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A principal component analysis approach to correcting the knee flexion axis during gait

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ABSTRACT

Accurate and precise knee flexion axis identification is critical for prescribing and assessing tibial and femoral derotation osteotomies, but is highly prone to marker misplacement-induced error. The purpose of this study was to develop an efficient algorithm for post-hoc correction of the knee flexion axis and test its efficacy relative to other established algorithms. Gait data were collected on twelve healthy subjects using standard marker placement as well as intentionally misplaced lateral knee markers. The efficacy of the algorithm was assessed by quantifying the reduction in knee angle errors. Crosstalk error was quantified from the coefficient of determination (r^2) between knee flexion and adduction angles. Mean rotation offset error (α_o) was quantified from the knee and hip rotation kinematics across the gait cycle. The principal component analysis (PCA)-based algorithm significantly reduced r^2 ($p<0.001$) and caused $\alpha_{o,knee}$ to converge toward $11.9 \pm 8.0°$ of external rotation, demonstrating improved certainty of the knee kinematics. The within-subject standard deviation of $\alpha_{o,hip}$ between marker placements was reduced from $13.5 \pm 1.5°$ to $0.7 \pm 0.2°$ ($p<0.001$), demonstrating improved precision of the knee kinematics. The PCA-based algorithm performed at levels comparable to a knee abduction–adduction minimization algorithm (Baker et al., 1999) and better than a null space algorithm (Schwartz and Rozumalski, 2005) for this healthy subject population.

1. Introduction

Gait analysis provides critical data on patient dynamic functionality upon which orthopedic surgeons rely for pre- and post-intervention assessments (Filho et al., 2008; Lofterod and Terjesen, 2008; Saraph et al., 2002; Wren et al., 2011). Alternative forms of assessment, such as static magnetic resonance imaging (MRI), physical examination, or visual analysis, do not provide accurate and precise quantification of a patient's capabilities during dynamic activities. Among patients with suspected tibial or femoral torsion, knee and hip kinematics are a critical component of de-rotation osteotomy decisions (Aminian et al., 2003; DeLuca et al., 1997; Ounpuu et al., 2002). Gait analysis is consulted to identify whether surgery is required to create neutral alignment of the lower extremity segments during stance phase and reduce off-axis loading of the knee (Bennett et al., 1985; Stefko et al., 1998). These surgeries are invasive, expensive, and require lengthy recovery periods (Krengel and Staheli, 1992; Staheli et al., 1985), placing significant weight on the validity and reliability of the measured gait kinematics.

Motion-capture marker misplacement has previously been identified as the largest source of between-laboratory kinematic variability – accounting for up to 75% of the overall variance (Gorton et al., 2009) – as well as within-laboratory variability (Kadaba et al., 1989). Therefore, improving the validity and reliability of gait kinematics by addressing human marker placement error is critical to improving the internal validity of gait analyses. Derotation osteotomy decisions depend specifically on the placement of the anatomical markers that define the knee rotation axis. Misplacement of these markers can lead to mean rotation offset error of the hip and knee as well as crosstalk between knee flexion and adduction angles (Baker et al., 1999; Kadaba et al., 1990; Piazza and Cavanagh, 2000), and, ultimately, ineffective or harmful surgical interventions. A method is needed to consistently and reliably ensure correct identification of the knee flexion axis.

Other knee flexion axis correction techniques have been explored in the literature, including iterative, statistical, and hardware-based approaches (Baker et al., 1999; Charlton et al.,

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2004; Ehrig et al., 2007; Halvorsen et al., 1999; Schache et al., 2006; Schwartz and Rozumalski, 2005), but none have been reduced to routine clinical practice. This may, in part, be due to lack of practicality. Some methods were designed or are generally implemented as ad-hoc techniques, identifying the functional knee axis using range-of-motion (ROM) trials (i.e. squats or passive knee flexion) prior to data collection (Groen et al., 2012; Schwartz and Rozumalski, 2005). These functional techniques are less susceptible to human error than a traditional marker-only approach, but have the disadvantage that pediatric, cognitively impaired, very weak, or limited-mobility patients are often incapable of or unwilling to participate in additional components of a data collection. A more practical approach for such patients is to implement an algorithm that utilizes the standard walking trials as a means to identify the true axes of joint rotation. The Baker et al. (1999) knee abduction–adduction minimization method has demonstrated success as a post-hoc application; utilizing the subject's walking trials to correct the knee flexion axis after data collection. In direct comparison studies, this algorithm reduced kinematic errors (Baker et al., 1999; Schache et al., 2006) and matched ultrasound measurements of tibial torsion in healthy subjects (Fullenkamp, 2005). A potential weakness of this algorithm is its inefficient design; it identifies the correct axis based on iterative guessing rather than statistical optimization. Although the resulting computational demand is negligible with modern computing power, the precision of the correction is dependent on the number of iterations of the algorithm. A more pressing weakness of this algorithm is its self-stated ineffectiveness in patient populations with large abduction–adduction motion (Baker et al., 1999). What is lacking, therefore, is an efficient knee flexion axis correction method that is not iterative, that is designed to distinguish flexion motion even under conditions of increased abduction–adduction ROM, and that can be applied post-hoc to gait data alone.

