. These metrics were selected based on previously reported kinematic descriptors [13], [24], [73]. Each metric was then analysed using LMMs to evaluate the main effects of sex, method, and their interaction following the model formula described in Equation 2.

Agreement between the two motion capture systems was further assessed using Bland-Altman limits of agreement (LOA). The Bland-Altman analysis quantified the mean difference between the MB and ML measurements and estimated the limits within which 95% of the differences between the systems are expected to lie [96]. This analysis assumed that both measurement methods exhibit similar precision, that the precision remains constant across trials, and that bias between the methods is consistent [97]. To further validate the normality assumption underlying the Bland-Altman analysis, the kurtosis of the distributions of differences between MB and ML measurements was assessed. A kurtosis value of approximately 3 was interpreted as a mesokurtic distribution, supporting the validity of the LOA estimates [98]. The bias and LOA were then compared before and after LCS alignment to quantify the increase in agreement between MB and ML data, which served to answer the second objective listed in 1.2.2 Objective 2.

4. Results

Prior to running statistical tests, sensitivity analyses were performed to determine what model had the most agreement to the Theia3D v2025.1.6 model with 3 DOF at the knee; this included which MB hip joint had the greatest agreement with the ML HJC, and whether adding kinematic constraints to the MB model decreased RMSD. The complete results of the sensitivity analysis can be found in Appendix E. The results of the sensitivity analysis established that the model defined the hip using the Harrington method had the lowest mean offset between joint centres of 4.07 ± 1.29

mm compared to 4.91 ± 1.60 mm using the functional hip joint and 5.79 ± 1.53 mm using the Bell regression equations. Additionally, aligning the MB LCSs to the ML LCS by applying a mean transformation matrix saw a reduction of kinematic RMSD from $5.61 \pm 4.11^\circ$ to $3.12 \pm 1.88^\circ$. Based on the results of the sensitivity analysis, the MB model selected for all subsequent analyses implemented the Harrington HJC and pelvis definition, and model kinematics were not constrained.

4.1 Anthropometric Measurements and Joint Centre Comparison

During the first step of the analysis, a skeletal model in Visual3D was scaled using a static A-pose trial. The scaled model was then applied to all other trials from the same participant. Doing so ensures that the joint centres are at the same position relative to their segment across all trials. This is assuming the marker clusters used to define the segment orientation have not moved. Theia3D creates and scales a new skeletal model for each trial. The outcomes of the kinematic analyses are largely impacted by the position of the joint centres and the motion capture method's ability to accurately define the joint centre across multiple trials. These joint centres were then used to define anthropometric measurements such as the pelvis-to-hip distance, inter-hip width, femur length and segment lengths. To evaluate the consistency of each motion capture system, within-trial and between-trial comparisons of anthropometric measurements were performed. These results can be found in Appendix D.

Next, how well each motion capture approach can capture the anthropometric measurements for specific populations was evaluated. Table 3 presents a comparison of mean segment length measurements (mean $\pm$ SD) between MB and ML motion capture methods for male and female participants, as well as the results of the LMM.

For pelvis-to-hip distance, mean values for males were 12.6 cm for both methods. Females had a shorter overall pelvis-to-hip distance, and the ML measurements were less than MB. The inter-hip widths between methods were very comparable, measuring approximately 17 cm in males and 16.3 cm in females. Thigh lengths saw the most amount of deviation between motion capture approaches. MB measured longer overall femurs compared to ML. Shank lengths were almost identical across methods with average tibial lengths of 40.1 cm and 36.7 cm for males and females, respectively.

From the LMM, significant ($p < 0.001$) main effects of sex were observed for pelvis-to-hip distance, hip width, thigh length, and shank length. Significant main effects of methods were observed for pelvis-to-hip distance and thigh length. Hip width and shank length did not show a significant method effect. Significant interaction effects were observed for the pelvis-to-hip distance, hip width, and thigh length. No interaction effect was observed for shank length.

Post-hoc analyses examining sex differences within each method showed that both MB and ML approaches found significant differences in hip width and shank length. Only the MB data found significant differences in thigh length, whereas only the ML data had significant difference in pelvis-to-hip distance. Post-hoc analyses looking at the method differences within each sex found significant differences for pelvis-to-hip distance and thigh length in female participants, and for hip width and thigh lengths in males. No significant method differences are observed for shank length in either sex.

Table 3. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from segment lengths. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size.

Segment	Pelvis-to-Hip Distance (cm)	Hip Width (cm)	Thigh Length (cm)	Shank Length (cm)
Male				
MB	12.6 ± 0.8	17.1 ± 1.1	42.2 ± 2.8	40.1 ± 2.8
ML	12.6 ± 0.4	17.0 ± 0.9	39.3 ± 2.3	40.1 ± 2.5
Female				
MB	12.1 ± 0.8	16.3 ± 1.0	40.3 ± 2.1	36.6 ± 1.8
ML	11.7 ± 0.5	16.4 ± 1.0	38.3 ± 1.9	36.8 ± 1.7
P-values				
Sex	0.01	0.004	0.006	< 0.001
Method	< 0.001	0.127	< 0.001	0.311
Interaction	< 0.001	0.004	< 0.001	0.354
Post-Hoc: Sex Differences Within Method, p-value (effect size)				
MB	0.084 (0.59)	0.040 (0.72)	0.021 (0.68)	< 0.001 (1.23)
ML	< 0.001 (1.98)	< 0.001 (1.40)	0.122 (0.44)	< 0.001 (1.28)
Post-Hoc: Method Differences Within Sex, p-value (effect size)				

Female	< 0.001 (0.62)	0.188 (-0.16)	< 0.001 (1.00)	0.311 (-0.11)
Male	0.418 (0.07)	< 0.001 (-0.56)	< 0.001 (1.12)	0.767 (0.02)

The results of the joint offsets are presented in Table 4. A negative offset signifies that the ML joint centre is lateral, anterior, or proximal to the MB joint centre. For the pelvis, the mean medial and anterior offsets were less than 1 cm. The main source of difference between pelvic joint centres appeared in the axial plane, with an average offset of 2.94 cm, indicating that the MB method consistently estimates the pelvis centre as proximal to the ML pelvis centre. The hip also had very little offset in the medial direction. However, greater than 2 cm of HJC offset was measured in both antero-posterior planes and axial planes, indicating the ML approach estimates the hip posterior and distal to the MB hip. For the knee less than 0.5 cm of differences was observed in each direction, resulting in a total offset of 1.39 cm. The ankle saw a greater difference than the knee, with less than 0.71 cm of difference measured in each direction resulting in an average offset of 1.46 cm. Figure 5 presents the offset between joint centres in a female participant.

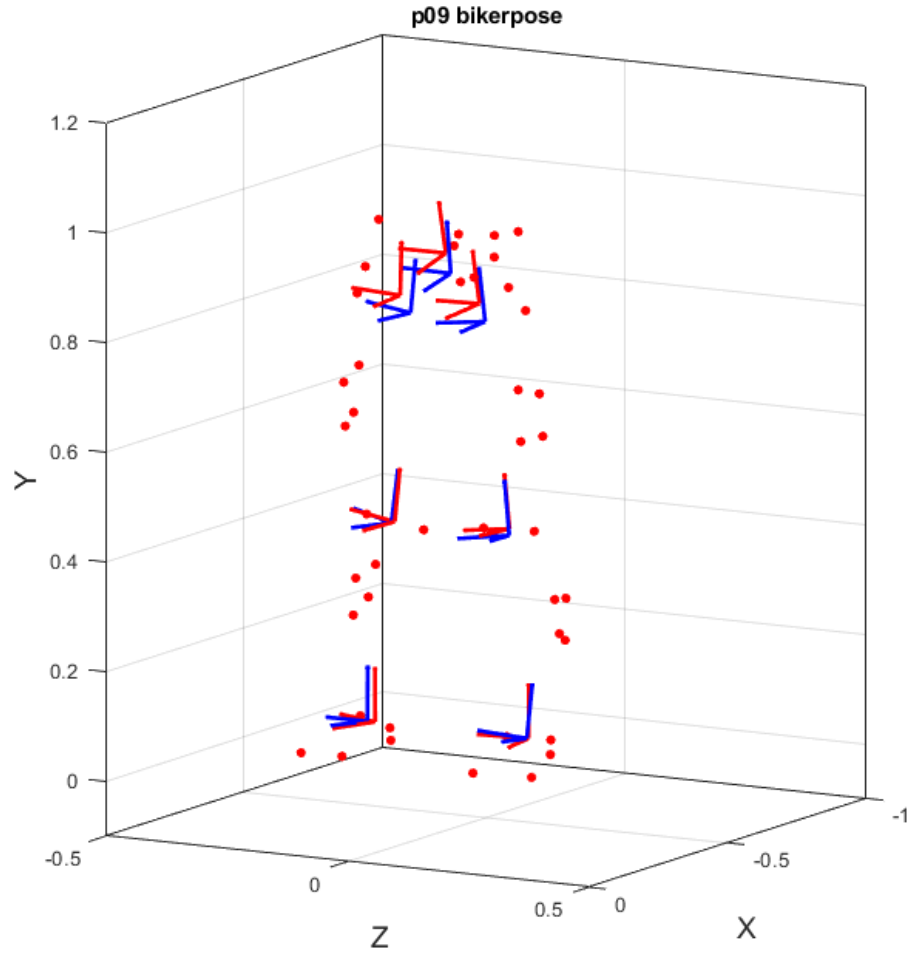


Figure 5. Joint centre offset between raw MB and ML data in one female participant

Table 4. Mean offset between MB and ML joint centres across all participants and trials

Joint	Anterior (cm)	Medial (cm)	Axial (cm)	Total Offset (cm)
Pelvis	0.86 ± 1.1	0.01 ± 0.4	2.94 ± 1.6	3.37 ± 1.4
Hip	2.26 ± 1.4	-0.30 ± 0.9	2.69 ± 1.7	4.02 ± 1.3
Knee	0.46 ± 1.1	-0.46 ± 0.3	0.16 ± 0.8	1.39 ± 0.7
Ankle	0.65 ± 0.8	-0.71 ± 0.3	0.52 ± 0.8	1.46 ± 0.7

In addition to comparing the mean offsets in each plane, the total offsets between the joint centres for each trial were used to create a LMM that evaluated whether the RMSD between MB and ML joint centres is affected by sex. Table 5 presents the p-values, as well as the Cohen's d effect size for each joint, where negative effect size indicates that female participants had larger joint offsets

than male. Statistically significant differences and large negative effect sizes between sexes were observed for the pelvis, right hip, and right ankle. The left hip was the only joint to show a significant difference and a positive effect size, although the effect size was small ($d = 0.38$). No statistically significant differences were observed for the right knee, left knee, or left ankle, despite large negative effect sizes reported for these joints ($d < -0.92$).

Table 5. p-values and Cohen's d effect sizes for comparisons of differences in joint centre offsets between sexes. Orange and yellow indicates that a very significant ($p < 0.001$) or a significant ($p < 0.05$) difference was found.

Joint	P value	Effect Size
Pelvis	0.03	-1.19
Right Hip	0.01	-1.22
Left Hip	0.01	0.38
Right Knee	0.27	-0.92
Left Knee	0.50	-1.07
Right Ankle	< 0.001	-1.44
Left Ankle	0.12	-1.14

4.2 Segment Alignment

Table 6 and Table 7 presents the mean RMSD, bias, and LOA across the walking and DVJ trials of male and female participants for both the raw data and the comparison of the MB data rotated to align with the ML LCS. Here the results are presented as “Bias $\pm$ LOA”. A smaller absolute bias indicates a smaller systematic difference between systems, while a smaller RMSD and narrower LOA indicate lower overall and trial-to-trial disagreement, respectively.

Aligning the MB LCSs to ML data resulted in a decrease of RMSD and bias across most joints and planes of motion in both sexes, although magnitudes varied. In males, pelvic list, tilt, knee adduction, knee rotation, and ankle flexion showed relatively small changes following alignment, with RMSD reductions of less than 1° . Hip flexion, hip adduction, hip rotation, and knee flexion RMSD decreased by $1\text{--}2^\circ$. In females, the largest raw differences occurred in pelvis list, hip rotation, and ankle rotation, with alignment reducing RMSD substantially. Hip rotation RMSD decreased from 8.73° to 3.48° and ankle rotation from 16.14° to 6.11° . Ankle eversion RMSD increased slightly in females after alignment to 6.87° , but bias decreased from -5.29° to -1.59° .

For all other joint, RMSD and bias decreased after alignment, with bias often approaching zero, reflecting a reduction in systematic differences between MB and ML systems.

A similar pattern was observed during DVJ. Alignment reduced RMSD for almost all joints, with the largest improvements generally occurring in the pelvis, hip, and ankle rotation. For example, pelvic tilt RMSD decreased from 9.04° to 5.01° in males and from 10.07° to 4.08° in females. Hip rotation RMSD decreased from 7.62° to 4.65° in males and from 8.70° to 4.32° in females. The largest reduction in the DVJ was observed for female ankle rotation, where RMSD decreased from 11.39° to 5.28°. Alignment also substantially reduced bias for several variables, including knee rotation and ankle adduction

Table 6. RMSD (°), bias (°), and LOA (°) between MB and ML kinematics in raw data and aligned MB data in males and females.

	Raw – Male		Aligned – Male		Raw – Female		Aligned – Female	
Joint	RMSD	Bias ± LOA	RMSD	Bias ± LOA	RMSD	Bias ± LOA	RMSD	Bias ± LOA
Pelvic List	1.74	-0.94 ± 2.9	1.47	0.02 ± 2.9	3.48	3.06 ± 3.3	1.66	0.00 ± 3.3
Pelvis Tilt	2.06	0.59 ± 3.9	1.97	0.01 ± 3.9	2.31	0.31 ± 4.5	2.29	0.01 ± 4.5
Pelvic Rotation	0.77	-0.10 ± 1.5	0.73	-0.04 ± 1.4	1.70	-1.22 ± 2.3	1.44	-0.74 ± 2.4
Hip Flexion	4.57	-3.69 ± 5.3	2.76	-0.01 ± 5.5	2.97	-1.48 ± 5.1	2.84	-0.12 ± 5.6
Hip Adduction	2.25	-0.32 ± 4.4	1.80	-0.38 ± 3.5	2.33	1.50 ± 3.5	1.25	-0.27 ± 2.4
Hip Rotation	3.20	1.40 ± 5.7	2.92	-0.39 ± 5.7	8.73	7.92 ± 7.3	3.48	0.15 ± 6.9
Knee Flexion	2.36	-1.65 ± 3.3	1.61	0.12 ± 3.2	3.99	-3.48 ± 3.3	1.85	0.03 ± 3.6
Knee Valgus	1.24	0.89 ± 1.7	1.42	0.10 ± 2.8	2.02	1.39 ± 2.9	1.46	0.29 ± 2.8
Knee Rotation	4.18	2.89 ± 6.0	2.95	0.38 ± 5.8	2.31	0.71 ± 4.3	2.31	0.10 ± 4.5
Ankle Flexion	1.83	1.37 ± 2.4	1.18	0.52 ± 2.1	1.58	0.44 ± 3.0	1.31	0.68 ± 2.2

Ankle Everson	5.72	-5.38 ± 3.9	5.16	-1.11 ± 9.9	5.87	-5.29 ± 5.1	6.87	-1.59 ± 13.2
Ankle Rotation	12.71	-11.75 ± 9.5	4.94	-1.95 ± 9.0	16.14	-15.02 ± 11.7	6.11	-2.82 ± 10.7

Table 7. *RMSD (°), bias(°), and LOA(°) between MB and ML kinematics in raw data and aligned MB data in males and females.*

Joint	Raw – Male		Aligned – Male		Raw – Female		Aligned – Female	
	RMSD	Bias ± LOA	RMSD	Bias ± LOA	RMSD	Bias ± LOA	RMSD	Bias ± LOA
Pelvic List	1.74	-0.94 ± 2.9	1.47	0.02 ± 2.9	3.48	3.06 ± 3.3	1.66	0.00 ± 3.3
Pelvis Tilt	2.06	0.59 ± 3.9	1.97	0.01 ± 3.9	2.31	0.31 ± 4.5	2.29	0.01 ± 4.5
Pelvic Rotation	0.77	-0.10 ± 1.5	0.73	-0.04 ± 1.4	1.70	-1.22 ± 2.3	1.44	-0.74 ± 2.4
Hip Flexion	4.57	-3.69 ± 5.3	2.76	-0.01 ± 5.5	2.97	-1.48 ± 5.1	2.84	-0.12 ± 5.6
Hip Adduction	2.25	-0.32 ± 4.4	1.80	-0.38 ± 3.5	2.33	1.50 ± 3.5	1.25	-0.27 ± 2.4
Hip Rotation	3.20	1.40 ± 5.7	2.92	-0.39 ± 5.7	8.73	7.92 ± 7.3	3.48	0.15 ± 6.9
Knee Flexion	2.36	-1.65 ± 3.3	1.61	0.12 ± 3.2	3.99	-3.48 ± 3.3	1.85	0.03 ± 3.6
Knee Valgus	1.24	0.89 ± 1.7	1.42	0.10 ± 2.8	2.02	1.39 ± 2.9	1.46	0.29 ± 2.8
Knee Rotation	4.18	2.89 ± 6.0	2.95	0.38 ± 5.8	2.31	0.71 ± 4.3	2.31	0.10 ± 4.5
Ankle Flexion	1.83	1.37 ± 2.4	1.18	0.52 ± 2.1	1.58	0.44 ± 3.0	1.31	0.68 ± 2.2
Ankle Everson	5.72	-5.38 ± 3.9	5.16	-1.11 ± 9.9	5.87	-5.29 ± 5.1	6.87	-1.59 ± 13.2
Ankle Rotation	12.71	-11.75 ± 9.5	4.94	-1.95 ± 9.0	16.14	-15.02 ± 11.7	6.11	-2.82 ± 10.7

Figure 6 and Figure 7 present the mean waveforms, before and after alignment, across all participants and the Pearson correlation coefficient between the MB and ML approaches for the left lower limb joints during walking. Alignment of the LCSs substantially reduced the RMSD and bias between the MB and ML kinematics; however, Pearson correlation coefficients showed minimal change. Hip flexion, knee flexion, and ankle flexion had very strong correlation between the motion capture systems with correlation coefficients between 0.98 and 1.0. Pelvis rotation and hip adduction had strong correlation with a value of 0.87, whereas knee adduction had a lower correlation of 0.76. A medium correlation was observed for pelvis list and ankle inversion with coefficients of 0.60 and 0.57, respectively. Correlation coefficients less than 0.5 suggest poor agreement between systems and was observed for hip rotation, knee rotation, and ankle rotation [99]. Additionally, both pelvic tilt, hip rotation, knee rotation and ankle rotation saw negative correlation which indicate an inverse relationship between the joint angles.

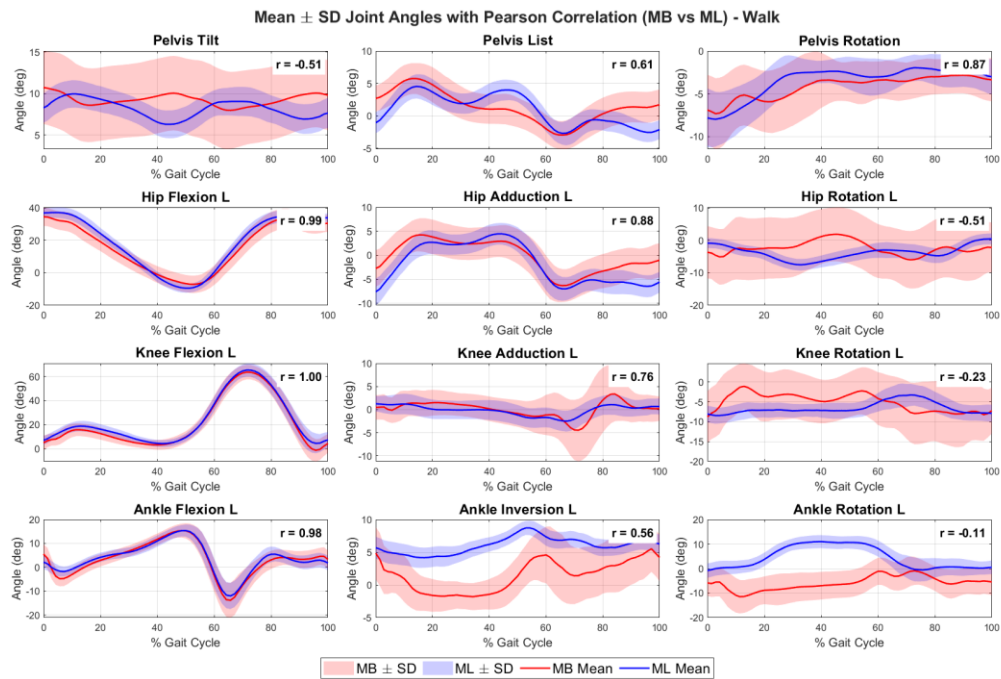


Figure 6. Mean kinematic waveforms of left leg during the walk for raw MB and ML data across all participants

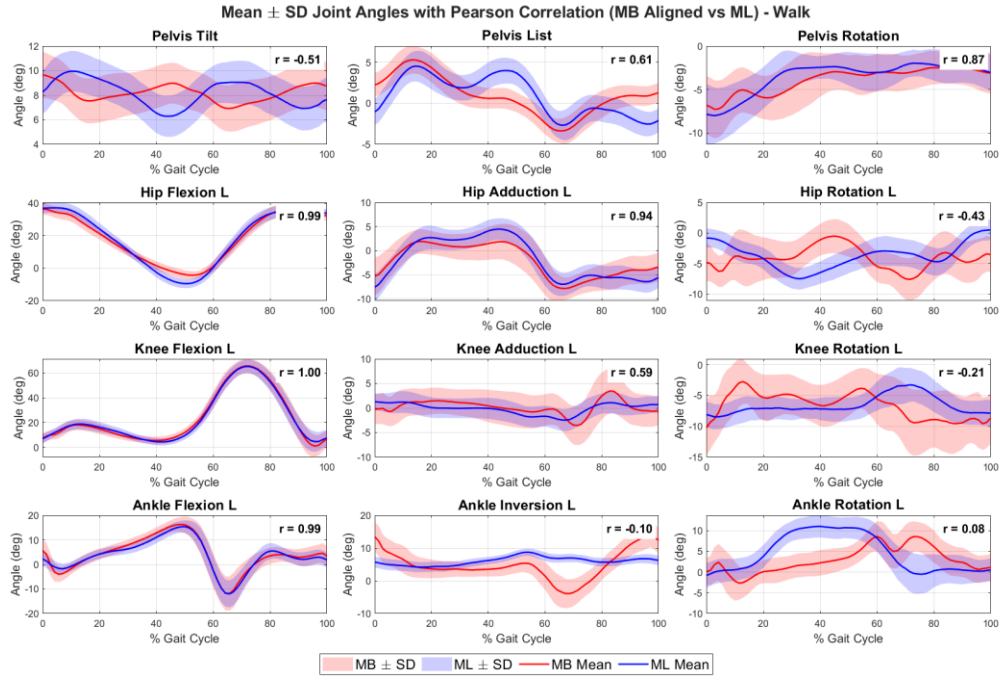


Figure 7. Mean kinematic waveforms of left leg during the walk for aligned MB and ML data across all participants

4.3 Kinematic Comparison – Walking Trials

The next step in the analysis of the kinematics involved comparison of the joint angles at key events when separated into groups. This analysis serves to determine if both motion capture systems identify the same sex-based kinematic differences. First, different joint angle and kinematic metrics were extracted including the ROM, minimum angle, maximum angle, and initial contact angle. Because gait is often not symmetric, the left and right joints were kept separate. This also helped to determine if sex-based differences were measured in both legs. Once these metrics were obtained, the mean values of each metric were measured for each sex, linear mixed models were created for dependent variable ($DV \sim \text{Sex} + \text{Method} + \text{Sex} * \text{Method} + (1|\text{Participant})$). The linear mixed model was used to determine if there was a significant main effect of sex, method, or their interaction.

Table 8 presents the descriptive values of the pelvis angles during the walk, and the p-values obtained from the linear mixed model. In the sagittal plane, pelvis tilt showed no significant main effects of sex. However, sex-within-method post-hoc analyses revealed that sex-based differences in the ML data were found, with females exhibiting larger pelvis tilt ROM of 5.15° and a smaller minimum angle of 5.27° anterior tilt. Significant main effects of method were found for minimum

angle, maximum angle, and ROM, with post-hoc analyses indicating significant differences for most metrics in both male and female groups. In female participants, MB measured smaller ROM than ML, but the opposite is true in male participants. Interaction effects were also found for all metrics, indicating that method-related differences were sex-dependent. Figure 8a illustrates the mean pelvis angle waveforms grouped by sex and method.

In the frontal plane, significant main effects of sex were observed for pelvic list ROM, minimum angle, and maximum angle, with females having larger amplitudes compared to males. These sex-based ROM differences were evident in both systems, as supported by significant post hoc comparisons within MB and ML. But ML was the only method to find sex-based differences in minimum and maximum angles. Additionally, the minimum angle was the only metric that found significant method difference, with method-within-sex comparisons revealing that ML had a significant smaller minimum angle in females, whereas this angle was significantly larger in males. Sex-by-method interactions were also present for all pelvic list metrics, suggesting that the magnitude of these differences varied between MB and ML. In females, the pelvic list metrics were very comparable between methods, as a result no significant differences were observed between methods for ROM and maximum angle; however, the male metrics found very significant differences across all pelvic list metrics. Figure 8b shows the mean pelvis tilt angles for each group.

For pelvis rotation, no significant main effect of sex was observed for pelvis rotation metrics. However, significant sex-by-method interactions were observed for ROM and maximum pelvis rotation. Within sex analyses found significant difference between methods across sexes. Figure 8c presents the mean pelvis rotation by sex and method.

