Proximal placement of lateral thigh skin markers reduces soft tissue artefact in Plug-in-Gait knee axis estimates during normal gait

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Abstract

A primary source of measurement error in gait analysis is soft tissue artefact. Hip and knee angle measurements, used regularly to guide clinical decisions, are particularly affected due to pervasive soft tissue on the femur. However, despite several studies of thigh marker artefact it remains unclear how lateral thigh marker height affects results using the popular Plug-in Gait model. We compared Plug-in Gait hip and knee joint angles for ten healthy subjects estimated using a proximal- and distal-third thigh marker placement and found significant differences. Relative to the distal marker, the proximal marker produced 37% less varus-valgus range and 50% less hip rotation range, suggesting that it produced less soft-tissue artefact in knee axis estimates. Knee flexion was also significantly affected due to knee centre displacement. Based on an analysis of the Plug-in Gait knee axis definition and two different numerical optimization of the thigh rotation offset parameter, we show that the proximal marker reduced sensitivity to soft-tissue artefact by decreasing collinearity between the points defining the femoral frontal plane and reducing anteroposterior movement between the knee and thigh markers. This study demonstrates that Plug-in Gait thigh marker height can have a considerable influence on outcomes used for clinical decision-making.

Keywords: gait analysis, biomechanical modelling, motion capture

Word count: 3 040

Introduction

Measurements of hip and knee joint angles are used regularly in gait analysis to make clinical decisions. However, since these measurements are conducted using surface-mounted markers, movement of soft tissue relative to the underlying bone presents a considerable challenge to the validity of these key outcomes[1,2]. The femur, which is common to both joints, is particularly prone to soft-tissue artefact as it is enveloped by muscles of considerable bulk along most of its length[3,4]. Therefore, researchers are exploring ways of reducing soft-tissue artefacts when tracking the femur to ensure measurement accuracy.

The anatomical frame of the femur is typically defined using the hip joint centre, the knee joint centre and the knee flexion-extension axis[5]. Incorrect hip and knee centre estimates result in misalignment of the primary longitudinal axis of the femur, which propagates to the sagittal and frontal angles of the hip and knee. The secondary knee axis can only be misaligned in the transverse plane, resulting in offsets to hip and knee rotation[6,7], although this also leads to cross-talk between frontal and sagittal plane motions of the knee[8]. Therefore, efforts to minimize errors in hip and knee angles are either aimed at directly reducing soft tissue artefact in measured marker motion, or at reducing its propagation within the biomechanical model used to estimate the knee axis and joint centres.

Despite developments in functional modelling techniques for tracking joint centres and axes[9-11], improvements to traditional models such as Plug-in-Gait[12] are still desirable as they remain widely used. Plug-in Gait tracks the femoral frontal plane using a hip centre estimated relative to pelvic markers[13], a knee marker on the lateral femoral epicondyle and a lateral thigh marker. The knee centre is then estimated to lie on the knee axis in the estimated frontal plane, half a knee width from the knee marker, such that the resultant knee axis and longitudinal axis are perpendicular. Therefore, incorrect anteroposterior positioning of the thigh marker results in both knee axis misalignment and knee centre displacement[14].

Misalignment of the frontal plane due to thigh marker misplacement is corrected in Plug-in Gait using a thigh rotation offset parameter. This represents the rotation of the measured thigh marker required to position it in the true frontal plane. The offset can be estimated using a mechanical knee alignment device or a numerical optimization that minimizes knee varus valgus motion[15]. While the optimization approach has been shown to improve test-retest reliability compared to knee alignment devices[6], thigh rotation offsets cannot compensate for dynamic artefacts regardless of estimation method. By extension, numerical methods are susceptible to error due to thigh and knee marker artefacts during optimization movements. Therefore, numerical optimization over different a limited phase of the gait cycle may produce better results than using the whole gait cycle and comparisons could be used to detect where soft tissue artefact is occurring. This has not been adequately explored.

