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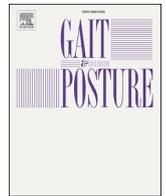
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Full length article

How to compare knee kinetics at different walking speeds?



Evvy Meinders^{a,*}, Marjolein J. Booiij^a, Josien C. van den Noort^{a,b}, Jaap Harlaar^{c,d}

^a Amsterdam UMC, Vrije Universiteit Amsterdam, Dept. Rehabilitation Medicine, Amsterdam Movement Sciences, de Boelelaan 1117, Amsterdam, Netherlands

^b Amsterdam UMC, Univ. of Amsterdam, Dept. Radiology and Nuclear Medicine, Medical Imaging Quantification Center, Amsterdam Movement Sciences, Meibergdreef 9, Amsterdam, Netherlands

^c Delft University of Technology, Dept BioMechanical Engineering, Mekelweg 2, Delft, Netherlands

^d Erasmus Medical Center, Dept Orthopedics, Rotterdam, Netherlands

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ABSTRACT

Background: Walking speed is a confounding factor in biomechanical analyses of gait, but still many studies compare gait biomechanics at comfortable walking speed (CWS) that is likely to differ between groups or conditions. To identify gait deviation unrelated to walking speed, methods are needed to correct biomechanical data over the gait cycle for walking speed.

Research question: How to compare knee kinetics over the gait cycle at different walking speeds?

Methods: 22 asymptomatic subjects walked on a dual-belt treadmill at CWS and 4 fixed speeds. Knee moments in sagittal (KFM) and frontal plane (KAM) were calculated via inverse dynamics. The net moment differences between CWS and fixed speed were expressed as a root-mean-square error (RMSE) normalized to the range of the variable. Two methods to correct for walking speed were compared. In method 1, KFM and KAM values were estimated based on interpolation between speeds at each percentage of the gait cycle. In method 2, principal component analysis was used to extract speed related features to reconstruct KFM and KAM at the speed of interest. The accuracy of both methods was tested using a leave-one-out cross validation.

Results: Walking speed influenced the magnitude and shape of KFM and KAM. To account for these speed influences using both methods, leave-one-out cross validation showed low normalized RMSE (< 5 %), with little difference between the two methods. RMSE for both reconstruction methods were up to 60 % lower than the RMSE between CWS and fixed speed.

Significance: Both methods could accurately correct knee kinetics over the gait cycle for the effects of walking speed. Walking speed dependency should be incorporated in each gait laboratory's reference dataset to be able to identify gait deviations unrelated to gait speed.

1. Introduction

Biomechanical analysis of gait is a method to objectively quantify musculoskeletal functions during normal and pathological gait [1]. Its potential is to support clinicians to identify, understand and quantify problems of walking, to inform their decision making. Walking speed is a factor that strongly affects the magnitude and shape of biomechanical waveforms in asymptomatic gait, already shown by Winter in 1983 [2–8]. Still, many studies compare biomechanical variables between groups or conditions when walking at different speeds. Often walking speed is standardized at comfortable walking speed (CWS); however that might be significantly different [9]. Using biomechanical analyses

at CWS, any difference found can either be due to actual difference in walking patterns or due to differences resulting from walking at a different speed [10,11]. For example, lower peak knee flexion moment during stance is seen in patients with knee osteoarthritis and total knee replacement [12,13], but is also seen in asymptomatic controls walking at a comparably low speed [4]. Therefore, to identify gait deviations unrelated to gait speed, biomechanical comparisons should be performed at matching walking speeds.

Previous studies have included walking speed as a covariate in the statistical analysis for biomechanical peak values [12,14,15]. Also, Lelas and colleagues analysed the relationship between speed and biomechanical peak values [16], such that peak values for a given speed can be

* Corresponding author.

E-mail addresses: evvy.meinders@griffithuni.edu.au (E. Meinders), m.booiij@amsterdamumc.nl (M.J. Booiij), j.vandennoort@amsterdamumc.nl (J.C. van den Noort), J.Harlaar@tudelft.nl (J. Harlaar).

