Journal of Biomechanics 41 (2008) 2219–2226

JOURNAL OF BIOMECHANICS

www.elsevier.com/locate/jbiomech
www.JBiomech.com

Anatomical frame identification and reconstruction for repeatable lower limb joint kinematics estimates

Marco Donati, Valentina Camomilla*, Giuseppe Vannozzi, Aurelio Cappozzo

Department of Human Movement and Sport Sciences, Istituto Universitario di Scienze Motorie, Piazza Lauro de Bosis, 15 00194 Roma, Italy

Accepted 17 April 2008

Abstract

The quantitative description of joint mechanics during movement requires the reconstruction of the position and orientation of selected anatomical axes with respect to a laboratory reference frame. These anatomical axes are identified through an ad hoc anatomical calibration procedure and their position and orientation are reconstructed relative to bone-embedded frames normally derived from photogrammetric marker positions and used to describe movement. The repeatability of anatomical calibration, both within and between subjects, is crucial for kinematic and kinetic end results. This paper illustrates an anatomical calibration approach, which does not require anatomical landmark manual palpation, described in the literature to be prone to great indeterminacy. This approach allows for the estimate of subject-specific bone morphology and automatic anatomical frame identification. The experimental procedure consists of digitization through photogrammetry of superficial points selected over the areas of the bone covered with a thin layer of soft tissue. Information concerning the location of internal anatomical landmarks, such as a joint center obtained using a functional approach, may also be added. The data thus acquired are matched with the digital model of a deformable template bone. Consequently, the repeatability of pelvis, knee and hip joint angles is determined. Five volunteers, each of whom performed five walking trials, and six operators, with no specific knowledge of anatomy, participated in the study. Descriptive statistics analysis was performed during upright posture, showing a limited dispersion of all angles (less than 3 deg) except for hip and knee internal–external rotation (6 deg and 9 deg, respectively). During level walking, the ratio of inter-operator and inter-trial error and an absolute subject-specific repeatability were assessed. For pelvic and hip angles, and knee flexion-extension the inter-operator error was equal to the inter-trial error—the absolute error ranging from 0.1 deg to 0.9 deg. Knee internal–external rotation and ab-adduction showed, on average, inter-operator errors, which were 8% and 28% greater than the relevant inter-trial errors, respectively. The absolute error was in the range 0.9–2.9 deg.

Keywords: Anatomical calibration; Anatomical landmarks; Biomechanics; Human movement analysis; Stereophotogrammetry

1. Introduction

Movement analysis using rigid-body modelling requires the definition of local orthogonal systems of axes associated with each bony segment of interest. The systems of axes defined using photogrammetric markers associated with the bone of interest are referred to as marker cluster frames, they are used to describe the instantaneous global bone-pose and, normally, they have an arbitrary and non-repeatable pose relative to the bone. In order for the six variables that are estimated for the joint kinematics description to be both intra- and inter-subject repeatable, they must be calculated using the pose of local frames that are themselves repeatable. The latter frames are based on selected anatomical features of the bone and are referred to as anatomical frames (AF).

The position and orientation of the AF relative to a marker cluster frame is obtained using the positional information of anatomical landmarks (AL) in the latter frame and a deterministic or statistical geometric rule (calibrated anatomical system technique (CAST) protocol—as described in Cappozzo (1984) and Cappozzo et al. (1995)). Superficial ALs are identified by palpation; the location of internal ALs is estimated using statistical models (Della Croce et al., 2005) or, as may be the case with joint centers, using a functional

*Corresponding author. Tel.: +39 06 36733522; fax: +39 06 36733517.
E-mail address: valentina.camomilla@iusm.it (V. Camomilla).

0021-9290/$ - see front matter © 2008 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2008.04.018

approach (Cappozzo, 1984). Several papers report information concerning the precision of locating both internal and palpable ALs by this anatomical calibration procedure and of estimating the pose of the relevant AFs (Della Croce et al., 2005). How these uncertainties propagate to joint kinematics and kinetics has also been extensively investigated (Ramakrishnan and Kadaba, 1991; Della Croce et al., 1999; Stagni et al., 2000). With reference to pelvis and lower limb bones, the dispersion of AL position and AF orientation, as identified by different operators, may exhibit a root mean square value in the range 10–25 mm and 3–10 deg, respectively (Della Croce et al., 1999). This literature highlights that, using the current "state of the art" methods, the scarce repeatability with which palpable ALs are identified may lead to concealing both the intra- and inter-individual differences sought in clinical practice as well as in basic research. The relevant debate published in the Journal of Pediatric Orthopedics in 2003 (Noonan et al., 2003; Gage, 2003; Wright, 2003) further emphasizes the importance of the repeatability issue as associated with the use of gait analysis data in clinical decision making.