The primary purpose of this study was to develop an efficient algorithm to correct the knee flexion axis post-hoc. The secondary purpose was to quantify the efficacy of this method as compared to two established algorithms. We evaluated the efficacy of the algorithm by analyzing gait data from healthy subjects walking at self-selected speeds with standard and misplaced lateral knee markers. By comparing the knee and hip kinematics before and after the knee flexion axis correction for each marker placement, we addressed three specific hypotheses. We hypothesized first that the correction algorithm would improve the knee flexion axis orientation certainty, as evidenced by the minimization of crosstalk between knee flexion and adduction. Second, we hypothesized that the algorithm would improve the knee flexion axis precision, as evidenced by decreased within-subject variability of the mean knee and hip rotation offsets. Finally, we hypothesized that the algorithm would perform better as a post-hoc correction method than previously-established algorithms, as evident by less error and greater precision. Due to the importance of mitigating additional functional trials with challenging patient populations, all algorithms under comparison – regardless of intended design – were applied post-hoc to standard gait trials to establish the relative effectiveness under these conditions.

2. Theoretical basis of PCA correction algorithm

In healthy and many pathologic knees, the majority of motion occurs about the flexion axis. Healthy knees have a flexion–extension ROM of approximately 60° and a range of abduction–adduction of approximately 6° during gait, based on bone pin studies (Lafortune et al., 1992). As a comparative example, the knee flexion–extension ROM of children with cerebral palsy, a patient population with a high prevalence (~13%) of rotational malalignment (Wren et al., 2005), may be reduced to 25° in conditions of crouch (Lin et al., 2000). Abduction–adduction ROM has not been identified as deviating from the healthy individuals in this population (Bell et al., 2002). We aimed to leverage the fact that knee flexion–extension ROM is greater than knee abduction–adduction ROM as a means to identify the flexion axis. The greater knee flexion ROM results in a dominant trajectory of the shank relative to the thigh, which can be identified statistically using Principal Component Analysis (PCA) (Pearson, 1901).

Traditionally, the knee flexion axis is defined as perpendicular to the long axis of the thigh, in the plane of the knee joint center (KJC), hip joint center (HJC), and a technical marker placed on the lateral thigh or knee (Baker et al., 1999; Kadaba et al., 1990). Misplacement of the technical marker in the superior–inferior direction has no effect on the final flexion axis orientation because it does not alter the plane definition. Anterior–posterior misplacement, however, generates crosstalk between knee flexion and adduction (Kadaba et al., 1990; Piazza and Cavanagh, 2000). On this basis, we can optimize algorithm efficiency by isolating the correction to the transverse plane of the thigh as identified by the HJC and KJC landmarks. Furthermore, by assuming a circular knee centered at the midpoint of the medial and lateral epicondyle markers, we can reduce the degrees of freedom to a single parameter: the angle between the uncorrected, or *marker-based*, knee flexion axis and the corrected, or *motion-based*, knee flexion axis.

Motion about the knee is tracked with the highest sensitivity by using the most distal marker on the shank segment, i.e. the lateral ankle marker. To isolate the data to the plane of interest, the trajectory of the lateral ankle marker can be projected onto the transverse plane of the *marker-based* thigh coordinate system. The first principal component of this trajectory identifies the direction of greatest variance of ankle motion about the knee during gait, which is considered the direction of flexion–extension. Therefore the second principal component – the vector perpendicular to the first principal component in the transverse plane – represents the axis about which flexion–extension occurs, (i.e. the *motion-based* flexion axis).