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estimated and compared. However, by analysing only discrete values in the gait cycle, such as peak values, specific gait events need to be chosen prior to analysis and such preoccupation might hide valuable differences in gait biomechanics. Analysis of the whole gait cycle gives a comprehensive view of the differences in gait biomechanics.

The most straightforward method to make a valid comparison of gait biomechanics at matching speed, is by imposing a specific fixed walking speed. However, in some cases this imposed speed cannot be achieved, especially for patients with disturbed gait patterns. Additionally, pathological gait is often investigated at CWS, because these gait patterns relate best to the gait of the patient during daily life. When investigating pathological gait at CWS, these gait patterns can best be compared to gait of asymptomatic controls at matched walking speeds. However, it is not practical to collect gait data for asymptomatic controls at every CWS measured in the patient group, and therefore a method is needed to easily compare patients and asymptomatic controls at matched speed while patients can still walk at CWS.

Two possible methodologies exist to correct biomechanical data over the gait cycle. Fukuchi et al. [17] proposed a method to predict gait data over the whole gait cycle by analysing the relationship between gait data and speed at each time sample independently. For synthetic gait in computer animations, Glardon et al. [18] used principal component analysis to find specific features over the gait cycle dependent on speed. The advantage of this second method is that changes in waveform characteristics over time due to speed, such as timing or peak-to-peak differences, can be captured via specific features and are therefore incorporated in the reconstructed data.

In studying knee function in patients with knee osteoarthritis or total knee replacement the knee flexion moment (KFM) and knee adduction moment (KAM) are often used surrogates for knee joint loading [19,20]. Abnormal knee joint loading is seen as an important factor in the progression of knee osteoarthritis [21,22], and restoring normal knee joint loading is a goal of rehabilitation after TKR surgery. Many studies report differences in KFM and KAM at CWS between asymptomatic controls and (i) individuals with knee osteoarthritis or (ii) after total knee replacement surgery, ignoring the effect of walking speed on the biomechanics of knee function [23]. The aim of this study was to correct KFM and KAM over the whole gait cycle for effects of walking speed, as well as to compare two different methodologies.

2. Methods

2.1. Subjects & protocol

Lower limb data of 22 asymptomatic controls (13 males; Age: 65.9 ± 5.6 years; Height: 1.75 ± 0.1 m; BMI: 25.7 ± 3.1 kg/m²) were used in this study. The subjects walked at five walking speeds, besides comfortable walking speed (CWS, normalized speed ranged between 0.29 – 0.49), four fixed speeds were imposed based on normalized walking speed of [0.2, 0.3, 0.4 and 0.5]. Fixed walking speed was calculated for each subject, using the formula:

$$v = v_{nor} * \sqrt{(g \cdot L)} \quad (1)$$

with v fixed walking speed of subject, v_{nor} normalized walking speed ($v = [0.2, 0.3, 0.4$ and $0.5]$), g gravitational acceleration, and L leg length [24].

Subjects walked in a virtual environment on a dual-belt instrumented treadmill (2.20 m belt length), in the GRAIL (Gait Real-time Analysis Interactive Lab, MotekForceLink B.V., Netherlands). Familiarization trials of at least 5 min were performed before data collection started. The participants walked for 3 min at CWS and for 2 min at each fixed speed.

The protocol was approved by the Scientific and Ethical Review Board of the faculty of Behavioral and Movement Sciences of the VU Amsterdam and the Amsterdam UMC. All subjects signed an informed consent before measurements started.

2.2. Data acquisition

During each walking trial, 3D marker trajectories were captured via an InfraRed optoelectronic movement recording system with wireless, light-reflecting markers (Vicon, Oxford, UK; sample frequency = 100 Hz). For this study, 26 markers were placed on the subject, according to the CAST model [25], to reconstruct the position and orientation of the body segments (feet, lower legs, upper legs, pelvis, trunk) in space. Additionally, ground reaction forces (GRF) were measured via two 6D force plates (MotekForceLink B.V., the Netherlands; sample frequency = 1000 Hz), one for each belt and synchronized with motion data at 100 Hz.

