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Review

Human movement analysis using stereophotogrammetry Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics

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Abstract

Estimating the effects of different sources of error on joint kinematics is crucial for assessing the reliability of human movement analysis. The goal of the present paper is to review the different approaches dealing with joint kinematics sensitivity to rotation axes and the precision of anatomical landmark determination. Consistent with the previous papers in this series, the review is limited to studies performed with video-based stereophotogrammetric systems. Initially, studies dealing with estimates of precision in determining the location of both palpable and internal anatomical landmarks are reviewed. Next, the effects of anatomical landmark position uncertainty on anatomical frames are shown. Then, methods reported in the literature for estimating error propagation from anatomical axes location to joint kinematics are described. Interestingly, studies carried out using different approaches reported a common conclusion: when joint rotations occur mainly in a single plane, minor rotations out of this plane are strongly affected by errors introduced at the anatomical landmark identification level and are prone to misinterpretation. Finally, attempts at reducing joint kinematics errors due to anatomical landmark position uncertainty are reported. Given the relevance of this source of errors in the determination of joint kinematics, it is the authors' opinion that further efforts should be made in improving the reliability of the joint axes determination.

Keywords: Stereophotogrammetry; Human movement; Anatomical landmark; Joint kinematics

1. Introduction

Human movement analysis using stereophotogrammetric measurements and rigid body modelling requires the definition of systems of axes associated with each bony segment incorporated in the model. The systems of axes defined from body-surface marker positions are referred to here as marker cluster technical frames (CTF), while those defined from anatomical landmark (AL) positions are referred to as anatomical frames (AF). The differences of the features of the CTFs and the AFs have been described in the first paper of this series [1]. A major issue in human movement analysis is the identification of ALs and the reconstruction of their position in a selected set of axes, namely the AL calibration [1–3]. ALs can be either internal or subcutaneous and in general the determination of their location lacks accuracy and precision. This affects AF position and orientation precision and, consequently, the estimation and interpretation of joint kinematics and kinetics.

This paper reviews the information available in the literature regarding the precision and accuracy of the determination of the location of both internal and palpable ALs, and thus have the relevant AFs, as well as the sensitivity of joint kinematics variables to AF precision and accuracy. Given the relevance of AF axes determination in allowing the correct interpretation of joint kinematics, definitions of AFs aimed at reducing joint kinematics sensitivity to AL uncertainty are also discussed.

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2. Determination of subcutaneous palpable AL locations

The incorrect location of subcutaneous bony ALs through palpation can be caused by three main factors: (1) the palpable ALs are not points but surfaces, sometimes large and irregular; (2) a soft tissue layer of variable thickness and composition covers the ALs; (3) the identification of the location of the ALs depends on which palpation procedure was used.

White et al. [4] first reported on the repeatability of the determination in vitro of pelvic and lower limb AL positions. Problems associated with precision and accuracy in AL identification were also illustrated by Small et al. [5] with reference to the hand and wrist. In that study a Roentgen-photogrammetric technique was used both in vivo and on cadavers. Their study was the first dealing with this issue in in vivo experiments. Unfortunately, the number of studies dealing with this problem using in vivo stereophotogrammetric measurements is still limited. Recently, Piazza and Cavanagh [6] focused their investigation on the medial (ME) and lateral (LE) femoral epicondyles with the goal of estimating the range of variability of the axis passing through them used to define the knee flexion/extension axis. Ten examiners were used to locate ALs. Although not explicitly reported, from their results an AL position variability of about 10 mm can be inferred. The authors highlighted that AL position uncertainty and consequently the erroneous determination of AF axes may result in the observation of physiological knee motions such as the screw-home mechanism (external rotation of the tibia as knee extends) even when such motion did not occur, leading to erroneous clinical interpretations of the estimation.

Della Croce et al. [7] studied the precision of lower limb ALs position determination, its effects on AF orientation determination, and the effects of errors in AF orientation on joint kinematics. Intra- and inter-examiner AL precision values were determined from subjects with skin marker clusters attached to the pelvis and lower limb segments by physical therapists who had gait laboratory experience. The physical therapists were asked to palpate the ALs listed in Table 1 using detailed directions [8]. In the same table, relevant precision values are reported expressed in an AF obtained from the mean values of AL positions ($\overline{\mathrm{AF}}$). Intra-examiner precision was higher than the inter-examiner precision. This was interpreted as being caused by the different examiner interpretations of the procedure for locating ALs. Greater trochanter (GT) variation was the largest among the femoral ALs (root mean square value up to 18 mm). Tibial ALs were, on average, the most precise. In the intra-examiner observations, the locations of the metatarsal ALs were reasonably repeatable, especially in the sagittal plane.

A recent study carried out by Rabuffetti et al. [9] estimated the variability in identifying the location of ALs distributed over the whole body using the same method proposed in [7]. Rabuffetti et al. estimated the precision of AL identification of a so-called “self-marking” procedure (the subject under examination performed AL identification and calibration on his/her own body) to be used for experimental

Table 1
Intra- and inter-examiner precision of the palpable anatomical landmark position components in the relevant mean anatomical frame obtained by [7] (*x*, *y*, *z*, 3D*) and [9] (3D*)