Table 8. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from pelvis kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Pelvis Tilt (°)											
ROM	4.23 (1.41)	5.15 (1.03)	4.23 (1.14)	3.66 (0.99)	0.988	< 0.001	< 0.001	0.995 (0.00)	< 0.001 (-1.51)	< 0.001 (- 0.75)	0.00777 (0.53)
Min	5.73 (1.46)	5.27 (1.67)	6.63 (1.66)	6.68 (1.39)	0.090	0.003	0.017	0.0981 (0.55)	0.0114 (1.02)	0.00371 (0.29)	0.725 (- 0.03)
Max	9.95 (1.85)	10.42 (1.82)	10.86 (1.28)	10.34 (1.50)	0.111	0.001	< 0.001	0.102 (0.71)	0.908 (- 0.05)	0.00182 (-0.26)	< 0.001 (0.37)
Pelvis List (°)											
ROM	10.44 (2.31)	10.94 (2.11)	8.56 (2.31)	7.19 (1.37)	0.007	0.062	< 0.001	0.0228 (- 0.82)	< 0.001 (-2.74)	0.0912 (- 0.22)	< 0.001 (0.72)
Min	-4.21 (1.38)	-4.60 (1.15)	-3.36 (1.25)	-2.79 (0.72)	0.021	0.020	< 0.001	0.0642 (0.68)	< 0.001 (2.51)	0.0324 (0.30)	< 0.001 (- 0.56)
Max	6.23 (1.50)	6.34 (1.54)	5.20 (1.54)	4.40 (1.10)	0.036	0.546	< 0.001	0.0534 (- 0.67)	< 0.001 (-1.76)	0.592 (- 0.07)	< 0.001 (0.60)
Pelvis Rotation (°)											
ROM	8.79 (2.81)	7.89 (2.87)	7.99 (2.43)	8.51 (2.87)	0.360	0.024	0.012	0.33 (- 0.33)	0.495 (0.22)	0.0332 (0.32)	0.166 (- 0.19)
Min	-8.88 (2.83)	-8.12 (3.14)	-8.33 (2.59)	-8.67 (3.11)	0.558	0.063	0.056	0.512 (0.21)	0.577 (- 0.18)	0.0798 (- 0.26)	0.373 (0.12)
Max	-0.09 (0.10)	-0.23 (0.51)	-0.34 (1.01)	-0.16 (0.49)	0.211	0.105	0.010	0.334 (- 0.24)	0.646 (0.15)	0.0419 (0.39)	0.0832 (- 0.22)

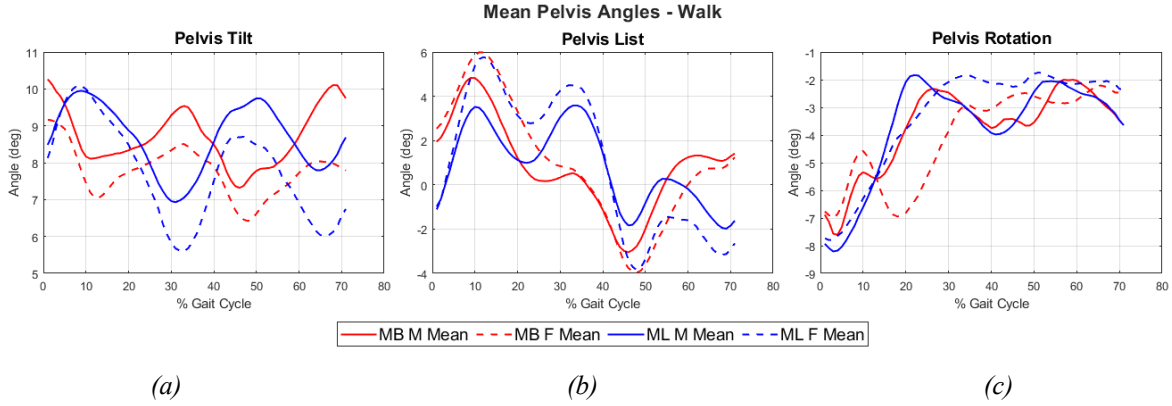


Figure 8. Mean pelvis angles during walk for male and female participants

Table 9 summarizes the hip kinematic metrics during walking for male and female participants. For hip flexion, significant main effects of method were observed for ROM, minimum, and initial contact angles in both the left and right limbs. Method-within-sex analyses also revealed that ML had a larger ROM, and initial contact angle, and a smaller minimum angle than MB. A significant main effect of sex was found for the right hip flexion ROM, with post hoc analyses showing that only the MB method found that female participants had significantly larger ROM. Significant sex-by-method interactions were identified for the right hip flexion maximum angle and initial contact angle, as well as for the left hip flexion minimum angle. Figure 9a shows the mean hip flexion waveforms for left and right hips.

For hip adduction, significant main effects were found for ROM and maximum adduction angle, with females achieving greater amplitudes compared to males. A main effect of sex was also observed in the right initial contact angle. Post-hoc sex-within-method analyses showed that both methods found significant differences between the sexes for ROM and maximum angles; however, only MB found significance in initial contact angle. Main effects of method were found for maximum left hip adduction angle and left initial contact angle, with method-by-sex comparisons revealing that ML had significantly larger peak hip adduction angles than MB with greater difference observed in male data. Figure 9b shows the mean hip adduction waveforms.

For hip internal rotation, significant main effects of method were observed across all metrics for both limbs, sex-by-method analyses also revealed that ML had significantly less rotation across all metrics and both sexes. Significant main effects of sex were observed for hip rotation minimum and maximum angles, as well as for right hip rotation ROM. Post hoc analyses of sex-within-

method revealed that significant differences were observed within both MB and ML data for the maximum hip flexion angle in both limbs, indicating that females have significantly larger maximum hip rotation angles. Whereas only ML identified significant sex-based differences of the angle at IC. Finally, significant sex-by-method interactions were present for right hip rotation maximum angle and bilaterally for hip rotation initial contact angle. Figure 9c shows the mean hip rotation waveforms.

Table 9. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from hip kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Right Hip Flexion (°)											
ROM	44.48 (3.46)	48.71 (3.00)	41.27 (4.57)	46.59 (4.88)	0.026	< 0.001	0.093	0.032 (-0.70)	0.158 (-0.43)	< 0.001 (-1.31)	< 0.001 (-1.12)
Min	-6.54 (2.82)	-11.88 (2.78)	-4.98 (2.57)	-10.58 (2.41)	0.097	< 0.001	0.592	0.128 (0.61)	0.168 (0.54)	< 0.001 (1.90)	< 0.001 (2.25)
Max	37.94 (2.74)	36.83 (2.47)	36.29 (4.34)	36.01 (3.90)	0.185	< 0.001	0.021	0.212 (-0.38)	0.491 (-0.21)	< 0.001 (0.43)	0.181 (0.07)
IC	32.84 (2.25)	33.90 (1.77)	34.20 (3.46)	34.53 (3.22)	0.157	< 0.001	0.048	0.182 (0.39)	0.515 (0.19)	< 0.001 (-0.53)	0.226 (-0.10)
Left Hip Flexion (°)											
ROM	43.02 (4.36)	48.59 (2.58)	42.31 (3.52)	46.92 (4.57)	0.609	< 0.001	0.112	0.63 (-0.20)	0.211 (-0.37)	< 0.001 (-1.55)	< 0.001 (-1.13)
Min	-4.95 (2.39)	-10.45 (2.44)	-4.55 (2.07)	-9.23 (2.28)	0.714	< 0.001	0.036	0.73 (0.19)	0.189 (0.53)	< 0.001 (2.28)	< 0.001 (2.15)
Max	38.07 (3.33)	38.14 (2.52)	37.76 (3.18)	37.68 (3.71)	0.733	0.824	0.758	0.737 (-0.10)	0.642 (-0.12)	0.859 (-0.03)	0.753 (0.02)
IC	33.48 (2.88)	35.13 (1.89)	35.48 (2.63)	36.07 (2.96)	0.057	< 0.001	0.003	0.0729 (0.76)	0.414 (0.32)	< 0.001 (-0.68)	0.00451 (-0.21)
Right Hip Adduction (°)											
ROM	15.32 (2.03)	15.05 (2.03)	12.26 (2.70)	11.88 (1.48)	< 0.001	0.452	0.831	< 0.001 (-1.14)	< 0.001 (-2.14)	0.506 (0.13)	0.218 (0.17)
Min	-10.74 (1.77)	-10.40 (1.83)	-10.54 (2.24)	-9.29 (1.30)	0.726	0.171	0.012	0.726 (0.09)	0.0437 (0.85)	0.193 (-0.19)	< 0.001 (-0.68)
Max	4.58 (2.24)	4.66 (1.98)	1.72 (1.74)	2.59 (1.45)	< 0.001	0.768	0.034	< 0.001 (-1.65)	< 0.001 (-1.42)	0.802 (-0.04)	< 0.001 (-0.55)

IC	-6.34 (2.36)	-6.06 (1.89)	-4.85 (1.84)	-6.76 (1.17)	0.006	0.382	< 0.001	0.034 (0.81)	0.266 (-0.60)	0.431 (-0.13)	< 0.001 (1.25)
Left Hip Adduction (°)											
ROM	15.24 (4.28)	15.62 (2.35)	12.60 (1.68)	12.20 (1.34)	0.007	0.266	0.106	0.046 (-1.57)	< 0.001 (-2.54)	0.32 (-0.11)	0.180 (0.26)
Min	-10.01 (3.44)	-9.56 (2.46)	-9.53 (1.62)	-8.46 (1.58)	0.662	0.081	0.092	0.722 (0.30)	0.171 (0.69)	0.137 (-0.15)	< 0.001 (-0.66)
Max	5.23 (2.24)	6.06 (1.88)	3.08 (1.76)	3.74 (1.85)	0.001	0.001	0.624	0.005 (-1.22)	< 0.001 (-1.25)	0.002 (-0.40)	0.006 (-0.37)
IC	-4.91 (3.11)	-6.11 (2.32)	-3.40 (2.82)	-6.37 (1.85)	0.072	0.001	< 0.001	0.152 (0.54)	0.719 (-0.14)	< 0.001 (0.44)	< 0.001 (1.24)
Right Hip Rotation (°)											
ROM	13.11 (2.58)	9.22 (2.48)	11.32 (3.05)	8.54 (1.28)	0.010	< 0.001	0.060	0.058 (-0.59)	0.318 (-0.53)	< 0.001 (1.54)	< 0.001 (1.19)
Min	-9.76 (1.84)	-8.28 (1.30)	-10.89 (2.00)	-9.28 (1.71)	0.039	< 0.001	0.703	0.0797 (-0.57)	0.0599 (-0.59)	< 0.001 (-0.93)	< 0.001 (-0.87)
Max	3.36 (1.69)	0.94 (1.84)	0.43 (2.67)	-0.74 (0.99)	< 0.001	< 0.001	0.003	< 0.001 (-1.10)	0.001 (-1.70)	< 0.001 (1.37)	< 0.001 (0.58)
IC	-3.45 (2.33)	0.13 (1.41)	-3.15 (3.62)	-1.03 (1.02)	0.724	< 0.001	0.007	0.863 (0.08)	0.005 (-1.14)	< 0.001 (-1.86)	< 0.001 (-0.80)
Left Hip Rotation (°)											
ROM	12.25 (3.56)	9.16 (1.35)	11.90 (3.13)	8.35 (1.19)	0.657	< 0.001	0.484	0.808 (-0.11)	0.074 (-0.68)	< 0.001 (1.15)	< 0.001 (1.49)
Min	-9.62 (2.61)	-7.55 (1.70)	-11.02 (2.22)	-8.26 (1.38)	0.027	< 0.001	0.091	0.0913 (-0.63)	0.182 (-0.51)	< 0.001 (-0.94)	< 0.001 (-1.49)
Max	2.64 (2.09)	1.60 (1.07)	0.88 (2.04)	0.10 (1.75)	0.002	< 0.001	0.530	0.014 (-0.86)	0.005 (-0.86)	< 0.001 (0.62)	0.003 (0.41)
IC	-4.14 (2.81)	0.51 (1.11)	-4.66 (1.64)	-0.78 (1.50)	0.377	< 0.001	0.053	0.523 (-0.32)	0.00985 (-0.86)	< 0.001 (-2.17)	< 0.001 (-2.46)

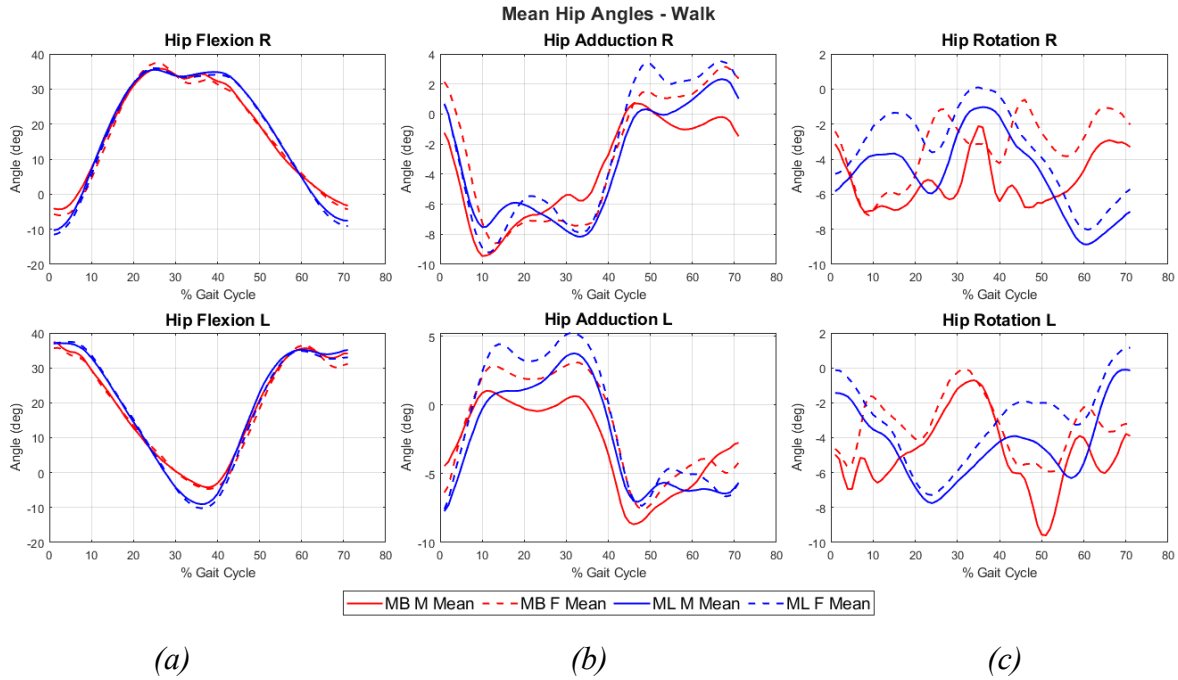


Figure 9. Mean hip angle waveforms during walking

Table 10 presents the results of the LMM for the knee. In general, across both MB and ML approaches, female participants achieved greater knee flexion and knee adduction amplitudes, whereas male participants had greater knee rotation amplitudes. In the sagittal plane, a significant main effect of sex was observed in right knee flexion ROM. However, neither method identified a significant difference during sex-within-method analyses of ROM and only a medium effect size was calculated. Significant main effects of methods were observed across almost all metrics. While the post hoc analyses found similar significant differences when comparing the methods within each sex, effect sizes revealed that the differences were typically more significant within the female group. A significant sex-by-method interaction was found bilaterally in knee flexion ROM. Figure 10a presents the mean knee flexion angles during walking tasks.

In the frontal plane, significant main effects of sex were observed in the right knee adduction ROM and minimum knee adduction angle. Post hoc analyses of sex-within-method showed that both MB and ML methods found significant differences in the minimum angle. However, only the ML method observed significant differences between sexes bilaterally for the minimum adduction angles, maximum adduction angles, and angles at IC. The amplitudes of the ML knee adductions metrics were significantly lower than the MB results, as it shown by significances being found for

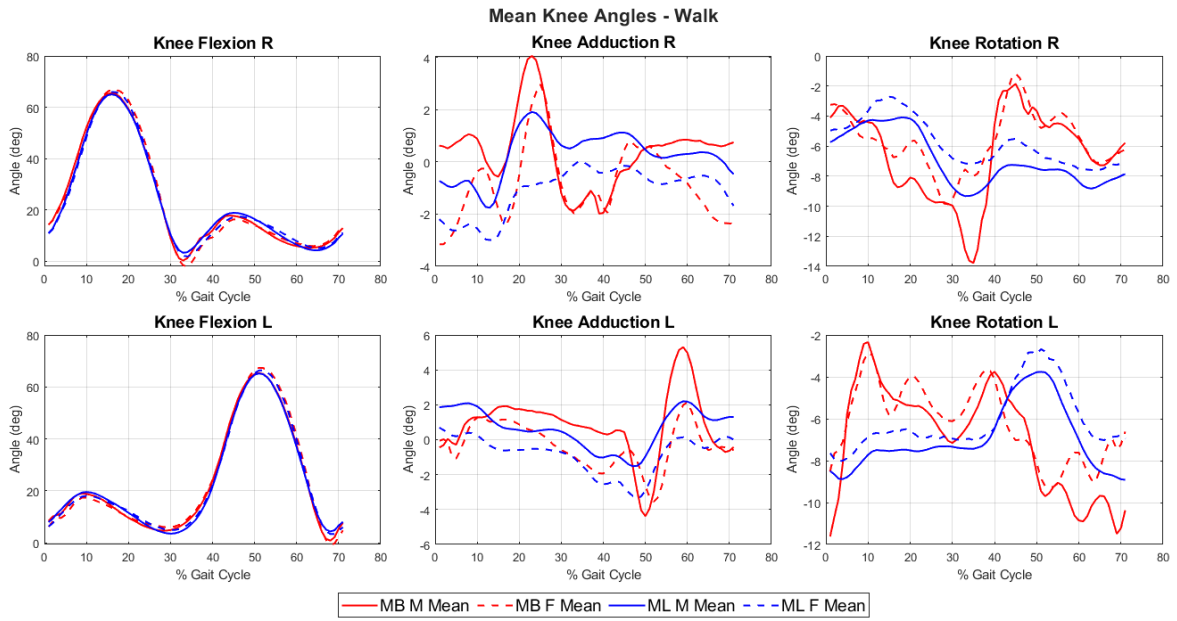
all metrics of method-within-sex comparisons, except for the left adduction initial contact angle in the female group. Figure 10b shows the mean knee adduction angles for female and male groups.

In the transverse plane, significant main effects of sex were observed bilaterally for minimum knee rotation angles (maximum external rotation) and initial contact angle, as well as for left knee rotation ROM. Post hoc analyses of sex-by-method comparisons revealed that significant differences were only observed between sexes in both MB and ML data for knee rotation initial contact angles. Knee rotation metrics saw the male participants having larger rotation amplitudes compared to females. The MB results also had greater amplitudes compared to ML. As a result, a main effect of method was observed across all metrics, with further method-by-sex comparisons revealing significant differences between methods across sexes. Significant effects of the sex-by-method interactions were also observed bilaterally for the maximum rotation angle, and for the right initial contact angle and left knee ROM. Figure 10c presents the mean knee rotation angles during walking tasks.

Table 10. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from knee kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

DV	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Right Knee Flexion (°)											
ROM	71.56 (5.13)	66.23 (4.21)	67.71 (7.17)	64.09 (5.02)	0.048	< 0.001	0.025	0.0915 (-0.54)	0.207 (- 0.43)	< 0.001 (1.14)	< 0.001 (0.58)
Min	-3.73 (3.39)	0.64 (2.63)	-1.45 (5.11)	1.54 (3.10)	0.080	< 0.001	0.006	0.145 (0.45)	0.407 (0.29)	< 0.001 (-1.44)	< 0.001 (-0.71)
Max	67.83 (4.28)	66.87 (4.00)	66.26 (5.00)	65.63 (4.90)	0.347	0.007	0.498	0.361 (- 0.31)	0.45 (- 0.25)	0.0155 (0.23)	0.040 (0.13)
IC	8.24 (2.63)	7.17 (2.38)	9.34 (3.56)	8.33 (3.14)	0.305	0.001	0.894	0.339 (0.31)	0.263 (0.37)	< 0.001 (0.43)	0.003 (0.30)
Left Knee Flexion (°)											
ROM	69.99 (3.99)	65.38 (3.55)	67.18 (4.98)	64.07 (4.24)	0.064	< 0.001	0.037	0.108 (- 0.56)	0.371 (- 0.31)	< 0.001 (1.22)	< 0.001 (0.67)
Min	-2.02 (3.29)	1.55 (2.58)	-1.41 (4.45)	1.57 (2.05)	0.709	< 0.001	0.253	0.777 (0.14)	0.837 (0.01)	< 0.001 (-1.21)	< 0.001 (-0.86)
Max	67.97 (4.18)	66.92 (3.76)	65.77 (4.36)	65.64 (4.51)	0.134	0.004	0.073	0.154 (- 0.50)	0.347 (- 0.28)	0.00533 (0.26)	0.703 (0.03)
IC	6.92 (3.28)	6.97 (2.68)	9.20 (2.91)	8.86 (3.53)	0.061	0.851	0.312	0.0562 (0.78)	0.139 (0.54)	0.861 (- 0.02)	0.179 (0.10)
Right Knee Adduction (°)											
ROM	12.40 (4.23)	5.45 (2.01)	10.75 (3.00)	4.88 (1.58)	0.034	< 0.001	0.138	0.167 (- 0.55)	0.357 (- 0.36)	< 0.001 (2.10)	< 0.001 (2.45)
Min	-6.84 (2.95)	-4.12 (1.99)	-4.85 (1.80)	-2.40 (1.97)	0.004	< 0.001	0.576	0.0192 (1.10)	0.0166 (0.87)	< 0.001 (-1.08)	< 0.001 (-1.30)
Max	5.56 (2.53)	1.33 (1.28)	5.89 (2.67)	2.48 (1.40)	0.649	< 0.001	0.079	0.804 (0.12)	0.0176 (0.82)	< 0.001 (2.11)	< 0.001 (1.60)

IC	-0.98 (2.33)	-0.33 (1.83)	-1.11 (2.93)	0.82 (1.09)	0.804	0.024	0.002	0.826 (-0.04)	0.0428 (1.05)	0.014 (-0.31)	< 0.001 (-0.87)
Left Knee Adduction (°)											
ROM	11.81 (3.16)	5.71 (1.67)	11.93 (2.72)	5.24 (1.89)	0.913	< 0.001	0.363	0.924 (0.04)	0.373 (-0.25)	< 0.001 (2.41)	< 0.001 (2.86)
Min	-6.52 (2.45)	-3.91 (1.38)	-5.76 (2.96)	-2.16 (2.36)	0.263	< 0.001	0.056	0.406 (0.25)	0.0121 (0.74)	< 0.001 (-1.31)	< 0.001 (-1.35)
Max	5.30 (2.61)	1.81 (1.21)	6.16 (2.65)	3.09 (1.84)	0.209	< 0.001	0.329	0.342 (0.33)	0.0268 (0.70)	< 0.001 (1.72)	0.000 (1.35)
IC	-0.25 (2.78)	0.26 (1.43)	-0.33 (3.11)	1.60 (1.75)	0.850	0.078	0.001	0.863 (-0.03)	0.0303 (0.76)	0.11 (-0.23)	< 0.001 (-0.76)
Right Knee Rotation (°)											
ROM	15.59 (3.25)	6.92 (1.73)	17.01 (3.43)	7.05 (1.68)	0.066	< 0.001	0.063	0.243 (0.41)	0.965 (0.07)	< 0.001 (3.33)	< 0.001 (3.69)
Min	-13.89 (3.10)	-8.54 (1.85)	-15.62 (2.19)	-9.95 (2.05)	0.021	< 0.001	0.537	0.0539 (-0.79)	0.0491 (-0.69)	< 0.001 (-2.10)	< 0.001 (-2.67)
Max	1.70 (2.57)	-1.61 (2.89)	1.39 (3.42)	-2.90 (2.36)	0.710	< 0.001	0.063	0.778 (-0.09)	0.147 (-0.54)	< 0.001 (1.21)	< 0.001 (1.46)
IC	-5.75 (4.16)	-7.11 (2.22)	-10.53 (3.68)	-8.83 (1.95)	< 0.001	0.009	< 0.001	0.00129 (-1.30)	0.0262 (-0.88)	0.0269 (0.41)	< 0.001 (-0.58)
Left Knee Rotation (°)											
ROM	14.86 (3.17)	7.05 (2.33)	16.75 (3.33)	6.95 (1.88)	0.002	< 0.001	0.020	0.077 (0.57)	0.946 (-0.05)	< 0.001 (2.81)	< 0.001 (3.62)
Min	-14.33 (2.50)	-8.76 (1.75)	-15.90 (2.93)	-9.86 (1.94)	0.035	< 0.001	0.352	0.0926 (-0.53)	0.116 (-0.57)	< 0.001 (-2.58)	< 0.001 (-2.43)
Max	0.53 (2.08)	-1.71 (2.55)	0.85 (3.13)	-2.91 (2.67)	0.665	< 0.001	0.009	0.748 (0.10)	0.243 (-0.45)	< 0.001 (0.97)	< 0.001 (1.29)
IC	-6.99 (3.58)	-7.45 (1.57)	-10.30 (2.88)	-8.82 (2.10)	< 0.001	0.249	0.001	0.00451 (-1.15)	0.0417 (-0.66)	0.335 (0.16)	< 0.001 (-0.59)



(a)

(b)

(c)

Figure 10. Mean knee angle waveforms during walking

Table 11 displays the results of the LMM for the ankle. In the sagittal plane, female participants achieved more ankle flexion compared to males. A significant main effect of sex was found bilaterally in ankle flexion ROM and in left minimum flexion angle. Sex-by-method post-hoc analyses showed that both MB and ML found significant differences between sex in left ankle flexion ROM and minimum angle. Only the ML method found significant differences in the right ankle flexion ROM, MB measured a large effect size of -0.80 for the same metric. Ankle flexion ROM was approximately 2 degrees larger in MB kinematics compared to ML. A significant main effect of method was measured bilaterally for ankle flexion ROM, minimum angle, right initial contact angle, and left maximum angle. Method-by-sex post hoc analyses showed significant differences between methods for most metrics, across sexes. Figure 11a presents the mean ankle flexion during walking tasks.

In the frontal plane, ankle inversion/eversion in females was greater than males. Significant main effects of sex were observed bilaterally in ROM, and initial contact angle, as well as in right minimum inversion angle. Post-hoc sex-within-method analyses found no significant differences between sexes for either method; however, medium effect sizes ($d > 0.6$) were calculated in MB for right ankle ROM, and right and left initial contact angles. The amplitudes achieved in ankle inversion kinematics are much smaller in the ML results, resulting in a significant main effect of methods being found across all metrics. Significant interactions were also observed bilaterally in ankle inversion ROM, minimum angle, and initial contact angle. Figure 11b presents the mean ankle inversion during walking tasks.

Lastly, in the transverse plane, no significant main effects of sex were found. In general, the MB method estimated smaller minimum angles, while the ML method measured larger maximum angles. These differences resulted in significant main effects of method being found in minimum ankle rotation angle, left ankle ROM, and right ankle maximum angles. Method-by-sex post-hoc analyses revealed significant difference between methods in both sexes; however, larger effect sizes were calculated in the female group. Interaction effects were found for initial contact angles and left ankle rotation ROM. Figure 11c presents the mean ankle internal rotation during walking tasks.