The authors acknowledge some similarities of the proposed correction algorithm to that developed by Halvorsen et al. (1999), which uses the secondary and tertiary principal components of the instantaneous marker displacements to identify the knee flexion axis. The proposed PCA algorithm differentiates itself in two main ways. First, we have simplified the 3D eigenvalue problem addressed by Halvorsen et al. (1999) to 2D using the most reliable marker-based joint axis – the longitudinal axis (Schache et al., 2006) – with the purpose of increasing the algorithm's robustness under conditions of noise. Second, the data input is the instantaneous marker positions rather than the instantaneous marker displacement vectors, which are highly susceptible to noise due to the short (< 0.5 s) time span between samples.

This PCA-based algorithm was developed and initially tested using custom in-house software (MATLAB, The MathWorks, Natick, MA). The algorithm was subsequently built into Visual3D (C-Motion Inc., Germantown, MD).

3. Methods

A convenience sample of 12 healthy subjects (5 male; age 25.2 ± 3.0 years; BMI 22.1 ± 2.1 kg/m^2) participated in this study. The protocol was approved by the Mayo Clinic Institutional Review Board and written informed consent was obtained from all participants prior to data collection.

Marker trajectories (Fig. 1) were recorded as subjects walked at a self-selected speed (120 Hz, Motion Analysis Corporation, Santa Rosa, CA). Four walking trials were recorded for three marker placement conditions. A modified Helen Hayes marker set (Davis et al., 1991) was placed by a licensed physical therapist with

4.5 years of experience in the gait laboratory after testing and confirming that the subjects had no knee laxity.

The anatomical coordinate system of the thigh was defined during a static trial from the locations of the hip joint center (HJC), knee joint center (KJC), and lateral knee marker (Fig. 1). The lateral knee marker was placed in three different positions for each limb: standard, anterior (3.3 cm), and posterior (3.3 cm) with respect to standard (Fig. 2). This method was similar to a previous study (Fullenkamp, 2005). The stated distance was chosen because it represents the expected worst case scenario of marker misplacement as 3.3 cm is the width of the reflective marker base. The technical coordinate system of the thigh, based on three thigh markers (TH1, TH2, TH3; Fig. 1), was used to track the anatomical thigh coordinate system during the dynamic trials.

The anatomical coordinate system of the shank was defined in the static trial from the KJC, the lateral ankle marker, and the medial ankle marker (Fig. 1). The technical coordinate system of the shank, used to track the anatomical coordinate system during the dynamic trials, was defined from the ankle joint center, the lateral ankle marker, and a shank marker (Shank1; Fig. 1).

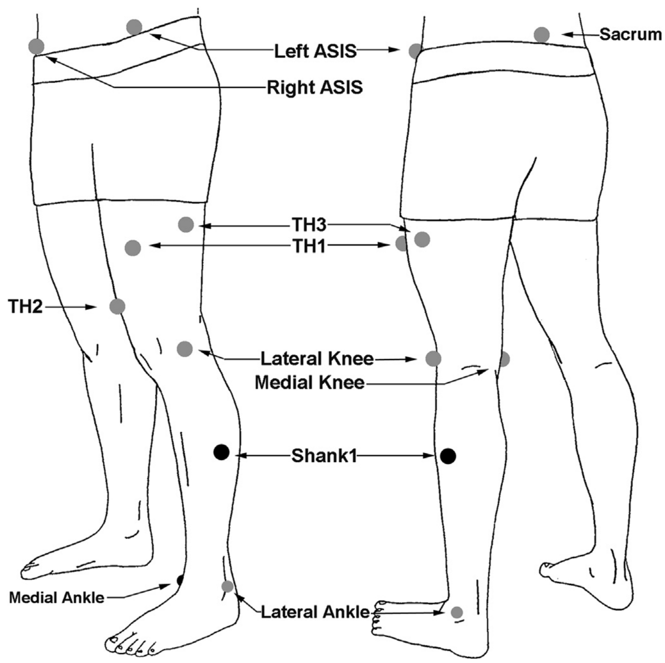

Fig. 1. Three-dimensional trajectories of nineteen lower extremity markers were recorded. All 19 markers were used to calculate gait kinematics. The location of the hip joint center (HJC) was calculated from the left and right anterior superior iliac spine (ASIS) markers and the sacrum marker (mid-point of the posterior superior iliac spines) using the Harrington hip model (Harrington et al., 2007). The KJC was defined as the midpoint between the medial and lateral knee markers, placed on the medial and lateral femoral epicondyles, respectively. The ankle joint center was defined as the midpoint between the medial and lateral ankle markers, placed on the medial and lateral malleoli, respectively. Shown in gray are the markers used for the PCA correction algorithm for a single limb.