Bones	Anatomical landmark	AL	Intra-examiner (mm)					Inter-examiner (mm)				
			x	*y*	*z*	3D	3D*	*x*	*y*	*z*	3D	3D*
Pelvis	Left anterior superior iliac spine	LASIS	3.4	4.0	11.0	12.2	5.6	3.5	7.0	12.4	14.7	13.2
	Right anterior superior iliac spine	RASIS	10.0	11.5	14.5	21.0		12.4	15.2	15.0	24.7	
	Left posterior superior iliac spine	LPSIS	2.8	8.3	7.5	11.5	11.3	9.5	10.8	14.6	20.5	14.9
	Right posterior superior iliac spine	RPSIS	5.7	10.7	4.6	13.0		8.6	15.7	17.1	24.8	
Femur	Prominence of greater trochanter external surface	GT	12.2	11.1	7.0	17.9	10.4	12.8	9.8	7.2	17.7	19.2
	Medial epicondyle	ME	5.1	5.0	6.7	9.8		8.2	9.5	8.0	14.9	
	Lateral epicondyle	LE	3.9	4.9	7.8	10.0	10.5	9.5	13.5	9.8	19.2	14.6
	Antero-lateral ridge of patellar surface groove	LP	3.8	3.9	7.8	9.5		8.8	7.2	9.7	14.9	
	Antero-medial ridge of patellar surface groove	MP	5.2	2.4	10.8	12.2		4.2	2.6	17.9	18.6	
	Most distal point of lateral condyle	LC	4.7	3.4	2.9	6.5		7.7	5.0	9.8	13.4	
	Most distal point of medial condyle	MC	4.4	1.4	4.4	6.4		5.3	5.5	11.9	14.1	
Tibia fibula	Prominence of tibial tuberosity	TT	1.2	1.8	4.3	4.8	7.2	1.9	7.2	9.1	11.8	9.2
	Apex of the head of fibula	HF	3.3	3.3	3.3	5.7		6.1	8.4	4.9	11.5	
	Most medial ridge of medial tibial plateau	MMP	3.4	4.4	6.6	8.6		12.1	6.6	14.1	19.7	
	Most lateral ridge of medial tibial plateau	MLP	8.0	2.1	5.6	10.0		7.4	6.7	9.6	13.8	
	Distal apex of medial malleolus	MM	2.2	2.6	6.6	7.4		9.9	6.2	9.9	15.3	
	Distal apex of lateral malleolus	LM	2.6	2.4	5.7	6.7	9.2	9.3	7.1	12.1	16.8	8.6
Foot bones	Upper ridge of calcaneus posterior surface	CA	7.0	4.9	5.7	10.3		9.1	9.2	9.8	16.2	
	Dorsal aspect of first metatarsal head	FM	2.6	3.2	6.9	8.0		9.1	9.7	16.9	21.5	
	Dorsal aspect of second metatarsal head	SM	2.2	6.3	6.0	9.0		8.2	7.1	8.7	13.9	
	Dorsal aspect of fifth metatarsal head	VM	0.7	2.0	6.5	6.8	7.0	3.9	8.0	10.0	13.4	8.5

motion analysis applications in space. They also estimated the intra- and inter-examiner precision of AL identification performed by movement analysis experts, and the accuracy of the “self-marking” procedure. The last quantity was estimated for each AL by computing the distance between the AL position mean values obtained by the experts and by the “self-marking” operators. The accuracy of the “self-marking” procedure provided interesting information, although some methodological concerns can be raised. In some cases, substantial differences in determining AL location between the expert and “self-marking” operators were found. For example, the GT location was found by the two groups of operators to be more than 20 mm apart and all pelvic ALs were determined with more than 10 mm difference. This showed that “self-marking” is likely to be a biased operation. Intra- and inter-operator precision values obtained in this study can be compared to the results of Della Croce et al. [7], as shown in Table 1 which reports the figures for the lower limb ALs for both studies. Inter-operator precision values obtained in the two studies were highly consistent. Intra-operator values obtained in [9] are in some cases more precise, probably because the data refer to “self-marking” operators whose proprioceptive and tactile feedbacks can provide additional guidance in locating ALs. In general, the overall consistency of the figures reported in the two studies strengthens their reliability. The results reported in the table can also be used as guidelines for the choice of the ALs most suitable for AF definition so that the least precise ALs, such as the GT, do not play a key role in the definition of the relevant AF.

3. Determination of internal AL locations: hip and knee joint centres

Those ALs not representing palpable bony prominences are referred to as ‘internal’. Among the lower limb internal ALs, the geometric centres of the femoral head and of the acetabulum are used the most. In human movement analysis, the corresponding articular surface areas are assumed to have spherical shapes and a common centre, therefore the hip is assumed to be a ball-and-socket joint. The accuracy and precision with which the hip joint centre (HJC) location is estimated are crucial for error propagation to the kinematic and kinetic measurements of the hip and knee joints [10–16].

The HJC location can be estimated using either a functional or a prediction approach. The former, originally proposed in 1984 [2], suggests that the HJC is the pivot point of a 3D relative movement between the femur and pelvis. Position and orientation data of these body segments can be estimated using stereophotogrammetry, and several different analytical methods can be adopted to calculate the coordinates of the centre of the mutual rotation in both reference frames. Recent experimental work performed with a stereophotogrammetric system and a ball-and-socket mechanical joint [17] has demonstrated that the collection of an adequate hip range of motion is more important for a reliable estimation of the HJC than the type of motion. It was shown that, in the absence of soft tissue artifacts, the error in determining the pivot point location can reach 5 mm when performing a 30° rotation and 10 mm when performing a 15° rotation. In a later study, the same authors tested the use of the functional method both on limited range ad hoc 3D movements and on gait trials [18]. They showed the estimation of the HJC being satisfactory when obtained with the former and unsatisfactory when extracted from the latter. Other authors have focused on the mathematical approach to determine the HJC location, and two novel algorithms were proposed [19,20]. Recently, it was shown that under certain conditions the two above-mentioned algorithms are identical [21]. The functional method requires acquisition of an additional task in the gait analysis session, and can only be applied to patients who have reasonable hip motion, and it is affected by the soft tissue artifacts. Nevertheless, with these limitations, it remains, at present, the only clinically feasible method potentially able to detect subject-specific location of the HJC, the alternative imaging-based techniques being inconvenient in most clinical settings.