Table 11. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from ankle kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Right Ankle Flexion (°)											
ROM	34.94 (6.40)	32.84 (5.15)	31.38 (4.51)	29.42 (3.46)	0.029	0.001	0.873	0.0527 (-0.80)	0.0155 (- 0.99)	0.002 (0.36)	< 0.001 (0.49)
Min	-19.51 (6.08)	-17.83 (5.19)	-16.62 (3.94)	-15.22 (3.35)	0.073	0.001	0.685	0.0951 (0.74)	0.0786 (0.78)	0.002 (- 0.30)	0.003 (- 0.38)
Max	15.43 (3.18)	15.01 (3.12)	14.77 (3.39)	14.20 (2.72)	0.494	0.144	0.709	0.519 (- 0.19)	0.394 (- 0.30)	0.172 (0.13)	0.033 (0.18)
IC	3.62 (4.35)	0.88 (3.34)	1.54 (4.89)	0.33 (3.35)	0.125	< 0.001	0.010	0.185 (- 0.43)	0.604 (- 0.17)	< 0.001 (0.71)	0.004 (0.29)
Left Ankle Flexion (°)											
ROM	33.91 (4.94)	31.69 (3.25)	29.67 (5.21)	27.06 (4.44)	0.004	< 0.001	0.609	0.0143 (-0.82)	< 0.001 (-1.04)	< 0.001 (0.53)	< 0.001 (0.54)
Min	-16.53 (4.62)	-14.95 (3.52)	-13.18 (5.09)	-12.07 (3.92)	0.020	0.002	0.517	0.0423 (0.66)	0.0211 (0.74)	0.00484 (-0.38)	0.0223 (- 0.24)
Max	17.39 (2.38)	16.74 (1.88)	16.49 (3.12)	14.99 (2.59)	0.256	0.018	0.026	0.311 (- 0.29)	0.023 (- 0.68)	0.0111 (0.30)	< 0.001 (0.52)
IC	4.02 (3.49)	1.44 (2.40)	2.80 (4.28)	1.35 (3.03)	0.298	< 0.001	0.023	0.367 (- 0.28)	0.942 (- 0.03)	< 0.001 (0.86)	< 0.001 (0.39)
Right Ankle Inversion (°)											
ROM	27.43 (6.79)	5.70 (1.08)	23.62 (6.35)	5.41 (0.97)	0.004	< 0.001	0.004	0.111 (- 0.60)	0.339 (- 0.30)	< 0.001 (4.47)	< 0.001 (4.01)
Min	-7.67 (4.27)	4.73 (0.91)	-5.64 (3.56)	4.23 (1.59)	0.007	< 0.001	0.002	0.126 (0.57)	0.291 (- 0.31)	< 0.001 (-4.02)	< 0.001 (-3.58)
Max	19.76 (4.82)	10.43 (0.92)	17.97 (4.45)	9.64 (1.46)	0.086	< 0.001	0.190	0.314 (- 0.40)	0.055 (- 0.54)	< 0.001 (2.69)	< 0.001 (2.52)

IC	16.03 (4.27)	7.19 (1.24)	13.04 (4.32)	6.61 (1.49)	0.003	< 0.001	0.001	0.0573 (-0.69)	0.247 (-0.39)	< 0.001 (2.82)	< 0.001 (1.99)
Left Ankle Inversion (°)											
ROM	22.51 (5.74)	5.50 (1.40)	20.20 (5.51)	5.60 (1.30)	0.035	< 0.001	0.031	0.257 (-0.42)	0.913 (0.08)	< 0.001 (4.07)	< 0.001 (3.65)
Min	-6.40 (3.84)	3.73 (1.11)	-4.77 (3.43)	3.37 (1.69)	0.034	< 0.001	0.006	0.236 (0.48)	0.52 (-0.21)	< 0.001 (-3.59)	< 0.001 (-3.01)
Max	16.10 (3.69)	9.23 (1.21)	15.44 (3.23)	8.97 (0.89)	0.353	< 0.001	0.505	0.555 (-0.21)	0.453 (-0.28)	< 0.001 (2.50)	< 0.001 (2.73)
IC	13.23 (3.81)	6.24 (1.03)	10.67 (3.78)	5.48 (1.74)	0.006	< 0.001	0.002	0.0614 (-0.68)	0.148 (-0.44)	< 0.001 (2.51)	< 0.001 (1.77)
Right Ankle Internal Rotation (°)											
ROM	17.57 (2.80)	17.05 (3.41)	16.59 (4.43)	16.21 (3.05)	0.312	0.365	0.863	0.368 (-0.22)	0.434 (-0.27)	0.304 (0.17)	0.544 (0.10)
Min	-6.48 (2.98)	-4.06 (3.57)	-6.57 (3.94)	-4.49 (2.94)	0.931	< 0.001	0.561	0.938 (-0.02)	0.703 (-0.15)	< 0.001 (-0.74)	< 0.001 (-0.60)
Max	11.09 (4.19)	12.99 (2.80)	10.02 (4.35)	11.72 (3.33)	0.349	< 0.001	0.757	0.415 (-0.25)	0.229 (-0.38)	< 0.001 (-0.53)	< 0.001 (-0.44)
IC	-0.62 (3.35)	0.82 (2.41)	-1.25 (3.92)	-1.38 (2.96)	0.527	0.003	0.019	0.603 (-0.16)	0.0209 (-0.74)	0.003 (-0.49)	0.778 (0.04)
Left Ankle Internal Rotation (°)											
ROM	17.90 (3.63)	15.55 (3.37)	16.47 (2.93)	15.92 (3.60)	0.180	< 0.001	0.012	0.207 (-0.49)	0.738 (0.11)	< 0.001 (0.67)	0.217 (0.17)
Min	-5.26 (3.87)	-3.29 (4.05)	-5.07 (2.59)	-4.08 (3.17)	0.814	< 0.001	0.167	0.821 (0.07)	0.579 (-0.25)	< 0.001 (-0.50)	0.0452 (-0.34)
Max	12.64 (2.25)	12.25 (2.14)	11.39 (3.66)	11.84 (2.86)	0.211	0.289	0.107	0.27 (-0.34)	0.721 (-0.14)	0.32 (0.18)	0.191 (-0.14)
IC	0.78 (3.12)	0.75 (1.93)	0.85 (3.30)	-0.45 (3.04)	0.948	0.931	0.018	0.996 (0.02)	0.216 (-0.39)	0.932 (0.01)	< 0.001 (0.41)

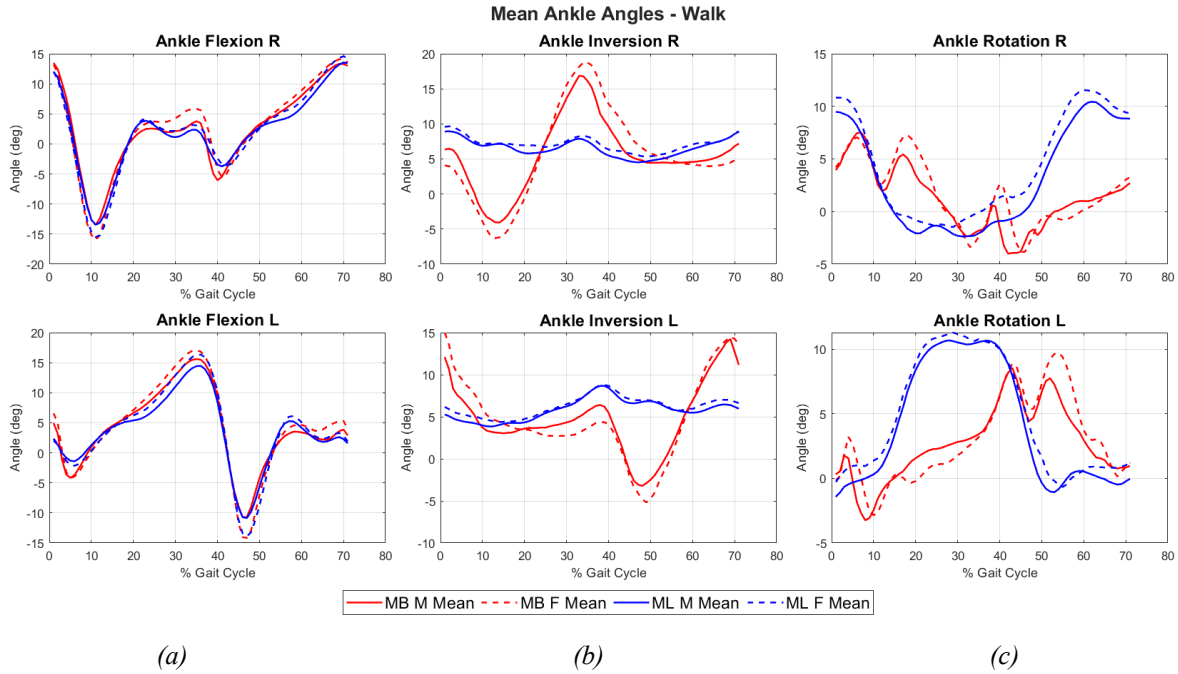


Figure 11. Mean knee angle waveforms during walking

4.4 Kinematic Comparison – Drop Vertical Jump Trials

The next step in the analysis was to evaluate the performance of the two motion capture systems during a more dynamic movement such as the DVJ. The same metrics of ROM, minimum, maximum, and initial contact angles were extracted from the data and evaluated against different groupings using LMMs.

Table 12 presents the results of the pelvis kinematics during DVJ. The sagittal plane saw the most amount of sex-based and method-based differences observed. Both MB and ML found that females had significantly larger pelvic tilt ROM compared to males. This is further reflected by the significant main effect of sex found across all metrics. Post hoc sex-by-method comparisons revealed that both MB and ML methods found significant differences in ROM and maximum angles, while only MB found difference in minimum pelvic tilt. In ROM, both methods had good agreement; however, positive bias towards the MB system is observed in minimum and maximum angles. As a result, significant main effects of method, and sex-by-method interactions were observed for minimum and maximum angles. Post-hoc method-within-sex analyses revealed that significant differences between methods were observed in both sexes.

For pelvic list a main effect of sex, and an interaction between sex and method were found in ROM and minimum pelvic list angles. However, post hoc sex-within methods analyses did not reveal significant differences between sexes for either method, nor was a large effect size calculated. While a main effect of method was not found for any metric, method-within-sex comparisons found significant differences between methods in both male and female data. In pelvic rotation, a significant main effect of method was identified in pelvic rotation ROM and minimum angle. Figure 12 presents the mean pelvis kinematics for female and male participants during DVJ tasks.

Table 12. Means, standard deviations, main effect *p*-values, and post-hoc analysis *p*-values and effect sizes from pelvis DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample *t*-tests and Cohen's *d* effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Pelvis Tilt											
ROM	16.99 (4.75)	15.27 (4.96)	12.87 (6.13)	10.16 (4.95)	0.006	0.057	0.440	0.0211 (- 0.67)	0.00404 (-1.03)	0.0195 (0.36)	0.0115 (0.49)
Min	9.22 (3.46)	7.67 (3.14)	6.09 (4.65)	7.80 (3.68)	0.015	< 0.001	< 0.001	0.0247 (- 0.67)	0.91 (0.04)	< 0.001 (0.47)	< 0.001 (- 0.41)
Max	26.21 (4.57)	22.94 (5.50)	18.97 (6.11)	17.97 (6.99)	< 0.001	< 0.001	0.033	< 0.001 (-1.19)	0.032 (- 0.71)	< 0.001 (0.65)	0.245 (0.15)
Pelvis List											
ROM	4.93 (1.68)	4.31 (1.67)	7.29 (5.24)	4.28 (1.68)	0.002	0.287	0.004	0.081 (0.45)	0.943 (- 0.02)	0.0386 (0.37)	< 0.001 (0.77)
Min	-1.64 (1.87)	-1.60 (1.48)	-2.95 (3.83)	-1.25 (1.62)	0.045	0.940	0.007	0.15 (- 0.34)	0.423 (0.22)	0.906 (- 0.02)	0.00272 (-0.58)
Max	3.29 (2.18)	2.71 (1.77)	4.34 (3.84)	3.03 (1.61)	0.098	0.238	0.297	0.173 (0.27)	0.524 (0.20)	0.0411 (0.29)	0.0422 (0.44)
Pelvis Rotation											
ROM	6.92 (4.14)	5.06 (3.37)	5.99 (2.16)	4.69 (2.09)	0.379	< 0.001	0.228	0.411 (- 0.43)	0.704 (- 0.18)	< 0.001 (0.49)	< 0.001 (0.61)
Min	-7.03 (4.19)	-5.40 (3.41)	-6.49 (2.19)	-5.21 (2.19)	0.614	< 0.001	0.464	0.634 (0.24)	0.847 (0.09)	< 0.001 (-0.43)	< 0.001 (- 0.58)
Max	-0.11 (0.18)	-0.35 (0.82)	-0.50 (0.99)	-0.53 (0.87)	0.064	0.105	0.312	0.0326 (- 0.39)	0.44 (- 0.21)	0.0507 (0.40)	0.867 (0.03)

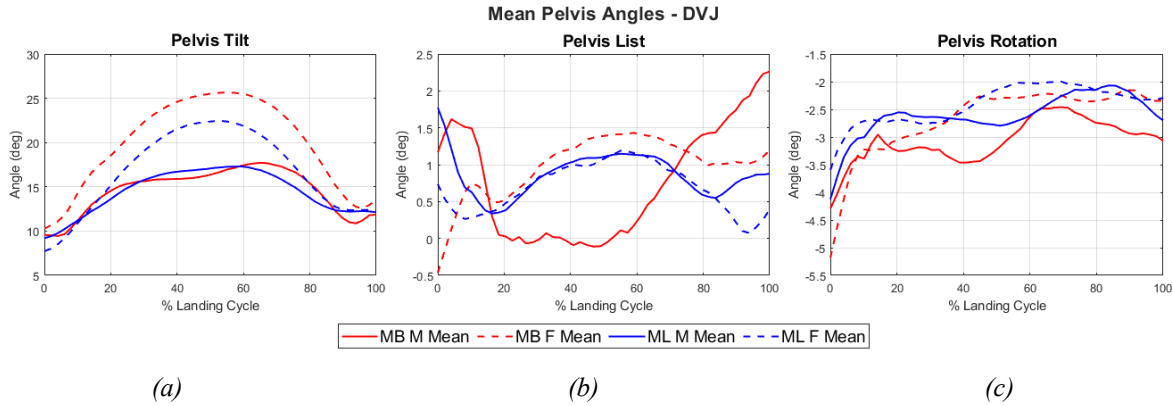


Figure 12. Mean pelvis angle waveforms during DVJ task

Table 13 presents the results of the hip kinematics during DVJ. In the sagittal plane, a significant main effect of method was observed in initial contact angles and left minimum hip flexion. Post-hoc analyses within the sexes revealed significant difference between methods were found across sexes.

In the frontal plane, main effects of sex were not identified. Main effects of method were observed in hip adduction ROM, minimum adduction and initial contact angles. Post-hoc method-within-sex analyses found significant difference between methods in females for all metrics that had a main effect of method, whereas only some of the metrics had significant difference between methods in male data.

In the transverse plane, a significant main effect of sex was found in left hip rotation ROM. Further sex-by-method comparisons did not reveal a significant difference between sexes in either MB or ML, however a medium effect size of 0.5 was calculated in both. Sex-within method comparisons also showed that males landed with significantly smaller left hip rotation (greater external rotation) compared to females in ML. Main effects of method were identified across all metrics except for right hip initial contact angle. Post-hoc analyses showed very significant differences between methods in both sexes in all metrics except for right hip initial contact angle in females. Significant interaction effects were also observed in left hip rotation maximum hip rotation and right hip initial contact angles. Figure 13 presents the mean hip angle waveforms during DVJ.

Table 13. Means, standard deviations, main effect *p*-values, and post-hoc analysis *p*-values and effect sizes from hip DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample *t*-tests and Cohen's *d* effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Right Hip Flexion (°)											
ROM	71.11 (9.35)	71.13 (10.03)	68.49 (19.45)	70.24 (17.35)	0.601	0.990	0.449	0.622 (-0.13)	0.861 (-0.05)	0.988 (-0.00)	0.351 (-0.09)
Min	13.17 (7.43)	14.39 (3.89)	12.03 (6.77)	13.92 (3.68)	0.543	0.087	0.504	0.642 (-0.17)	0.712 (-0.13)	0.0795 (-0.21)	0.0114 (-0.35)
Max	84.28 (8.42)	85.52 (10.96)	80.52 (17.16)	84.16 (17.76)	0.452	0.313	0.170	0.435 (-0.22)	0.801 (-0.08)	0.193 (-0.13)	0.0145 (-0.21)
IC	32.09 (4.67)	28.87 (3.88)	33.12 (9.11)	31.64 (7.51)	0.644	< 0.001	0.119	0.669 (0.11)	0.161 (0.37)	< 0.001 (0.75)	0.057 (0.18)
Left Hip Flexion (°)											
ROM	70.95 (8.78)	71.12 (10.07)	65.46 (19.36)	69.00 (18.10)	0.278	0.919	0.142	0.304 (-0.28)	0.691 (-0.12)	0.905 (-0.02)	0.0574 (-0.19)
Min	12.75 (6.91)	14.12 (3.41)	12.39 (7.52)	14.09 (4.24)	0.854	0.047	0.737	0.889 (-0.05)	0.982 (-0.01)	0.0241 (-0.25)	0.0294 (-0.28)
Max	83.70 (7.40)	85.23 (10.64)	77.86 (16.24)	83.09 (17.81)	0.226	0.208	0.032	0.2 (-0.36)	0.69 (-0.12)	0.143 (-0.17)	< 0.001 (-0.31)
IC	37.45 (5.61)	33.34 (4.38)	38.15 (11.42)	36.23 (10.45)	0.807	< 0.001	0.109	0.815 (0.06)	0.278 (0.28)	< 0.001 (0.82)	0.0631 (0.18)
Right Hip Adduction (°)											
ROM	10.44 (3.30)	7.30 (2.04)	9.73 (3.24)	7.35 (3.06)	0.324	< 0.001	0.369	0.462 (-0.22)	0.948 (0.02)	< 0.001 (1.14)	< 0.001 (0.76)
Min	-13.96 (3.53)	-10.92 (3.55)	-13.23 (3.98)	-11.77 (3.84)	0.552	< 0.001	0.020	0.567 (0.18)	0.501 (-0.22)	< 0.001 (-0.86)	0.00136 (-0.37)
Max	-3.52 (2.90)	-3.62 (2.62)	-3.50 (4.20)	-4.42 (3.46)	0.987	0.830	0.206	0.988 (0.00)	0.415 (-0.23)	0.812 (0.04)	0.0689 (0.24)

IC	-10.60 (3.40)	-8.70 (3.58)	-10.21 (4.00)	-9.99 (4.65)	0.772	< 0.001	0.006	0.753 (0.10)	0.379 (-0.28)	< 0.001 (-0.54)	0.645 (-0.05)
Left Hip Adduction (°)											
ROM	8.79 (2.75)	7.00 (3.19)	9.43 (4.08)	8.09 (3.14)	0.487	0.003	0.593	0.582 (0.16)	0.306 (0.35)	0.0021 (0.60)	0.0346 (0.37)
Min	-10.81 (3.26)	-9.26 (3.16)	-11.08 (4.71)	-10.94 (4.02)	0.825	0.002	0.046	0.846 (-0.06)	0.19 (-0.42)	0.00324 (-0.48)	0.761 (-0.03)
Max	-2.02 (2.62)	-2.25 (3.07)	-1.66 (4.10)	-2.85 (3.53)	0.748	0.562	0.096	0.762 (0.09)	0.607 (-0.17)	0.574 (0.08)	0.0032 (0.31)
IC	-6.39 (2.23)	-5.10 (2.24)	-6.23 (3.46)	-6.26 (3.17)	0.859	0.001	0.016	0.867 (0.05)	0.218 (-0.37)	0.00109 (-0.58)	0.936 (0.01)
Right Hip Internal Rotation (°)											
ROM	15.50 (3.78)	8.26 (2.42)	16.90 (6.05)	9.51 (2.99)	0.197	< 0.001	0.883	0.347 (0.23)	0.186 (0.42)	< 0.001 (2.28)	< 0.001 (1.55)
Min	-11.54 (3.52)	-7.25 (1.82)	-12.04 (3.72)	-7.29 (1.73)	0.524	< 0.001	0.520	0.655 (-0.14)	0.942 (-0.02)	< 0.001 (-1.53)	< 0.001 (-1.64)
Max	3.96 (3.73)	1.01 (2.39)	4.86 (5.59)	2.22 (2.54)	0.370	< 0.001	0.760	0.511 (0.16)	0.157 (0.47)	< 0.001 (0.94)	0.00268 (0.61)
IC	-3.86 (3.84)	-4.30 (2.81)	-2.71 (3.59)	-4.93 (1.92)	0.240	0.340	0.007	0.346 (0.32)	0.443 (-0.33)	0.338 (0.13)	< 0.001 (0.77)
Left Hip Internal Rotation (°)											
ROM	15.23 (3.81)	6.94 (2.09)	18.20 (5.97)	8.46 (2.90)	0.007	< 0.001	0.148	0.075 (0.50)	0.0692 (0.52)	< 0.001 (2.70)	< 0.001 (2.08)
Min	-12.91 (5.31)	-6.80 (2.14)	-13.79 (5.40)	-8.31 (2.90)	0.509	< 0.001	0.449	0.644 (-0.16)	0.0856 (-0.52)	< 0.001 (-1.51)	< 0.001 (-1.26)
Max	2.32 (4.24)	0.14 (1.88)	4.41 (6.36)	0.14 (2.57)	0.085	0.002	0.032	0.249 (0.33)	0.994 (0.00)	< 0.001 (0.67)	< 0.001 (0.88)
IC	-2.53 (4.78)	-4.77 (2.69)	-4.86 (4.33)	-7.67 (3.12)	0.053	< 0.001	0.462	0.144 (-0.54)	0.00333 (-0.93)	< 0.001 (0.58)	< 0.001 (0.74)

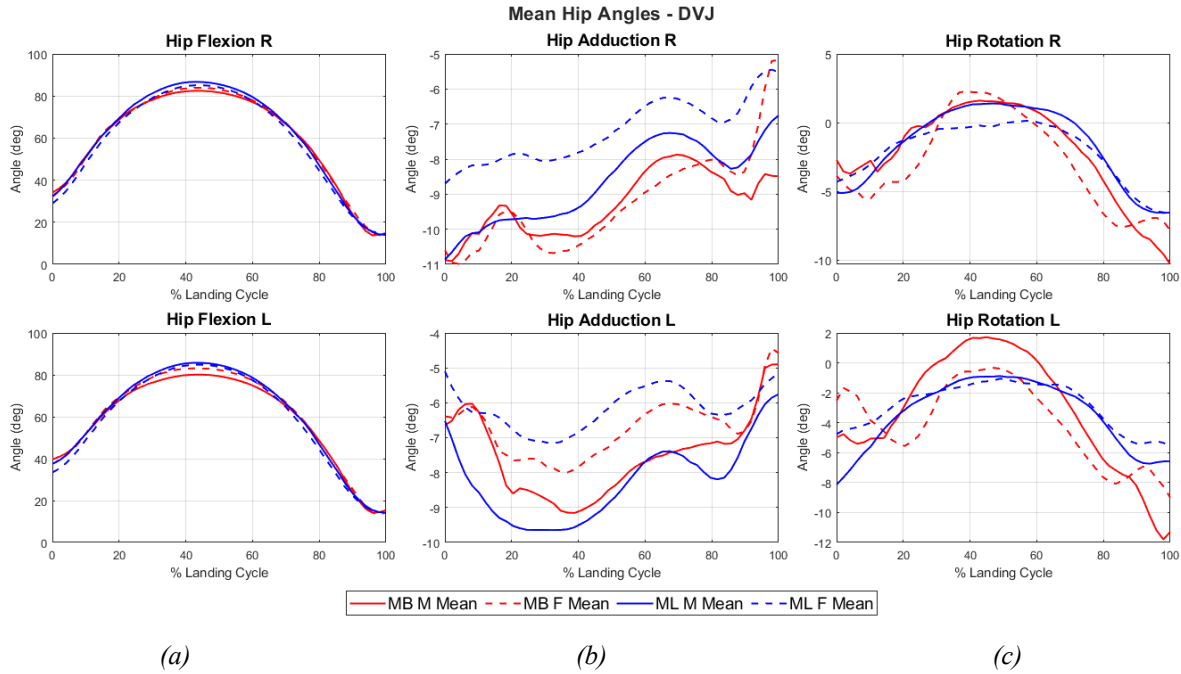


Figure 13. Mean hip angle waveforms during DVJ task

Table 14 presents the data from the knee kinematics during DVJ. In the sagittal plane, the knee flexion ROM is very comparable between methods; however, main effects of methods are observed bilaterally for minimum and maximum knee flexion, indicating bias between the approaches. Post-hoc method-within-sex analyses further revealed significance between MB and ML approaches in both male and female data. However larger effect sizes suggest greater discrepancies between the methods within female participants. Significant main effect of sex was found in left minimum knee flexion. Sex-within-method investigation discovered that ML found a significant difference between sexes bilaterally, whereas MB only identified a difference in the left knee. Figure 14a illustrates the mean knee flexion waveforms.

In the frontal plane, significant main effects of sex were found in knee adduction ROM and minimum knee adduction. Sex-within method analyses found that MB found that females had significantly larger ROM and smaller minimum hip adduction (greater hip abduction) compared to males bilaterally. ML also found that females had significantly smaller minimum hip adduction, although a significant difference in ROM was not observed. However, ML did find that males has significantly larger maximum knee adduction in the right leg. Main effects of method were found across all metrics except for left initial contact angle. Post-hoc analyses within each sex found that the ROM and amplitudes achieved by ML was significantly smaller than by MB methods across

sexes, although the effect sizes were larger in female data. Figure 14b illustrates the mean knee adduction waveforms.

For knee internal rotation, main effects of sex were observed in ROM, right maximum angles, and left minimum angles. Within method comparisons revealed that significantly larger ROM, maximum rotation, and significantly smaller internal rotation (greater external rotation) was observed in females by MB. Significant main effect of method was observed across all metrics except for right initial contact angle, and method-within-sex comparisons revealed that overall greater discrepancies were found in the data from female participants, as indicated by larger effect sizes. Figure 14c illustrates the mean knee rotation waveforms.