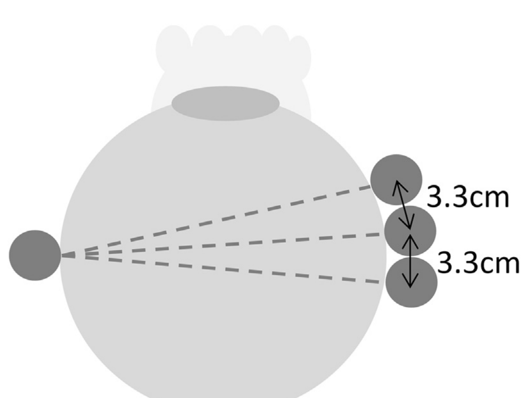

Fig. 2. Model of the transverse cross-section of a right knee with all three lateral knee markers. Marker-based flexion axes are indicated by the dashed lines.

Two established algorithms for correctly identifying the knee flexion axis were selected for efficacy comparison: a null space (NS) algorithm (Schwartz and Rozumalski, 2005) and a knee abduction–adduction minimization (KAAM) algorithm (Baker et al., 1999). All three algorithms were built into Visual3D, which was used for data processing. Independent thigh and shank segments were generated for each of the algorithms under comparison. The walking trials were used as “functional trials” for both the KAAM and NS methods. Knee and hip joint angles, normalized to the gait cycle, were exported for each marker placement condition and correction algorithm. Each subject-limb was treated independently.

For each trial, systematic kinematic errors associated with knee marker misplacement were assessed to evaluate the efficacy of the correction algorithm. Crosstalk was quantified from the coefficient of determination (r^2) between knee adduction and flexion throughout the gait cycle. Additionally, the mean knee and hip rotation offsets ($\alpha_{o,knee}$ and $\alpha_{o,hip}$) were quantified as the mean of the rotation kinematics across the gait cycle. The main effects and interactions of knee marker placement (standard, anterior, and posterior) and algorithm (uncorrected, PCA, KAAM, NS) on r^2, $\alpha_{o,knee}$, and $\alpha_{o,hip}$ were assessed using linear mixed models with fixed (placement, algorithm) and random (limb, $n=24$) effects.

The changes in within-limb variability of $\alpha_{o,knee}$ and $\alpha_{o,hip}$ were assessed to evaluate the precision of the corrected flexion axis. For each individual limb and correction algorithm, the standard deviations of $\alpha_{o,knee}$ and $\alpha_{o,hip}$ across all walking trials and marker placements were calculated. The effect of the correction algorithm on the standard deviations of $\alpha_{o,knee}$ and $\alpha_{o,hip}$ was then compared using separate one-way analyses of variance.

All post-hoc comparisons were made with a Tukey–Kramer HSD adjustment. Statistics were evaluated using JMP (The SAS Institute Inc., Cary, NC). The significance level was defined as $p<0.05$.

4. Results

The KAAM and PCA correction methods reduced crosstalk between knee flexion and adduction regardless of marker placement. Both correction algorithms significantly reduced the coefficient of determination (r^2) relative to uncorrected for all marker placement conditions and were not statistically different from each other ($p<0.001$; Fig. 3). The KAAM algorithm-derived r^2 for the posterior marker placement condition was significantly higher than for the standard ($p=0.021$) or anterior ($p=0.043$) conditions (Fig. 3). The NS algorithm reduced r^2 for the anterior marker placement condition ($p<0.001$), did not change r^2 for the posterior marker placement condition ($p=0.433$), and increased r^2 for the standard marker placement ($p<0.001$; Fig. 3).

The KAAM and PCA algorithms corrected the knee and hip rotation kinematics of the anterior and posterior marker placement conditions to be consistent with the rotation kinematics of the standard marker placement condition. The mean knee and hip rotation offsets of the data corrected using the PCA and KAAM algorithms were not significantly different from the mean rotation offsets of the uncorrected standard marker placement data or from each other ($p \geq 0.24$; Figs. 4 and 5). However, the mean knee and hip rotation offsets of the posterior marker placement condition corrected with the NS algorithm were significantly different from the mean rotation offsets of the uncorrected standard marker placement data ($p<0.001$; Figs. 4 and 5). The combined mean knee rotation offset averaged across trials, limbs, and placements was 11.5 ± 7.5 (mean $\pm$ SD) degrees of external rotation with the PCA algorithm, 11.9 ± 8.0 with the KAAM algorithm, and 18.8 ± 11.0 with the NS algorithm.