The prediction approach uses regression equations with pelvis and anthropometric measurements as independent variables (Fig. 1). Regression coefficients (Table 2) were obtained by using imaging techniques based on relatively small samples of living adult males [10,22–26] or by direct measurements on a larger sample of cadaver specimens [27]. Those provided by Bell et al. [25] and Davis et al. [26] are currently the most widely used. The most complete set of equations presented in [27] requires the palpation of the pubic symphysis landmark, which may not be acceptable in the routine clinical context. These prediction approaches are based on a very limited and specific population of subjects, and their application to every clinical situation is therefore unsafe. However, similar regression equations are implemented in the major clinical gait software packages, yet little is known about these and about relevant validation work. An original geometrical method has been recently proposed to address movement analysis of seated posture, where only the antero-lateral aspect of the subject can be visible to the cameras [28]. A HJC 3D location is preliminarily obtained using the

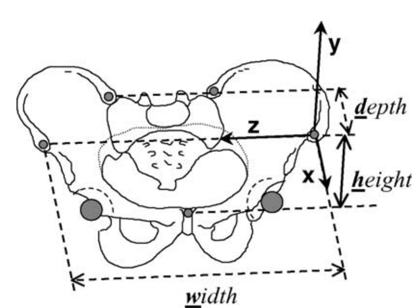

Fig. 1. Anatomical landmarks (ASISs and PSISs as small grey circles, HJC as larger grey circles) and geometrical measurements on which most of the prediction methods for the estimation of the centre of the acetabulum are based. The most widely used methods are summarised in Table 1.

Table 2
Regression equations of the main prediction methods for HJC location estimation expressed in millimetres, based on anthropometric measurements on both male (M) and female (F) adult subjects

Method	Sample	*x*	*y*	*z*
[25]	7 M	$-0.19w$	$-0.30w$	$0.14w$
[26]	25	$-0.95a + 0.031l - 4$	$-0.31a - 0.096l + 13$	$0.5w - 0.055l + 7$
[27]	35 F and 30 M*	$-0.34d$	$-0.79h$	$0.14w$
[31]	11 M	$-0.31d$	$-0.096l$	$0.38w$

In only one case [27] measurements were obtained on cadavers. Relevant anatomical reference frame and significance of *w* (width), *d* (depth), and *h* (height) are depicted in Fig. 1. The following are for estimation of the homolateral, i.e. left, hip. For the method of Davis et al. [26], the following definitions are also necessary: *l* (leg length) = distance between ASIS and homolateral medial malleolus; *a* = antero-posterior component of the distance between a point approximating the hip centre and the ipsilateral ASIS.

prediction approach proposed in [27]. When posterior pelvic markers are lost, only the medio-lateral HJC coordinate remains determined, and the remaining two coordinates are constrained on a circle of known radius. By intersecting such circle with the sphere representing all possible HJC positions in space with radius equal to the known distance between the HJC and the lateral epicondyle and centred in the latter, two solution points are found, one of which can be easily discarded. The predictions in the limited view condition compare well with the preliminary estimation with a full set of pelvic markers. Critical pubic symphysis landmark calibration is still necessary.

An attempt was made to combine kinematic data gathered from gait trials with morphology-based HJC location estimations aimed at enhancing the latter without the need for an ad hoc motion trial [29]. Subject-specific HJC location was estimated using regression equations for an initial ‘guess’ to be used in an optimization function in which (a) distances between the hip and knee joint centres are taken from the subject gait analysis data, (b) anthropometric measurements are taken directly from the subject, and (c) morphological parameters were previously derived from statistical analysis of measurements on standard frontal radiograms of the pelvis. These data are collected and properly weighted, and an iterative optimization algorithm performed to identify the solution. The repeatability of the combined method was reported to be better than that of its isolated parts, although the validation was limited by the lack of knowledge of the true HJC positions.

Several experimental works compared the performance of prediction and functional methods. Initially, the prediction method was claimed to provide more accurate estimations than the functional method [25]. In seven normal male volunteers, two sets of regression equations [22,25] resulted in 36 and 19 mm mean estimation errors, respectively, in locating the HJC, whereas the functional method resulted in a mean error of 38 mm, ranging from 14 to 65 mm. McGibbon et al. [30] compared a novel functional method based on the determination of the axis of rotation of the hip during a sit-to-stand trial of one subject, with prediction methods [22–24,27] by determining ex vivo the HJC location of the same subject. The authors showed that their functional method was remarkably more accurate than the tested prediction methods. In a more recent work using 11 normal male volunteers [31], the functional method limited the mean estimation error to 12 mm, performing better than two popular prediction methods [25,26], which produced mean errors of about 23 and 21 mm, respectively. The better performance of the functional method can be, in part, justified by the use of plastic shells aimed at improving the rigidity between the pelvis and the relevant marker cluster. Standard radiostereometry was used for the first time to measure HJC actual location with respect to a system of reference constructed using the anterior superior iliac spines. From this, a further set of equations was provided, as in Table 2. The estimations based on these two prediction methods were also found to be consistently biased, antero-superiorly in the method proposed in [25] and antero-inferiorly and medially in the method proposed in [26].

The performance of three prediction methods in identifying the HJC location [22,25,27] was also tested in 10 healthy subjects using corresponding measurements on frontal radiograms and assuming the HJC to be at the same antero-posterior location as the greater trochanter [32]. Unfortunately, only the resulting hip joint moment data during gait were reported for the different estimations. It was claimed that the estimation closest to the assumed known position was that of Andriacchi et al. [22], but only when adjusted by the authors, with a mean location error of 7 mm too far medially and 8 mm too far superiorly. A preliminary study [33] assessed the reliability of eight prediction methods by comparing their estimations with actual measurements obtained from CT scans in two children with known pathology. Three-dimensional errors ranged from 9 to 105 mm. Another preliminary study [34] assessed the accuracy of popular regression equations [26] in 5 adults and 15 children, including 5 with cerebral palsy (CP), by comparing the predictions with 3D measurements obtained from magnetic resonance imaging analysis. The maximum discrepancy between predicted and measured HJC locations was 40 mm in a normal child and 85 mm in a child with CP. Mean HJC location errors were found to be significantly higher in children with CP (55 mm) than in normal children (22 mm) and adults (17 mm). This study strongly pointed out the necessity for specific regression parameters to match age, gender, the subjects’ anthropometrics and possibly pathology better.