Recently, an anatomical calibration method was designed to enhance repeatability with respect to the conventional CAST approach (Donati et al., 2007). This procedure, named UP-CAST, was applied to the femur and entailed the determination of the positions of a large number of unlabelled points (UPs) located over all prominent parts of the surface of the bone and the matching of a deformable digital model of a template bone to them. The resulting digital bone model provides an estimate of the position of all relevant ALs in the marker cluster frame. The accurate in-vivo identification of ALs through palpation is therefore not required, since the template bone already embeds information on the ALs location.

This paper presents a movement analysis protocol that applies the mentioned UP-CAST calibration to the pelvis, femur, and shank bones. Repeatability, as the absolute prerequisite for any protocol to become the language of science in a clinical context, was assessed for the location of landmarks of the above-mentioned bones and for the estimate of pelvis, hip, and knee joint 3D kinematics.

2. Materials and methods

2.1. Digital bone templates

Digital models of a femur, a tibia, and a fibula of an adult male subject were made available from the Visible Human Project data. For the pelvis, given the more evident gender-based differences characteristic of this bone (Kepple et al., 1998), both a male and a female template were used. The ALs listed in Fig. 1 were identified on these digital models using the written and pictorial instructions delivered in the Vakhum EU project (Van Sint Jan, 2007).

2.2. Data acquisition

Five adult able-bodied volunteers were selected in order to represent both genders, with a body mass index in the normal range

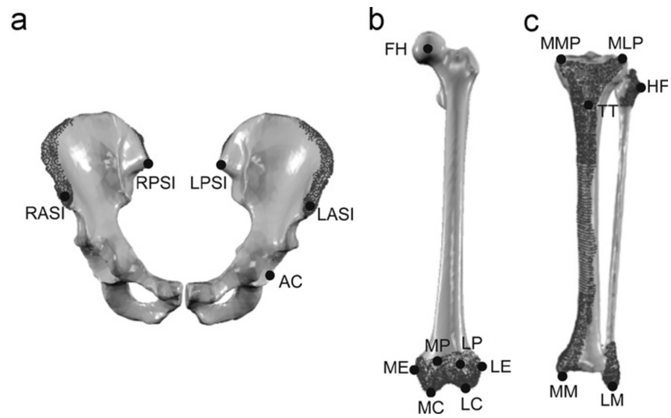

Fig. 1. Position and acronyms of the selected landmarks. (a) *Pelvis:* left and right anterior superior iliac spines (LASI, RASI), left and right posterior superior iliac spines (LPSI, RPSI), center of the acetabulum (AC). (b) *Femur:* lateral and medial epicondyles (LE, ME), antero-medial (MP) and antero-lateral (LP) ridge of the patellar grove, lateral and medial most distal point of the condyles (LC, MC), femoral head (FH). (c) *Tibia and Fibula:* tibial tuberosity (TT), lateral and medial malleoli (LM, MM), head of the fibula (HF), most lateral and medial points of the tibial plateau (MLP, MMP). Areas digitized using UP-CAST are indicated with darker points.

Table 1
Subjects' gender, body mass index and skin-fold thickness measured in the indicated sites

Subj	Gender	BMI (kg/m^2)	Iliac crest (mm)	Front thigh (mm)	Medial calf (mm)
1	M	23.9	11.1	11.8	11.3
2	M	21.6	7.4	5.1	11.2
3	M	19.4	11.4	14.9	8.8
4	F	20.4	21.0	13.3	25.7
5	F	21.8	16.4	5.7	23.6

(BMI = 19–24 kg/m^2). Skin-fold thickness measurements were carried out in order to highlight inter-subject differences in soft tissue layers over the lower limb, which, of course, BMI cannot supply (Table 1).

The UP-CAST protocol was carried out by six technicians who had no specific competences in AL identification through palpation. Since intra- and inter-examiner precision of the calibration have been shown not to be significantly different (Donati et al., 2007), only inter-examiner precision was assessed.

For anatomical calibration purposes, the pelvis, thigh, and shank of the volunteers were fitted with plate-mounted clusters of markers. Their geometry followed the recommendations given in Cappozzo et al. (1997) and their locations were chosen to minimally interfere with the anatomical calibration procedure. Areas located on the sacrum (Fig. 2a), the central thigh, and the postero-lateral shank were used. Based on these clusters, cluster frames were constructed (CF-UP).