The within-limb variabilities of the mean knee and hip rotation offsets, quantified from the standard deviation across marker placement conditions and trials for each limb, were reduced using all three algorithms. The mean knee and hip rotation offset variability were reduced by the NS algorithm ($p=0.036$ and $p=0.013$, respectively), the KAAM algorithm ($p<0.001$), and the PCA algorithm ($p<0.001$), relative to the uncorrected data (Fig. 6). The mean knee and hip rotation offset variability corrected using the KAAM and PCA algorithms were significantly different from those corrected using the NS algorithm ($p<0.001$) but not from each other ($p=0.986$; Fig. 6).

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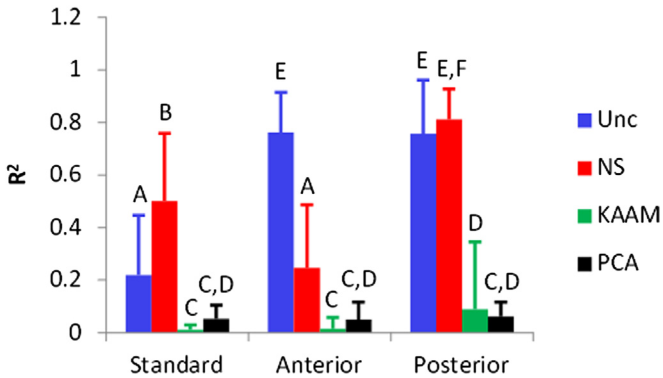

Fig. 3. Mean (SD) coefficient of determination (r^2) between knee flexion and adduction compared between marker placement conditions and correction algorithms. Means with different letters are statistically different ($p<0.05$).

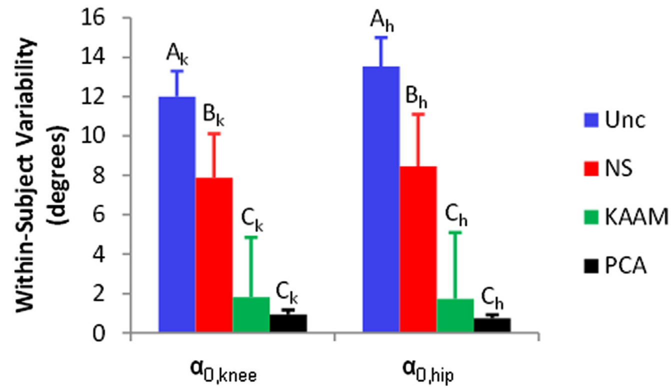

Fig. 6. Mean (SD) within-limb variability of mean knee ($\alpha_{o,knee}$) and hip ($\alpha_{o,hip}$) rotation offsets across marker placement conditions and trials, compared between algorithms. Means with different letters are statistically different ($p<0.05$).

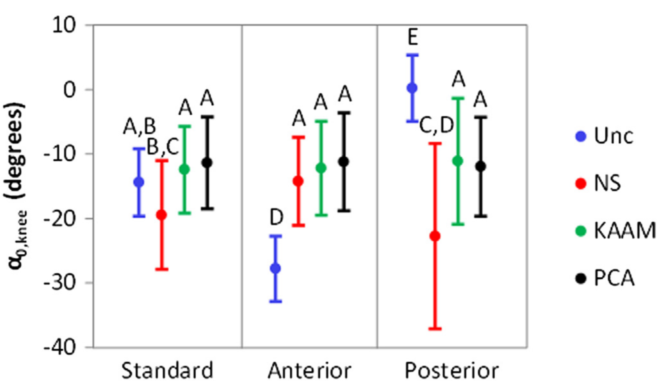

Fig. 4. Mean (SD) knee rotation offset ($\alpha_{o,knee}$) for each marker placement condition and correction algorithm. Means with different letters are statistically different ($p<0.05$).