Even though from a modelling perspective the height of the thigh marker on the segment does not affect Plug-in Gait outcomes, thigh marker artefact may vary with proximodistal positioning of the thigh marker. Studies have found that proximodistal placement affects thigh marker movement relative to the femur during gait, although these did not assess the propagation of thigh marker artefact to hip and knee angles[3,16]. This is important to know because Plug-in Gait knee axis misalignment results from relative anteroposterior movement between the thigh and knee markers and not from individual marker artefacts. The height of the thigh marker may also affect marker artefact propagation in Plug-in Gait by influencing the collinearity between the hip centre, thigh marker and knee marker. Less collinearity results in less joint angle artefact for a given amount of thigh marker artefact. Although this principle also underlies the use of thigh wand markers, the potential benefits of wands may be negated by additional motion of the wand base[17]. However, the relationship between collinearity and thigh skin marker height has not been explored in the literature.

The purpose of this study was to compare the effect of placing Plug-in Gait lateral thigh skin markers at two different heights on the segment (proximal-third and distal-third). Our primary question was (Q1) in comparison to a distal-third marker, does the use of a proximal-third thigh marker result in differences in hip rotation and knee flexion angles? Furthermore, if so, we asked which of the two thigh markers demonstrates less (Q2) soft-tissue artefact in knee varus valgus angles (Q3) collinearity between the hip centre, knee marker and thigh marker and (Q4) sensitivity to phase of the gait cycle used for numerical optimization of thigh rotation offsets.

Methods

Ten healthy, conveniently selected subjects (7 male and 3 female) participated in the study (age: 36.7 (SD 10.2) years, height: 1.71 (SD 0.1) m, weight: 73.1 (SD 20.4) kg, BMI: 24.6 (SD 4.5) kg.m^{-2}). Ethics support was obtained from the institution's Ethics Committee and all subjects gave informed consent for data collection in writing.

Kinematic data of subject walking was recorded at 200 Hz for all subjects using a Vicon MX system (Vicon, Oxford Metrics Group, Oxford). Testing was performed using Vicon Nexus software (version 1.8.5) and the Plug-in-Gait model. Data was collected for 10 barefoot strides per subject (5 on each side) during self-selected walking speed (1.4 ± 0.14 m.s^{-1}). Marker placement for the Plug-in-Gait lower-limb marker set was performed by a trained gait analyst. Skin mounted markers (not wands) were used. Markers were placed on the distal-third of the thigh segment approximately 70% of the distance from the greater trochanter to the lateral epicondyle, as described in the Plug-in Gait manual (Figure 1a). A second thigh marker was also placed on the proximal-third of the thigh segment approximately 30% of the distance from the greater trochanter to the lateral epicondyle.

Marker trajectories were smoothed using the Vicon Woltring filter routine (MSE = 15mm) and gait events were extracted from the foot marker kinematics. Thereafter we created two copies of the dataset, one with the proximal thigh marker labelled and the other with the distal thigh marker labelled (Figure 1b). Joint angles were then calculated twice for each thigh marker using two different thigh rotation offset values (details to follow). For each of the four datasets, we calculated unique shank rotation offset and tibial torsion values for the Plug-in Gait model using ankle markers attached to the medial malleoli during a static trial.

To answer our primary research question (Q1), we compared differences in hip and knee joint angles for the proximal and distal thigh marker data sets using Baker's standard thigh rotation offset optimization over the whole gait cycle. Specifically, we analysed differences in joint angle range, mean, maximum and minimum values over the gait cycle as these are commonly assessed in gait analysis. We answered our second question (Q2) by quantifying soft-tissue artefact using varus-valgus range, variance and correlation with knee flexion (square of Pearson correlation coefficient). This approach is based on the assumption that a healthy knee operates like a hinge joint during normal walking and thus experiences negligible true varus-valgus motion. We assessed the collinearity of the two thigh markers for our third question (Q3) by calculating the perpendicular distance of the thigh markers relative to the line joining the hip centre and the knee marker. This was done in quiet standing during the static calibration trial.