Table 14. Means, standard deviations, main effect p-values, and post-hoc analysis p-values and effect sizes from knee DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample t-tests and Cohen's d effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Right Knee Flexion (°)											
ROM	96.69 (9.59)	97.69 (9.30)	97.76 (17.04)	97.53 (15.27)	0.814	0.489	0.548	0.82 (0.06)	0.971 (- 0.01)	0.492 (- 0.11)	0.874 (0.01)
Min	-3.13 (7.13)	1.24 (4.46)	0.64 (8.52)	4.82 (6.17)	0.090	< 0.001	0.870	0.145 (0.44)	0.046 (0.58)	< 0.001 (-0.74)	< 0.001 (-0.56)
Max	93.57 (7.71)	98.93 (8.79)	98.40 (12.57)	102.35 (12.50)	0.190	< 0.001	0.342	0.18 (0.38)	0.357 (0.27)	< 0.001 (-0.65)	< 0.001 (-0.31)
IC	26.48 (6.75)	27.01 (6.17)	27.76 (7.23)	27.52 (6.98)	0.560	0.566	0.554	0.566 (0.18)	0.8 (0.07)	0.609 (- 0.08)	0.762 (0.03)
Left Knee Flexion (°)											
ROM	97.19 (8.83)	97.25 (8.89)	94.06 (17.33)	97.16 (16.41)	0.497	0.967	0.139	0.499 (- 0.18)	0.984 (- 0.01)	0.968 (- 0.01)	0.0267 (-0.18)
Min	-3.71 (6.04)	1.55 (3.58)	2.27 (10.21)	5.61 (6.97)	0.013	< 0.001	0.105	0.0346 (0.59)	0.0326 (0.58)	< 0.001 (-1.06)	< 0.001 (-0.38)
Max	93.48 (6.48)	98.80 (7.94)	96.33 (12.09)	102.77 (13.19)	0.425	< 0.001	0.443	0.394 (0.24)	0.292 (0.30)	< 0.001 (-0.73)	< 0.001 (-0.51)
IC	35.03 (6.90)	34.16 (6.67)	38.09 (13.05)	37.55 (13.19)	0.361	0.541	0.870	0.345 (0.23)	0.291 (0.26)	0.413 (0.13)	0.752 (0.04)
Right Knee Adduction (°)											
ROM	15.51 (3.35)	6.19 (2.77)	10.86 (3.15)	5.89 (1.73)	< 0.001	< 0.001	< 0.001	< 0.001 (-1.48)	0.707 (- 0.17)	< 0.001 (3.03)	< 0.001 (1.96)
Min	-8.17 (2.85)	-5.26 (2.46)	-4.36 (2.50)	-3.39 (2.53)	< 0.001	< 0.001	0.003	< 0.001 (1.52)	0.0291 (0.74)	< 0.001 (-1.09)	0.0174 (-0.39)

Max	7.35 (3.38)	0.93 (1.36)	6.50 (3.34)	2.50 (1.89)	0.281	< 0.001	< 0.001	0.458 (-0.25)	0.00682 (0.83)	< 0.001 (2.49)	< 0.001 (1.47)
IC	1.20 (3.39)	-0.54 (1.74)	1.45 (2.42)	0.46 (1.62)	0.747	< 0.001	0.116	0.807 (0.10)	0.0991 (0.61)	< 0.001 (0.64)	< 0.001 (0.48)
Left Knee Adduction (°)											
ROM	15.63 (3.87)	6.15 (2.13)	12.58 (4.44)	6.30 (1.98)	< 0.001	< 0.001	< 0.001	0.0382 (-0.69)	0.809 (0.08)	< 0.001 (3.04)	< 0.001 (1.82)
Min	-9.15 (3.73)	-4.57 (1.57)	-6.23 (3.40)	-3.28 (2.18)	0.001	< 0.001	0.018	0.0225 (0.86)	0.0277 (0.59)	< 0.001 (-1.60)	< 0.001 (-1.03)
Max	6.48 (3.24)	1.58 (1.78)	6.35 (3.88)	3.02 (2.71)	0.883	< 0.001	0.014	0.913 (-0.04)	0.0722 (0.53)	< 0.001 (1.88)	< 0.001 (0.99)
IC	1.21 (3.53)	0.63 (1.93)	2.81 (4.25)	2.12 (2.69)	0.135	0.145	0.862	0.248 (0.38)	0.0721 (0.56)	0.182 (0.21)	0.0618 (0.19)
Right Knee Internal Rotation (°)											
ROM	22.41 (6.86)	13.30 (3.19)	17.01 (5.11)	14.10 (3.54)	< 0.001	< 0.001	< 0.001	0.0129 (-1.06)	0.489 (0.22)	< 0.001 (1.70)	< 0.001 (0.66)
Min	-17.76 (4.06)	-11.24 (2.01)	-16.23 (3.25)	-12.67 (3.03)	0.104	< 0.001	< 0.001	0.206 (0.47)	0.0997 (-0.47)	< 0.001 (-2.04)	< 0.001 (-1.14)
Max	4.65 (4.36)	2.06 (3.42)	0.78 (4.75)	1.43 (4.04)	0.006	< 0.001	< 0.001	0.0183 (-0.82)	0.628 (-0.16)	< 0.001 (0.66)	0.218 (-0.15)
IC	-9.94 (4.68)	-7.16 (1.97)	-9.52 (4.68)	-8.07 (3.32)	0.725	< 0.001	0.096	0.776 (0.09)	0.344 (-0.27)	< 0.001 (-0.77)	0.00208 (-0.36)
Left Knee Internal Rotation (°)											
ROM	20.60 (6.16)	11.62 (3.22)	15.49 (4.73)	12.22 (4.36)	0.001	< 0.001	< 0.001	0.00924 (-1.08)	0.655 (0.14)	< 0.001 (1.83)	< 0.001 (0.72)
Min	-19.35 (4.02)	-10.51 (2.40)	-16.54 (3.30)	-10.74 (2.02)	0.002	< 0.001	< 0.001	0.0203 (0.85)	0.749 (-0.12)	< 0.001 (-2.67)	< 0.001 (-2.12)
Max	1.25 (4.25)	1.11 (3.46)	-1.05 (4.98)	1.48 (4.49)	0.103	0.805	0.001	0.17 (-0.46)	0.793 (0.08)	0.77 (0.04)	< 0.001 (-0.53)
IC	-6.46 (5.13)	-6.25 (2.52)	-8.61 (4.48)	-6.30 (2.71)	0.081	0.699	0.007	0.194 (-0.48)	0.952 (-0.02)	0.726 (-0.05)	< 0.001 (-0.63)

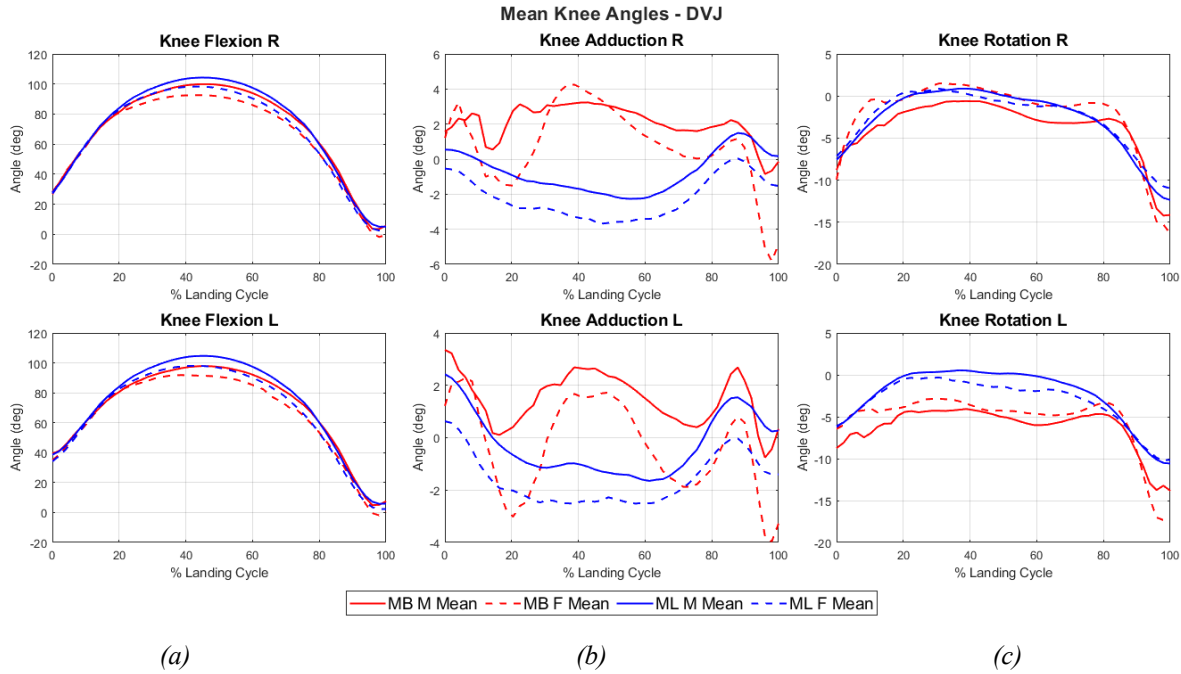


Figure 14. Mean knee angle waveforms during DVJ task

Table 15 presents the kinematic data for the ankle joint during DVJ tasks. For ankle flexion, a significant main effect of sex was found in ROM and minimum angles. Sex-within-method comparisons revealed that the MB method identified that females had significantly greater ROM and smaller minimum flexion (greater plantar flexion) for both metrics bilaterally, whereas the ML approach only found a difference in right ankle ROM. Main effects of method were found bilaterally across all metrics, with MB estimating larger amplitudes. Significant method differences within each sex were identified in both male and female data. Interaction effects were also observed in minimum flexion angles and right ankle ROM. Figure 15a presents mean ankle flexion waveforms during DVJ.

In the frontal plane, the MB system measured larger ankle inversion amplitudes compared to ML. As a result, main effects of method were found bilaterally in ROM, maximum adduction and adduction at IC, as well as in right minimum adduction. Within sex analyses further revealed that comparable differences and effect sizes were observed across sexes. A main effect of sex was not observed; however, ML did find that peak ankle inversion was significantly bilaterally. Figure 15b presents the mean ankle inversion waveforms during DVJ.

In the transverse plane, main effects of sex were identified for right ankle internal rotation ROM, maximum, and initial contact angles. Post-hoc within method comparisons further showed that

significantly larger maximum internal rotation and initial contact angles were observed in female participants. Main effects of sex were observed for minimum, maximum, and initial contact angles. Post-hoc analyses revealed that significant differences between methods were observed in both sexes, although greater differences were observed in females for initial contact angles specifically. Sex and method interactions were also observed for right ankle rotation ROM, maximum and initial contact angles. Figure 15c illustrates the mean ankle rotation waveforms during DVJ.

Table 15. Means, standard deviations, main effect *p*-values, and post-hoc analysis *p*-values and effect sizes from ankle DVJ kinematics. Main effects were evaluated with a linear mixed model, post-hoc analyses were evaluated with two sample *t*-tests and Cohen's *d* effect size. Orange and yellow indicate very significant ($p < 0.001$) and significant ($p < 0.05$) effect. Negative effect sizes indicate that female has larger mean than males in gender-within-methods comparison, or that ML has larger mean in method-within-gender comparison.

	Mean (SD) Female - MB	Mean (SD) Female - ML	Mean (SD) Male - MB	Mean (SD) Male - ML	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)	Method p within F (d)	Method p within M (d)
Right Ankle Flexion (°)											
ROM	76.55 (7.14)	63.74 (4.92)	69.00 (9.13)	58.24 (9.63)	0.006	< 0.001	0.093	0.00839 (-0.83)	0.0426 (-0.57)	< 0.001 (2.09)	< 0.001 (1.15)
Min	-43.22 (5.83)	-32.12 (4.37)	-38.04 (7.44)	-29.81 (7.33)	0.015	< 0.001	0.011	0.0174 (0.70)	0.265 (0.32)	< 0.001 (-2.16)	< 0.001 (-1.11)
Max	33.33 (3.87)	31.62 (3.39)	30.97 (4.73)	28.44 (6.10)	0.145	< 0.001	0.198	0.121 (- 0.50)	0.0703 (-0.52)	< 0.001 (0.47)	< 0.001 (0.46)
IC	-17.59 (8.73)	-11.69 (5.58)	-16.94 (13.21)	-11.33 (8.15)	0.835	< 0.001	0.854	0.863 (0.05)	0.878 (0.04)	< 0.001 (-0.81)	< 0.001 (-0.51)
Left Ankle Flexion (°)											
ROM	76.90 (8.06)	62.28 (4.48)	68.92 (9.40)	58.01 (8.44)	0.003	< 0.001	0.002	0.0109 (- 0.85)	0.0819 (-0.51)	< 0.001 (2.24)	< 0.001 (1.22)
Min	-43.01 (6.37)	-30.02 (3.42)	-36.86 (8.71)	-27.83 (7.74)	0.009	< 0.001	< 0.001	0.0216 (0.71)	0.304 (0.28)	< 0.001 (-2.54)	< 0.001 (-1.10)
Max	33.89 (3.93)	32.25 (3.23)	32.06 (4.47)	30.18 (5.27)	0.214	< 0.001	0.705	0.218 (- 0.41)	0.181 (- 0.39)	< 0.001 (0.46)	< 0.001 (0.39)
IC	-12.65 (7.74)	-5.97 (4.88)	-11.97 (11.53)	-5.36 (8.04)	0.802	< 0.001	0.958	0.83 (0.06)	0.775 (0.08)	< 0.001 (-1.03)	< 0.001 (-0.66)
Right Ankle Inversion (°)											
ROM	13.89 (5.94)	6.15 (1.90)	12.14 (4.40)	6.07 (2.00)	0.108	< 0.001	0.097	0.318 (- 0.40)	0.888 (- 0.04)	< 0.001 (1.75)	< 0.001 (1.78)
Min	0.32 (2.95)	3.83 (1.80)	0.81 (3.01)	2.76 (2.49)	0.525	< 0.001	0.011	0.618 (0.16)	0.126 (- 0.43)	< 0.001 (-1.44)	< 0.001 (-0.71)
Max	14.20 (4.58)	9.97 (1.29)	12.94 (3.66)	8.83 (1.75)	0.154	< 0.001	0.879	0.333 (- 0.34)	0.028 (- 0.65)	< 0.001 (1.26)	< 0.001 (1.44)

IC	6.48 (4.14)	7.80 (1.17)	8.43 (3.61)	7.49 (1.64)	0.029	0.004	0.001	0.146 (0.54)	0.526 (-0.19)	0.0112 (-0.43)	0.0182 (0.34)
Left Ankle Inversion (°)											
ROM	12.45 (4.46)	7.90 (3.07)	11.89 (4.03)	7.18 (3.22)	0.593	< 0.001	0.861	0.672 (-0.14)	0.483 (-0.22)	< 0.001 (1.19)	< 0.001 (1.29)
Min	0.85 (2.58)	1.23 (2.88)	0.82 (2.83)	0.60 (3.70)	0.966	0.474	0.430	0.964 (-0.01)	0.584 (-0.17)	0.454 (-0.14)	0.702 (0.06)
Max	13.30 (4.33)	9.13 (1.89)	12.71 (3.05)	7.78 (1.64)	0.498	< 0.001	0.252	0.621 (-0.19)	0.0173 (-0.82)	< 0.001 (1.25)	< 0.001 (2.01)
IC	8.33 (3.97)	6.72 (1.28)	7.26 (3.93)	5.90 (1.30)	0.198	0.002	0.737	0.381 (-0.27)	0.0412 (-0.64)	0.00256 (0.54)	0.00859 (0.47)
Right Ankle Internal Rotation (°)											
ROM	18.13 (4.96)	18.21 (4.24)	15.61 (4.34)	18.10 (3.84)	0.033	0.924	0.033	0.0899 (-0.58)	0.938 (-0.03)	0.927 (-0.02)	0.00156 (-0.61)
Min	-2.72 (2.93)	-7.92 (5.13)	-4.06 (3.65)	-9.78 (4.29)	0.309	< 0.001	0.502	0.233 (-0.37)	0.266 (-0.43)	< 0.001 (1.24)	< 0.001 (1.43)
Max	15.42 (5.78)	10.29 (3.87)	11.55 (4.50)	8.32 (2.38)	0.004	< 0.001	0.034	0.0249 (-0.86)	0.0717 (-0.83)	< 0.001 (1.04)	< 0.001 (0.89)
IC	6.98 (4.12)	5.18 (3.64)	4.20 (2.89)	4.42 (3.19)	0.016	< 0.001	0.001	0.0209 (-0.96)	0.518 (-0.24)	< 0.001 (0.46)	0.635 (-0.07)
Left Ankle Internal Rotation (°)											
ROM	19.05 (5.19)	16.90 (4.79)	18.63 (4.21)	18.31 (4.15)	0.768	0.002	0.064	0.794 (-0.10)	0.362 (0.34)	0.00702 (0.43)	0.593 (0.08)
Min	-3.34 (2.89)	-8.41 (4.86)	-4.55 (3.82)	-10.31 (5.11)	0.401	< 0.001	0.312	0.293 (-0.32)	0.281 (-0.37)	< 0.001 (1.27)	< 0.001 (1.28)
Max	15.72 (5.32)	8.48 (2.25)	14.09 (3.77)	8.00 (3.27)	0.175	< 0.001	0.142	0.294 (-0.43)	0.6 (-0.15)	< 0.001 (1.77)	< 0.001 (1.73)
IC	8.89 (3.79)	6.55 (3.07)	6.75 (4.03)	5.35 (3.77)	0.076	< 0.001	0.176	0.105 (-0.53)	0.302 (-0.32)	< 0.001 (0.68)	0.00341 (0.36)

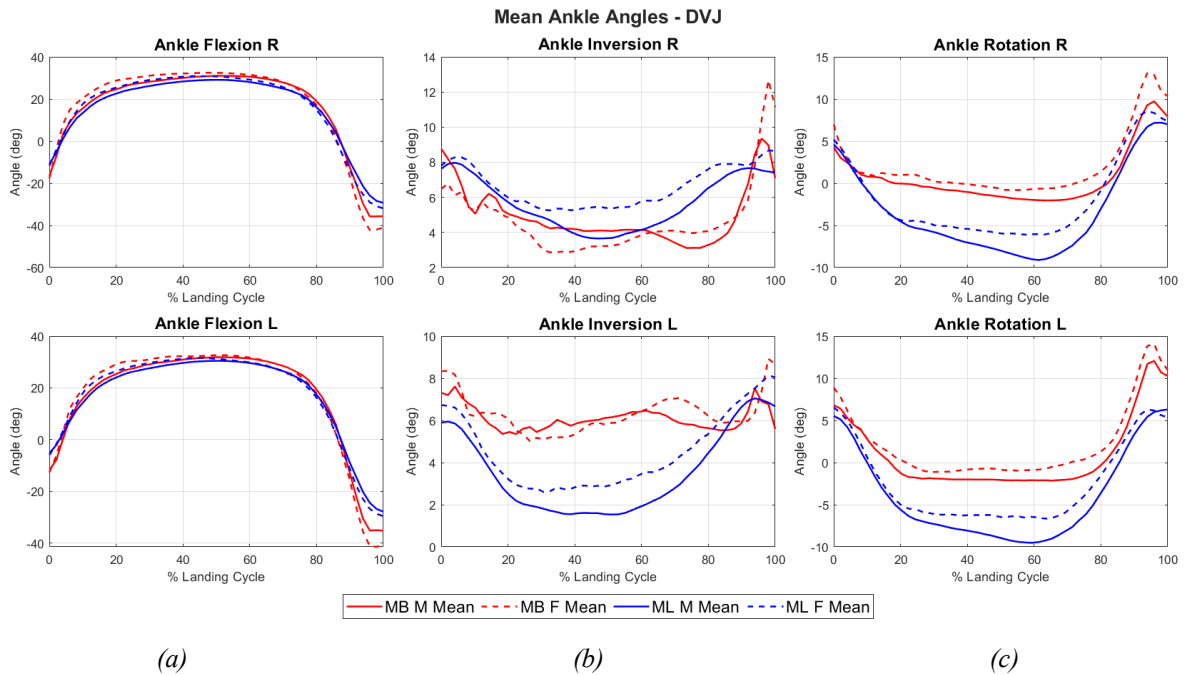


Figure 15. Mean ankle angle waveforms during DVJ task

4.5 Kinematic RMSD and Bias

In addition to the comparison of specific kinematic events, the RMSD and bias between motion capture approaches were used to determine if the differences are equivalent across sexes. This was evaluated by defining a LMM with a main effect of sex. Effect sizes and significance for each joint for RMSD and bias are listed in Table 16. Across all joints, the average difference between MB and ML data is larger in females than males, with means of $3.64 \pm 0.9^\circ$ and $3.28 \pm 1.0^\circ$, respectively. Significantly more RMSD was observed in female data for pelvis rotation, right ankle inversion and bilateral ankle rotation. However, moderate negative effect sizes ($d < -0.6$) were observed for right hip rotation, left hip flexion, and left ankle inversion, suggesting a tendency toward greater RMSD in females for these variables, even though these differences were not statistically significant.

Similarly, the average bias across joints was also larger in female participants with a mean of $-0.28 \pm 0.5^\circ$ compared to $-0.21 \pm 0.4^\circ$ in males. Significantly greater bias was found in pelvis rotation, right hip rotation, right ankle inversion, and right ankle rotation. Additionally, negative effect size

smaller than -0.6, indicating medium difference between groups were observed in right hip flexion, right ankle flexion, left knee adduction, and left ankle inversion.

Table 16. Mean, standard deviation, effect size, and significance of RMSD and Bias between MB and ML walking kinematics for male and female participants. Negative effect size indicates that mean of female data is greater than mean of male data. Yellow indicates that a significant difference ($p < 0.05$) was found.

Joint	RMSD (°)			Bias (°)		
	Mean (SD)	Mean (SD)	Effect Size	Mean (SD)	Mean (SD)	Effect Size
	Female	Male		Female	Male	
Pelvis Tilt	1.74 (0.33)	1.64 (0.17)	-0.59	-0.01 (0.04)	0.01 (0.05)	0.53
Pelvis List	2.49 (0.63)	2.13 (0.81)	-0.44	0.00 (0.02)	-0.00 (0.01)	-0.28
Pelvis Rotation	2.41 (0.72)	1.57 (0.63)	-1.32	-0.75 (0.78)	-0.05 (0.76)	-0.91
Hip Flexion R	3.22 (0.76)	3.14 (0.76)	-0.10	-0.13 (0.15)	-0.03 (0.18)	-0.60
Hip Adduction R	2.45 (1.28)	2.50 (1.44)	-0.12	-0.27 (0.58)	-0.37 (0.30)	-0.38
Hip Rotation R	4.49 (1.03)	3.86 (0.95)	-0.66	0.14 (0.61)	-0.40 (0.55)	-0.99
Knee Flexion	2.30 (0.58)	2.55 (2.01)	0.12	0.02 (0.10)	0.11 (0.15)	0.57
Knee Adduction R	3.26 (1.20)	2.74 (0.90)	-0.58	0.30 (0.48)	0.10 (0.53)	-0.37
Knee Rotation R	4.08 (1.16)	4.04 (1.18)	-0.03	0.09 (0.78)	0.37 (0.74)	0.37
Ankle Flexion R	2.07 (0.37)	2.03 (0.83)	-0.05	0.69 (0.29)	0.51 (0.25)	-0.70
Ankle Inversion R	7.33 (2.09)	5.76 (1.69)	-0.93	-1.59 (0.63)	-1.11 (0.63)	-0.75
Ankle Rotation R	6.80 (1.38)	5.60 (1.07)	-1.12	-2.81 (1.29)	-1.93 (0.89)	-0.99
Hip Flexion L	3.36 (0.86)	2.87 (0.73)	-0.67	-0.12 (0.17)	-0.02 (0.19)	-0.51
Hip Adduction L	2.52 (1.28)	2.86 (1.42)	0.36	-0.40 (0.50)	-0.50 (0.30)	0.33
Hip Rotation L	4.32 (1.30)	4.08 (0.93)	-0.25	-0.27 (0.61)	-0.47 (0.59)	0.34
Knee Flexion L	2.16 (0.42)	2.26 (1.95)	0.05	0.04 (0.12)	0.06 (0.11)	0.20
Knee Adduction L	3.21 (1.12)	2.86 (0.83)	-0.14	0.52 (0.64)	0.16 (0.57)	-0.63
Knee Rotation L	4.38 (1.22)	4.29 (1.34)	-0.06	-0.10 (0.77)	0.01 (0.67)	-0.17
Ankle Flexion L	1.94 (0.30)	2.20 (0.75)	0.35	0.50 (0.28)	0.41 (0.24)	-0.36
Ankle Inversion L	6.01 (1.63)	5.01 (1.52)	-0.66	-1.54 (0.75)	-1.09 (0.57)	-0.79
Ankle Rotation L	6.72 (1.49)	5.58 (1.08)	-1.05	-1.79 (1.03)	-1.39 (0.93)	-0.43

The mean RSMD for male and female participants, as well as the effect sizes for each joint during DVJ are presented in Table 17. Across all joints, the average RMSD for male and female participants was $4.34 \pm 2.6^\circ$ and $4.26 \pm 1.8^\circ$, respectively. Significant differences in RMSD were observed between sexes in pelvic list and right hip flexion, where the differences were greater in

males. In addition to the joints that found significant p-values, a large negative effect size ($d < -0.8$) was found in left ankle flexion, indicating larger differences in females.

The average bias across joints was also larger in male participants compared to females, with means of $-0.61 \pm 3.83^\circ$ and $0.36 \pm 2.29^\circ$, respectively. Yet significant difference in bias between sexes was only found for pelvic tilt and left knee flexion where the bias was greater in females.

Table 17. Mean, standard deviation, effect size and significance of RMSD and Bias between MB and ML DVJ kinematics for male and female participants. Negative effect size indicates that mean of female RMSD is greater than mean of male data. Yellow indicates that a significant difference ($p < 0.05$) was found.

Joint	RMSD			Bias		
	Mean (SD) Female	Mean (SD) Male	Effect Size	Mean (SD) Female	Mean (SD) Male	Effect Size
pelvis tilt	4.08 (1.96)	5.01 (2.82)	0.33	3.13 (2.25)	0.09 (4.08)	-0.75
pelvis list	1.31 (0.44)	2.50 (2.19)	0.54	0.30 (0.69)	0.00 (1.73)	-0.17
pelvis rotation	1.24 (0.38)	1.41 (0.49)	0.35	-0.25 (0.47)	-0.50 (0.70)	-0.36
hip flexion r	4.10 (1.41)	5.98 (3.15)	0.60	1.34 (2.57)	-1.12 (4.47)	-0.55
hip adduction r	3.69 (2.03)	3.19 (1.30)	-0.39	-2.02 (2.65)	-0.55 (2.49)	-0.59
hip rotation r	4.32 (1.27)	4.65 (2.10)	0.16	-0.88 (2.70)	-0.58 (3.00)	-0.10
knee flexion r	4.20 (1.56)	3.56 (1.02)	-0.63	-2.42 (1.72)	-1.81 (1.32)	-0.47
knee adduction r	5.73 (2.14)	4.41 (2.25)	-0.59	3.02 (3.22)	2.39 (2.93)	-0.22
knee rotation r	4.51 (1.78)	3.86 (1.24)	-0.53	0.39 (2.85)	-0.89 (2.53)	0.51
ankle flexion r	4.79 (1.26)	4.41 (1.23)	-0.31	0.77 (0.85)	1.14 (1.32)	0.28
ankle Inversion r	3.55 (1.24)	3.32 (0.96)	-0.25	-1.87 (1.83)	-0.82 (2.09)	-0.51
ankle rotation r	5.28 (1.57)	5.17 (1.90)	-0.06	3.88 (2.39)	3.84 (2.07)	-0.02
hip flexion l	4.51 (1.61)	6.06 (3.86)	0.40	1.31 (2.84)	-1.81 (4.68)	-0.67
hip adduction l	3.14 (1.75)	3.47 (1.42)	0.23	-0.58 (2.82)	0.73 (2.64)	0.50
hip rotation l	4.80 (1.74)	5.10 (2.20)	0.14	-1.14 (3.33)	-0.01 (3.38)	-0.34
knee flexion l	5.10 (3.89)	4.62 (1.24)	-0.39	-2.64 (2.05)	-2.81 (1.16)	0.14
knee adduction l	4.91 (2.15)	4.11 (2.13)	-0.38	1.29 (3.69)	1.51 (2.78)	0.08
knee rotation l	4.99 (1.98)	5.67 (2.64)	0.26	-2.23 (2.95)	-3.78 (3.79)	0.41
ankle flexion l	5.66 (3.42)	4.56 (1.23)	-0.89	-0.37 (1.64)	0.21 (1.53)	-0.38
ankle Inversion l	3.86 (1.74)	4.47 (1.97)	0.31	1.83 (2.36)	2.48 (2.57)	0.25
ankle rotation l	5.72 (2.03)	5.64 (1.43)	-0.06	4.63 (2.15)	4.78 (1.65)	0.09

5. Discussion

The purpose of this research project was to investigate two research questions relating to the comparison of MB and ML motion capture systems. The first objective was to evaluate whether both motion capture systems identify sex-based differences in anthropometrics and kinematic differences as reported in literature. Significant main effects of sex were observed for pelvis-to-hip distance, hip width, thigh length, and shank length. These findings align with well-established evidence that males exhibit longer lower limb segments, while pelvis-related dimensions do not scale uniformly with overall body size [29], [30], [57], [59]. In the kinematic analyses, across walking and DVJ tasks and across methods the results largely support the findings of the literature review that sex-based differences are most consistently expressed in the frontal planes, particularly at the pelvis, hip, and knee. Additionally, MB tends to estimate larger adduction angles, while ML estimates smaller rotation angles, especially at the knee.

The second objective was to evaluate the extent to which alignment of the LCS reduces angle differences and bias between MB and ML data and assess whether the amount of angle RMSD and bias differs between sexes. The results of the segment alignment protocol demonstrate that a substantial portion of the kinematic differences between systems can be attributed to differences in the LCS orientation, particularly at the knee and ankle joints, which had the least amount of differences between motion capture in joint centre estimations. Across all joints, the mean RMSD and bias were reduced from $3.12^\circ \pm 2.8^\circ$ and $0.25^\circ \pm 0.1^\circ$ to $2.12^\circ \pm 2.3^\circ$ and $0.01^\circ \pm 0.1^\circ$ following alignment. Transverse planes had the greatest amount of initial difference and the largest reduction in RMSD between systems was observed in right and left ankle rotation which decreased from 14.38° and 11.76° to 5.48° and 5.51° . These findings support recent literature suggesting that direct comparisons between MB and ML pipelines without accounting for measurement differences may overestimate systematic variance and limit interpretability [32]. The comparison of joint centre estimation from MB and ML approaches revealed significant differences and large negative effect sizes for almost all lower limb joints indicating larger differences between motion capture approaches in female participants. These results translated to the comparisons of the kinematic data which saw medium or large negative effect sizes for many joints and planes of motion across both walking and DVJ trials suggesting that the agreement between the ML and MB systems may be lower when evaluating female participants. However, the comparison of RMSD between

kinematics from each approach through entire trials did not find many significant differences between female and male data.