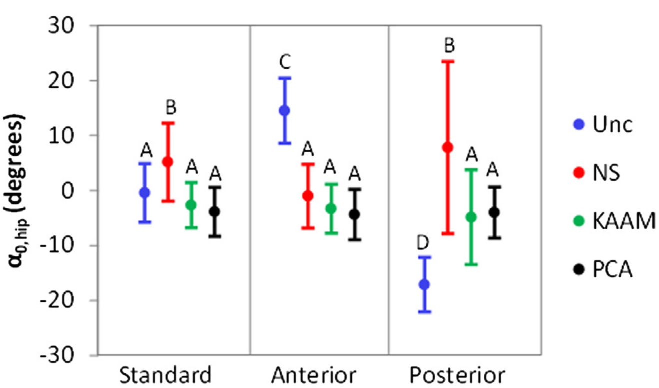

Fig. 5. Mean (SD) hip rotation offset ($\alpha_{o,hip}$) for each marker placement condition and correction algorithm. Means with different letters are statistically different ($p<0.05$).

Visual inspection of the individual limb data aligned with the combined group results. A similar qualitative reduction of errors was observed using the PCA and KAAM algorithms (Fig. 7). The PCA and KAAM algorithm kinematics were visibly consistent with each other and across the three marker placement conditions, with the exception of a shifted mean flexion–extension offset between marker placement conditions (Fig. 7).

5. Discussion

This study demonstrated that a PCA-based algorithm can identify the correct knee flexion axis following standard gait collection. The importance of this is the elimination of additional functional trials, which compromise the precious and limited attention span of many patients. We hypothesized that the algorithm would correct downstream kinematic errors associated with opposite extremes of knee flexion axis misalignment, decrease within-limb variability of the knee and hip rotation angles for three marker placement conditions, and demonstrate superiority as a post-hoc correction method over established algorithms. The PCA algorithm was found to reduce kinematic errors as demonstrated by reduced crosstalk between knee flexion and adduction and consistent mean knee and hip rotation offset across marker placement conditions. The algorithm also decreased within-limb variability of the knee and hip rotation angles as evidenced by a reduced standard deviation in the mean rotation offsets across trials and marker placement conditions. These results supported the study hypotheses and manifested improvement in both certainty and precision of the knee flexion axis using the PCA correction algorithm. Finally, the PCA algorithm was found to be superior when compared to the NS method applied post-hoc (Schwartz and Rozumalski, 2005) but did not demonstrate improvement over the KAAM method (Baker et al., 1999), contrary to the hypothesis. It should be noted that the NS algorithm would likely perform better using a functional knee flexion–extension trial, according to its original design (Schwartz and Rozumalski, 2005). However, this analysis was not in the scope of the current study.

It is well established that crosstalk between knee flexion and adduction, quantified as the correlation between the two kinematic variables, is an indicator of a misidentified knee flexion axis and downstream kinematic errors (Baker et al., 1999; Kadaba et al., 1990; Piazza and Cavanagh, 2000; Schache et al., 2006). Both the PCA and KAAM algorithms significantly reduced the crosstalk for all three marker placement conditions. The crosstalk results were not significantly different between these two methods. However, the PCA algorithm corrected data were more consistent across marker placement conditions; the KAAM algorithm demonstrated a weaker ability to reduce crosstalk for the posterior marker placement condition compared to the anterior or standard placement conditions. These data indicate that these methods are equally effective at minimizing crosstalk in a healthy young adult population. In comparison to the PCA and KAAM algorithms, the NS correction algorithm performed poorly in reducing crosstalk under all marker placement conditions. In fact, crosstalk of the corrected data was increased under the standard (mean $R^2=0.50$) and posterior (mean $R^2=0.81$) marker placement conditions, indicating that this algorithm may not be effective for post-hoc correction applications.