Finally, in addressing the fourth research question (Q4) we compared the change in hip and knee joint angles for each thigh marker as assessed for Q1 to those obtained when optimizing the thigh rotation offset over the mid-stance phase of the gait cycle. The mid-stance optimization phase was defined as the time from maximum stance phase knee flexion until minimum stance knee flexion. The rationale for choosing the mid-stance phase is that when Baker's method is used to optimize over the whole gait cycle then the thigh rotation offset is typically optimal for mid-swing (to reduce cross-talk error near peak knee flexion). Therefore, under the assumption that knee flexion is a primary driver of marker artefact, we chose the phase of the gait cycle near minimum knee flexion while still allowing for sufficient flexion range of motion to detect cross-talk.

We calculated group mean and standard deviations of all outcomes chosen for Q1, Q2, Q3 and Q4 and performed significance testing using students T-tests. All P-values were calculated for two-tailed distributions with paired measurements for each subject’s leg (P-values of 0.05 were taken as significant). Therefore, our effective sample size was twenty (10 left and 10 right legs). For visual inspection purposes, we plotted mean knee flexion, knee varus-valgus and hip rotation curves for each of the four data sets (Figure 1b) over the gait cycle – time normalised to 51 points. Group variability for each joint angle was assessed using one standard deviation above and below the mean curve at each point in the gait cycle.

Results

Our primary finding (Q1) was that the two different thigh marker placements had a marked effect on hip rotation and knee flexion results when using the standard whole gait cycle optimization (Figure 3a). Significant differences were observed for all hip rotation, knee flexion and knee varus-valgus outcomes except minimum knee flexion (Table 1). Relative to the proximal marker, distal marker hip rotation exhibited a nearly consistent external bias during the stance phase and a notably larger range of motion during the swing phase (Figure 2a). This resulted in a reduction of 17° in both hip rotation range and mean external angle for the proximal marker (Table 1). Knee flexion was increased throughout the gait cycle for the distal marker, especially in the stance phase where minimum flexion was 6° larger, although knee flexion range was reduced by 4° (Figure 2a).

We also found that the knee varus-valgus results for the proximal thigh marker demonstrated significantly less soft tissue artefact regardless of optimization strategy used (Q2). This can be observed qualitatively by the relative flatness of the varus-valgus traces using the two thigh markers (Figure 2a+d). Varus-valgus range, variance and cross were reduced by 37%, 54% and 31% respectively using the proximal marker and a whole gait cycle optimization, although the effect on cross-talk was not significant (Table 1).

In relation to Q3, we found that there was significantly less collinearity between the proximal marker and the hip centre and knee marker. The perpendicular distance of the proximal marker from the line joining the hip centre and the knee marker (80 ± 9 mm) was significantly larger than that found for the distal marker (37 ± 8 mm).

In answer to our last question (Q4), we found that the proximal marker showed noticeably less sensitivity to the two optimization strategies used than the distal marker. The difference in thigh rotation offset values was 1.1°, which was insignificant and effected negligible change in proximal marker hip and knee joint angles (Figure 2b). All differences in hip rotation and knee flexion outcomes were smaller than 2° for the proximal marker, and none were significant (Table 1). There was a greater significant difference between thigh rotation offsets for the distal thigh marker (8.9°, $p < .001$) which resulted in appreciable changes in hip and knee angles (Figure 2c). While there was almost no effect on the range of hip rotation and knee flexion using the mid-stance optimization, hip rotations and knee varus-valgus for the distal marker were more neutral in the stance phase and knee flexion was reduced throughout the gait cycle (Table 1). When compared to the relatively unchanged proximal marker results, this can be clearly seen in that the offsets differences demonstrated for the whole gait cycle optimization (Figure 2a) were eliminated from the stance phase using the mid-stance optimization (Figure 2d).