5.1 Anthropometric measurements and joint centre comparison

Across motion capture methods, males exhibited greater thigh and shank lengths than females, which is consistent with previous studies that have reported longer femoral and tibial dimensions across diverse populations. The observed hip width ranging from 16.3 cm to 17.1 cm across methods and sexes further supports the notion that pelvis geometry is sex-dependent rather than stature-dependent. This agrees with imaging and radiographic studies demonstrating sex-specific variations in pelvis widths and HJC positioning relative to the mid-line of the pelvis [12], [63]. The magnitude of the main effects of sex observed in this study for shank length and hip width, suggest that both motion capture systems are generally sensitive to biologically meaningful anthropometric measurements.

The segment lengths measured in this study were comparable to those reported in the literature. The mean femoral length measured using the MB system was approximately 42.2 ± 2.8 cm in males and 40.3 ± 2.1 cm in females. These values fall very close to the lengths reported by Moosa et al of 43.6 ± 1.9 cm and 40.2 ± 2.2 cm [59]. The slightly longer femoral lengths reported by Moosa et al., despite their shorter mean population height, can likely be attributed to methodological differences. Moosa et al. measured from the distal lateral femoral condyle to the top of the femoral head, whereas the present study defined the femoral length based on the joint centre locations. In contrast, the ML system measured femoral lengths that were shorter than those stated in the literature, and only the MB system found significant differences between the sexes for femoral length. Given that sex-based differences in femoral lengths have been well established because they are linked to stature, this suggests that Theia3D may underrepresent femoral length differences between sexes.

For tibial length, previous studies have also reported that shank length is linked to stature and reported significant sex-based differences [58], [60]. In the present study, both the MB and ML approaches found significant differences between sexes, with no significant method effect observed. In contrast, hip width measurements showed greater variability in the literature. While Hara et al. reported finding no statistically significant differences in inter-hip width between skeletally mature males and females, Alazzawi et al. reported significant differences in the distance

from the centre of the femoral head to the pelvic midline, a measure equivalent to half the inter-hip width [12], [30]. Alazzawi et al. reported mean distances of 90.6 ± 5.6 mm for males and 87.8 ± 5.6 mm for females [12]. In this study, both MB and ML found significant differences between the hip widths, although both methods measured shorter widths than those reported by Alazzawi, likely reflecting differences in population characteristics and measurement approaches [12].

Despite both motion capture approaches agreeing on the presence of sex-based anthropometric differences, significant main effects of method were observed for pelvis-to-hip distance and thigh length, as well as significant sex-by-method interactions for pelvis-to-hip distance, hip width, and thigh length. These findings indicate that MB and ML systems do not estimate all anthropometric measures equivalently, and that discrepancies are not uniform across sexes.

Both motion capture methods demonstrated excellent within-participant consistency, with high ICC values and low coefficients of variation across trials (Appendix D – Within-trial and between-trial comparison). ML motion capture showed near-zero variation within individual trials, reflecting the rigid enforcement of segment lengths within each generated skeletal model. However, between-trial variability was slightly greater for the ML system, particularly for pelvis and thigh-related measures. These distinctions highlight a significant methodological consideration that while ML system may enforce strong consistency within a single trial, generating a new skeletal model for each trial can introduce additional variability. In contrast the MB approach, applies a single model across all trials for a given participant, resulting in better between-trial consistency.

The MB system consistently estimated longer femoral lengths than the ML system for both males and females. Because no significant method effects or sex-by-method interaction were observed in shank length and small systematic joint centre offsets were observed at the knee and ankle, differences in the HJC estimation are the most likely source of disagreement in femoral length. The reduced systematic offset at the knee and ankle compared to the hip and pelvis likely reflect the presence of clearer visual landmarks and reduced soft-tissue artefacts around the distal joints.

Analysis of the joint centre offsets further support this interpretation. The largest systematic offset between the methods occurred in the axial direction at the pelvis, where the ML pelvis centre located approximately 2.94 cm distal to the MB pelvis centre. A comparable distal offset of 2.69 cm is also observed at the HJC. Because both the pelvis origin and the HJC are shifted distally by

a similar magnitude, the results suggest that the ML pelvis is positioned lower relative to the MB pelvis, rather than the HJC alone being estimated differently.

The modelling approached used by the two systems support this interpretation. In the MB system, the pelvis was defined using markers placed on the anterior-superior iliac spines and the posterior-superior iliac spines to define the Visual3D composite pelvis. Subsequently, the hip joints were estimated using the Harrington method. The Theia3D model also uses a Visual3D composite pelvis. While Theia3D does not reveal what method is used to estimate the HJC, the Bell method has previously been recommended in past software versions [17], [46], [100]. Because the pelvis-to-hip distance is very comparable between systems, it is reasonable to assume that Theia3D defines the HJC relative to the pelvis, rather than other hip related landmarks.

Together, these findings suggest that the shorter femoral lengths observed in the ML system are primarily due to Theia3D consistently estimating the pelvis landmarks to be located lower relative to the pelvis landmarks identified through palpation. Because the HJC are estimated relative to the pelvis origin, this downward shift of the pelvis resulted in shorter estimated femoral segment lengths.

With respect to the RMSD between the joint centre estimations for MB and ML systems, significant sex-based differences were observed at the pelvis, right hip, left hip, and right ankle. Although not all joints demonstrated statistically significant differences in RMSD between sexes, large negative effect sizes were observed for all joints, ranging from -0.92 to -1.44, apart from the left hip. These effect sizes indicate consistently larger RMSD values in female participants. As a result, the lack of significance at some joints likely reflects the limited statistical power rather than the absence of a meaningful sex-based difference.

5.2 Segment Alignment

Prior to alignment, females generally demonstrated larger RMSD and bias than males in several joints, particularly for pelvic list, hip rotation, and ankle rotation. However, this pattern was not consistent across all variables. Following the alignment of the MB LCSs to the corresponding ML LCS, RMSD and bias were generally reduced across both sexes and movement tasks. During walking, substantial reductions were observed for female hip rotation (8.73° to 3.48°) and ankle rotation (16.14° to 6.11°), while during the DVJ notable reductions were observed for pelvic tilt,

hip rotation, knee rotation, and ankle rotation. Female ankle rotation RMSD decreased from 11.39° to 5.28° , while female pelvic tilt RMSD decreased from 10.07° to 4.08° . These findings indicate that differences in LCS orientation contribute substantially to the observed disagreement between the MB and ML systems.

The reduction in bias was particularly evident at the hip and pelvis where raw offsets of several degrees were reduced to values approaching zero. For example, during walking, male hip flexion bias decreased from -3.69° to -0.01° , while female hip rotation bias decreased from 7.92° to 0.15° . These findings suggest that some of the systematic differences observed between MB and ML systems may reflect differences in segment coordinate system definitions rather than differences in the underlying movement. These results also highlight that systematic offsets commonly reported in literature overestimate bias between the systems. The 2024 Apollo update to Theia3D introduced a pelvis definition with approximately 10° of anterior tilt, aligning the ML pelvis orientation more closely with the Visual3D composite pelvis and reducing hip flexion RMSD to approximately 5° [46]. The present findings extend these observations by demonstrating that explicit alignment of the MB and ML LCSs can further reduce RMSD and systematic bias across multiple joints and movement tasks. However, LCS alignment should primarily be considered as a methodological approach for validation and cross-system comparison.

Capturing out-of-plane motion is particularly challenging for the ML system. Theia3D relies on computer vision-based pose estimation. While this approach performs well for large, sagittal plane motions, such as knee flexion during walking, it is less sensitive to smaller amplitude movements in the frontal and transverse planes. Several studies have reported that Theia3D has underestimated the magnitude of range of motion in frontal and transverse planes when compared to the MB system [8], [17]. These discrepancies are often attributed to challenges in resolving subtle segment rotations from video data, occlusion of anatomical landmarks, and limited visual cues for axial rotation [18]. As a result, even when the waveforms appear similar in shape, the absolute magnitude of the joint excursion may reflect limitations in both measurement approaches rather than true differences

Additionally, the use of a single average correction matrix per joint assumes a constant rotational offset throughout the movement, which may not fully capture the dynamic differences in segment tracking between systems. As such, residual disagreement in these planes likely reflects a

combination of measurement noise, task-dependent tracking differences, and limitations of static alignment approaches.

5.3 Kinematic sex-based differences

5.3.1 Walking

The literature review identified pelvis list, hip adduction, hip internal rotation, and ankle dorsiflexion as the most consistent sex-specific kinematic differences during walking. The present results showed that both MB and ML motion capture found larger pelvis list, hip adduction, and hip internal rotation in females indicating that these characteristics are stable sex-specific differences during walking. Much like the literature, sex-specific differences in the sagittal plane were mixed, with neither system consistently detected larger hip or knee flexion in females.

Sex differences at the knee were less consistent than those observed at the pelvis and hip. Both methods identified greater frontal-plane knee motion in females; however, the magnitude and statistical significance of these differences varied between systems. These findings agree with Phinyomark et al. who saw significant difference at peaks and during the landing; however, the ROM estimated by ML is noticeably smaller ($\sim 5^\circ$ vs $\sim 10^\circ$) [54]. For internal knee rotation, both methods found that males achieved significantly greater external rotation at initial contact compared to females. While previous studies comparing knee internal rotation angles have found greater angles in males, significance has not been established [44], [52], [103].

Sex-based kinematic differences at the ankle were less consistently reproduced than those observed at the proximal joints. Although both systems generally indicated greater ankle dorsiflexion in females, markerless detected a larger number of statistically significant differences, suggesting greater method sensitivity at the ankle.

In addition to the joint movements where both or neither motion capture system found significant differences between sexes, there were also instances that saw disagreement. Pelvic tilt did not show a main effect of sex at the group level, which is consistent with much of the literature that has reported non-significant differences in pelvic tilt ROM or peak amplitudes during walking [13], [24], [28], [71]. However, sex-within-method post-hoc analyses revealed that the ML system identified sex-based differences, with females demonstrating greater pelvic tilt ROM and a more

posterior pelvic tilt. This suggests that small sagittal-plane pelvis differences may be more sensitive to methodological differences than the larger frontal-plane characteristics.

Pelvis rotation demonstrated a different pattern. Previous literature generally reports larger rotational ROM in females [13], [28], [71]. In the present study, only MB found a larger mean ROM in females (8.79° vs 7.99°), whereas ML saw that males achieved larger ROMs (7.89° vs 8.51°). This reversal indicates that subtle transverse-plane sex differences may depend on the motion capture approach and should therefore be interpreted cautiously.

Overall, both motion capture systems reproduced the principal sex-specific kinematic characteristics of walking described in previous literature, particularly greater frontal- and transverse-plane pelvis and hip motion in females. Differences between methods were primarily confined to smaller or more subtle knee, ankle, and pelvic rotation variables, suggesting that both systems can identify the major sex-specific movements during gait, but caution should be used when choosing a system to evaluate subtle frontal and transverse plane differences.

5.3.2 DVJ

Compared with walking, the DVJ produced larger and more pronounced sex-based kinematic differences, consistent with previous literature. Both MB and ML motion capture identified greater anterior pelvic tilt, increased peak knee valgus, and larger ankle dorsiflexion ROM in females, indicating that both systems successfully captured the principal sex-specific landing characteristics reported in previous studies. The agreement between methods for these variables suggests that markerless motion capture can identify the largest and most clinically relevant sex differences associated with landing mechanics.

In contrast, several sex-specific differences consistently reported in the literature were not reproduced by either motion capture approach. Although previous studies have shown that females demonstrate greater hip adduction during DVJ, neither MB nor ML identified significant differences between sexes. Likewise, neither system saw sex-specific differences in hip flexion or hip rotation. However, both systems found a medium effect size for hip rotation suggesting that there is likely a difference between groups, but that the sample was not large enough to establish significance. These findings suggest that sex differences at the hip during DVJ are less robust than

anticipated in the present cohort and may be influenced by participant characteristics or experimental protocol.

At the knee, both systems consistently identified greater peak knee valgus in females, one of the most widely reported biomechanical characteristics associated with female landing mechanics. However, agreement was less consistent for other knee variables. Only MB detected greater knee valgus ROM in females, whereas only ML identified greater peak knee adduction. Similarly, neither method detected differences in knee valgus at initial contact despite moderate effect sizes observed in the markerless data, suggesting that the study may have been underpowered to detect these smaller differences. In the transverse plane, MB identified greater knee rotational ROM and peak rotations in females, whereas previous studies have generally reported greater internal rotation in males. This discrepancy indicates that rotational knee kinematics remain highly sensitive to methodological and experimental differences.

At the ankle, both methods identified greater sagittal-plane ROM in females, consistent with previous reports of increased ankle dorsiflexion during landing. In contrast, frontal- and transverse-plane ankle differences were less consistently reproduced. Markerless alone identified greater ankle inversion in females, whereas only MB detected greater internal ankle rotation. These findings suggest that although sagittal-plane ankle kinematics are reproduced consistently across methods, smaller rotational and frontal-plane differences remain method dependent.

Overall, both motion capture systems reproduced the principal sex-specific landing characteristics described in the literature, particularly greater anterior pelvic tilt, knee valgus, and ankle dorsiflexion in females. However, expected differences at the hip were not consistently observed, and agreement between methods decreased for smaller frontal- and transverse-plane movements, particularly rotational kinematics. These findings suggest that Theia3D and marker-based motion capture can identify the major sex-specific movement strategies during landing but should be interpreted more cautiously when evaluating subtle hip and rotational joint mechanics.

5.4 Kinematic method-based differences

MB and ML motion capture differ fundamentally in how joint kinematics are estimated, and systematic differences between the two approaches have been reported previously, often attributing the differences to offsets in segment orientation and joint centre estimation [17], [22], [44].

Although the present study standardized data collection through consistent lighting, tight-fitting clothing, and marker-placement by a single researcher, significant main effects of method remained across both walking and DVJ. Additionally, most of the bias between systems caused by unaligned LCSs was removed through rotation of the LCSs; however, ROM differences persisted in frontal and transverse planes, with mean RMSD and bias reduced from $3.12^\circ \pm 2.8^\circ$ and $0.25^\circ \pm 0.1^\circ$ to $2.12^\circ \pm 2.3^\circ$ and $0.01^\circ \pm 0.1^\circ$ following alignment. While both systems often identified similar directional trends in sex-based kinematics, the absolute magnitudes differed, and these discrepancies were frequently sex-dependent.

To contextualize the performance of both MB and ML systems, comparison to biplane radiography provides a valuable perspective. One study evaluating knee kinematics in healthy young adults found an average ROM during gait of $63.7 \pm 15^\circ$, $3.7 \pm 6^\circ$, and $11.5 \pm 10^\circ$ in sagittal, frontal, and transverse planes [87]. When comparing to those observed by ML in the present study, which estimated ROMs of $65.4 \pm 3.6^\circ$, $5.5 \pm 1.5^\circ$, and $7.0 \pm 2.1^\circ$ (female left knee) in flexion, adduction, and rotation respectively. While the sagittal plane ROM shows excellent agreement with the gold standard, both the mean and variability differ substantially in non-sagittal planes. The frontal plane ROM is slightly elevated compared to biplane radiography, though the standard deviation is considerably smaller (1.5° vs 6°), likely reflecting kinematic constraints within the Theia3D model that limit extreme values. For knee rotation, however, the ROM reported by Gale et al. is noticeably larger than what was estimated by ML, with ML only capturing a small portion of the rotational amplitude observed via biplane radiography. The MB system estimated knee rotation ROM of $14.9 \pm 3.2^\circ$, which is much closer to that of Gale et al. and exceeded the biplane radiography values. MB also overestimated knee adduction ROM at $11.8 \pm 3.2^\circ$ compared to 3.7° via biplane radiography. These overestimations by the MB system can likely be explained by soft-tissue artefacts, combined with the fact that kinematic constraints were not added to the MB model. The absence of physiological joint constraints allows the MB system to track marker motion that may not represent true bone motion, particularly in planes with small ROM where soft-tissue displacements become proportionally more influential. Additionally, kinematic cross-talk caused by the misalignment of LCS, has been shown to produce non-uniform error in frontal and transverse plane joint angles [67], [104], [105]. Errors caused by kinematic cross-talk can be present in MB and ML data; however, in the present study cross-talk error is expected to be larger in MB because the LCS defined using the raw marker trajectories are subsequently rotated to

aligned with the corresponding ML coordinate system, a step that can amplify any underlying LCS mis-alignment.

5.4.1 Sex x Method interactions

One of the principal findings of this study was the prevalence of significant sex-by-method interactions across both walking and DVJ. These interactions demonstrate that differences between MB and ML motion capture cannot be explained by a simple systematic offset. Instead, the magnitude and direction of disagreement varied with participant sex and movement task, indicating that the agreement between the systems can be influenced by sex. Although interactions were observed throughout the lower limb, they were most common at the pelvis, hip, knee, and ankle in the frontal and transverse planes.

During walking, interactions were most frequently observed at the pelvis and hip. Females generally demonstrated larger method differences for pelvic tilt, hip flexion, and hip rotation, whereas pelvic list and some frontal-plane hip variables showed larger discrepancies in males. Despite these isolated exceptions, the overall pattern indicated greater disagreement between MB and ML in female participants, particularly for variables describing pelvis orientation and rotational hip motion. Because these joints also exhibit the largest anatomical sex differences, the findings suggest that underlying morphology may influence markerless joint centre estimation and segment orientation.

During DVJ, sex-dependent method differences became more pronounced at the hip, knee, and ankle. Females generally exhibited larger discrepancies for hip adduction, knee flexion, knee valgus, and ankle flexion, whereas only a few variables demonstrated larger differences in males. The greater joint excursions and higher angular velocities associated with landing likely amplify differences between MB and ML, particularly in frontal- and transverse-plane movements where markerless motion capture has previously shown reduced agreement.

Several factors may contribute to these findings. Segment length analysis demonstrated larger method differences in thigh length and pelvis-to-hip distance in females, while joint centre analyses identified larger offsets at the pelvis, hip, and ankle. Because frontal- and transverse-plane kinematics are particularly sensitive to joint centre location and segment orientation, these anatomical differences are likely propagated into larger sex-dependent kinematic discrepancies.

The concentration of interactions at the pelvis and hip further supports this interpretation, as these regions exhibit the greatest anatomical differences between males and females.

Consistent with the interaction analyses, RMSD also demonstrated sex-dependent differences. During walking, females generally exhibited larger RMSD at the pelvis and ankle, whereas during DVJ the pattern became more variable, with larger RMSD observed in males for several variables. This may partly be explained by the higher BMI observed in male participants (26.0 vs 23.5), which could have created an increased challenge for accurate marker placement and greater visual obstruction of the pelvis and hip during DVJ, especially at peak loading. Despite these task-specific differences, the overall trend remained toward larger disagreement in females, supporting the conclusion that method agreement is influenced by participant sex.

The findings that disagreement was most pronounced in the frontal and transverse planes aligned with prior literature demonstrating reduced agreement between MB and ML systems for movements outside the sagittal plane [11], [32], [37], [107], [108]. Because the movements in these planes are less pronounced than flexion, they are particularly sensitive to joint centre estimation and segment orientation differences. The segment length analysis supports this interpretation. Significant sex-by-method interactions were observed for pelvis-to-hip distance and thigh length, and ML estimated shorter thigh lengths relative to MB in both sexes, though the magnitude of difference was greater in females. Additionally, joint centre offset analyses revealed significantly greater differences and negative effect sizes in females at the pelvis, right hip, and right ankle. These differences in joint centre estimations may disproportionately influence kinematics in females.

5.4.2 RMSD differences

In addition to comparing the methods during specific kinematic events, the overall RMSD between methods were analysed. For the walk, significant differences in RMSD were observed in pelvis rotation, right ankle rotation and right ankle inversion, which all found large negative effect sizes ($d < -0.93$). Additionally, medium negative effect sizes ($d < -0.66$) were found for left and right hip flexion and left ankle inversion. In contrast, during DVJ, significant differences in RMSD between sexes and positive effect sizes were found for pelvis list and right hip flexion ($d < 0.6$), as well as left ankle flexion ($d = -0.89$). RMSD increased in both sexes during landing tasks due to increased speed and complexity of the movement. However, the significant increased observed in

males may be related to the higher average BMI of the male participants. Overall, the mean effect size in both walking and DVJ is negative. While not significant, it highlights the trend that differences are greater in female participants.

5.5 Implications

The finding that differences between MB and ML kinematics are not evenly distributed between males and females has important implications for research and clinical applications. If ML motion capture estimates smaller frontal and transverse plane motion in one sex relative to the other, studies comparing groups may underestimate between-group differences. This is particularly relevant for variables such as hip internal rotation or knee adduction, where differences can be small in magnitude but clinically meaningful. Attenuation of these differences could increase the likelihood of type II errors or reduced observed effect sizes, potentially influencing how sex-specific movement strategies are interpreted.

These results also suggest that findings from MB studies may not directly translate to ML-derived data. In the present study, method choice influenced not only the magnitude of observed sex-based differences, but in some cases whether those differences reached statistical significance. As such, assumptions of interchangeability between systems may not be appropriate without prior validation in the specific population and movement. As ML systems such as Theia3D become more widely adopted, researchers and clinicians should be cautious when comparing results across methodologies. If agreement between systems differs across sexes, predictive models developed using ML-derived data may not generalize equally across populations.

5.6 Limitations

Several limitations should be considered when interpreting these findings. First, the present sample consisted of 28 participants (14 per sex). While this sample size is adequate to detect moderate-to-large effects, it lacked power to identify smaller sex-specific differences in measurements. A larger sample would likely provide more stable estimates of variance and allow clearer identification of subtle interactions between sex and method. This highlights the importance of evaluating potential sex-specific biases when adopting new motion capture technologies, particularly when studying variables that are known to differ between males and females.

Secondly, the healthy and predominantly young adult sample (27 ± 4 years) limits generalisability to children, older adults, or clinical populations. Additionally, although the male and female groups were representative of expected anthropometric differences within healthy adults, significant differences in height, weight, and BMI existed between groups. Consequently, body size represents a potential confounding factor when interpreting the observed sex-specific differences as the independent effects of sex were not isolated from the anthropometric characteristics. To explore this limitation, a supplementary analysis was performed on a subset of similarly sized participants. Results of this analysis are presented in Appendix H – LMM Results of Similarly Sized Participants. Despite the reduced variability in height and BMI, many of the principal sex effect and sex-by-method interactions effects remained, suggesting that body size alone does not explain the observed findings. However, the substantially smaller sample size limited statistical power of the analysis. Additionally, the study compared ML against MB rather than a true gold standard such as biplane radiography. While MB served as an appropriate reference given its widespread use and the existence of established sex-based findings in the literature, both systems are subject to error – MB through soft tissue artifact and marker placement variability, ML through the limitations discussed above.

There were also methodological limitations related to data acquisition. During collection, four of the eight Vicon Vue cameras were mounted at a height of 2.5 meters, which is at the upper end of the range Theia3D recommends for camera setup. The increased downward viewing angle may have contributed to partial visual obstruction of the pelvis and thigh segments, potentially influencing pelvis and HJC estimations and contributing to the shorter femur lengths estimated by ML. Additionally, the sampling frequency was limited to 80 Hz to allow for synchronized MB, ML, and Move4D acquisition. While sufficient for walking, this frequency may have limited temporal resolution during the DVJ particularly during high-velocity phases such as landing.

There are also challenges with marker data quality. Larger gaps were observed in the frontal pelvis markers during peak loading of the DVJ, which may have influenced MB kinematic estimates in the frontal and transverse planes. These tracking inconsistencies could have contributed to some of the variability observed between systems.

Model definition differences represent another important limitation. The sensitivity analysis identified the MB model that best aligned with Theia3D joint centres and kinematics; however, the

selected MB model did not include kinematic constraints. In contrast, Theia3D incorporates active joint constraints, though the exact constraint parameters beyond DOF are not publicly available. The consistently smaller frontal and transverse plane amplitudes estimated by ML may reflect reduced sensitivity, model-imposed limits, or a combination of both. Without full transparency regarding model constraints, it is difficult to determine the extent to which disagreement arises from tracking error versus modelling assumptions.

Finally, in post-processing, it was assumed that the orientation difference between MB and ML LCSs remained constant throughout each trial. While this assumption allowed for practical alignment of the systems, it is unlikely that inter-system differences are constant across dynamic movements.

6. Conclusion

6.1 Summary

With the constant improvement of ML motion capture technologies, there is increasing demand in deploying these machine-learning based motion capture systems such as Theia3D in both laboratory and field settings. However, female participants continue to be underrepresented in biomechanical research, and some vision-based machine-learning systems have demonstrated population-dependent performance in other domains. As a machine-learning based system it is important to evaluate whether Theia3D and MB systems demonstrate consistent agreement across sexes.

This project was conducted to evaluate the performance of Theia3D ML motion capture relative to MB motion capture during a simple walking task and a more dynamic DVJ. The objectives were to compare anthropometric measurements and kinematics of the lower-limbs between systems to values reported by previous studies evaluating sex-specific differences, and to compare the agreement between the systems to determine whether differences were consistent across sexes. To improve comparison of the systems, LCS were aligned using a mean transformation matrix that reduced systematic segment orientation bias prior to kinematic comparison. To answer these questions, a data collection was conducted with 28 participants, performing walking and DVJ, for

three consecutive repetitions, and simultaneously tracked by both the MB and ML motion capture systems.

Alignment of the segment coordinate systems substantially reduced overall RMSD and bias between methods, demonstrating that a considerable portion of the disagreement arises from modelling and coordinate system definitions rather than tracking error alone. However, it is important to ensure that MB segment coordinate systems are properly defined when creating the model to reduce kinematic cross-talk once the coordinate systems are transformed.

For sex-based differences, both systems demonstrated sensitivity to biologically meaningful anthropomorphic differences. MB and Theia3D identified significant sex-based differences in inter-hip width and shank length consistent with prior literature. However, systematic differences in femoral length estimates were observed, and sex-by-method interactions indicated that these differences were not uniform across sexes. Joint centre offset analyses further revealed larger systematic offsets at the pelvis and hip, particularly in female participants.

Both MB and Theia3D reproduced several sex-specific movement patterns reported in the literature. During walking, both systems identified greater pelvis list, hip adduction, and hip internal rotation in females, while during DVJ both detected greater anterior pelvic tilt, knee valgus, and ankle dorsiflexion in females. These findings demonstrate that Theia3D is capable of identifying the major sex-specific movement strategies observed during common functional tasks. However, MB motion capture identified a greater number of significant differences in frontal and transverse plane kinematics, whereas Theia3D generally estimated smaller rotational amplitudes and was less consistent in detecting more subtle sex-specific differences.

One of the principal findings of this study was the presence of frequent sex-by-method interactions, demonstrating that differences between MB and Theia3D cannot be explained by a simple systematic offset. Instead, agreement between systems depended on participant sex, movement task, and joint plane. Females generally exhibited larger joint centre offsets, greater kinematic RMSD, and larger method differences at the pelvis, hip, and ankle, suggesting that agreement between the systems is not sex neutral. These findings emphasize that validation of the Theia3D markerless system should extend beyond overall accuracy metrics and evaluate performance separately across male and female populations.