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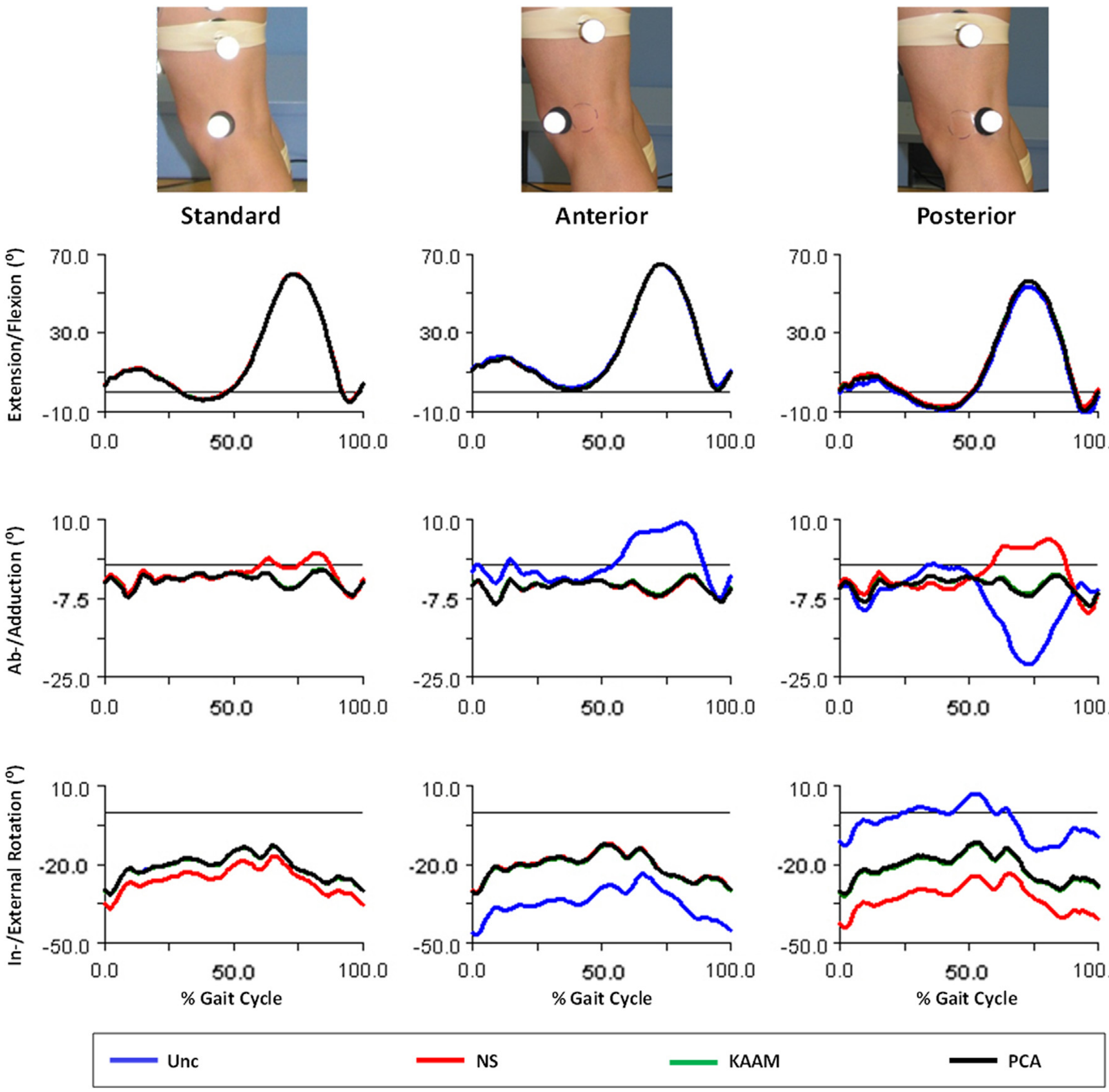

Fig. 7. Representative sample of one subject limb's knee kinematics averaged over four gait cycles for each marker placement (columns) and plane of motion (rows). PCA algorithm kinematics overlay the KAAM algorithm kinematics in each case. PCA algorithm kinematics also overlay the NS algorithm kinematics for the case of anterior marker placement.

The mean knee rotation offset converged toward approximately 12° of external rotation using both the PCA and KAAM correction algorithms, consistent with previous studies. A CT study of both limbs of 80 healthy subjects reported a mean external tibial torsion of 15° (Sestan et al., 2008). No significant differences in the mean hip and knee rotation offsets were found between the uncorrected placement, the PCA correction, and the KAAM correction for the standard marker placement conditions. This is somewhat surprising, given that crosstalk was reduced for the standard marker placement condition. This finding may be partially attributable to the fact that marker placement errors can occur randomly in either the anterior or the posterior direction and the resulting mean rotation offset errors may, therefore, be averaged out across individuals and go undetected by this measure.

An important component of the current study that was first implemented by Fullenkamp (2005), was the use of multiple marker placement conditions. These marker placement conditions represented opposite extremes of misplacement, which allowed for comparison of kinematic variability reduction between correction algorithms. All three algorithms succeeded in significantly decreasing the within-limb variability of $\alpha_{o,knee}$ and $\alpha_{o,hip}$, but the NS method was less effective than either the KAAM or PCA methods. This points to improvements in flexion axis precision and has direct implications on clinical evaluation of tibial and femoral torsion (DeLuca et al., 1997). Although the corrected mean hip and knee rotation offset variability were not statistically different using the KAAM and PCA algorithms, the standard deviation of the variability differed visibly between the two groups. It appears that, for some subjects, the PCA correction algorithm may be better at achieving precision. Further study is needed to identify the cases for which this may be true.