Discussion

We compared the effect of placing the lateral thigh marker at different heights (distal- and proximal-third) on Plug-in-Gait hip and knee kinematics during walking. We found that the use of these two thigh markers results in appreciable differences in joint angle results (Q1). Relative to the distal marker, the proximal marker significantly reduces soft-tissue artefact in varus-valgus angles (Q2), collinearity of the points defining the femoral frontal plane (Q3) and sensitivity to different thigh rotation offset optimization strategies (Q4). This suggests that a proximal-third thigh marker gives better estimates of hip rotation during walking. The varus-valgus results obtained with the mid-stance optimization reveal that proximal and distal marker artefacts are very similar during early and mid-stance but significantly larger for the distal marker during late-stance and swing. This not only manifests in a large hip rotation artefact during swing, but also notable stance phase bias errors in the distal marker results when optimizing over the whole gait cycle. These observations suggest that the choice of thigh marker height and optimization strategy are important inter-related factors that can have a considerable influence on outcomes and normal reference datasets used for clinical decision-making in gait analysis laboratories.

The findings of this study are directly opposed to reports that proximal thigh marker placement leads to underestimation of hip rotation range[17-19]. However, these studies measured a wide range of hip rotation with fixed knee flexion in exercises specifically designed to achieve this whereas our study tested walking where the opposite conditions apply (wide range of knee flexion and minimal hip rotation). Our study suggests that a distal thigh marker leads to over-estimation of hip rotation range during walking, which was also found by Schache et al. in a study of soft-tissue artefacts during gait[3]. This reinforces the review of Leardini et al.[2] which emphasized that soft-tissue artefact is task dependent and highlights the dangers of extrapolating from results conducted on other movements to recommendations for gait analysis. Our hip rotation results for the proximal marker are very similar to recently published reference data from two internationally regarded gait analysis laboratories – both of which use mechanical knee alignment devices[20]. This suggests that whole gait cycle numerical optimization produces comparable results when using a proximal-third skin marker but not when using a distal one. Therefore, where numerical optimization over the whole gait cycle is preferred for estimating the thigh rotation offset, consideration should be given to rotational artefacts and it may be preferable to use a proximal thigh marker. Alternatively, if significant soft-tissue artefact is observed using a chosen thigh marker after applying whole gait cycle optimization, the mid-stance optimization may improve analysis of the stance phase. Moreover, when collecting normative datasets – of which the standard deviations are used to assess clinical cases - careful consideration should be given to the choice of optimization strategy that will be used as this appears to appreciably influence group variability (Figure 2b-c). It should be noted, however, that the large swing phase artefacts observed for the distal marker cannot be corrected using a knee alignment device.

All the observed differences in hip and knee angles for the two thigh markers can be attributed the effect of marker artefact, thigh rotation offset and collinearity to Plug-in-Gait estimates of the knee axis and knee centre (Figure 3). The proximal marker produced low knee varus-valgus range throughout the gait cycle and very similar results for both optimizations (Figure 2b), suggesting that relative anteroposterior displacement of knee marker and proximal thigh marker was either masked by the larger perpendicular distance (Figure 3a) or negligible (Figure 3b). In contrast, the marked difference in distal marker results for the two optimizations suggests that there was increased displacement of the distal marker relative to the knee marker between stance and swing. This is reflected in the large artefact observed in distal marker hip rotation during swing, which appears to correlate with knee flexion. It is known from fluoroscopy studies that the knee marker moves posteriorly in relation to the femoral epicondyle as the knee flexes during walking[16,21]. Root-mean-square (RMS) values of this movement were estimated to be 10mm by Akbarshahi et al.[16] and 7mm by Tsai et al.[21] (note that range of motion is approximately four times the RMS value). Distal- and mid-third lateral thigh markers are reported to move less. If this is true, mid-stance optimization would cause an internal rotation of the knee axis in swing (Figure 3c). This would lead to increased internal hip rotation in swing, as well as increased knee valgus and decreased knee flexion due to cross-talk – all of which was observed for the distal marker (Figure 2d). In contrast, optimization over the whole gait cycle would minimize cross-talk near peak knee flexion (Figure 3d), over-estimating external hip rotation during stance and increasing knee varus due to cross-talk. Again, this was observed for the distal marker although anterior displacement of the knee centre (relative to the knee centre position for a mid-stance optimized) masked the cross-talk effect, increasing (instead of decreasing) knee flexion during stance (Figure 2a).