The positions of unlabelled points (UPs), located over all prominent parts of the selected bones, were determined with respect to the relevant CF-UP, using a wand equipped with a cluster of three markers and a sphere on the tip that rolls over the surface to be digitized (Fig. 2a). The position of the tip of the wand relative to the wand markers was determined through a stereophotogrammetric calibration procedure. The accuracy of this calibration was within 1 mm. Only body segment areas where the soft tissue layer over the bone exhibited the smallest thickness were digitized so that the surfaces determined could be associated to the bone. Specifically, the areas acquired were: for the pelvis, around the iliac spines and the iliac crest; for the femur, on the condyles and the prominent

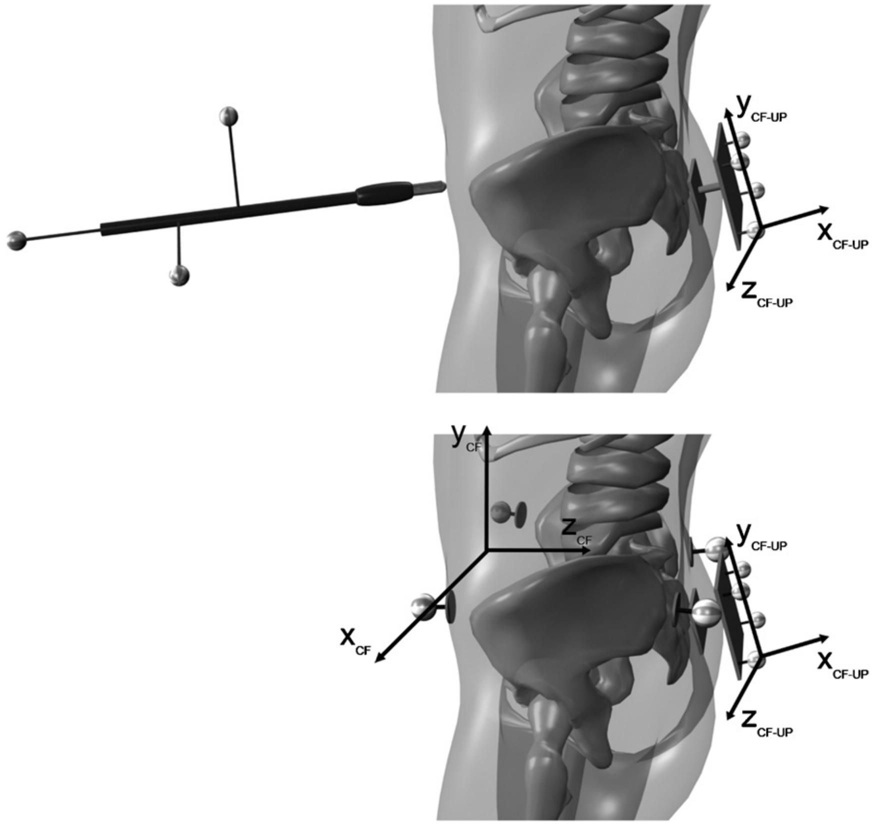

Fig. 2. (a) Technical cluster frame (CF-UP) that minimally interferes with the anatomical calibration. The wand is also shown. (b) Technical cluster frame (CF) that, during movement, is less affected than the CF-UPs by inertial effects and soft tissue artifacts.

patellar groove; for the tibia, on the tibial tuberosity along the anterior crest down to the medial malleolus; for the fibula, around the head and the lateral malleolus (Fig. 1). The subjects were asked to assume an upright posture during the pelvis calibration and to keep their knee flexed at 90° with the aid of a stool when calibrating the femur, tibia, and fibula.

The CF-UPs were not adequate for movement tracking purposes because they were prone to large artifact movements that could not possibly be compensated for using post-processing optimization techniques (Soderkvist and Wedin, 1993). Thus, skin marker sets were used, aiming at reducing the negative effects of soft tissue artifacts. In particular, four markers were located on the pelvic iliac spines (Fig. 2b), four on the central thigh, two on the antero-medial surface of the tibia, one close to the medial malleolus and one close to the styloid process of the fibula. Using these markers, non-rigid clusters were obtained and used to define cluster frames, hereafter referred to as CFs.