The current results of the KAAM method for the standard placement condition (r^2 reduced to 0.01 ± 0.01) were comparable to those demonstrated in the literature for standard marker placement of 20 adult subjects during gait (r^2 reduced to 0.02 ± 0.03) (Schache et al., 2006). This provides credibility to the study setup, data processing, and results. It should also be noted both for the current results as well as the previously published results that the "standard" marker – even when placed by an experienced physical therapist – frequently results in erroneous flexion axis definition. In the current study, an r^2 of 0.22 ± 0.23 was found for standard marker placement on young (less than 34) and healthy (BMI less than 26) subjects. In a previous study, an r^2 of 0.36 ± 0.60 was found for standard marker placement on young (mean of 22), healthy (mean BMI of 22.2) subjects (Schache et al., 2006). These results reinforce the importance of using a reliable correction algorithm.

A limitation of the current study is that although the KJC location was substantially affected by the marker misplacement (as evidenced by the shifted mean flexion–extension offset), two of the three tested algorithms were not designed to correct the KJC. In a normal clinical setting, this degree of marker misplacement is not anticipated; nor is marker misplacement expected to be isolated to the lateral knee marker. However, the unreliability of the mean flexion extension offset should be considered when applying these algorithms and care should be taken in interpreting knee hyperextension or flexion contracture. Further improvements would include better identification of the anterior-posterior position of the KJC.

Subsequent work is needed to compare the effectiveness of the current algorithms under various challenging conditions of pathologic knees. We expect to find greater algorithm differentiation under pathologic conditions than was observed in the healthy population, particularly when comparing the PCA and KAAM algorithms. One example is patients with frontal plane knee laxity, which may be due to conditions such as collagen disorders (e.g. Ehlers–Danlos syndrome) or impact-induced ligament injury. Frontal plane knee laxity is a self-reported weak area of the KAAM method (Baker et al., 1999), but we hypothesize that the proposed PCA algorithm will be robust in these patients as long as an adequate ratio of flexion–extension to abduction–adduction excursion is retained. A similarly challenging application is patients with restricted flexion–extension excursion, such as individuals with arthritis (Brinkmann and Perry, 1985) or toe-walkers (Davids et al., 1999). The algorithms should also be tested in overweight patients with redundant tissue, where marker placement over bony landmarks is more challenging (Besier et al., 2003) and increased skin motion artifact is expected (Shultz et al., 2009). Comparing the correction algorithms in these kinds of challenging patient populations will provide greater insight into the strengths and weaknesses of each algorithm.

Another PCA-based flexion axis correction method has demonstrated a significant crosstalk reduction in a population of older adults (Baudet et al., 2014). The Baudet algorithm is distinctive from the current PCA algorithm as the input is the three Euler angles as opposed to transformed data projected into a 2D plane. Statistically significant crosstalk reduction was demonstrated, but it is difficult to ascertain the magnitude of the change given that the final values were rounded. Furthermore, no direct comparison was made to other algorithms in the literature. Further analysis of this method is required to determine its effectiveness relative to the currently proposed PCA algorithm.

We have provided evidence for increased certainty and precision of the knee kinematics with the PCA correction algorithm over uncorrected kinematics, which indicates improvement in validity and reliability of the data. By demonstrating a statistically significant and meaningful reduction in downstream kinematic errors, we have shown improvement in the *certainty* of the knee flexion axis. Further study is needed to quantify the true accuracy of the algorithm by comparing the corrected gait analysis-based kinematics with a gold standard of fluoroscopy-based kinematics (Li et al., 2008). We have also demonstrated that the *precision* of the flexion axis is likely improved by the algorithm, as the within-limb variability of the knee rotation angle was decreased. This decrease in the observed variance between tests also demonstrates increased test-retest *reliability* (Streiner and Norman, 2006). All of these improvements combined provide evidence for the *validity* of the PCA algorithm measurements for surgical assessments such as de-rotational osteotomy. With this correction, much greater weight can be placed on the measurements for surgical interventions and outcome evaluations in the future.

Conflict of interest statement

The authors have no conflict of interest to report.

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