This study was limited to a relatively small group of subjects within a low and relative narrow range of body mass index. Furthermore, since knee marker soft-tissue artefact is correlated to knee flexion, cases where knee flexion range is reduced (due to injury or pathology) or increased (as in running gait) will produce very different knee marker soft-tissue artefact to that of healthy walking. These findings are therefore not necessarily applicable to other movements, gait populations or group anthropometrics. The results are also only relevant to the standard Plug-in-Gait protocol where knee centre estimation is performed using the thigh marker and where the knee marker is measured and not reconstructed virtually using a technical cluster on the thigh. It is also worth noting that the knee centre will still be displaced whichever thigh marker is used - due to knee marker displacement - leading to soft-tissue artefact in knee flexion which cannot be investigated further from the data collected for this study. It may be that models that are less dependent on the knee marker are required to improve accuracy in measuring the position of the knee joint. It should also be noted that this analysis is based on using skin markers. The use of proximal wand markers may decrease collinearity and reduce sensitivity to soft-tissue artefact still further. However, the varus-valgus range was already consistently low in this study using the skin marker, and any additional beneficial effect would have to be balanced against the potential for increased movement of the wand marker in relation to the bone.

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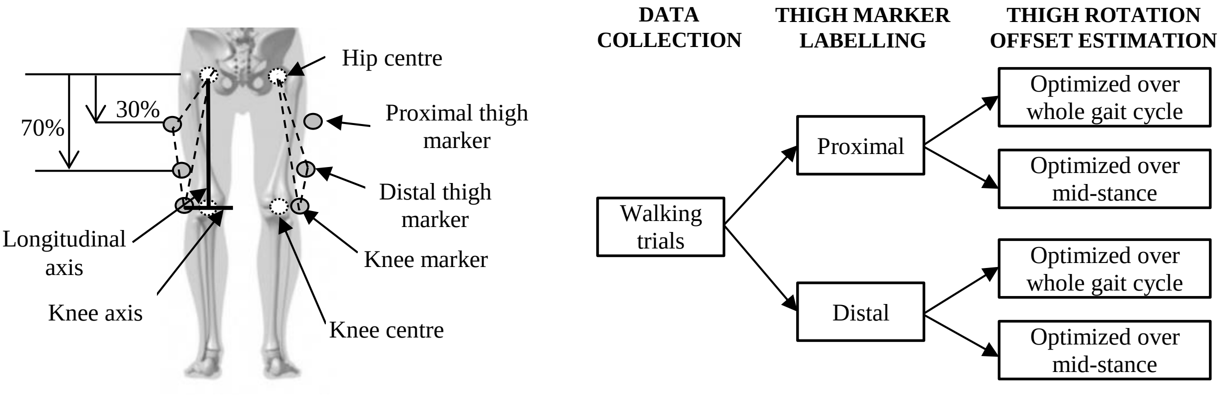
Captions