From a practical perspective, the results indicate that Theia3D is well suited for identifying broad sex-specific movement patterns during walking and landing tasks. However, practitioners should exercise greater caution when interpreting the magnitude of sex differences in frontal- and transverse-plane kinematics, particularly at the pelvis and hip where agreement with MB was less consistent and frequently sex dependent. Where possible, studies investigating subtle biomechanical differences between groups should validate findings against established reference methods.

Finally, these findings have important implications for both biomechanics research and clinical applications. Populations in which frontal and transverse plane motion is of primary interest, including individuals with knee osteoarthritis, ankle instability, joint hypermobility, cerebral palsy, or other musculoskeletal disorders, often exhibit clinically meaningful movement differences that are relatively small in magnitude. The present results suggest that these subtle kinematic differences may be influenced by both the motion capture method and participant sex. As Theia3D motion capture continues to become more widely adopted, ensuring that the system is validated across diverse populations will be essential for accurate and equitable assessment of human movement.

6.2 Future Work

Several areas for future research emerge from the findings of this study. First, further investigation into factors that could influence Theia3D performance are required. While this study examined the main effects of sex, other participant characteristics such as height, weight, BMI, body composition, and individual anatomical morphology may also contribute to the observed differences between marker-based and markerless motion capture. Future studies should aim to isolate the influence of these variables using larger and more diverse participant cohorts, allowing researchers to determine whether the observed sex-dependent differences are primarily driven by biological sex itself or by associated anthropometric characteristics.

Beyond participant-specific characteristics, the population-level generalisability should also be considered. The present study focused on healthy young adults; however, agreement between systems may vary across populations with different ages, ethnicities, or clinical conditions. Future research should therefore evaluate Theia3D across diverse demographic groups, including participants with joint injuries or movement impairments, particularly conditions associated with

clinically meaningful alterations in frontal and transverse plane kinematics, to determine whether Theia3D can accurately capture deviations from typical movement patterns.

Expanding comparisons to additional movements would strengthen the understanding of task-dependent agreement. Activities with established sex-specific kinematic patterns such as running, single-leg landings, or change-of-direction manoeuvres, may provide further insight into how differences between systems vary across movement demands. Examining movements under fatigue conditions may also be valuable, as fatigue is known to alter neuromuscular control and joint mechanics. Together, these future directions would help clarify the contexts in which ML motion capture can be confidently applied and where caution remains necessary.

Further investigation into model-based differences between the Theia3D model and ground-truth skeletal measurement is also warranted. Neither MB nor ML motion capture can be considered a true representation of skeletal kinematics. MB systems are influenced by marker placement error and soft-tissue artefact, while ML systems are affected by occlusion, image resolution, and assumptions embedded within trained models. As such, future validation studies should compare Theia3D outputs to imaging-based reference standards, such as dynamic biplane radiography, to better quantify joint specific differences. These comparisons would also allow researchers to determine whether differences vary by sex or by task, particularly in movement involving complex frontal and transverse plane mechanics.

Finally, methodological refinements may also improve comparability between systems. For example, the Visual3D MB skeletal model could be refined to incorporate frontal and transverse plane joint constraints that better align with the modelling assumptions used in Theia3D. Reducing differences in model definition may help isolate tracking-related error from model-related variance.

7. References

- [1] E. S. Cowley, A. A. Olenick, K. L. McNulty, and E. Z. Ross, “‘Invisible Sportswomen’: The Sex Data Gap in Sport and Exercise Science Research,” *Women Sport Phys. Act. J.*, vol. 29, no. 2, pp. 146–151, Oct. 2021, doi: 10.1123/wspaj.2021-0028.
- [2] E. van der Kruk, “BIASMECHANICS: Does an unconscious bias still persist in biomechanics, positioning males as the default in human research? A meta-analysis on the Journal of Biomechanics 2024 publications,” *J. Biomech.*, vol. 181, p. 112560, Mar. 2025, doi: 10.1016/j.jbiomech.2025.112560.
- [3] G. R. Booth, P. A. Crompton, and G. P. Siegmund, “The Lack of Sex, Age, and Anthropometric Diversity in Neck Biomechanical Data,” *Front. Bioeng. Biotechnol.*, vol. 9, Aug. 2021, doi: 10.3389/fbioe.2021.684217.
- [4] M. Venkat, “The first female crash test dummy has only now arrived,” *NPR*, Nov. 01, 2022. Accessed: Jan. 25, 2026. [Online]. Available: <https://www.npr.org/2022/11/01/1133375223/the-first-female-crash-test-dummy-has-only-now-arrived>
- [5] J. T. Costello, F. Bieuzen, and C. M. Bleakley, “Where are all the female participants in Sports and Exercise Medicine research?,” *Eur. J. Sport Sci.*, vol. 14, no. 8, pp. 847–851, Nov. 2014, doi: 10.1080/17461391.2014.911354.
- [6] A. M. Wasiak and J. H. Challis, “Trends in gender representation of women as authors and participants over five decades of the journal of biomechanics,” *J. Biomech.*, vol. 193, p. 113027, Dec. 2025, doi: 10.1016/j.jbiomech.2025.113027.
- [7] F. Varcin and M. G. Boocock, “The accuracy, validity and reliability of Theia3D markerless motion capture for studying the biomechanics of human movement: A systematic review,” *Artif. Intell. Med.*, vol. 173, p. 103332, Mar. 2026, doi: 10.1016/j.artmed.2025.103332.
- [8] L. Wade, L. Needham, P. McGuigan, and J. Bilzon, “Applications and limitations of current markerless motion capture methods for clinical gait biomechanics,” *PeerJ*, vol. 10, p. e12995, Feb. 2022, doi: 10.7717/peerj.12995.
- [9] J. LaChance, W. Thong, S. Nagpal, and A. Xiang, “A Case Study in Fairness Evaluation: Current Limitations and Challenges for Human Pose Estimation”.
- [10] R. M. Kanko, E. Laende, W. S. Selbie, and K. J. Deluzio, “Inter-session repeatability of markerless motion capture gait kinematics,” *J. Biomech.*, vol. 121, p. 110422, May 2021, doi: 10.1016/j.jbiomech.2021.110422.
- [11] N. A. Wills, I. J. Eghobamien, V. I. Nwaokoro, and D. N. Pamukoff, “The influence of motion capture method, sex, and body mass index on lower extremity kinematics and intersession reliability during a drop-vertical jump,” *J. Sports Sci.*, vol. 43, no. 16, pp. 1629–1640, Aug. 2025, doi: 10.1080/02640414.2025.2516309.
- [12] S. Alazzawi, M. H. Field, N. V. Bardakos, M. A. R. Freeman, and R. E. Field, “The position of the centre of the femoral head relative to the midline of the pelvis: A consistent landmark in total knee replacement surgery,” *The Knee*, vol. 19, no. 6, pp. 827–831, Dec. 2012, doi: 10.1016/j.knee.2012.04.010.
- [13] D. A. Bruening, R. E. Frimenko, C. D. Goodyear, D. R. Bowden, and A. M. Fullenkamp, “Sex differences in whole body gait kinematics at preferred speeds,” *Gait Posture*, vol. 41, no. 2, pp. 540–545, Feb. 2015, doi: 10.1016/j.gaitpost.2014.12.011.

- [14] L. Mündermann, S. Corazza, and T. P. Andriacchi, “The evolution of methods for the capture of human movement leading to markerless motion capture for biomechanical applications,” *J. NeuroEngineering Rehabil.*, vol. 3, no. 1, p. 6, Mar. 2006, doi: 10.1186/1743-0003-3-6.
- [15] S. L. Colyer, M. Evans, D. P. Cosker, and A. I. T. Salo, “A Review of the Evolution of Vision-Based Motion Analysis and the Integration of Advanced Computer Vision Methods Towards Developing a Markerless System,” *Sports Med. - Open*, vol. 4, no. 1, p. 24, Jun. 2018, doi: 10.1186/s40798-018-0139-y.
- [16] K. Song, T. J. Hullfish, R. Scattone Silva, K. G. Silbernagel, and J. R. Baxter, “Markerless motion capture estimates of lower extremity kinematics and kinetics are comparable to marker-based across 8 movements,” *J. Biomech.*, vol. 157, p. 111751, Aug. 2023, doi: 10.1016/j.jbiomech.2023.111751.
- [17] R. M. Kanko, E. K. Laende, E. M. Davis, W. S. Selbie, and K. J. Deluzio, “Concurrent assessment of gait kinematics using marker-based and markerless motion capture,” *J. Biomech.*, vol. 127, p. 110665, Oct. 2021, doi: 10.1016/j.jbiomech.2021.110665.
- [18] D. L. Benoit, M. Damsgaard, and M. S. Andersen, “Surface marker cluster translation, rotation, scaling and deformation: Their contribution to soft tissue artefact and impact on knee joint kinematics,” *J. Biomech.*, vol. 48, no. 10, pp. 2124–2129, Jul. 2015, doi: 10.1016/j.jbiomech.2015.02.050.
- [19] A. Rajagopal, C. L. Dembia, M. S. DeMers, D. D. Delp, J. L. Hicks, and S. L. Delp, “Full-Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait,” *IEEE Trans. Biomed. Eng.*, vol. 63, no. 10, pp. 2068–2079, Oct. 2016, doi: 10.1109/TBME.2016.2586891.
- [20] E. A. Steffensen, F. Magalhães, B. A. Knarr, and D. C. Kingston, “Comparison of markerless and marker-based motion capture of gait kinematics in individuals with cerebral palsy and chronic stroke: A case study series,” *Res. Sq.*, p. rs.3.rs-2557403, Feb. 2023, doi: 10.21203/rs.3.rs-2557403/v1.
- [21] J. Poomulna, B. A. Knarr, V. Dutt, and D. C. Kingston, “Comparison of gait deviation index (GDI) and gait variability index (GVI) measured by marker-based and markerless motion capture systems in children with cerebral palsy (CP),” *Gait Posture*, vol. 115, pp. 7–13, Jan. 2025, doi: 10.1016/j.gaitpost.2024.10.018.
- [22] I. Coll, M. P. Mavor, T. Karakolis, R. B. Graham, and A. L. Clouthier, “Validation of Markerless Motion Capture for Soldier Movement Patterns Assessment Under Varying Body-Borne Loads,” *Ann. Biomed. Eng.*, vol. 53, no. 2, pp. 358–370, Oct. 2024, doi: 10.1007/s10439-024-03622-w.
- [23] J. Outerleys, A. Mihic, V. Keller, E. Laende, and K. Deluzio, “Markerless motion capture provides repeatable gait outcomes in patients with knee osteoarthritis,” *J. Biomech.*, vol. 168, p. 112115, May 2024, doi: 10.1016/j.jbiomech.2024.112115.
- [24] R. Mullerpatan, T. Shetty, and B. Agarwal, “Influence of age and gender on gait kinematics of pelvis and hip in healthy adults aged 19–60 years,” *Front. Bioeng. Biotechnol.*, vol. 12, Jan. 2025, doi: 10.3389/fbioe.2024.1515583.
- [25] S. Ko, M. I. Tolea, J. M. Hausdorff, and L. Ferrucci, “Sex-specific differences in gait patterns of healthy older adults: Results from the Baltimore Longitudinal Study of Aging,” *J. Biomech.*, vol. 44, no. 10, pp. 1974–1979, Jul. 2011, doi: 10.1016/j.jbiomech.2011.05.005.
- [26] “Differences in Lower Extremity Kinematics Between a Bilateral Drop-Vertical Jump and A Single-Leg Step-down.” Accessed: Dec. 30, 2025. [Online]. Available: <https://www.jospt.org/doi/epdf/10.2519/jospt.2007.2202>

- [27] A. T. Peebles, L. C. Dickerson, K. E. Renner, and R. M. Queen, "Sex-based differences in landing mechanics vary between the drop vertical jump and stop jump," *J. Biomech.*, vol. 105, p. 109818, May 2020, doi: 10.1016/j.jbiomech.2020.109818.
- [28] L. K. Smith, J. L. Lelas, and D. C. Kerrigan, "Gender Differences in Pelvic Motions and Center of Mass Displacement during Walking: Stereotypes Quantified," *J. Womens Health Gend. Based Med.*, vol. 11, no. 5, pp. 453–458, Jun. 2002, doi: 10.1089/15246090260137626.
- [29] C. A. Hedt *et al.*, "Sex-related Anthropometrics in a Lower-Body Mobility Assessment Among Professional Soccer Athletes," *Int. J. Sports Phys. Ther.*, vol. 17, no. 3, Apr. 2022, doi: 10.26603/001c.32595.
- [30] R. Hara, J. McGinley, C. Briggs, R. Baker, and M. Sangeux, "Predicting the location of the hip joint centres, impact of age group and sex," *Sci. Rep.*, vol. 6, no. 1, p. 37707, Nov. 2016, doi: 10.1038/srep37707.
- [31] C. Bleakley, N. Klempel, J. Wagemans, F. Netterström-Wedin, and J. Smoliga, "Statistical Power in Musculoskeletal Research: A Meta-Review of 266 Randomised Controlled Trials," *Sports Med. - Open*, vol. 11, no. 1, p. 134, Nov. 2025, doi: 10.1186/s40798-025-00908-8.
- [32] C. Antognini, A. Ortigas-Vásquez, C. Knowlton, M. Utz, A. Sauer, and M. A. Wimmer, "Comparison of markerless and marker-based motion analysis accounting for differences in local reference frame orientation," *J. Biomech.*, vol. 185, p. 112683, May 2025, doi: 10.1016/j.jbiomech.2025.112683.
- [33] E. M. Walker, M. Nelson, M. D. Drew, S. M. Krammer, and T. N. Brown, "Tibial compression during sustained walking with body borne load," *J. Biomech.*, vol. 133, p. 110969, Mar. 2022, doi: 10.1016/j.jbiomech.2022.110969.
- [34] S. D'Souza, T. Siebert, and V. Fohanno, "A comparison of lower body gait kinematics and kinetics between Theia3D markerless and marker-based models in healthy subjects and clinical patients," *Sci. Rep.*, vol. 14, p. 29154, Nov. 2024, doi: 10.1038/s41598-024-80499-8.
- [35] E. Van Der Kruk and M. M. Reijne, "Accuracy of human motion capture systems for sport applications; state-of-the-art review," *Eur. J. Sport Sci.*, vol. 18, no. 6, pp. 806–819, Jul. 2018, doi: 10.1080/17461391.2018.1463397.
- [36] S. Lopez, C. Johnson, N. Frankston, E. Ruh, M. McClincy, and W. Anderst, "Accuracy of conventional motion capture in measuring hip joint center location and hip rotations during gait, squat, and step-up activities," *J. Biomech.*, vol. 167, p. 112079, Apr. 2024, doi: 10.1016/j.jbiomech.2024.112079.
- [37] R. M. Kanko, J. B. Outerleys, E. K. Laende, W. S. Selbie, and K. J. Deluzio, "Comparison of concurrent and asynchronous running kinematics and kinetics from marker-based motion capture and markerless motion capture under two clothing conditions," *Bioengineering*, preprint, Feb. 2023. doi: 10.1101/2023.02.22.529537.
- [38] G. E. Gorton, D. A. Hebert, and M. E. Gannotti, "Assessment of the kinematic variability among 12 motion analysis laboratories," *Gait Posture*, vol. 29, no. 3, pp. 398–402, Apr. 2009, doi: 10.1016/j.gaitpost.2008.10.060.
- [39] N. Nishikawa, S. Watanabe, and K. Yamamoto, "Comparison of kinematics between markerless and marker-based motion capture systems for change of direction maneuvers," *J. Biomech.*, vol. 192, p. 112965, Nov. 2025, doi: 10.1016/j.jbiomech.2025.112965.
- [40] "Theia3D Apollo Release Part 3: Model DOFs | Theia Markerless." Accessed: Jan. 25, 2026. [Online]. Available: <https://www.theiamarkerless.com/blog/theia-2024-release-part-3-model-dofs>

- [41] “Theia Model Description | theia docs.” Accessed: Mar. 07, 2026. [Online]. Available: <https://docs.theiamarkerless.com/theia3d-documentation/theia-model-description>
- [42] S. Augustine, R. Sharir, G. Barton, R. Foster, and M. Robinson, “The effects of tight or loose-fitting clothing on markerless gait kinematics in adults,” *Gait Posture*, vol. 106, p. S14, Sep. 2023, doi: 10.1016/j.gaitpost.2023.07.021.
- [43] S. Riazati, T. E. McGuirk, E. S. Perry, W. B. Sihanath, and C. Patten, “Absolute Reliability of Gait Parameters Acquired With Markerless Motion Capture in Living Domains,” *Front. Hum. Neurosci.*, vol. 16, Jun. 2022, doi: 10.3389/fnhum.2022.867474.
- [44] N. A. Wills, D. N. Pamukoff, I. J. Eghobamien, and V. I. Nwaokoro, “Influence Of Sex On Marker To Marker-less Motion Capture Evaluations Of Gait Biomechanics: 2712,” *Med. Sci. Sports Exerc.*, vol. 56, no. 10S, pp. 1006–1007, Oct. 2024, doi: 10.1249/01.mss.0001061252.45734.33.
- [45] P. Federolf *et al.*, “Validation of markerless (Theia3D™) against marker-based (Vicon™) motion capture data of postural control movements analyzed through principal component analysis,” *J. Biomech.*, vol. 189, p. 112831, Aug. 2025, doi: 10.1016/j.jbiomech.2025.112831.
- [46] “Theia3D Apollo Release Part 2: Model Definitions | Theia Markerless.” Accessed: Jan. 12, 2026. [Online]. Available: <https://www.theiamarkerless.com/blog/theia-2024-release-part-2-model-definitions>
- [47] “Theia3D Apollo Release Part 1: Updated Validation | Theia Markerless.” Accessed: Jan. 25, 2026. [Online]. Available: <https://www.theiamarkerless.com/blog/theia-2024-release-part-1-updated-validation>
- [48] “V3D Composite Pelvis [HAS-Motion Software Documentation].” Accessed: Jan. 14, 2026. [Online]. Available: https://has-motion.com/wiki/doku.php?id=visual3d:documentation:modeling:segments:v3d_composite_pelvis
- [49] A. C. da Hora, S. P. Pedrozo, A. C. da Hora, and S. P. Pedrozo, “Beyond Algorithmic Bias,” in *Mitigating Bias in Machine Learning*, 1st Edition., C. A. Berry, B. Marshall, C. A. Berry, and B. Marshall, Eds., McGraw-Hill Education, 2025. Accessed: Jan. 09, 2026. [Online]. Available: <https://www.accessengineeringlibrary.com/content/book/9781264922444/chapter/chapter1>
- [50] “Exposing Limitations in Fairness Evaluations: Human Pose Estimation – Sony AI,” Sony AI – Unleash Human Imagination and Creativity with AI. Accessed: Jan. 09, 2026. [Online]. Available: <https://ai.sony/blog/Exposing-Limitations-in-Fairness-Evaluations-Human-Pose-Estimation/>
- [51] “Seven Types Of Data Bias In Machine Learning.” Accessed: Jan. 12, 2026. [Online]. Available: <https://www.telusdigital.com/insights/data-and-ai/article/7-types-of-data-bias-in-machine-learning>
- [52] S. Hudson, M. Barwood, C. Low, J. Wills, and M. Fish, “A systematic review of the physiological and biomechanical differences between males and females in response to load carriage during walking activities,” *Appl. Ergon.*, vol. 114, p. 104123, Jan. 2024, doi: 10.1016/j.apergo.2023.104123.
- [53] Shogo Takano, Yoshitaka Iwamoto, Junya Ozawa, and Nobuhiro Kito, “Relationship Between Hip Frontal Dynamic Joint Stiffness and Frontal and Transverse Plane Hip Kinematics During Gait: Sex Differences,” *J. Sport Rehabil.*, vol. 30, no. 3, pp. 475–481, Mar. 2021, doi: 10.1123/jsr.2020-0037.

- [54] A. Phinyomark, S. T. Osis, B. A. Hettinga, D. Kobsar, and R. Ferber, "Gender differences in gait kinematics for patients with knee osteoarthritis," *BMC Musculoskelet. Disord.*, vol. 17, no. 158, p. NA-NA, Apr. 2016, doi: 10.1186/s12891-016-1013-z.
- [55] E. S. Chumanov, C. Wall-Scheffler, and B. C. Heiderscheit, "Gender differences in walking and running on level and inclined surfaces," *Clin. Biomech.*, vol. 23, no. 10, pp. 1260–1268, Dec. 2008, doi: 10.1016/j.clinbiomech.2008.07.011.
- [56] M. Sakaguchi, H. Ogawa, N. Shimizu, H. Kanehisa, T. Yanai, and Y. Kawakami, "Gender differences in hip and ankle joint kinematics on knee abduction during running," *Eur. J. Sport Sci.*, vol. 14, pp. S302–S309, Jan. 2014, doi: 10.1080/17461391.2012.693953.
- [57] A. A. Ahmed, "Estimation of sex from the lower limb measurements of Sudanese adults," *Forensic Sci. Int.*, vol. 229, no. 1, p. 169.e1-169.e7, Jun. 2013, doi: 10.1016/j.forsciint.2013.04.005.
- [58] M. Elnady, naira girgis, A. Elseidy, shaimaa Hassanein, and shireen slima, "THE POTENTIAL ROLE OF SOME LOWER LIMB PARAMETERS IN STATURE ESTIMATION," *Egypt. J. Forensic Sci. Appl. Toxicol.*, vol. 25, pp. 79–93, Jun. 2025, doi: 10.21608/ejfsat.2025.335458.1354.
- [59] S. S. Moosa *et al.*, "Sexual dimorphic parameters of femur: a clinical guide in orthopedics and forensic studies," *J. Med. Life*, vol. 14, no. 6, pp. 762–768, 2021, doi: 10.25122/jml-2021-0022.
- [60] F. M. Nor, N. Abdullah, A.-M. Mustapa, L. Qi Wen, N. A. Faisal, and D. A. A. Ahmad Nazari, "Estimation of stature by using lower limb dimensions in the Malaysian population," *J. Forensic Leg. Med.*, vol. 20, no. 8, pp. 947–952, Nov. 2013, doi: 10.1016/j.jflm.2013.09.006.
- [61] M. D. Harris, E. C. Behrman, K. R. Lohse, M. C. Shepherd, B. A. MacWilliams, and E. M. Mannen, "Can predictive and functional methods locate the dysplastic hip joint center?," *Gait Posture*, vol. 124, p. 109982, Feb. 2026, doi: 10.1016/j.gaitpost.2025.109982.
- [62] A. Leong, "Sexual dimorphism of the pelvic architecture: A struggling response to destructive and parsimonious forces by natural & mate selection," *McGill J. Med.*, vol. 9, no. 1, 2006, doi: 10.26443/mjm.v9i1.602.
- [63] A. S. Augustine, A. M. Ewen, D. T. Wallace, and F. Picard, "Variations in pelvic dimensions: An anatomical and computed tomography study," *Clin. Anat.*, vol. 31, no. 7, pp. 981–987, 2018, doi: 10.1002/ca.23273.
- [64] H. Kainz, C. P. Carty, L. Modenese, R. N. Boyd, and D. G. Lloyd, "Estimation of the hip joint centre in human motion analysis: A systematic review," *Clin. Biomech.*, vol. 30, no. 4, pp. 319–329, May 2015, doi: 10.1016/j.clinbiomech.2015.02.005.
- [65] M. E. Harrington, A. B. Zavatsky, S. E. M. Lawson, Z. Yuan, and T. N. Theologis, "Prediction of the hip joint centre in adults, children, and patients with cerebral palsy based on magnetic resonance imaging," *J. Biomech.*, vol. 40, no. 3, pp. 595–602, Jan. 2007, doi: 10.1016/j.jbiomech.2006.02.003.
- [66] N. M. Fiorentino, M. J. Kutschke, P. R. Atkins, K. B. Foreman, A. L. Kapron, and A. E. Anderson, "Accuracy of Functional and Predictive Methods to Calculate the Hip Joint Center in Young Non-pathologic Asymptomatic Adults with Dual Fluoroscopy as a Reference Standard," *Ann. Biomed. Eng.*, vol. 44, no. 7, pp. 2168–2180, Jul. 2016, doi: 10.1007/s10439-015-1522-1.
- [67] F. Leboeuf, A. Barre, K. Aminian, and M. Sangeux, "On the accuracy of the Conventional gait Model: Distinction between marker misplacement and soft tissue artefact errors," *J. Biomech.*, vol. 159, p. 111774, Oct. 2023, doi: 10.1016/j.jbiomech.2023.111774.

- [68] F. Leboeuf, J. Reay, R. Jones, and M. Sangeux, “The effect on conventional gait model kinematics and kinetics of hip joint centre equations in adult healthy gait,” *J. Biomech.*, vol. 87, pp. 167–171, Apr. 2019, doi: 10.1016/j.jbiomech.2019.02.010.
- [69] R. M. Ehrig, M. O. Heller, S. Kratzenstein, G. N. Duda, A. Trepczynski, and W. R. Taylor, “The SCoRE residual: A quality index to assess the accuracy of joint estimations,” *J. Biomech.*, vol. 44, no. 7, pp. 1400–1404, Apr. 2011, doi: 10.1016/j.jbiomech.2010.12.009.
- [70] E. Gianzina, C. K. Yiannakopoulos, G. Kalinterakis, S. Delis, and E. Chronopoulos, “Gender-Based Differences in Biomechanical Walking Patterns of Athletes Using Inertial Sensors.,” *J. Funct. Morphol. Kinesiol.*, vol. 10, no. 1, p. NA-NA, Mar. 2025, doi: 10.3390/jfmk10010082.
- [71] C. L. Lewis, N. M. Laudicina, A. Khuu, and K. L. Loverro, “The Human Pelvis: Variation in Structure and Function During Gait,” *Anat. Rec.*, vol. 300, no. 4, pp. 633–642, 2017, doi: 10.1002/ar.23552.
- [72] S. Chen, G. Cheng, and Y. Lin, “Gender-based age differences in hip and knee kinematics of Chinese adults during walking and running.,” *J. Back Musculoskelet. Rehabil.*, vol. 33, no. 2, pp. 217–224, Mar. 2020, doi: 10.3233/BMR-181134.
- [73] D. C. Kerrigan, M. K. Todd, and U. Della Croce, “Gender differences in joint biomechanics during walking: normative study in young adults,” *Am. J. Phys. Med. Rehabil.*, vol. 77, no. 1, pp. 2–7, 1998, doi: 10.1097/00002060-199801000-00002.
- [74] S. Araki *et al.*, “Sex differences in age-related differences in joint motion during gait in community-dwelling middle-age and older individuals,” *Gait Posture*, vol. 103, pp. 153–158, Jun. 2023, doi: 10.1016/j.gaitpost.2023.05.009.
- [75] A. A. Al-kharaz and A. K. Chong, “Gender differences in ankle kinematics of adults during gait,” *J. Sex- Gend.-Specif. Med.*, vol. 8, no. 3, pp. 147–153, Sep. 2022.
- [76] Y. Chun, J. P. Bailey, J. Kim, S.-C. Lee, and S. Y. Lee, “Sex and Limb Differences in Lower Extremity Alignment and Kinematics during Drop Vertical Jumps,” *Int. J. Environ. Res. Public Health*, vol. 18, no. 7, p. 3748, Apr. 2021, doi: 10.3390/ijerph18073748.
- [77] N. J. Romanchuk, M. J. Del Bel, and D. L. Benoit, “Sex-specific landing biomechanics and energy absorption during unanticipated single-leg drop-jumps in adolescents: implications for knee injury mechanics,” *J. Biomech.*, vol. 113, p. 110064, Dec. 2020, doi: 10.1016/j.jbiomech.2020.110064.
- [78] Y. Matsumoto, N. Ogihara, S. Kosuge, H. Hanawa, T. Kokubun, and N. Kanemura, “Sex differences in the kinematics and kinetics of the foot and plantar aponeurosis during drop-jump,” *Sci. Rep.*, vol. 13, p. 12957, Aug. 2023, doi: 10.1038/s41598-023-39682-6.
- [79] D. Sudo, M. Hosokawa, and Y. Maeda, “Effects of kinematics and kinetics parameters of the vertical drop jump test on differences between the sexes and dominant and non-dominant feet,” *Biomed. Hum. Kinet.*, vol. 16, no. 1, pp. 157–162, Apr. 2024, doi: 10.2478/bhk-2024-0016.
- [80] K. Kawaguchi *et al.*, “Sex-Based Differences in the Drop Vertical Jump as Revealed by Video Motion Capture Analysis Using Artificial Intelligence,” *Orthop. J. Sports Med.*, vol. 9, no. 11, p. 23259671211048188, Nov. 2021, doi: 10.1177/23259671211048188.
- [81] J. S. Chang, Y. H. Kwon, J. H. Choi, and H. S. Lee, “Gender Differences in Lower Extremity Kinematics and Kinetics of the Vertical Ground Reaction Force Peak in Drop-landing by Flatfooted Subjects,” *J. Phys. Ther. Sci.*, vol. 24, no. 3, pp. 267–270, 2012, doi: 10.1589/jpts.24.267.