In order to merge calibration and movement data, transformation matrices between CF-UPs and relevant CFs were determined. This was done using the global position of the markers of both clusters involved reconstructed while the volunteers assumed the same postures as during UPs digitization: upright posture for the pelvis markers and with the knee flexed for femur and tibia–fibula complex markers.

A further landmark, the center of the femoral head (FH), was estimated and included in the data set. Its location, assumed to coincide with the center of the acetabulum (AC), was determined in the femur and the pelvic CFs using the functional approach described in Camomilla et al. (2006). To this purpose, volunteers were asked to move their thigh relative to their pelvis by flexing, extending and circumducting it.

Similarly to what has been done in previous studies that aimed at assessing repeatability of gait data (Schwartz et al., 2004; Charlton et al., 2004), the volunteers were then asked to perform five walking tasks at a self-selected speed of progression.

Markers were tracked by a nine camera photogrammetric system (Vicon MX®) at 120 frames/s.

2.3. Data processing

Marker cluster frames were constructed using a least-squares approach (Soderkvist and Wedin, 1993). The position of AC was determined in the pelvic CF using a bias-compensated quartic best-fit algorithm (Gamage and Lasenby, 2002; Halvorsen, 2003). The UPs locations, in a first instance represented in the CF-UPs, were represented in the relevant CFs through the above-mentioned rigid transformation. Based on these experimental data, the selected bone templates, and the mathematical procedure briefly illustrated below, bone models, that approximated the subjects' bones, and the relevant AL position vectors in the CFs were estimated.

A non-isomorphic scaling and a re-orientation of the template aimed at matching it with the measured UPs in the CF were carried out. The coordinates of the superficial points of the subject bone model estimate, represented in the CF, were expressed as

$$ {}^{c}\hat{\mathbf{p}} = \mathbf{d} \cdot {}^{c}\mathbf{T}_{m} \cdot {}^{m}\mathbf{tp} \qquad (1) $$

where ${}^{m}\mathbf{tp}$ represents a template point defined in an arbitrary morphology frame, ${}^{c}\mathbf{T}_{m}$ is the transformation matrix which actuates the re-orientation of the template and $\mathbf{d}$ is a diagonal matrix which carries scale factors. The use of the latter matrix implies that the bone did not undergo torsion nor bending. These parameters were estimated through a first approximation registration and a subsequent minimization of a cost function, based on the direct Hausdorff distance between the template points and the UPs, which is an enhanced version of that proposed in Donati et al. (2007):

$$ f(\mathbf{d}, {}^{c}\mathbf{T}_{m}) = \frac{1}{|\mathrm{EUP}|flag} \sum_{\mathbf{up} \in \mathrm{EUP}} w_{\mathbf{up}} \left(\min_{\hat{\mathbf{p}} \in \mathrm{ETP}} \left| {}^{c}\hat{\mathbf{p}} - {}^{c}\mathbf{up} \right| \right) \qquad (2) $$

where ${}^{c}\mathbf{up}$ represents the position vector of the UPs in the CF, EUP and ETP are the ensemble of the UPs and the template points, respectively. To increase the robustness of the method, the w_{up} variable was included to account for the relative UP and template positions, ensuring that the template lies inside the UPs. The w_{up} variable doubles the minimal distances averaged in the Hausdorff distance only for the UPs laying inside the template surface, and it is equal to one for points outside it. The position of the points relative to the template surface is identified by calculating their distance from it, made positive for points inside the surface and negative for those outside (Hoppe et al., 1992). For bones characterized by an asymmetrical spatial distribution of the prominent areas, a fiducial point, namely FH and AC for femur and pelvis, respectively, was used. The *flag* variable weighs the cost function based on the use of the selected fiducial point. When this point is close to the relevant area of the template under analysis the *flag* value abruptly decreases the cost function value.

The above-mentioned procedure was applied on the different body segments, using segment-specific scaling criteria: for tibia and fibula scaling matrices were estimated separately, left and right iliac bones underwent the same non-isomorphic scaling, while the femur was scaled isomorphically, since only its distal morphology was available.

Each anatomical calibration provided three data sets made of the anatomical landmarks vectors ${}^{c}\hat{\mathbf{a}}_j$ for each body segment. For the purpose of result interpretation, these vectors were represented in three AFs associated with pelvis, femur and tibia–fibula complex and constructed as proposed in Cappozzo et al. (1995). A mean AF was determined for each bone or complex using the means of all the relevant ${}^{c}\hat{\mathbf{a}}_j$ vectors. Thereafter, vector transformation was applied to obtain the anatomical landmarks relative to the mean AFs, ${}^{\bar{a}}\hat{\mathbf{a}}_j$.