Two recent preliminary studies on the validation of the functional method [35,36] have reported contrasting results. The former pointed out that repeatability over hip rotation trials can be poor. This was also the case when a number of markers were tracked on the lateral thigh for the HJC estimation to be averaged, suggesting skin motion over the pelvis may be the most important source of error. The latter has reported promising repeatability over trials by using a brace to be fixed to the knee in full extension. A further preliminary work [37] has compared the performance of three traditional methods [2,20,26] in addition to an original combination of two of these [20,26] by using a CT scan measurement of the true location in one male subject. The best estimation was achieved from the latter combined method, but the accuracy obtained (11–14 mm) was found still not acceptable for the potential application the authors were pursuing, i.e. post-operative assessment of total hip replacement position. In a recent study, a different approach was used to test the functional method for the determination of the HJC location [38]. The intra- and inter-examiner repeatability of gait kinematics and kinetics obtained applying a model including the HJC location determined functionally was compared to that obtained by applying a model including regression equations for the HJC determination. It was shown that the functional approach provides gait variables with slightly higher repeatability.

In summary, the functional approach appears to be preferable when a considerable range of hip motion can be performed and which can still be satisfactory when the hip range of motion is limited. More effective algorithms for pivot point calculation can be examined in the future, although only a limited improvement is expected. At present, prediction methods are inadequate to address the wide range of pelvic morphology. The necessity to reexamine the traditional prediction techniques particularly when these are applied to children has been pointed out by several recent studies [33,34]. Finally, the reliability of anthropometric measurements, particularly with relatively inexperienced operators, has also been questioned [39].

The gait analysis community is still seeking a robust and detailed series of regression equations for HJC location. The use of the functional approach in large-scale experimental campaigns has also been suggested [31,40] and preliminarily performed [41]. This would provide kinematics-based estimations of HJC location while preventing large populations of volunteers from being exposed to radiation. The combined use of motor task kinematic data, to be collected in any case, and anthropometric measurements should be investigated to take advantage of both approaches, so that patients would not be asked to perform additional trials. All the current methods, however, are definitely expected to generate substantial errors in determining HJC location, and all the outcomes affected by these errors should therefore be considered very carefully in clinical decision making [16].

Less crucial, and therefore less studied, is the accurate determination of location of the centre of rotation of the knee (KJC). In fact, the KJC is defined when ball-and-socket joint models of the knee joint are used. Such models are only a rough approximation of the anatomy of the joint. However, the ball-and-socket model for the knee joint is commonly used in most commercial movement analysis software products for two main reasons: (1) it is easily implemented in a marker set-up and included in processing routines; (2) kinetic variables appear not to be very sensitive to the erroneous KJC location. It has been shown [42] that errors in the order of 10 mm in locating the KJC do not remarkably change the pattern of the knee moment time-histories. However, the authors noted that when changes in the sign of moments are of interest, such errors can misguide data interpretation. Additional errors in locating the KJC may be introduced if its location is determined during movement by using markers located on the distal thigh. The substantial movement of the skin over the distal femur during knee flexion may move the KJC of several centimeters [43,44]. Besier et al. [38] proposed a technique for the determination of the KJC location based on the identification of the mean helical axis obtained from a knee flexion/extension, reducing the errors introduced by the skin markers located on the distal part of the thigh. They showed that the technique improved the repeatability of kinetic and kinematic variables. Similarly to the method proposed for the determination of the HJC location, Frigo and Rabuffetti [29] determined the KJC combining anthropometric measurements to kinematic constraints imposed to data acquired during the execution of a gait trial. While the results showed a considerable improvement in both normal and pathologic subjects, concerns regarding the dependency of KJC location estimations on the trial kinematic data might be raised.

4. Determination of AF position and orientation

The knowledge of the AL positions in the relevant CTF allows for the definition of AFs and the determination of their orientation. A precise determination of AF orientation is crucial for joint kinematics reliability. Della Croce et al. [7] used the AF definitions proposed by Cappozzo et al. [3], which are in accordance with general standard directions of reference axes [45]. The experimental data reported in [7] provided observations of the AF position vectors and orientation matrices, and therefore the relevant orientation vectors [46], of the lower limb bones with respect to the relevant AF. Their precision figures are reported in Table 3. Pelvic and foot AF orientation errors were distributed roughly equally on the three axes. In contrast, the femoral and tibial AF orientation errors were concentrated mostly about their longitudinal (y) axis. This was associated with the shape of these bones and consequent AL cluster geometry, characterized by an overwhelming contribution of the longitudinal dimension with respect to the other two dimensions. It was pointed out that the values in Table 3 are affected not only by the precision of the relevant ALs, but also by the AF definition rule adopted.

Table 3
Intra- and inter-examiner precision of the indicated anatomical frame orientation components along the antero-posterior (θ_x), longitudinal (θ_y), and medio-lateral (θ_z) directions

Segment	Intra-examiner (°)			Inter-examiner (°)		
	θ_x	θ_y	θ_z	θ_x	θ_y	θ_z
Pelvis and hip	2.3	2.6	3.7	5.2	3.7	4.1
Femur	0.9	4.7	0.9	2.5	5.1	3.0
Tibia and Fibula	1.4	3.5	0.3	4.2	9.4	2.6
Foot	2.7	2.3	1.8	5.9	9.2	5.1

Piazza and Cavanagh [6] defined femur and tibia AFs using common definition rules [3]. They estimated the variability in determining the intercondylar axis orientation only and reported a mean angle of separation of 7.7°.

Other studies did not deal with the precision of the AL location determination and its effects on AF orientation precision, but simply hypothesized a certain error in determining anatomical axes orientation [13,15,47–49] with the goal of observing the consequent variations in joint kinematics or dynamics representation.