- [82] M. J. Decker, M. R. Torry, D. J. Wyland, W. I. Sterett, and J. Richard Steadman, "Gender differences in lower extremity kinematics, kinetics and energy absorption during landing," *Clin. Biomech.*, vol. 18, no. 7, pp. 662–669, Aug. 2003, doi: 10.1016/S0268-0033(03)00090-1.
- [83] S. C. Government of Canada, "Add/Remove data - Anthropometry measures of the household population." Accessed: Jul. 20, 2026. [Online]. Available: <https://www150.statcan.gc.ca/t1/tb11/en/cv.action?pid=1310031901>
- [84] F. Faul, E. Erdfelder, A.-G. Lang, and A. Buchner, "G*Power 3: A flexible statistical power analysis program for the social, behavioral, and biomedical sciences," *Behav. Res. Methods*, vol. 39, no. 2, pp. 175–191, May 2007, doi: 10.3758/BF03193146.
- [85] "(PDF) Determination of optimal filter parameters for filtering kinematic walking data using butterworth low pass filter," in *ResearchGate*, Accessed: Jan. 29, 2026. [Online]. Available: https://www.researchgate.net/publication/242203849_Determination_of_optimal_filter_parameters_for_filtering_kinematic_walking_data_using_butterworth_low_pass_filter
- [86] A. L. Bell, D. R. Pedersen, and R. A. Brand, "A comparison of the accuracy of several hip center location prediction methods," *J. Biomech.*, vol. 23, no. 6, pp. 617–621, Jan. 1990, doi: 10.1016/0021-9290(90)90054-7.
- [87] T. Gale and W. Anderst, "Knee Kinematics of Healthy Adults Measured Using Biplane Radiography," *J. Biomech. Eng.*, vol. 142, no. 101004, Jul. 2020, doi: 10.1115/1.4047419.
- [88] C. Belvedere, A. Leardini, S. Giannini, A. Ensini, L. Bianchi, and F. Catani, "Does medio-lateral motion occur in the normal knee? An *in-vitro* study in passive motion," *J. Biomech.*, vol. 44, no. 5, pp. 877–884, Mar. 2011, doi: 10.1016/j.jbiomech.2010.12.004.
- [89] Y. Ishii, H. Noguchi, M. Takeda, J. Sato, and S. Toyabe, "Anteroposterior Translation Does Not Correlate With Knee Flexion After Total Knee Arthroplasty," *Clin. Orthop. Relat. Res.*, vol. 472, no. 2, pp. 704–709, Feb. 2014, doi: 10.1007/s11999-013-3274-2.
- [90] S. Yang, S. P. Canton, M. V. Hogan, and W. Anderst, "Healthy ankle and hindfoot kinematics during gait: Sex differences, asymmetry and coupled motion revealed through dynamic biplane radiography," *J. Biomech.*, vol. 116, p. 110220, Feb. 2021, doi: 10.1016/j.jbiomech.2020.110220.
- [91] G. Atkinson and A. M. Nevill, "Statistical Methods For Assessing Measurement Error (Reliability) in Variables Relevant to Sports Medicine," *Sports Med.*, vol. 26, no. 4, pp. 217–238, 1998, doi: 10.2165/00007256-199826040-00002.
- [92] T. K. Koo and M. Y. Li, "A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research," *J. Chiropr. Med.*, vol. 15, no. 2, pp. 155–163, Jun. 2016, doi: 10.1016/j.jcm.2016.02.012.
- [93] D. Lakens, "Calculating and reporting effect sizes to facilitate cumulative science: a practical primer for t-tests and ANOVAs," *Front. Psychol.*, vol. 4, p. 863, Nov. 2013, doi: 10.3389/fpsyg.2013.00863.
- [94] J. Santos Nobre and J. da Motta Singer, "Residual Analysis for Linear Mixed Models," *Biom. J.*, vol. 49, no. 6, pp. 863–875, 2007, doi: 10.1002/bimj.200610341.
- [95] A. Loy, L. Follett, and H. Hofmann, "Variations of Q-Q Plots: The Power of Our Eyes!," *Am. Stat.*, vol. 70, no. 2, pp. 202–214, 2016.
- [96] D. Giavarina, "Understanding Bland Altman analysis," *Biochem. Medica*, vol. 25, no. 2, pp. 141–151, Jun. 2015, doi: 10.11613/BM.2015.015.

- [97] P. Taffé, “When can the Bland & Altman limits of agreement method be used and when it should not be used,” *J. Clin. Epidemiol.*, vol. 137, pp. 176–181, Sep. 2021, doi: 10.1016/j.jclinepi.2021.04.004.
- [98] “Distribution evaluation and normalizing.” Accessed: Jan. 31, 2026. [Online]. Available: <https://www-sciencedirect-com.proxy.bib.uottawa.ca/science/chapter/monograph/pii/B9780323917483000045>
- [99] “Correlation Coefficients: Appropriate Use and... : Anesthesia & Analgesia,” Ovid. Accessed: Jul. 20, 2026. [Online]. Available: <https://www.ovid.com/jnls/anesthesia-analgesia/fulltext/10.1213/ane.0000000000002864~correlation-coefficients-appropriate-use-and-interpretation>
- [100] “Full Body Model Description | theia docs.” Accessed: Feb. 19, 2026. [Online]. Available: <https://docs.theiamarkerless.com/theia3d-documentation/theia-model-description/full-body-model-description>
- [101] P. Toliopoulos, N. Hagemeister, A. Fuentes, F. Desmeules, and P.-A. Vendittoli, “Normal knee gait kinematics: Describing a normal cohort and illustrating differences between genders,” *Neurophysiol. Clin. Neurophysiol.*, vol. 46, no. 4, p. 283, Nov. 2016, doi: 10.1016/j.neucli.2016.09.114.
- [102] M. P. Kadaba, H. K. Ramakrishnan, and M. E. Wootten, “Measurement of lower extremity kinematics during level walking,” *J. Orthop. Res.*, vol. 8, no. 3, pp. 383–392, 1990, doi: 10.1002/jor.1100080310.
- [103] S. J. Piazza and P. R. Cavanagh, “Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment,” *J. Biomech.*, vol. 33, no. 8, pp. 1029–1034, Aug. 2000, doi: 10.1016/S0021-9290(00)00056-7.
- [104] N. Ito *et al.*, “Markerless motion capture: What clinician-scientists need to know right now,” *JSAMS Plus*, vol. 1, p. 100001, Oct. 2022, doi: 10.1016/j.jsampl.2022.100001.
- [105] I. Coll, “Validation of Markerless Motion Capture for the Assessment of Soldier Movement Patterns Under Varying Body-Borne Loads,” Thesis, Université d’Ottawa / University of Ottawa, 2023. doi: 10.20381/ruor-29079.

Appendix A – Participant anthropometrics and sizing

Table A. 1 Participant Anthropometrics and Sizing

Participant	Gender	Height (cm)	Weight (kg)	Age (yr)	BMI (kg/m ²)
P01	F	168	64	26	22.7
P02	F	178	61.2	24	19.3
P03	F	165	62	26	22.8
P04	F	163	68	24	25.6
P05	F	164	53.5	26	19.9
P06	M	188	83	28	23.5
P07	M	173	73.9	31	24.7
P08	M	187	76	25	21.7
P09	F	163	51	28	19.2
P10	M	176	67	28	21.6
P11	M	184	82	28	24.2
P12	M	186	93	21	26.9
P13	M	167	72.6	23	26.0
P14	F	164	50	38	18.6
P15	F	164	62.9	33	23.4
P16	M	184	93	25	27.5
P17	M	180	95.7	29	29.5
P18	F	164	72.6	27	27.0
P19	F	158	70.3	24	28.2
P20	M	184	86	25	25.4
P21	F	171	79.4	27	27.2
P22	F	168	63.9	25	22.6
P23	M	187	72.6	24	20.8
P24	M	168	81.6	40	28.9
P25	M	180	95	26	29.3
P26	F	174	77.1	26	25.5
P27	F	170	50	23	17.3
P28	M	188	90.7	25	25.7

Appendix B – Certificate of ethics approval

Université d'Ottawa

Bureau d'éthique et d'intégrité de la recherche

University of Ottawa

Office of Research Ethics and Integrity

H-06-25-11691 - REG-11691 - Certificat d'approbation éthique / Certificate of Ethics Approval

(English message follows)

Cher/Chère Ryan Graham,

Veuillez trouver ci-joint le certificat d'approbation éthique pour le projet intitulé «Investigating Sex-Based Differences in Lower-Limb Kinematics in Markerless Motion Capture».

Le certificat est valide jusqu'au : 06-07-2026

I wish you pleasant and fruitful research activities.

Recherche financée : veuillez faire suivre une copie du certificat au [Service de gestion de la recherche](#).

Si vous avez des questions, n'hésitez pas à communiquer avec le Bureau d'éthique à ethique@uottawa.ca ou en composant le 613-562-5387.

Vous pouvez voir votre demande en vous connectant à votre compte [eReviews](#).

Cordialement,

Germain Zongo
Responsable d'éthique en recherche

Ceci est une réponse automatisée, merci de ne pas répondre à ce courriel.

Dear Ryan Graham,

Please find attached the certificate of ethics approval for your research project titled "Investigating Sex-Based Differences in Lower-Limb Kinematics in Markerless Motion Capture".

This certificate is valid until: 06-07-2026

I wish you pleasant and fruitful research activities.

Funded research: A reminder that you must provide a copy of this certificate to [Research Management Services](#).

If you have any questions, please contact the Ethics Office at ethics@uottawa.ca or by telephone at 613-562-5387.

You can view your project at any time by logging into [eReviews](#).

Best regards,

Germain Zongo
Protocol Officer

This is an automated message. Please do not reply directly to this email.

Attachement(s) / Attachment(s)

550, rue Cumberland, pièce 154 550 Cumberland Street, Room 154
Ottawa (Ontario) K1N 6N5 Canada Ottawa, Ontario K1N 6N5 Canada

613-562-5387 • 613-562-5338 • ethique@uOttawa.ca / ethics@uOttawa.ca
www.recherche.uottawa.ca/deontologie | www.recherche.uottawa.ca/ethics

Appendix C – Research consent form



☎ 613-562-5853
613-562-5149
125 University Private
Ottawa ON K1N 6N5 Canada
www.uOttawa.ca

Research Consent Form

Research Project Title: Investigating Sex-Based Differences in Lower-Limb Kinematics in Markerless Motion Capture

Principal Investigator:

Dr. Ryan Graham
613-562-5800 X 1025; ryan.graham@uottawa.ca
University of Ottawa
Faculty of Health Sciences
Department of Human Kinetics
200 Lees Ave (585)
Ottawa, ON K1N6N5

Background and Purpose of the Study:

In Biomechanical research, women continue to be underrepresented as study participants and in model development. This imbalance limits our understanding of sex-based differences in movement and injury risk. Such gap may be particularly impactful in the application of machine learning based motion capture systems such as Theia3D. Despite its growing use in gait analysis for both clinical and performance applications, Theia3D has not been evaluated for sex-based bias. MOVE4D is an emerging markerless motion capture system that differentiates between men and women. This feature could be useful in the development of sex-specific models.

The purpose of this study is to investigate whether sex-based differences exist in Theia3D's estimation of joint centres and lower-limb kinematics by comparing markerless data against a gold standard marker-based motion capture system. Additionally, the findings will also contribute to the evaluation of the MOVE4D system in a biomechanical research context.

Description of Study Procedures:

You are invited to participate in a one-day motion-analysis procedure lasting approximately one hour at the University of Ottawa Movement Biomechanics and Analytics Laboratory (200 Lees Avenue, 231). A portion of data from this study will be used in Maxine Speers' master's thesis project. The study protocol includes a series of two movements: walking over a 5-metre distance and a drop-down vertical jump off a 30 cm high plyometric box. Both tasks will be performed three times. There will be a 30 second to one minute rest period between trials. Video recordings of the trials will be taken using three sets of camera systems: Vicon Vue for the markerless system, Vicon Vantage for the marker-based system and MOVE4D cameras.

You are asked to wear running shoes and tight-fitting shorts during the data collection. You are also asked to wear form-fitting clothing on their upper body; they may choose to wear either a tight shirt, a sports bra or no shirt – whichever option makes you most comfortable. Your choice will not affect your ability to participate. After completing the consent form, the researchers will place reflective markers on your skin and clothing using double-sided medical tape.

After completing the consent form, the researchers will place reflective markers on your skin and clothing using double-sided medical tape. Markers will be placed on anatomical landmarks by a female researcher. Upon request, a male researcher can be reached to place markers. Marker clusters and individual reflective markers will be placed on your upper and lower limbs, trunk, pelvis, and head.

Before the experimental trials can begin, you will be asked to perform two five-second static calibration trials (A-pose and biker-pose) and functional range of motion trials, which should take approximately one minute. These trials will be used to build an individualized biomechanical model. During the walking trials you will be asked to walk over force plates at a self-selected pace (distance of approximately 5 meters). For the drop-down vertical jump, you will be asked to step down from a plyometric box in a loaded squat, and then as quick as possible jump vertically as high as you can.

Possible Risks and Discomforts:

Your participation in this study will entail having reflective markers placed on anatomical landmarks on your arms, legs, feet, trunk, pelvis, and head. During removal of the adhesives, the researchers will make sure to remove them gently to minimize the discomfort. It is possible that minor discomfort is experienced from the tape, but it should be no different than what is felt when removing a Band-Aid.

Should you experience any major discomfort, please tell us immediately and seek primary care from a medical professional on campus (801 King Edward Ave, Tel.: 613-562-5498) or a medical professional of your choosing.

Benefits:

You will not gain any direct benefit from participation in this study. However, the data collected will contribute to the understanding of sex-based bias within markerless motion-capture systems as well as contribute to the validation of markerless motion capture systems for estimating kinematics.

Confidentiality and Privacy:

All personal information is kept confidential. Information gained from this study will be stored electronically and will need a password to access, which will only be known to Dr. Ryan Graham and the research team. Paper study records are stored in a locked cabinet in a locked office and will be destroyed after 5 years post-publication; electronic records without personally identifying information will be stored indefinitely. You will not be identified by name in any reports/presentations of the completed study, and your face will never be presented in any reports/presentations. Your anonymity will be strictly maintained – you will never be identified by name; only by an independent study number (e.g., P01).

Conservation of Data:

Information gained from this study such as video-recordings and marker-position data will be stored electronically, in a folder that will require a password to access, which will be known only to the research team. All hard copies of participant consent forms will be locked in a cabinet inside Ryan Graham's locked office. All data will be collected and analyzed on computers within the Movement Biomechanics and Analytics Laboratory. These computers are password protected at all times, where the password is known only to the faculty supervisor and other research

assistants who work in the lab. Data specific to this study will be placed in a password protected folder known only to the research team. Data containing identifying information will be disposed of 5 years post-publication as per APA standards. Electronic records without personally identifying information will be stored indefinitely to be used in further biomechanical analysis. After identifiable information is destroyed it will not be possible to withdraw from the study.

Compensation:

Compensation is not provided for your participation in this study.

Voluntary Participation:

You are under no obligation to participate and if you choose to participate, you can withdraw from the study at any time and/or refuse to answer any questions, without suffering any negative consequences. If you choose to withdraw, all data gathered until the time of withdrawal will be removed from the dataset and not used in the study.

Funding:

This research was funded by the University of Ottawa Research Chair held by Dr. Ryan Graham.

Questions about the Study:

You are free to ask questions at any time during the protocol and by contacting the principal investigator by email: Dr. Ryan Graham (ryan.graham@uottawa.ca). The ethical components of this research project have been approved by the University of Ottawa research ethics board. If you have any questions regarding the ethical conduct of this study, you may contact the Protocol Officer for Ethics in Research, University of Ottawa, Tabaret Hall, 550 Cumberland Street, Room 154, Ottawa ON, K1N 6N5. Tel.: (613) 562-5800 ext. 5387. Email: ethics@uottawa.ca.

There are two copies of the consent form, one of which is yours to keep.

Consent:

I have read this consent form, and I agree to participate in the procedures of this study.

Participant's name: _____ Date: _____
Participant's signature: _____ Date: _____

Investigator Statement (or Person Explaining the Consent):

I have carefully explained to the research participant the nature of the above research study. To the best of my knowledge, the research participant signing this consent form understands the nature, demands, risks and benefits involved in participating in this study. I acknowledge my responsibility for the care and well-being of the above research participant, to respect the rights and wishes of the research participant, and to conduct the study according to applicable Good Clinical Practice guidelines and regulations.

Researcher's name: _____ Date: _____
Researcher's signature: _____ Date: _____

Informed Consent to have Pictures and Videos Taken:

I consent to have pictures and videos taken of myself completing the experiment and understand that no pictures or videos will be taken at any point without consent. I also understand that if any of these pictures are used in a subsequent presentation or publication, my face and any other identifiers will be blurred. I cannot participate in the research study without consenting to have pictures taken.

Participant's name: _____ Date: _____
Participant's signature: _____ Date: _____

Informed Consent for Secondary Analysis of Data:

I consent to my data being used for secondary analysis after the completion of the initial study. I understand that my face and other identifiers will be blurred or removed from the data that is kept indefinitely.

Participant's name: _____ Date: _____
Participant's signature: _____ Date: _____

Future Participation:

- ☐ I am interested in being contacted to participate in future research performed by this laboratory (your email information will be saved in a password protected file).

Appendix D – Within-trial and between-trial comparison

The mean segment length, the standard deviation of length between frames, as well as coefficient of variation from each trial was extracted for each method. These data were calculated for each participant across all static and dynamic trials to evaluate the stability of each method within single trials, the average values are presented in Table D.1. Within a single trial, the ML method is very consistent, often achieving 0% variation and no deviation in segment lengths between frames. The MB had a slightly larger coefficient of variation, although it still performed very well with variation of less than 1% and standard deviations between 0.003 and 0.013 cm.

Between-trial comparisons were performed to evaluate how consistent each method is within trials of a single participant. For this analysis, the mean segment length from each trial within a single participant were used to calculate the interclass correlation (ICC), 95% CI, standard deviation and coefficients of variation. After, the results from each participant were averaged within each method. These results are presented in Table D.1. When looking at the between-trial comparisons, both methods had excellent ICC between 0.89 and 1.0 for MB and between 0.81 and 0.99 for ML. The MB method had slightly better consistency with an average ICC of 0.95 compared to 0.91 for the ML. The standard deviation and coefficient of variation were also better in MB results with mean values of 2.66 mm and 1.15%, contrasting the ML's 3.46 mm and 1.3 %.

Table D. 1 Mean segment lengths, standard deviations, coefficient of variations, and intraclass correlations for within-trial and between-trial comparisons of MB and ML data.

			Within-Trial		Between-Trial		
Segment	Method	Mean length (cm)	SD Frames (cm)	CV (%)	SD (mm)	CV (%)	ICC
Pelvis	MB	12.47	0.05	0.041	2.98	2.00	0.89
Hip R	ML	12.16	0.00	< 0.001	4.24	1.38	0.90
Pelvis	MB	12.46	0.05	0.039	2.21	1.76	0.91
Hip L	ML	12.16	0.00	< 0.001	1.68	1.38	0.90
Hip	MB	16.79	0.06	0.034	2.52	1.78	0.93
Width	ML	17.10	0.00	< 0.001	1.68	2.46	0.81
Thigh R	MB	43.07	0.05	0.011	1.48	0.67	0.98

	ML	40.76	0.00	< 0.001	2.75	0.79	0.97
Thigh L	MB	47.46	0.13	0.028	1.32	1.11	0.95
	ML	45.34	0.09	0.020	2.77	1.74	0.86
Shank R	MB	40.59	0.03	0.008	5.24	0.33	1.00
	ML	40.51	0.00	< 0.001	7.87	0.69	0.99
Shank L	MB	40.47	0.03	0.008	2.85	0.36	1.00
	ML	40.36	0.00	< 0.001	3.21	0.69	0.99

Appendix E – Sensitivity Analysis

Table E. 1 Joint centre offset between ML data and four MB models. Values are in cm

Joint	Functional Hip Joint	Bell	Harrington – Pelvis and Leg	Harrington – Pelvis Only
Pelvis	3.37 + 1.43	3.37 + 1.43	3.37 + 1.43	3.37 + 1.43
Hip	4.91 + 1.60	5.79 + 1.53	4.27 + 1.30	4.07 + 1.29
Knee	1.37 + 0.71	1.37 + 0.71	1.37 + 0.71	1.37 + 0.71
Ankle	1.50 + 0.72	1.50 + 0.72	1.50 + 0.72	1.50 + 0.72

Table E. 2 Kinematic comparison between ML data and five MB models. Values are in degrees

Joint	Functional Hip Joint (no alignment)	Bell (No Alignment)	Harrington – Pelvis and Leg (No Alignment)	Harrington - Pelvis Only (No Alignment)	Harrington – pelvis Only (After LCS Alignment)
Pelvis Tilt	5.47 ± 3.24	5.47 ± 3.24	5.47 ± 3.24	5.47 ± 3.24	2.29 ± 1.46
Pelvis List	2.58 ± 1.12	2.58 ± 1.12	2.58 ± 1.12	2.58 ± 1.12	1.65 ± 0.97
Pelvis Rotation	2.41 ± 1.02	2.41 ± 1.02	2.41 ± 1.02	2.41 ± 1.02	1.64 ± 0.86
Hip Flexion	7.57 ± 4.31	6.35 ± 3.57	5.84 ± 3.30	5.84 ± 3.30	3.07 ± 1.52
Hip Adduction	5.78 ± 2.60	3.25 ± 1.44	3.25 ± 1.42	3.25 ± 1.42	2.25 ± 1.40
Hip Rotation	8.18 ± 4.65	8.15 ± 4.37	8.14 ± 4.44	8.14 ± 4.44	3.67 ± 1.42
Knee Flexion	5.39 ± 2.69	5.71 ± 3.14	4.29 ± 2.42	4.29 ± 2.42	2.27 ± 1.43
Knee Adduction	5.39 ± 2.03	3.52 ± 2.09	3.53 ± 2.16	3.53 ± 2.16	3.88 ± 1.59
Knee Rotation	6.16 ± 2.57	6.43 ± 2.81	6.39 ± 2.82	6.39 ± 2.82	2.80 ± 1.45

Ankle Flexion	3.39± 1.51	3.39± 1.51	3.39± 1.51	3.39± 1.51	2.70 ±1.47
Ankle Inversion	13.11 ± 5.01	13.11 ± 5.01	13.11 ± 5.01	13.11 ± 5.01	4.92 ±2.15
Ankle Rotation	5.79 ± 1.79	5.79 ± 1.79	5.79 ± 1.79	5.79 ± 1.79	4.42 ±2.37
Mean of all joints	6.28 ± 4.12	5.80 ± 4.15	5.61 ± 4.11	5.61 ± 4.11	3.12 ± 1.88

Appendix F – Mean Correction Angles

Table F. 1 Mean correction angles of MB data for males and females. Values are in degrees.

	Female	Male
Pelvis Tilt	3.67 + 3.85	1.16 + 4.18
Pelvis List	0.63 + 1.50	0.38 + 2.69
Pelvis Rotation	-0.36 + 1.60	0.12 + 1.25
Hip Flexion	-1.09 + 4.41	-2.11 + 4.92
Hip Adduction	2.01 + 3.11	0.36 + 3.67
Hip Rotation	6.41 + 7.14	0.31 + 5.78
Knee Flexion	-4.10 + 3.24	-1.93 + 2.97
Knee Adduction	-0.28 + 1.69	0.05 + 0.93
Knee Rotation	1.31 + 3.65	3.07 + 2.54
Ankle Flexion	0.65 + 2.76	1.33 + 1.67
Ankle Inversion	-2.17 + 2.14	-2.19 + 2.11
Ankle Rotation	-13.85 + 4.96	-9.79 + 4.92

Appendix G – Q-Q plots

Q-Q plots created to evaluate normality of variables prior to statistical analysis.