The precision of the method was evaluated in terms of root mean square (RMS) error from the mean of all ${}^{\bar{a}}\hat{\mathbf{a}}_j$ vectors for each subject and bone. Three-dimensional AL position precisions were also calculated as the RMS of the norm of ${}^{\bar{a}}\hat{\mathbf{a}}_j - {}^{\bar{a}}\bar{\hat{\mathbf{a}}}_j$, i.e. the distances between each AL and its mean position.

For each gait trial and the six anatomical calibrations performed, hip and knee joint kinematics were computed using the Cardan angular convention (Grood and Suntay, 1983). The pelvis kinematics was computed by reconstructing the movements of the pelvic frame relative to the global frame oriented so that the positive x-, y-, and z-directions were directed forward along the progression, left, and the upward, respectively. The Cardan angular convention was used to derive the sequence tilt, obliquity, and rotation. Although another sequence has been shown to correspond more closely to the conventional clinical understanding of the pelvic angles (Baker, 2001), the Cardan convention was selected since it is more extensively used in the literature and allows for comparisons with previous repeatability studies.

Since hip and knee joint angles are affected by the AL location errors during both gait and upright posture, the two effects were investigated separately. For these joints, the time functions of the angles during movement were aligned with respect to the relative upright posture angles and descriptive statistics was performed on posture angles and represented using box plots. Repeatability of all angle time functions was then assessed as follows. Let $\phi(t)$ denote one of the gait angles aligned with the relative upright posture, and let the indices k and m denote operator and trial. Then, $\phi_{k,m}{}^{\text{subj}}$ is a gait angle for one subject (subj) associated with a single trial (m) and a single operator (k). The variable $\phi_{k,m}{}^{\text{subj}}(t)$ is time dependent. For each subject, the following parameters, modified from those defined in Schwartz et al. (2004), were derived:

$$\bar{\phi}_k^{\text{subj}}(t) = \frac{1}{N_{\text{trials}}} \sum_{m=1}^{N_{\text{trials}}} \phi_{k,m}^{\text{subj}}(t) \tag{3}$$

$$\bar{\phi}^{\text{subj}}(t) = \frac{1}{N_{\text{trials}}} \frac{1}{N_{\text{oper}}} \sum_{k=1}^{N_{\text{oper}}} \sum_{m=1}^{N_{\text{trials}}} \phi_{k,m}^{\text{subj}}(t) \tag{4}$$

The estimated standard errors of each ϕ are the standard deviation of the differences between ϕ and the relevant mean:

$$\sigma_{\phi(t)}^{\text{subj,trials}} = \sqrt{\frac{1}{N_{\text{trials}} N_{\text{oper}} - 1} \sum_{k=1}^{N_{\text{oper}}} \sum_{m=1}^{N_{\text{trials}}} \left(\phi_{k,m}^{\text{subj}}(t) - \bar{\phi}_k^{\text{subj}}(t)\right)} \tag{5}$$

$$\sigma_{\phi(t)}^{\text{subj,oper}} = \sqrt{\frac{1}{N_{\text{trials}} N_{\text{oper}} - 1} \sum_{k=1}^{N_{\text{oper}}} \sum_{m=1}^{N_{\text{trials}}} \left(\phi_{k,m}^{\text{subj}}(t) - \bar{\phi}^{\text{subj}}(t)\right)} \tag{6}$$

The estimated standard error of ϕ was also computed considering each trial singularly:

$$\sigma_{\phi(t)}^{\text{subj},i\text{th trial}} = \sqrt{\frac{1}{N_{\text{oper}} - 1} \sum_{k=1}^{N_{\text{oper}}} \left(\phi_{k,i}^{\text{subj}}(t) - \frac{1}{N_{\text{oper}}} \sum_{k'=1}^{N_{\text{oper}}} \phi_{k',i}^{\text{subj}}(t)\right)} \tag{7}$$

Mean estimated standard errors $\bar{\sigma}_\phi$ were obtained by averaging over time the defined $\sigma_{\phi(t)}$:

$$\bar{\sigma}_\phi^{\text{subj,var}} = \frac{1}{N_{\text{frames}}} \sum_{t=1}^{N_{\text{frames}}} \sigma_{\phi(t)}^{\text{subj,var}}, \quad \text{var} = \text{trials, oper}, i\text{th trial} \tag{8}$$