5. Joint kinematics sensitivity to erroneous determination of AL location and AF orientation

Given the crucial role of the HJC in locomotion analysis, the effects of the erroneous determination of its location have been investigated more than those of other AL locations [13,16]. Kadaba et al. [13] observed the effects on joint kinematics when the HJC position was made to vary analytically over a 20 mm range in all directions. They observed an offset in joint kinematics curves but not an effect on the relevant patterns throughout the gait cycle. Stagni et al. [16] extended the quantification of the propagation of the erroneous determination of HJC location to hip and knee kinetics, as assessed on a group of five able-bodied subjects during level walking. Hip and knee angle and moment components were estimated for each experimental trial first using a nominal estimate of the HJC position, then adding a set of 3D errors in HJC location taken within the range of the reported data [31]. It was observed that inaccuracies in the HJC coordinate estimates affected gait analysis results substantially. The hip moments showed the largest propagation error, particularly in the flexion/extension component. A HJC misplacement of 30 mm in the anterior–posterior direction generated a mean error on the flexion/extension moment of about 22% of its value. The ab/adduction moment was found to be the second largest affected quantity associated with medio-lateral HJC erroneous location. The effects of erroneous HJC location determination on knee angles and moments were found to be negligible.

Joint angle sensitivity to AF orientation variations has been shown to be high and particularly prejudicial to the reliability of those angles that undergo relatively small variations during movement (minor angles). At least four different approaches to the estimation of the above mentioned sensitivity are found in the literature: (1) an analytical approach (sensitivity analysis) [50]; (2) the analysis of errors determined from the measurement of controlled, and therefore known, joint kinematics of mechanical devices [6]; (3) experimental error data applied to simulated joint kinematics [7]; and (4) error simulation applied to human joint kinematics or dynamics [13,15,47–49].

Woltring [50], while proposing the orientation vector to describe joint kinematics, performed a mathematical sensitivity analysis of both the orientation vector and Cardan angles [51] to noisy angle data. It was shown that Cardan convention was affected by more “cross-talk” among components as the major angle increases. He concluded that the orientation vector was more suitable for describing joint kinematics than any Cardan convention. However, the use of this representation for joint kinematics has not been successful, mainly because its pseudo-vectorial nature without single-axis rotation mechanical equivalents is considered to be difficult to interpret in conventional physiological terms.

Piazza and Cavanagh [6] used two custom devices simulating knee kinematics with one and two degrees of freedom, respectively. The kinematics of the devices were known and imposed by the experimenters. Their test allowed for the estimation of the “cross-talk” among the angular components used to describe knee kinematics. Similar to the “cross-talk” resulting from the imperfect location of multi-axis sensors in a force platform, “cross-talk” may also describe the effect of the incorrect location of the rotation axes used to describe joint kinematics. Interestingly, some “cross-talk” was found even when no noticeable error was introduced in locating the fiducial “anatomical” landmarks of the devices. The authors concluded that joint kinematic representation is extremely sensitive to rotation axis location in space, and recommended a limited use of minor angle data.

Della Croce et al. [7] estimated the propagation of AL position precision to joint kinematics by simulating the joint movement. The effect of AF orientation errors was computed using the AF orientation observations of the proximal and distal segments of each lower limb joint, and aligning proximal and distal AF. A flexion motion for the hip and knee and dorsiflexion for the ankle were then simulated through an adequate rotation of the relevant distal AF about the medio-lateral axis of the proximal AF. This reproduced a situation similar to that found during gait, whereby one rotational component is significantly greater than the others. The three Cardan angles, as defined by Grood and Suntay [51], were calculated to describe the joint orientation in terms of flexion/extension, ab/adduction and internal/external rotations. These results are reported in Table 4 for joint angle precision when proximal and distal AF are aligned. Internal/external rotation components were the least precise. Precision propagation to knee ab/adduction and internal/external angles was shown to be dependent on the degree of knee flexion. The values of both ab/adduction and internal/external angles were considered to

Table 4
Intra- and inter-examiner precision of the joint angles during upright posture

Joint	Intra-examiner (°)			Inter-examiner (°)		
	Ab/adduction	Internal/external rotation	Flex/extension	Ab/adduction	Internal/external rotation	Flex/extension
Hip	2.5	5.3	3.9	5.2	5.6	5.0
Knee	1.7	5.8	1.0	5.2	10.4	3.7
Ankle	3.5	3.9	1.6	10.9	10.3	3.3

be large enough to affect the reliability of the intrinsically small values of these angles. The same did not hold true for hip and ankle.

Kadaba et al. [13] performed a sensitivity analysis of joint kinematics representation to variations of flexion/extension axis orientation. Joint kinematics data of one subject were used to perform the analysis. The knee flexion/extension axis was made to vary within a range of 30° and errors in knee internal/external and ab/adduction were plotted against the flexion/extension angle. This showed a dependency of the minor angle error on the degree of flexion. Among other observations, they concluded that ab/adduction and internal/external rotation angles must be interpreted with caution, especially at the knee. Ramakrishnan and Kadaba [15] performed the same sensitivity analysis on helical angles, which were found to be as sensitive to rotation axis precision as the Euler angles. Cheze [49] performed a similar test with the goal of identifying the joint kinematics representation the least sensitive to both AL location determination and skin artifacts. A time variant periodic error was added to the AL positions recorded during a gait cycle performed by a subject, to represent both skin movement artifacts and