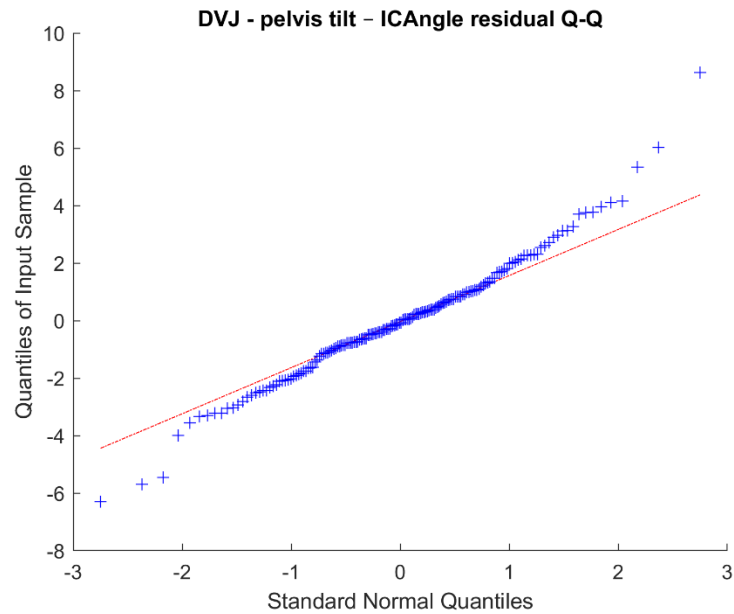


Figure G. 1 Q-Q Plot: DVJ – Pelvis Tilt – Initial Contact Angle

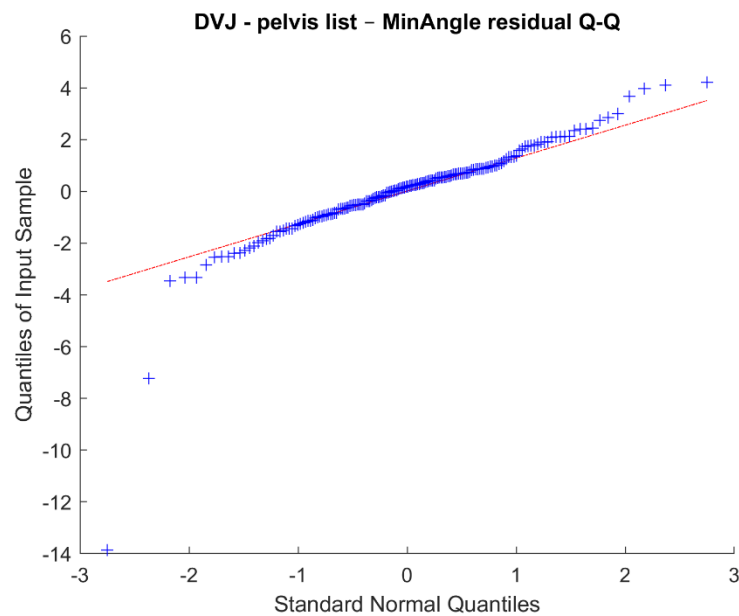


Figure G. 2 Q-Q Plot: DVJ – Pelvis List – Minimum Angle

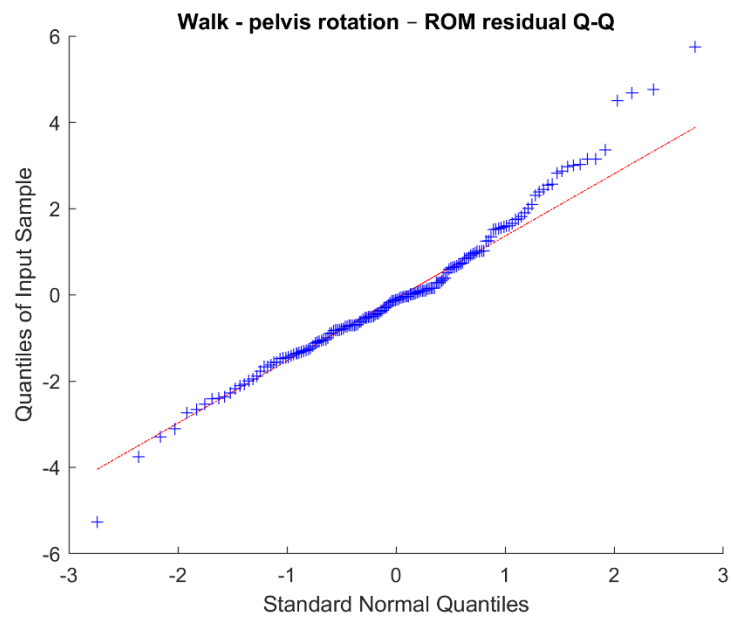


Figure G. 3 Q-Q Plot: Walk – Pelvis List – Minimum Angle

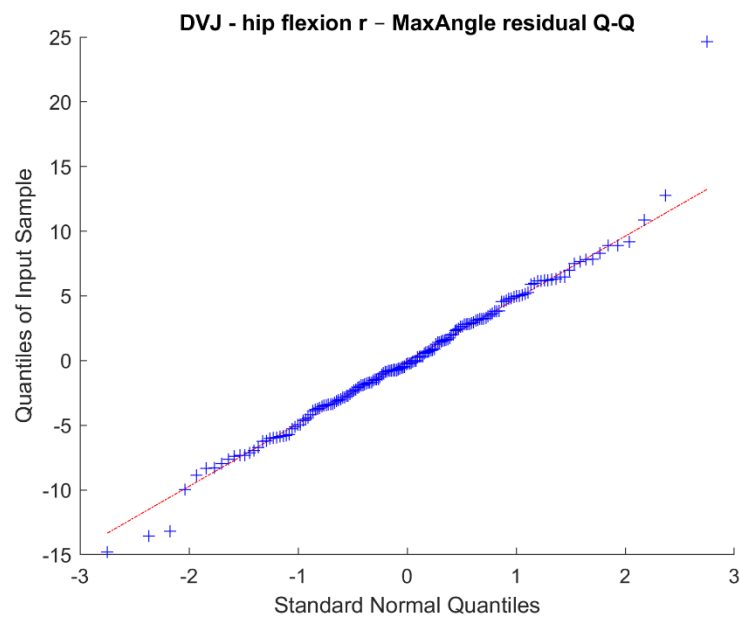


Figure G. 4 Q-Q Plot: DVJ – Hip Flexion R – Maximum Angle

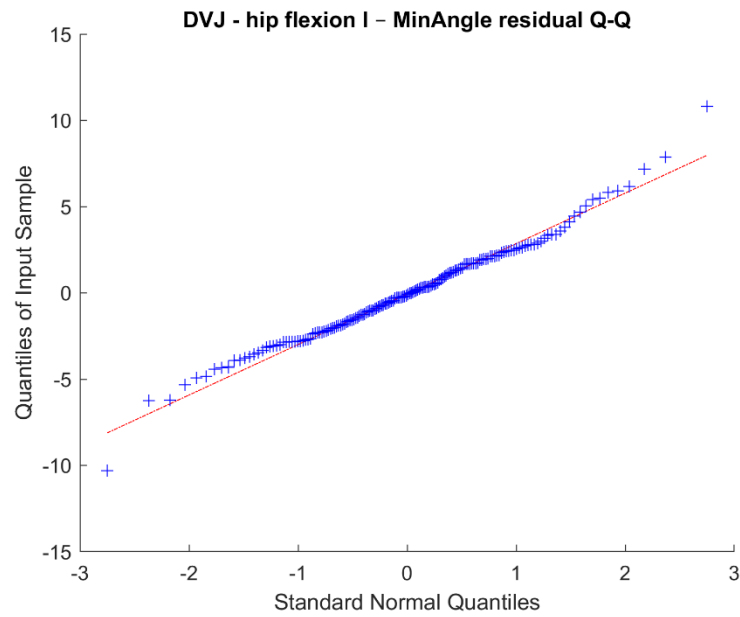


Figure G. 5 Q-Q Plot: DVJ – Hip Flexion L – Minimum Angle

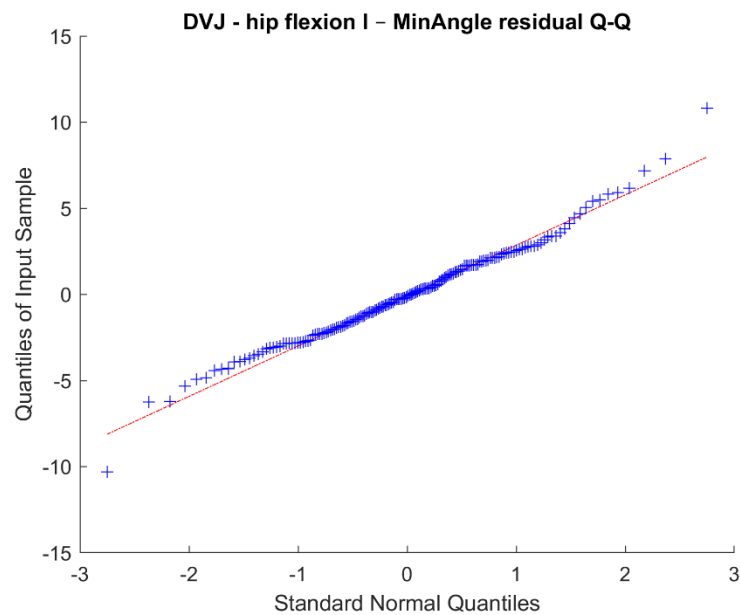


Figure G. 6 Q-Q Plot: Walk – Hip Flexion L – Minimum Angle

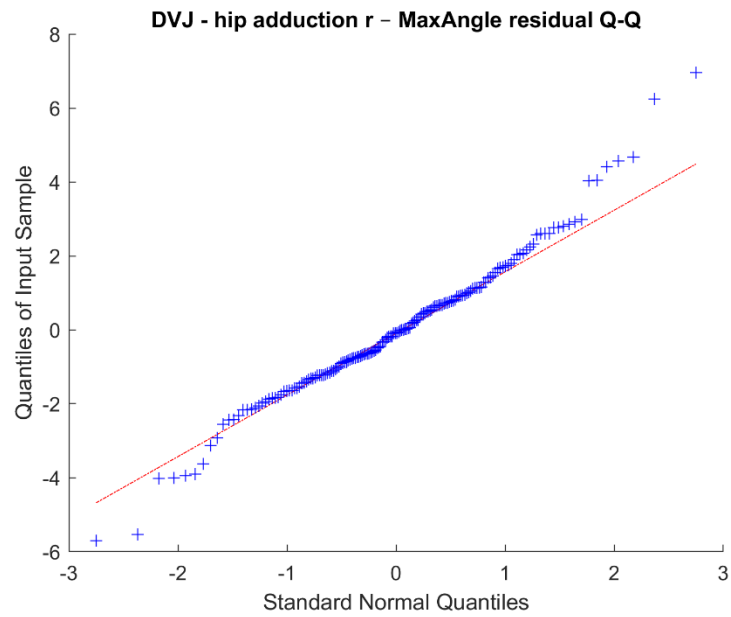


Figure G. 7 Q-Q Plot: DVJ – Hip Flexion L – Maximum Angle

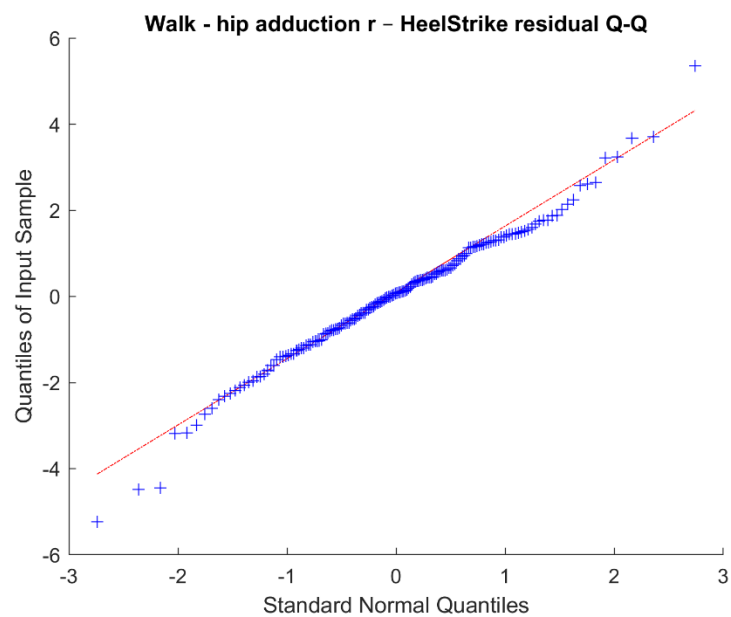


Figure G. 8 Q-Q Plot: Walk – Hip Adduction R – Initial Contact Angle

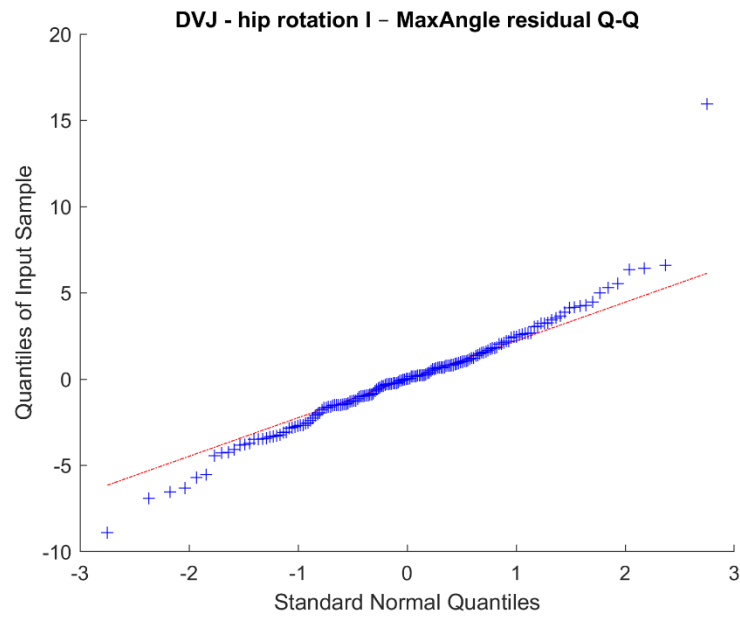


Figure G. 9 Q-Q Plot: DVJ – Hip Adduction R – Maximum Angle

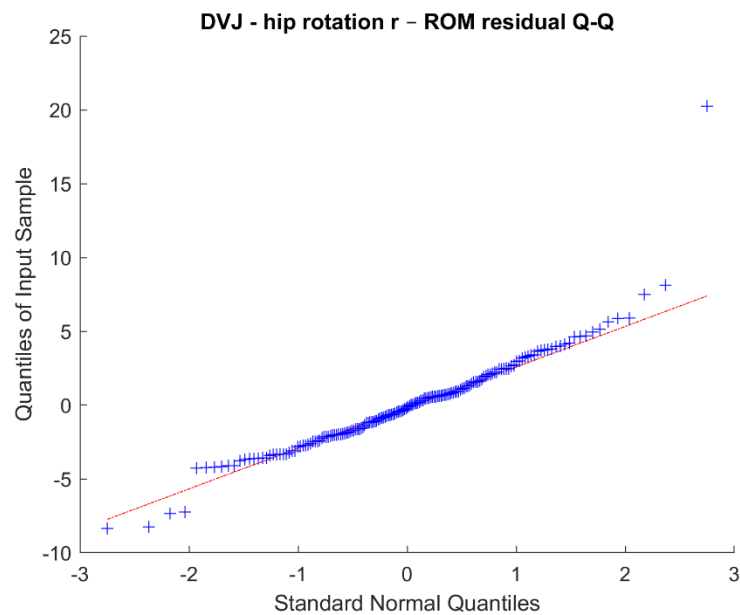


Figure G. 10 Q-Q Plot: DVJ – Hip Rotation R – ROM

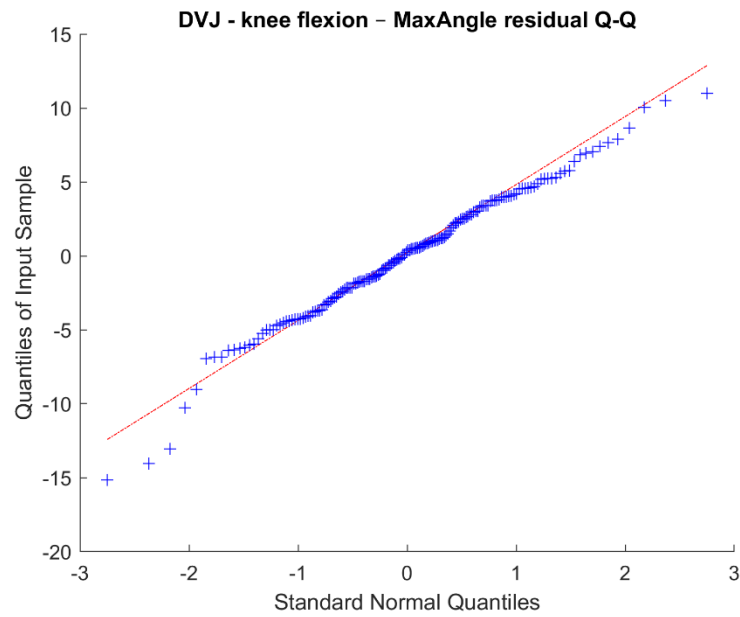


Figure G. 11 Q-Q Plot: DVJ – Knee Flexion R – Maximum Angle

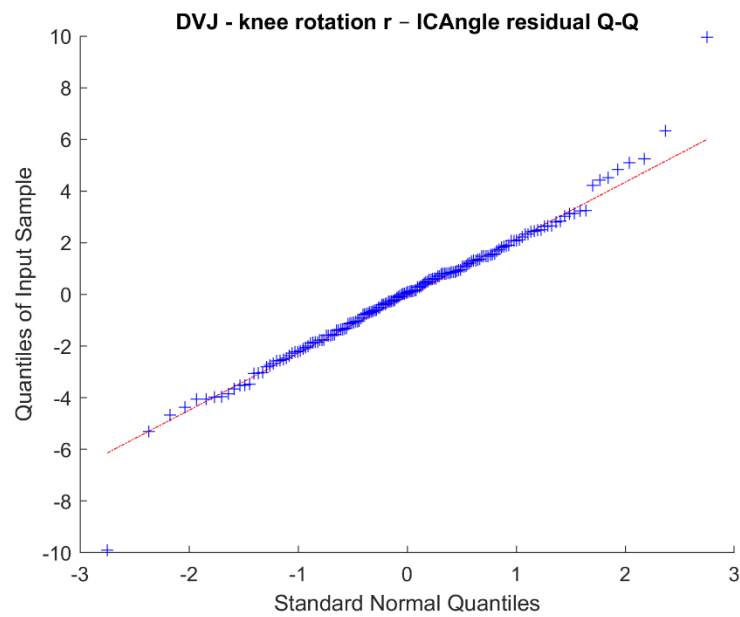


Figure G. 12 Q-Q Plot: DVJ – Hip Rotation R – Initial Contact Angle

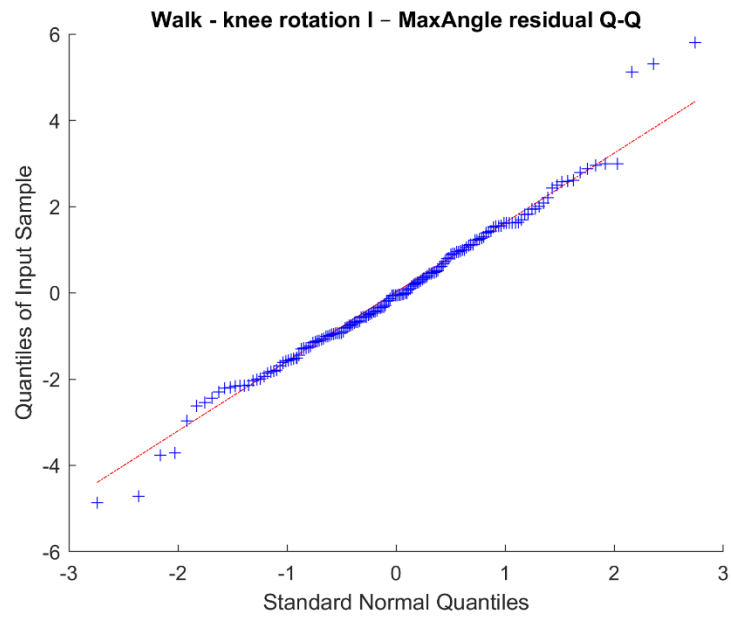


Figure G. 13 Q-Q Plot: Walk – Knee Rotation L – Maximum Angle

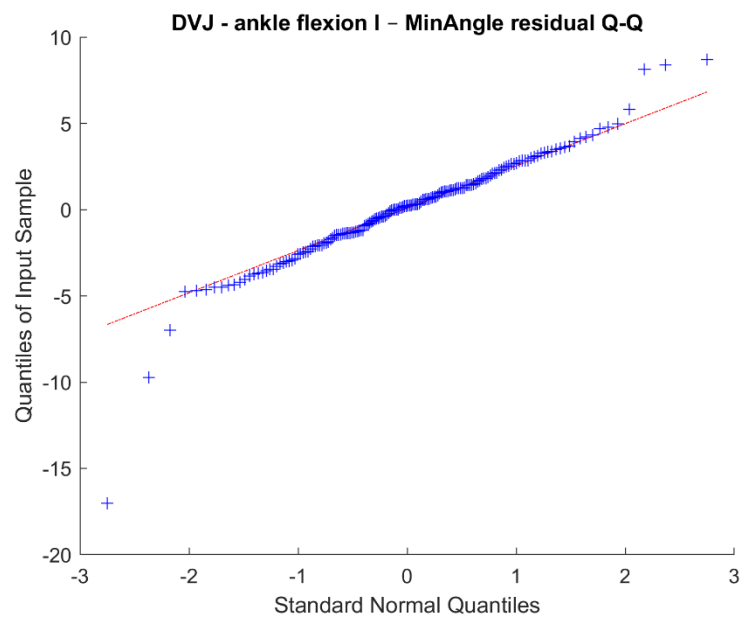


Figure G. 14 Q-Q Plot: DVJ – Ankle Flexion R – Minimum Angle

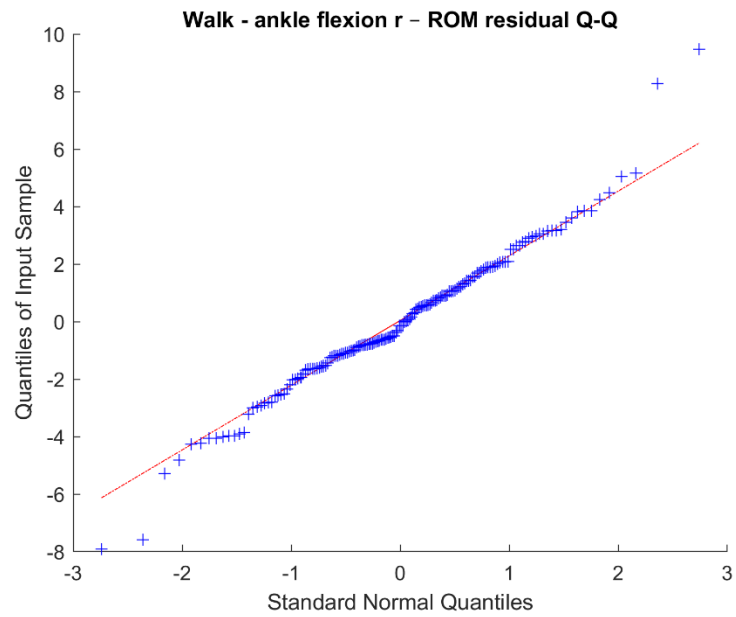


Figure G. 15 Q-Q Plot: Walk – Ankle Flexion R – ROM

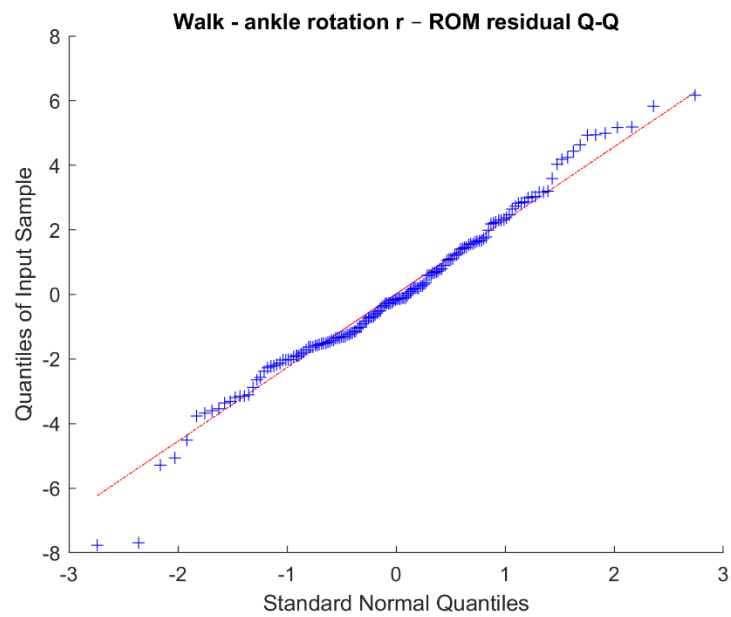


Figure G. 16 Q-Q Plot: Walk – Ankle Rotation R – ROM

Appendix H – LMM Results of Similarly Sized Participants

Due to the large amount of participant size variability, particularly between males and females, an analysis of a subset of participants were analysed using the same LMM to determine if the same sex-specific observations were observed when a portion height and weight variability was removed. Six participants (p01, p03, p07, p10, p13, p22, and p27) were selected as they had a height of 165-176 cm, and a BMI of 20-26. The results of the LMM are presented in Table H 1 and Table H 2.

Table H 1: LMM results for joint ROM during walking in similarly sized participants

Joint	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)
Pelvis Tilt	<0.001	0.002	<0.001	0.007	0.038
Pelvis List	0.650	0.064	0.011	0.747	0.110
Pelvis Rotation	0.207	0.033	0.021	0.152	0.208
Hip Flexion R	0.347	<0.001	0.477	0.389	0.532
Hip Adduction R	<0.001	<0.001	<0.001	<0.001	0.005
Hip rotation R	0.331	0.011	0.103	0.514	0.033
Hip Flexion L	0.190	<0.001	0.004	0.154	0.901
Hip Adduction L	0.523	0.177	0.688	0.677	0.403
Hip rotation L	0.014	0.921	<0.001	0.034	0.505
Knee Flexion R	0.481	<0.001	0.117	0.530	0.756
Knee Adduction R	0.002	<0.001	<0.001	0.022	0.745
Knee rotation R	0.172	<0.001	0.012	0.295	0.953
Knee Flexion L	0.255	<0.001	0.655	0.268	0.342
Knee Adduction L	0.176	<0.001	0.695	0.672	0.343
Knee rotation L	<0.001	<0.001	<0.001	0.002	0.161
Ankle Flexion R	0.526	0.094	0.584	0.543	0.740
Ankle Adduction R	0.550	<0.001	0.681	0.678	0.786
Ankle rotation R	0.037	0.920	0.242	0.135	<0.001
Ankle Flexion L	0.009	0.023	0.246	0.011	0.081
Ankle Adduction L	0.067	<0.001	0.160	0.255	0.682
Ankle rotation L	<0.001	0.013	0.579	0.014	0.007

Table H 2: LMM results for joint ROM during DVJ in similarly sized participants

Joint	P Value - Sex	P Value - Method	Interaction	Gender p within MB (d)	Gender p within ML (d)
Pelvis Tilt	0.826	0.855	0.022	0.868	0.028
Pelvis List	0.015	0.843	0.006	0.189	0.097
Pelvis Rotation	0.874	0.011	0.364	0.862	0.431
Hip Flexion R	0.335	0.503	0.562	0.314	0.693
Hip Adduction R	0.038	0.001	0.306	0.182	0.391
Hip rotation R	0.757	<0.001	0.898	0.828	0.757
Hip Flexion L	0.456	0.322	0.570	0.442	0.795
Hip Adduction L	0.003	0.593	0.014	0.042	0.144
Hip rotation L	<0.001	<0.001	<0.001	0.004	0.972
Knee Flexion R	0.041	0.262	0.324	0.044	0.164
Knee Adduction R	<0.001	<0.001	0.242	0.002	0.006
Knee rotation R	0.143	<0.001	<0.001	0.343	0.093
Knee Flexion L	0.257	0.964	0.727	0.272	0.148
Knee Adduction L	0.642	<0.001	0.309	0.741	0.085
Knee rotation L	0.659	<0.001	0.335	0.709	0.293
Ankle Flexion R	0.999	<0.001	0.341	0.998	0.515
Ankle Adduction R	0.049	0.003	0.025	0.215	0.833
Ankle rotation R	0.064	0.021	0.032	0.118	0.658
Ankle Flexion L	0.402	<0.001	0.032	0.480	0.787
Ankle Adduction L	0.817	<0.001	0.378	0.864	0.428
Ankle rotation L	0.143	0.018	<0.001	0.174	0.275

Despite removing some variability in height and BMI, several of the principal findings remained unchanged. Significant main effects of sex were still observed for pelvis tilt, right hip adduction, right and left knee rotation, right knee adduction, left ankle flexion, and bilateral ankle rotation during walking. Likewise, significant sex-by-method interactions persisted for multiple pelvis, hip, and knee variables. During DVJ a significant interaction effect was observed for several variables, most notably for pelvis tilt, hip rotation, and ankle rotation. Although some sex effects observed in the full cohort were no longer statistically significant, likely due to the substantially reduced sample size, the persistence of many key findings suggests that the sex-dependent differences

identified in this study cannot be explained solely by differences in participant height, weight, or BMI.