The repeatability was assessed in terms of the ratio of the inter-operator error to the inter-trial error ($\bar{\sigma}_\phi^{\text{subj,oper}} / \bar{\sigma}_\phi^{\text{subj,trials}}$). The inter-trial error is free of methodological errors, and thereby serves as an appropriate baseline for comparisons. To have an absolute reference in degrees, an alternative trial-independent repeatability, $\bar{\sigma}_\phi^{\text{subj}}$, was assessed:

$$\bar{\sigma}_\phi^{\text{subj}} = \frac{1}{N_{\text{trials}}} \sum_{i=1}^{N_{\text{trials}}} \bar{\sigma}_\phi^{\text{subj},i\text{th trial}} \tag{9}$$

3. Results

The precision of UP-CAST in locating the anatomical landmarks, ${}^{\bar{a}}\hat{\mathbf{a}}_j$, was evaluated for each bony segment as average over the five subjects (Table 2). Relevant values, along the three anatomical axes, ranged from 2.9 to 7.3 mm for the pelvis, from 2.6 to 7.2 mm for the femur and from 1.7 to 6.6 mm for the tibia and fibula.

On the pelvis, errors were slightly larger in the vertical direction. Anterior ALs resulted more dispersed than the posterior ones. Femoral medial ALs (ME, MP, MC) were

Table 2
Inter-examiner precision of the anatomical landmark position vector components (antero-posterior, AP, vertical, V; medio-lateral, ML) in the relevant mean AF. The 3D precision is also reported

	AP (mm)	V (mm)	ML (mm)	3D (mm)
Pelvis				
LASI	2.9	7.3	4.1	8.4
RASI	3.0	6.1	5.3	8.2
LPSI	3.8	5.3	2.8	6.9
RPSI	3.0	5.1	3.3	6.4
Femur				
LE	4.4	5.1	2.9	6.8
ME	7.2	4.8	3.2	8.5
LP	3.5	4.1	2.6	5.7
MP	6.1	4.0	3.1	7.4
LC	3.3	5.1	2.6	6.1
MC	4.9	4.8	4.1	7.4
Tibia and Fibula				
TT	1.7	5.2	3.3	6.2
HF	3.9	2.6	2.1	5.1
LM	2.9	3.7	2.1	4.8
MM	5.5	4.8	3.9	8.1
MLP	5.2	6.1	5.1	9.1
MMP	6.6	5.7	6.6	10.6

more dispersed along the anterior–posterior direction. With reference to the tibia–fibula complex, results confirmed HF as the most precise AL, as in Della Croce et al. (1999). Similar good performances were obtained for LM and for TT. Landmarks on the tibial plateau (MMP and MLP) were confirmed to be the most dispersed, however the relevant 3D error resulted lower than 11 mm.