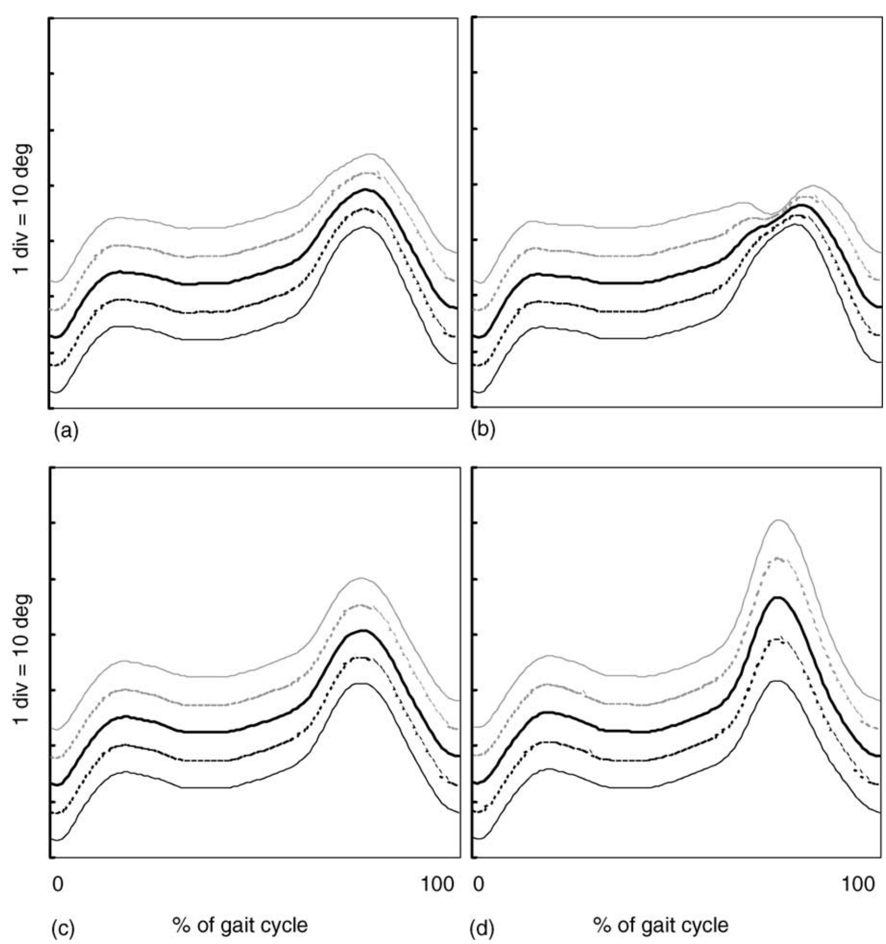

Fig. 2. Effects of proximal AF orientation variations on knee ab/adduction angle (adduction angles are positive) during a gait cycle. The proximal AF orientation is made to vary about the AF anterior/posterior axis: −10° (solid grey line), −5° (dotted grey line), +5° (dotted black line) and +10° (solid black line) with respect to the nominal orientation (thick solid line). Ab/adduction angle is calculated using four joint kinematics description methods: (a) the Cardanic convention introduced in [51]; (b) the non-orthogonal projections of the orientation vector (see [50] for details) on the joint axes defined by Grood and Suntay [51], as proposed by Meglan et al. [52]; (c) the orthogonal projections of the orientation vector on the proximal (thigh) AF [50]; and (d) the joint angles obtained following the geometric approach described by Paul [53].

AL location errors. Unfortunately, the study did not report the details of the analysis methods. Interestingly, however, the results showed that internal/external rotations were the most sensitive to the AL instantaneous position errors. The following figures (Figs. 2–5) extend the work of Fioretti et al. [47] in which the sensitivity to incorrect determination of the direction of the knee flexion/extension axis of four different methods used to describe joint kinematics during gait was tested (see Fig. 2 for details regarding the joint kinematics description methods). Figs. 2–5 illustrate the effects of AF orientation changes on knee kinematics obtained during the gait cycle of a healthy subject and described using the four methods mentioned above. In Figs. 2 and 3 the proximal AF is made to rotate about the anterior–posterior axis of the femur within a $\pm 10^\circ$ range. Fig. 2 shows the effects of the AF rotations on the ab/adduction angle calculated with the four methods (Fig. 2a–d). The four methods show about the same sensitivity to proximal AF rotations until the knee is maximally flexed. In this situation, methods (a) and (b) show a reduced sensitivity, while method (d) becomes more sensitive. Fig. 3 shows the effects of the same AF rotations reported in Fig. 2 on the internal/external angles calculated with the four methods. As expected, considering its definition, the method (d) internal/external angle is not affected by rotations about the AF anterior–posterior axis. The internal/external angles calculated with the other three methods are more sensitive when the knee is maximally flexed.

In Figs. 4 and 5 the knee proximal AF is made to rotate about its longitudinal axis within a $\pm 10^\circ$ range. Fig. 4 shows the sensitivity of ab/adduction calculated with the four methods to AF orientation changes. Similar to that found in Fig. 3, method (d) is not sensitive to rotations of the proximal AF about its longitudinal axis. Among the remaining three methods, method (c) is the least sensitive. The same comments as those for Fig. 2 can be extended to Fig. 5, which reports the sensitivity of the internal/external rotation to the proximal AF rotation. Additionally in method (d) the sensitivity of the angle component coincident with the axis about which the AF was made to rotate is higher than the sensitivity of the other methods. This disadvantage is counterbalanced by the fact that the other component analysed is insensitive to the same AF orientation change. The results have shown that no one of the four methods is the best choice for describing joint kinematics. However, a method has to be chosen if results of different tests are to be compared.