2.3. Post processing

Marker data and force plate data were filtered using a two-way second order Butterworth filter with cut-off frequency of 6 Hz. External knee flexion (KFM) and adduction (KAM) moments were calculated via inverse dynamics, using BodyMech (www.bodymech.nl). Knee joint moments were expressed in the distal segment coordinate frame and normalized to body weight. All time-series data was time-normalized, such that one full gait cycle was represented by 100-time samples, i.e. % of the gait cycle.

2.4. Analysis

The net moment differences between the variables at CWS and fixed speed were expressed as a root-mean-square error (RMSE) as a percentage of the range of KFM and KAM at fixed speed.

For reconstructing data at different speeds, kinetic data was used of $n = 22$ subjects containing p variables ($p = 2$, sagittal and frontal plane moment of the knee joint) for y strides ($y = 4$) at z speeds ($z = 5$), all strides containing k time samples ($k = 100$).

2.4.1. Per sample interpolation method (IP method)

The mean value of the variable at each time sample over *all* subjects and y strides was calculated for each speed. Subsequently, a $z-1$ order polynomial was fitted to find the relation between speed and the mean values of the variable for all subjects at that time sample. Consequently, we were able to estimate the variable at each time sample for the speed of interest representing the variable over the whole gait cycle for that speed.

2.4.2. PCA interpolation method (PCA method)

The data was stored in p matrices \mathbf{M} with size $(n * z * y) \times (k)$. The data was mean centered, after which a principal component analysis (PCA) space was computed for KFM and KAM separately. The PCA space could be defined by u basis vectors (first principal components, PCs), such that the motion data could be approximated from the original data, using the following formula:

$$\vec{\theta} \cong \vec{\theta}_0 + \sum_{i=1}^u \xi_i \vec{v}_i = \vec{\theta}_0 + \vec{\xi} \mathbf{V} \quad (2)$$

with, $\vec{\theta}$ the motion vector (KFM or KAM), $\vec{\theta}_0$ the center of the PCA space, $\vec{\xi}$ the PC scores (representation of the data on the principal components) and \mathbf{V} a vector matrix containing the principal components v of \mathbf{M} .

The number of PCs u was selected such that 90 % of the total variance of matrix \mathbf{M} was accounted for. For each PC, the mean of the PC scores (ξ_i) over multiple strides at each walking speed was calculated and the relation between the mean PC scores and speed was determined by fitting of a $z-1$ order polynomial. Consequently, the PC scores corresponding to each principal component could be estimated for all normalized speeds within the speed range of 0.2 to 0.5 (i.e. speed of