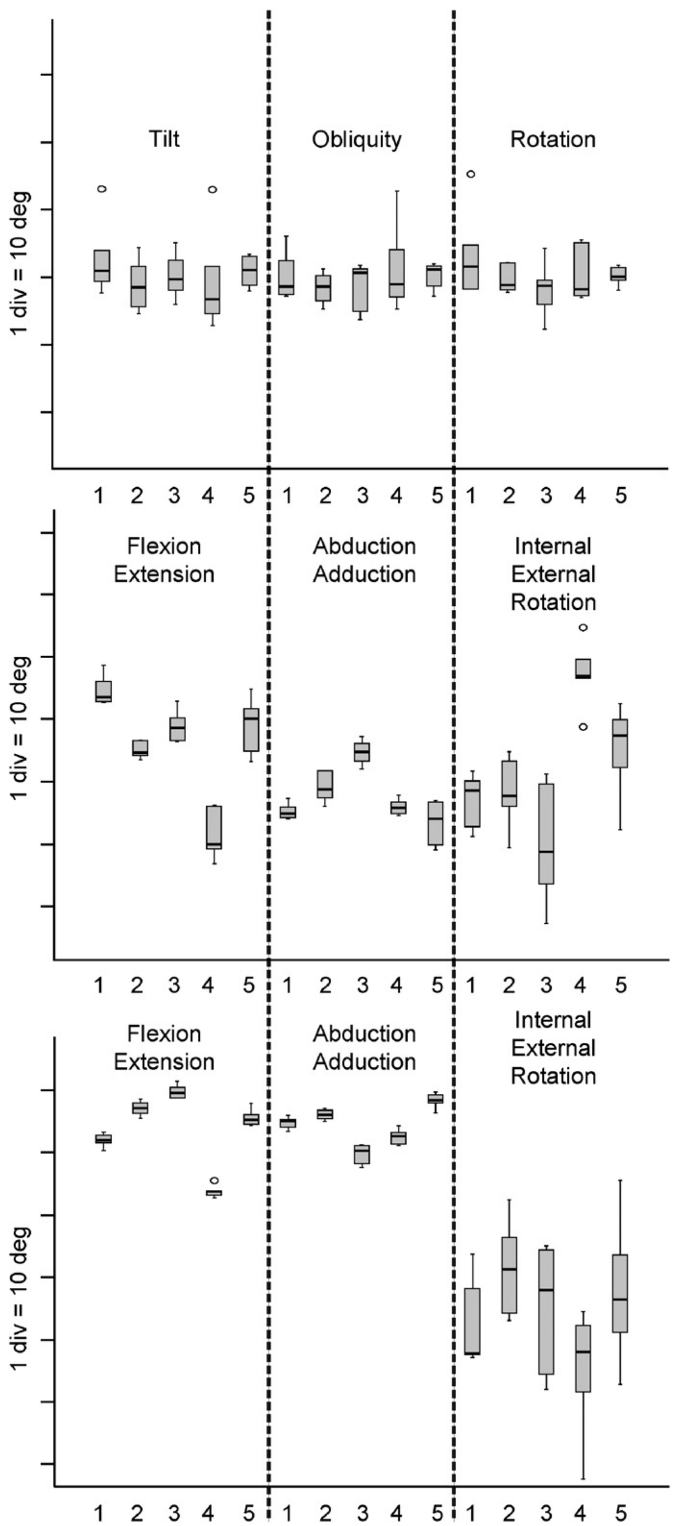

Fig. 3. Box plot descriptive statistics for the upright posture angles of all subjects.

The repeated anatomical calibration affected pelvis, hip and knee posture angles of all subjects in a similar way (Fig. 3). Hip and knee internal–external rotation errors underwent the largest variations, corresponding to ~15 deg of inter-quartile range (IQR). For the other angles the IQR was lower than 9 deg.

Repeatability of joint kinematics was visibly high, as reported in Fig. 4 for pelvis, hip and knee angles.

The inter-operator error was equal to the inter-trial error for pelvis and hip angles and for knee flexion extension. The $\bar{\sigma}_{\phi}^{\text{subj}}$ error ranged from 0.1 deg to 0.9 deg (Table 3). Knee internal–external rotation and ab-adduction showed, on average, $\bar{\sigma}_{\phi}^{\text{subj,oper}}$ errors of 8% and 28% greater than the relevant $\bar{\sigma}_{\phi}^{\text{subj,trials}}$, respectively. For these angles, the $\bar{\sigma}_{\phi}^{\text{subj}}$ error was between 0.9 deg and 2.9 deg (Table 3).

4. Discussion

The robustness of the UP-CAST method, modified and extended with respect to the version illustrated in Donati et al. (2007), was assessed. The method was shown to improve the precision in locating anatomical axes and, therefore, to reduce error propagation to 3D joint kinematics, and exhibited operative advantages with respect to anatomical landmark manual palpation.

The precision in locating anatomical landmarks of the pelvis, femur and tibia–fibula (maximal 3D error: 11 mm), reinforced the results previously obtained by Donati et al. (2007) and was better than that yielded by the conventional calibration as assessed by Della Croce et al. (1999) (maximal 3D error: 18 mm). In the latter work, the subjective interpretation associated with AL determination was addressed as the main source of error. The UP-CAST protocol overcomes this problem: the operators simply palpate the prominent bone areas and do not identify the ALs one by one. Moreover, the consequences of possible erroneous palpation of these areas are mitigated by the a-priori information included in the digital bone model. For this reason, the present approach is particularly appropriate for identifying ALs that are broad areas instead of mere points. For instance, the greatest improvements with respect to the conventional calibration (Della Croce et al., 1999) were obtained for pelvic ALs; the dispersion of these ALs, responsible for the larger errors in conventional calibration, became similar to that of the thigh and shank landmarks.