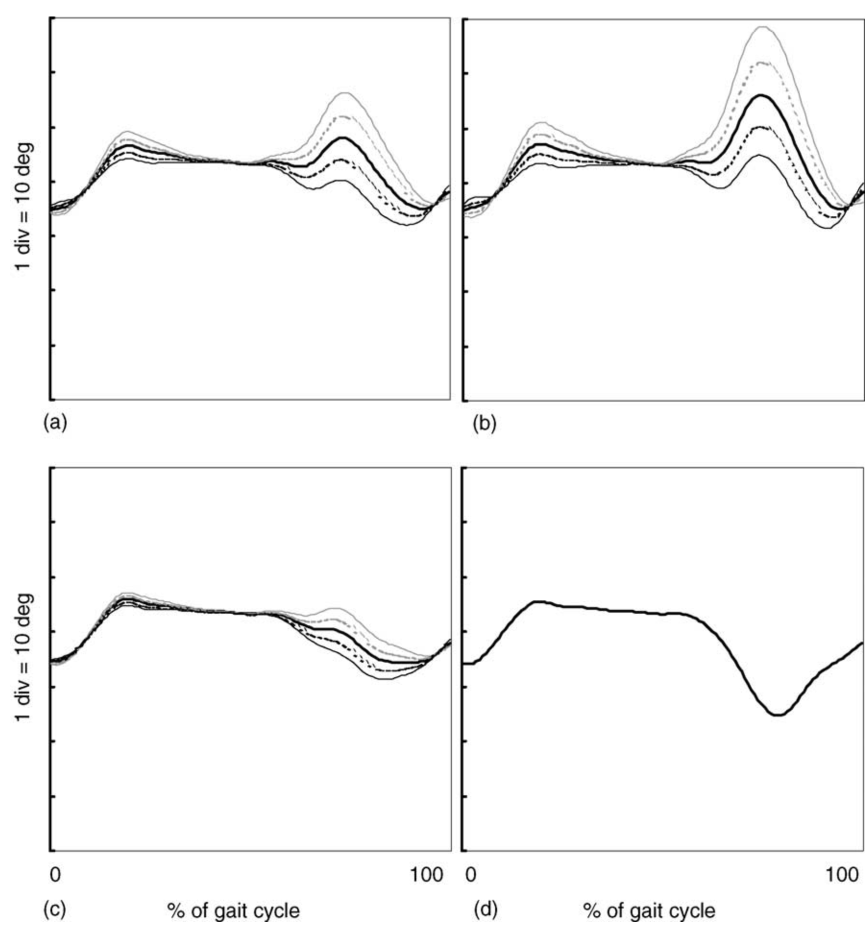

Fig. 3. Effects on knee internal/external rotation angle (internal rotations angles are positive) during a gait cycle of variations of the proximal AF orientation made to vary $\pm 10^\circ$ (increments of 5°) about the AF anterior/posterior axis. Internal/external angle is calculated using four joint kinematics description methods (see Fig. 2 for further details).

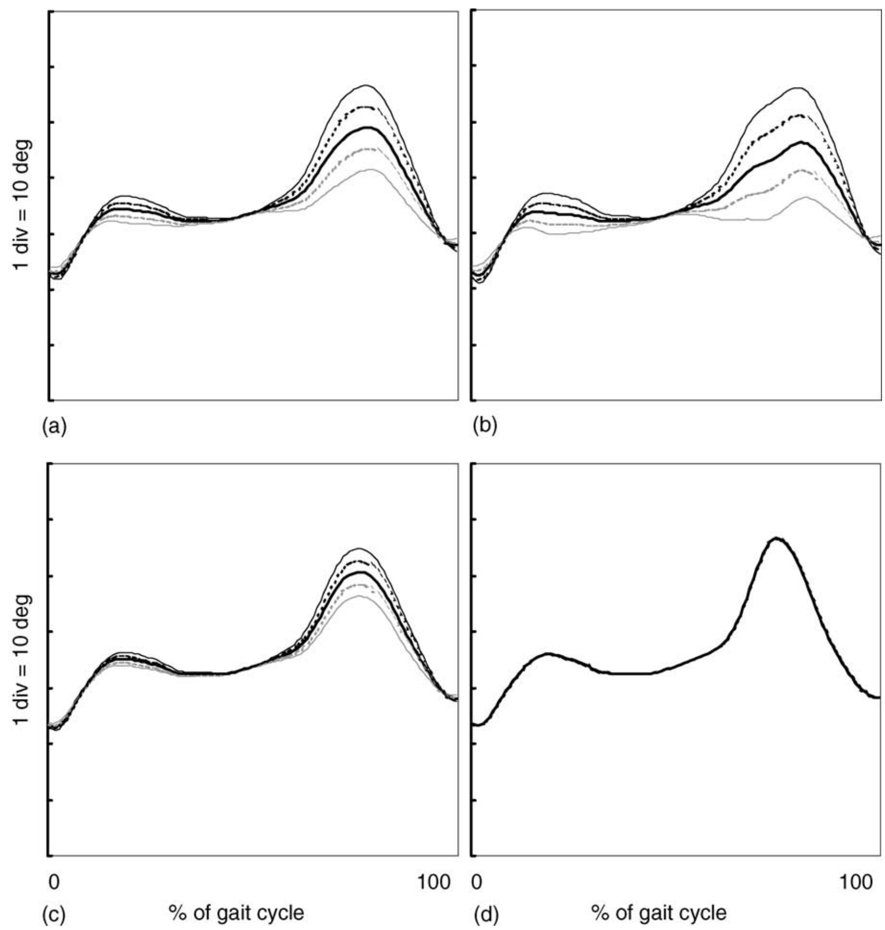

Fig. 4. Effects on knee ab/adduction angle (adduction angles are positive) during a gait cycle of variations of the proximal AF orientation made to vary ±10° (increments of 5°) about the AF longitudinal axis. Ab/adduction angle is calculated using four joint kinematics description methods (see Fig. 2 for further details).