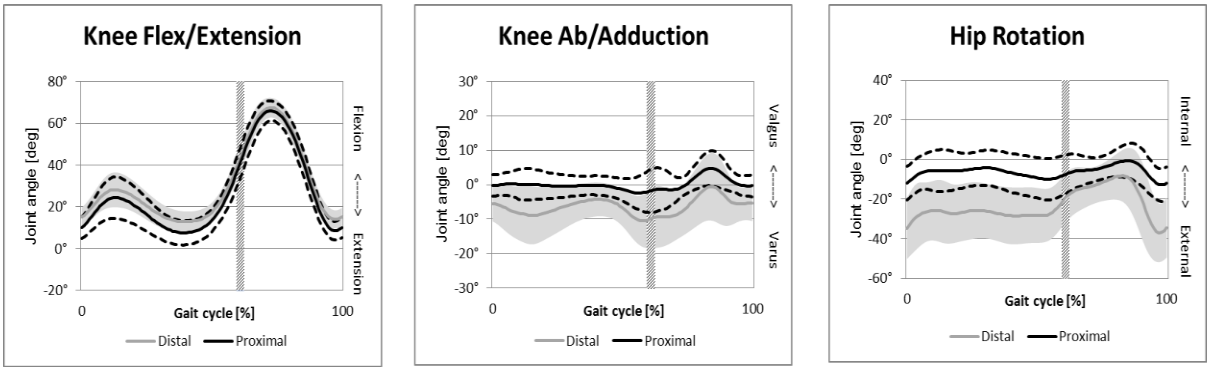
(a) (b)

Figure 1: Proximal and distal thigh marker (a) placement and (b) processing. Dashed lines in (a) illustrate the triangle of markers used to define the frontal plane of the femur in each case, solid lines show the joint axes.

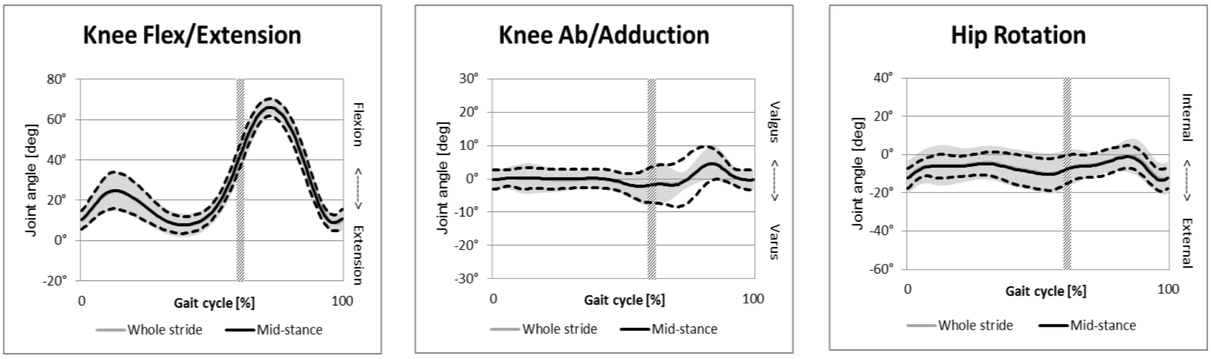
Table 1: Comparison of knee angle outcomes markers for both thigh rotation offset optimizations using of the proximal and distal thigh.

Outcome	**Whole gait cycle optimization**		**Mid-stance optimization**	
	Distal marker	**Proximal marker**	**Distal thigh**	**Proximal marker**
Hip rotation				
range (deg)	**34 ± 7**	**17 ± 4**	**35 ± 7**	**17 ± 4**
max (deg)	**-5 ± 13***	**2 ± 9**	**15 ± 6***	**1 ± 5**
mean (deg)	**-23 ± 13***	**-6 ± 8**	**-2 ± 7***	**-7 ± 5**
min (deg)	**-39± 14***	**-15 ± 8**	**-19 ± 8***	**-16 ± 6**
Knee flexion				
range (deg)	**56 ± 4**	**60 ± 4**	**56 ± 4**	**60 ± 3**
max (deg)	**68 ± 5***	**66 ± 5**	**63 ± 5***	**66 ± 4**
mean (deg)	**31 ± 5***	**27 ± 6**	**26 ± 5***	**27 ± 5**
min (deg)	**12 ± 5***	**6 ± 6**	7 ± 4*	6 ± 4
Varus-valgus				
range (deg)	**13± 4***	**10 ± 3****	**19 ± 6***	**12 ± 3****
variance (deg^2)	**14 ± 10***	**7 ± 5****	**39± 26***	**10 ± 5****
correlation to knee flexion (r^2)	0.13 ± 0.14*	0.09 ± 0.07**	0.61 ± 0.28*	0.43 ± 0.25**
mean (deg)	**-6 ± 7***	**0 ± 3**	**3 ± 4***	**0 ± 3**