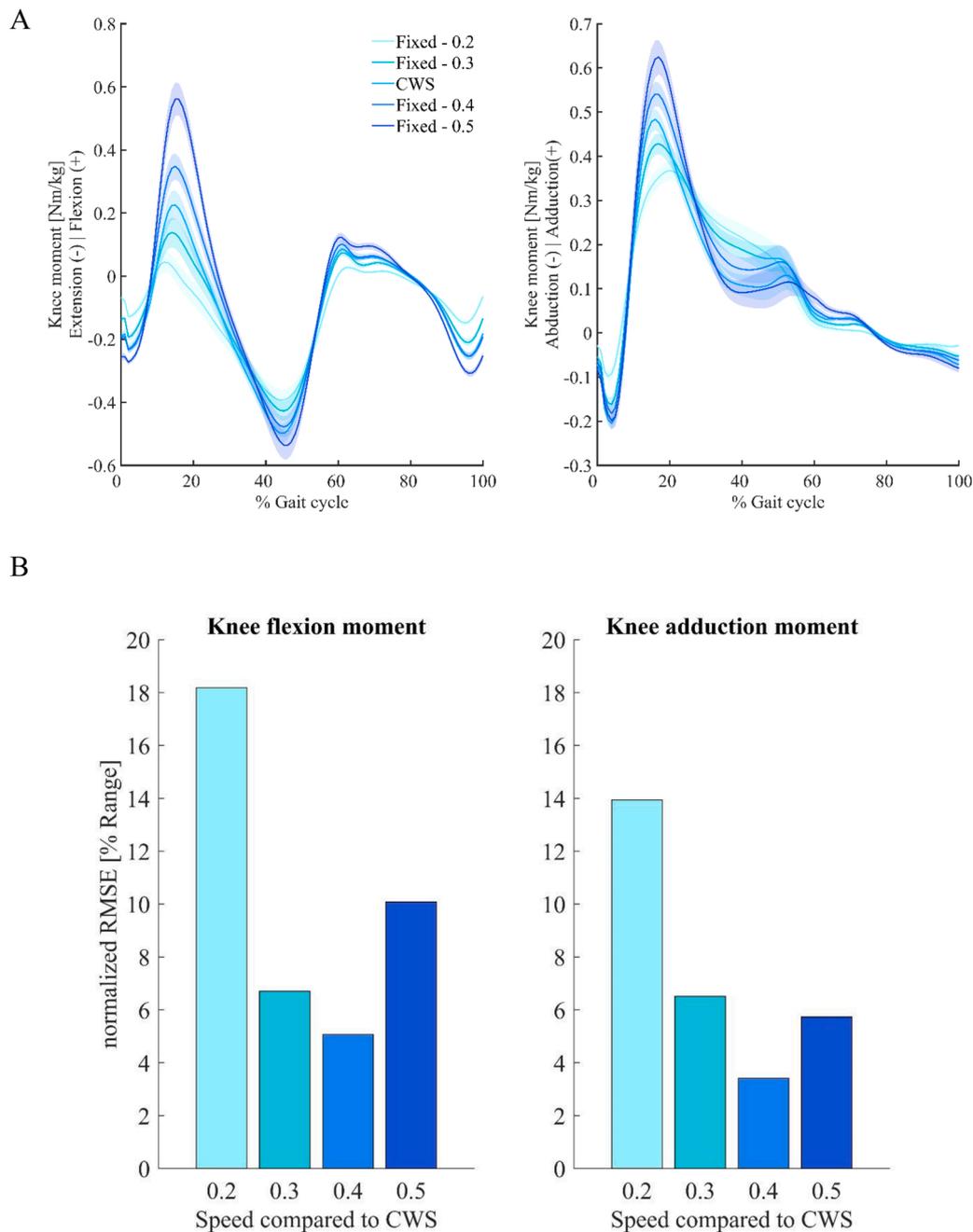


Fig. 1. (A) Knee flexion moment (KFM - left) and adduction moment (KAM - right) over the gait cycle for multiple normalized walking speeds. Shaded areas show the standard error of the mean. (B) Root mean square error between fixed speed and comfortable walking speed (CWS) normalized to the range of KFM and KAM for the fixed speed variable.

interest). Using the estimated PC scores ($\vec{\xi}$) and the corresponding PCs (V) in Eq. (2), kinetic data over the gait cycle for the asymptomatic controls at the speed of interest could be constructed.

2.5. Comparison of methods

To test the quality of both reconstruction methods a leave-one-out cross validation technique was used. In the leave-one-out cross validation, data of KFM and KAM at one speed for all participants was removed from the analysis, and the mean (reference) was tested against the reconstruction of KFM and KAM using both reconstruction methods using the data for KFM and KAM at the other 4 speeds. The RMSE over the gait cycle was calculated between the reconstructed variable and its

reference for both methods. The RMSEs were expressed as percentage of the total range (maximum - minimum) of the reference. Additionally, to visualize the effect of speed correction, the difference was calculated between (i) the RMSE of the variable at CWS and the reference variable and (ii) the RMSE of the reconstructed variable and the reference variable. This RMSE difference was expressed as a percentage of the RMSE of the variable at CWS and the reference variable.