6. Reduction of AL uncertainty effects on joint kinematics

As reported in the previous section, only a limited number of studies dealt with the effect of incorrect definition of joint axes, and therefore of AL location uncertainty, on joint kinematics description. Moreover, the conclusions of those studies were limited to a "warning" to the biomechanical community about using information regarding the minor angles of joint kinematics data. The problem can be approached in two ways: (1) reducing the uncertainty of the AL position determination and/or (2) reducing the effects of such uncertainty on the determination of segment orientation and therefore of joint kinematics. The first approach might be dealt with by using, in addition to rigorous palpation instructions, techniques for the registration of measured AL positions with a model of the AL distribution that can be obtained using imaging techniques. The second approach requires the reduction of AF sensitivity to the AL uncertainty. Recently, a first attempt in the latter direction was proposed by Della Croce et al. [54]. They tested 12 different definitions of the femur AF. Each of them used the position of up to eight femoral ALs and identified an AF using (1) simple geometric rules based on the position of different triplets of ALs, (2) a redundant number of ALs from which three points are calculated and used as input of the above mentioned geometric rules, and (3) a redundant number of ALs as input to optimization algorithms [1]. Sequences of "noisy" AL coordinates were generated by adding zero mean Gaussian sequences and standard deviation found in [7] to the mean AL positions and then used as an input to all the tested rules. The variability of the orientation of the AFs was then determined for each of the rules. This allowed for a selection of the rule with minimal sensitivity to AL position errors. This rule was among those based on an optimization algorithm using a redundant number of ALs. The authors, therefore, concluded that a higher number of ALs and more complex AF definition techniques making a combined use of CTFs and morphology technical frames, may contribute in reducing the errors in joint kinematics due to AL uncertainty. A different study [38] focused on the determination of the medio-lateral axis of the femur. The mean helical axis obtained from an ad hoc knee flexion/extension movement to define the femur AF was used and it was shown that the use of such axis in conjunction with a functional determination

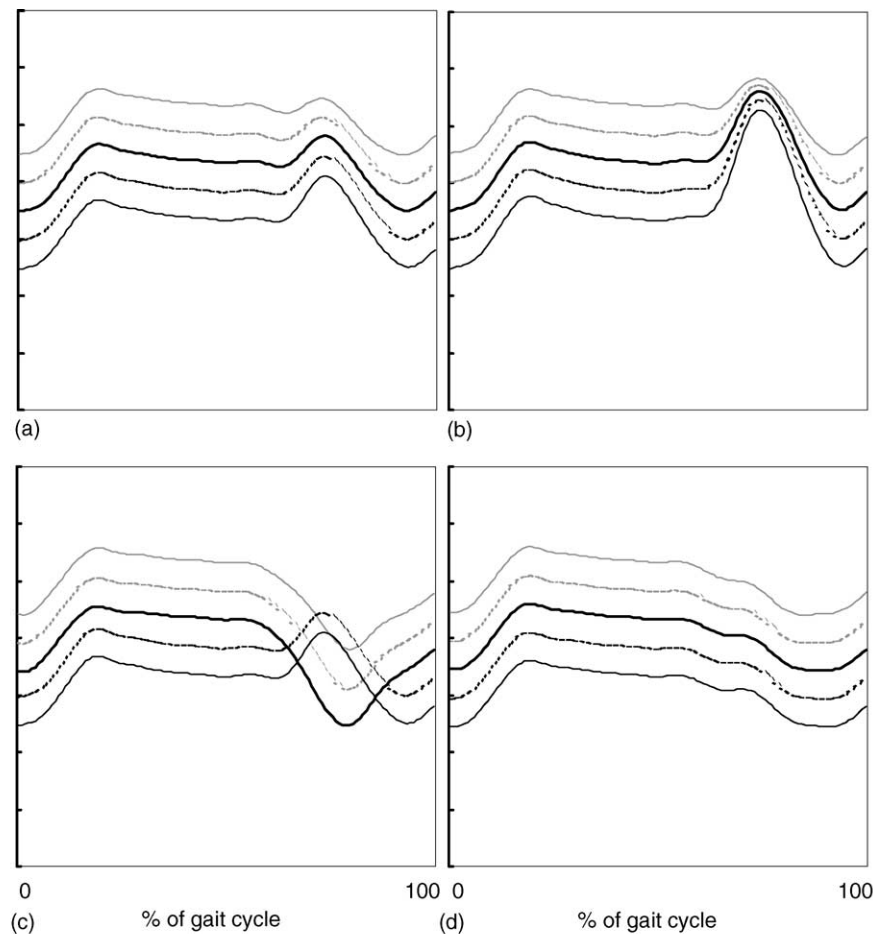

Fig. 5. Effects on knee internal/external rotation angle (internal rotation angles are positive) during a gait cycle of variations of the proximal AF orientation made to vary $\pm 10^\circ$ (increments of 5°) about the AF longitudinal axis. Internal/external angle is calculated using four joint kinematics description methods (see Fig. 2 for further details). The unexpected differences found in the waveforms obtained when method (c) was applied might be due to the fact that the values determined are not angles but projections of a pseudo-vector (the orientation vector) along the axes of the proximal AF.

of both HJC and KJC increased the repeatability of kinematic and kinetic variables observed during gait trials.

7. Conclusion

The studies reviewed in this paper have given quantitative descriptions of the precision of AL position determination and its effects on joint kinematics. Although following different approaches, these have consistently shown that reliability and interpretability of joint kinematics are largely dependent on the precision of the determination of AF orientation. The non-linear nature of this dependency renders the effects on joint kinematics unpredictable. Reduction of these errors can be obtained by improving the AL identification procedure. This reduction can be obtained, for instance, by using imaging techniques, by including in the AF definition a higher number of ALs than the three or four normally used, and by using AF definition rules less sensitive to AL uncertainty. Correct interpretation of joint kinematics remains limited to the major joint angles until the above mentioned improvements are achieved and a standard joint kinematics description method is established for every joint.

The data acquisition techniques developed for, and generally used in, clinical gait analysis often had the reduction of the number of markers and of the complexity of data acquisition as a favourable feature. This was, in part, dictated by the reduced number of cameras and by the limitations of the processing tools available in motion analysis laboratories in the past. Nowadays, with numerous motion analysis laboratories featuring more cameras and extremely advanced software, additional goals can be achieved with minimal complexity added to the acquisition technique. In this and in the previous papers of this series [1,44,55], it was shown that data reliability may be improved if more complex error-reduction-oriented techniques are used.

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