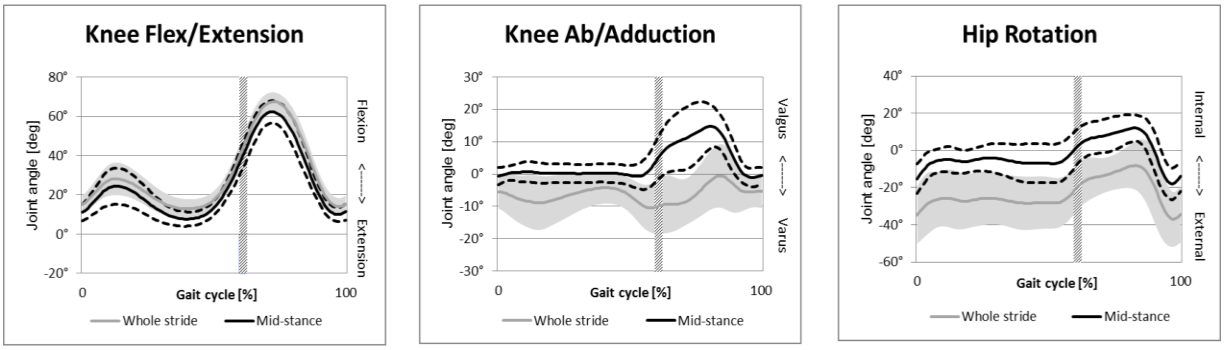
* significant differences between optimizations for the distal marker
** significant differences between optimizations for the proximal marker
bold significant differences between distal and proximal markers for a given optimization



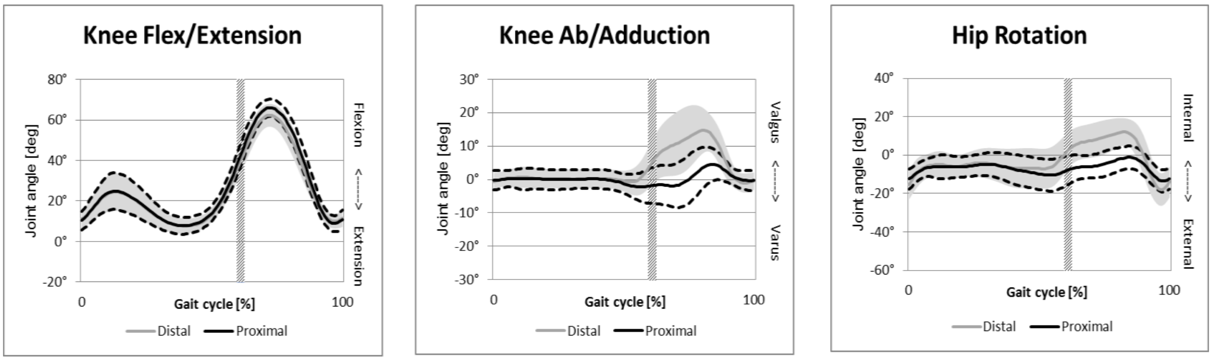
(a) Comparison of distal and proximal thigh marker results when optimizing thigh rotation offsets over the whole gait cycle



(b) Comparison of proximal thigh marker results using the two different thigh rotation offset optimization regions