3. Results

3.1. Effect of speed

The KFM peak and the KAM peak during loading response (~ 20 % gait cycle), and the knee extension moment peak during late stance (~

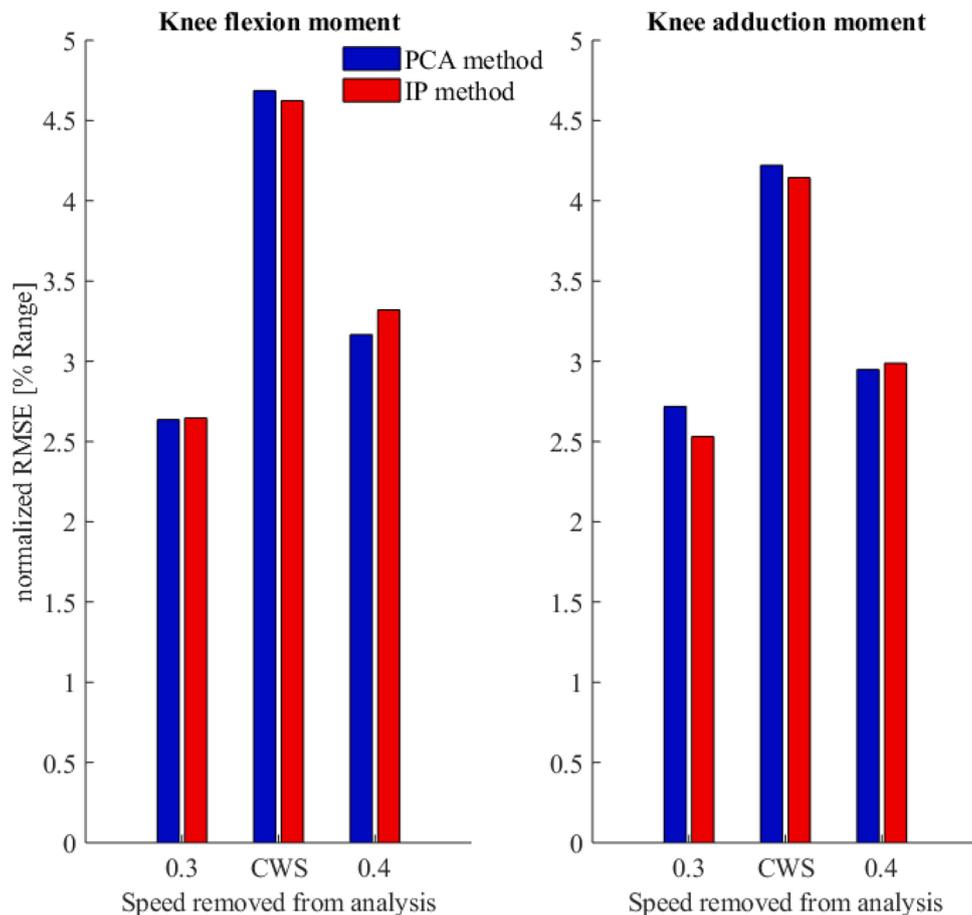


Fig. 2. The root-mean-square error (RMSE) of the reconstructed and reference value in the leave-one out cross-validation. The RMSE is presented as percentage of the range of the reference variable for knee flexion moment (KFM – left) and adduction moment (KAM – right) between the reconstructed and removed data for all speeds. The PCA method is shown in blue and the IP method in red.

45 % gait cycle) and late swing (~ 95 % gait cycle) increase with increasing walking speed (Fig. 1A). The normalized RMSE between the fixed speeds and CWS ranged between 3 % and 18 % of the range of the fixed speed variable and was highest for the lowest walking speed (Fig. 1B).

3.2. Comparison of both methods using a leave-one-out cross validation

Results for the cross-validation are shown for the three speeds for which interpolation was possible (0.3, CWS and 0.4). Low normalized RMSE values (<5 % of the range of the tested variable) are found for both methods when estimating KFM and KAM (Fig. 2). The normalized RMSE values ranged from 2.6 to 4.7 % and 2.5–4.2 % for KFM and KAM, respectively. Both methods showed similar average normalized RMSE values for both KFM (PCA method: 3.5 ± 1.1 %, IP method: 3.5 ± 1.0 %) and KAM (PCA method: 3.3 ± 0.8 %, IP method: 3.2 ± 0.8 %). For all speeds, the reconstructed variables fell within the standard deviation of the excluded variable.

Both reconstruction methods reduced the RMSE relative to the reference variable up to 60 % compared to the variable at CWS (Fig. 3). The RMSE of both reconstructions relative to the reference variable was in all cases lower than the RMSE of the CWS relative to the reference variable. Little difference was found between the PCA method and the IP method (1–3 % difference) for difference between the RMSE of the CWS and the RMSE of the reconstruction methods relative to the reference variable.

4. Discussion

The aim of this study was to correct KFM and KAM over the whole gait cycle for effects of walking speed. Two different methodologies (PCA and IP method) to correct for walking speeds were compared. Both methods were able to correct for walking speed and estimate the KFM and KAM with normalized RMSE values below 5 % of the range of the tested variable. Little difference was found in normalized RMSE values between both methods, showing that both methods are equally accurate in correcting the KFM and KAM over the gait cycle for walking speed. Correcting for walking speed reduced the RMSE with the fixed speed up to 60 % compared to using CWS, showing the importance of correcting biomechanical data for walking speed.

The IP method used in this study is similar to the method proposed by Fukuchi et al. [17], in which the relation between gait data and speed at each time sample was estimated using a first or second order polynomial. Fukuchi et al. showed RMSE of the KFM for multiple normalised speeds (ranging between 0.1–0.7) of about 0.1 Nm/kg, which we estimated to be around 16 % and 20 % of the range of the tested variable for a normalized speed of 0.3 and 0.4, respectively (ranges extracted from Fig. 2 Fukuchi et al.). These values reported by Fukuchi et al. are much higher than the values reported in this study (2.6 % and 3.3 % for speed 0.3 and 0.4 for the IP method respectively). In the study of Fukuchi et al. participants walked at speeds relative to their CWS and therefore not at the same walking speed. Therefore, the experimental KFM data at a relative speed contains KFM data at multiple speeds and may not have been a good reference, possibly explaining the higher RMSE values reported in their study. Moreover, Fukuchi et al. compared the