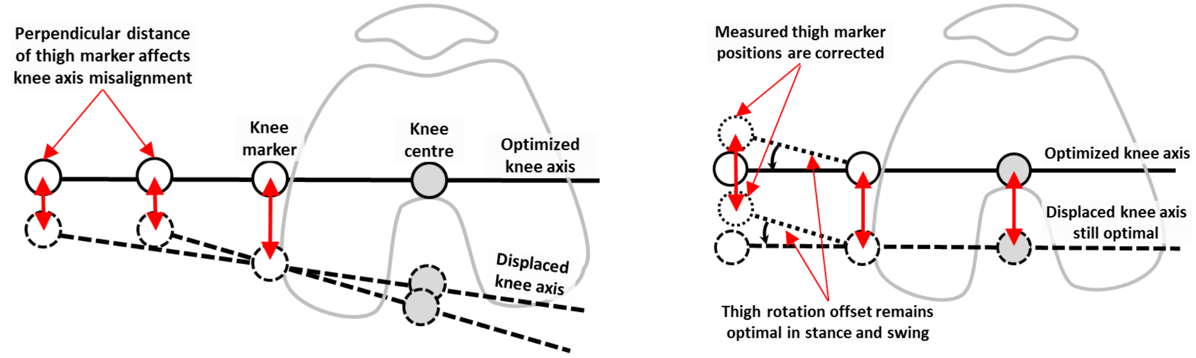


(c) Comparison of distal thigh marker results using the two different thigh rotation offset optimization regions

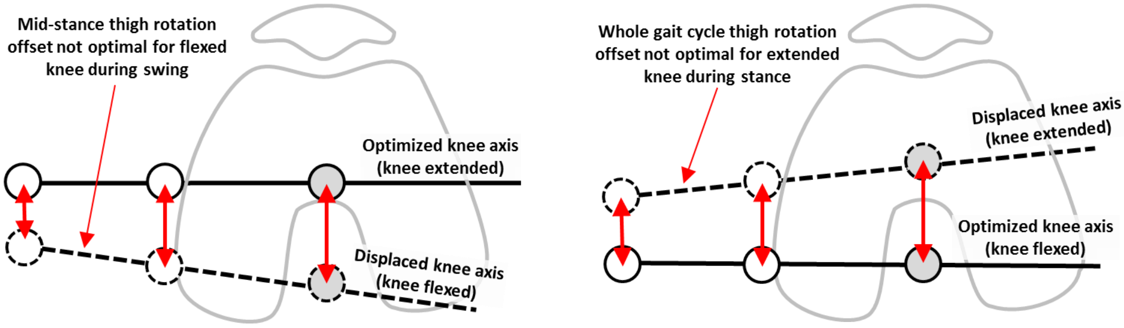


(d) Comparison of distal and proximal thigh marker results when optimizing thigh rotation offsets over mid-stance

Figure 2: Comparison of joint angles produced by the distal and proximal thigh markers when optimized over (a) the whole gait cycle and (d) mid-stance. The effect of the different optimizations on the (b) proximal and (c) distal markers is also shown. Note that differences in (b) and (c) are only due to thigh rotation offset values, whereas comparisons between markers are also affected by differences in marker artefact and collinearity.



(a) Effect of collinearity on unequal displacement (b) Effect of equal displacement on thigh rotation offset



(c) Mid-stance optimization with unequal displacement (d) Whole cycle optimization with unequal displacement

Figure 3: Transverse plane view of how thigh and knee marker artefact affects knee axis and knee centre definitions relative to the femur. As shown in (a), unequal marker displacement from the configuration optimized by the thigh rotation offset (solid circles and lines) results in both knee centre displacement and knee axis misalignment (dashed circles and lines) which the thigh rotation offset cannot correct. This knee axis misalignment is directly proportional to the difference in anteroposterior displacement and inversely proportional to the perpendicular distance of the thigh marker. If the displacement is equal, as in (b), there is still knee centre displacement but no knee axis misalignment. Measured thigh marker positions (dotted circles) are rotated correctly into the frontal plane relative to the knee marker throughout the gait cycle. However, as shown in (c), a mid-stance optimization would cause misalignment during the swing phase if marker displacements are unequal – whereas (d) shows how whole gait cycle optimization leads to reversed misalignment during stance for the same marker artefact.