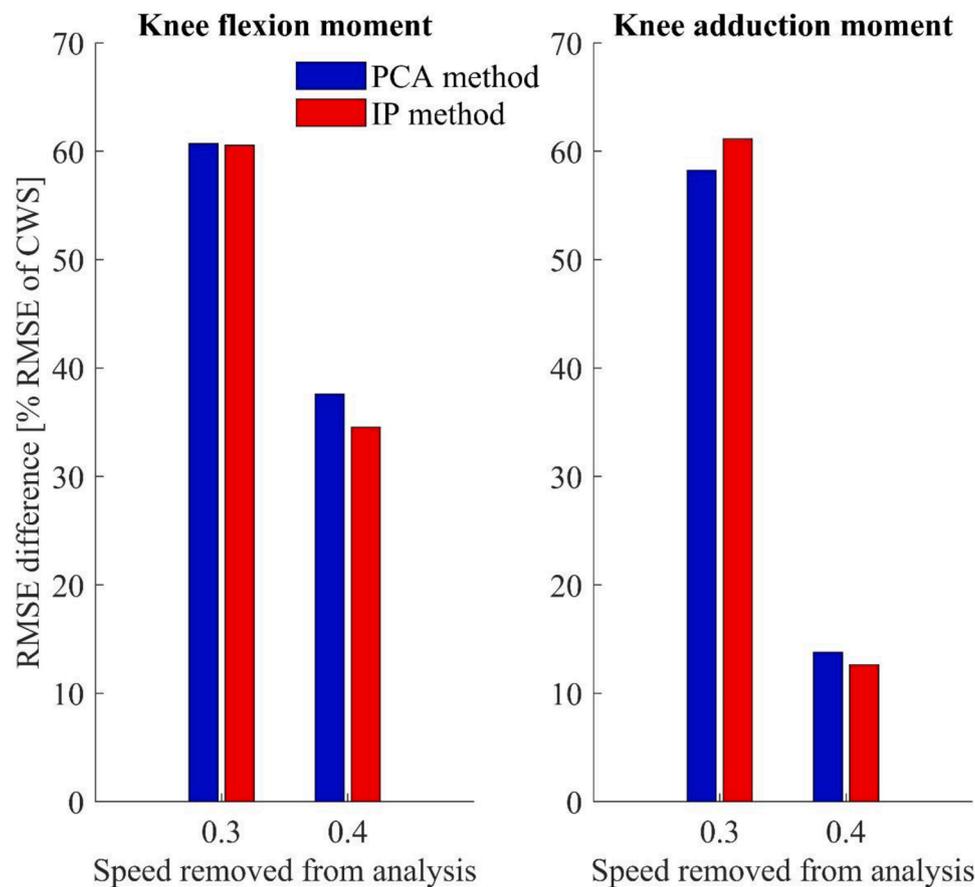


Fig. 3. The difference between the root-mean-square error (RMSE) of (i) the variable at comfortable walking speed (CWS) and the reference variable and (ii) the reconstructed variable and the reference variable. The RMSE difference is presented as percentage of the RMSE of CWS (i) for knee flexion moment (KFM – left) and adduction moment (KAM – right). The PCA method is shown in blue and the IP method in red.

reconstructed KFM with data for each participant, whereas in this study the reconstructed KFM was compared with the mean KFM for all participants. The greater variability in individual data compared to the mean is likely to increase the RMSE values and could also explain the higher RMSE values for KFM in Fukuchi's study compared to this study. It depends on the application scenario of walking speed correction, i.e. to compare group results or to compare individual results, which RMSE values apply.

The results of this study not only show that both reconstruction methods are able to accurately correct KFM and KAM for walking speed, but this speed correction will improve the comparison of controls versus slow walking patients substantially compared to a comparison at CWS. Limited differences between both reconstruction methods are seen. The advantage of the PCA method above the IP method is that both timing and magnitude differences can be incorporated in the reconstruction, whereas in the IP method time samples are tested independently.

In this study only data is presented for speeds where interpolation was possible since extrapolation leads to inaccurate estimates. Therefore, it is important to note that if one wants to collect a reference dataset, the walking speeds included in the dataset need to cover a large range of speeds, such that extrapolation is avoided when reconstructing data at patients walking speed.

4.1. Limitations

In this study only 5 speeds were used for analysis. Although the results of the cross-validation were good (normalized RMSE <5 % of the range of the tested variable), adding more speeds will likely improve the estimation of the biomechanical variable at each speed. Additionally, it

is important to keep in mind that PCA assumes that the data included is independent. Since multiple strides of the same person at different speeds are included, this assumption fails in our analysis. Other more complex PCA analyses (e.g. PCA in 3D) can be used in future studies, such that the assumption of independent data hold and the reconstruction will be improved. Also, including more individuals in the PCA analysis is likely to improve the accuracy of the reconstructed variable. Nevertheless, the current PCA method was already able to estimate the KFM and KAM accurately. Another limitation of this study is that we only looked at KFM and KAM. Speed effects are also shown in other joint moments, joint angles, ground reaction forces and muscle activation patterns [4,6,26,27]. Therefore, future research investigating other biomechanical variables than KFM and KAM may also benefit from this speed correction. Since the hip and ankle joint are commonly investigated in biomechanical analysis, we presented the results for the hip (frontal and sagittal plane) and ankle (frontal plane) in the supplementary materials.

5. Conclusion

This study compared principal component analysis and point interpolation methods to account for walking speed in net joint moments over the gait cycle. Both methods could accurately correct KFM and KAM for walking speed. Since speed affects biomechanical gait data used to inform clinical decision making, correcting for walking speed matters when investigating the effect of pathology on gait. Therefore, every gait laboratory is advised to collect data at multiple walking speeds when collecting a lab reference data set, such that the reference set can be used for comparisons at matched walking speed for a range of

patient groups.

Declaration of Competing Interest

The authors report no declarations of interest.

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Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:<https://doi.org/10.1016/j.gaitpost.2021.06.004>.

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