As expected, the upright posture angles were affected by AL misidentification and implied a systematic error in the angle time functions during gait. The propagation of AL misidentification to posture angles, while confirming previous results (Della Croce et al., 1999), showed a moderate dispersion (less than 3 deg) for the pelvis and for flexion extension and ab-adduction of both the hip and knee. For these joints, internal–external rotation posture angles had a much higher variability (6 deg and 9 deg,

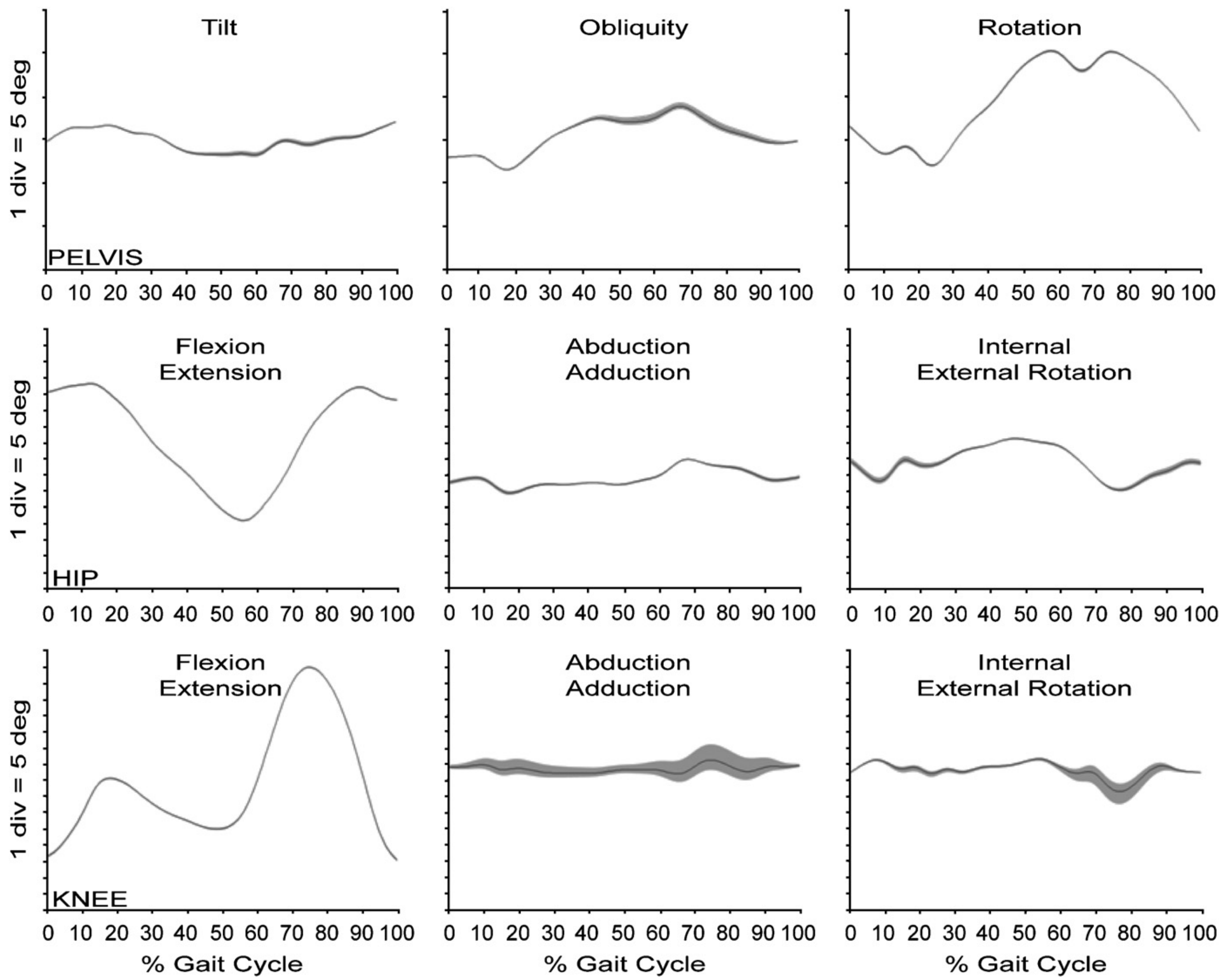

Fig. 4. Inter-operator variability of the joint kinematics of one subject and one gait cycle: minimum, maximum and average over all operators.

Table 3
The $\bar{\sigma}_{\phi}^{subj}$ error is given for each subject and each angle under analysis (ob: obliquity; rot: rotation; fe: flexion extension; aa: abduction–adduction; ie: internal–external rotation)

Subj	Pelvis (deg)			Hip (deg)			Knee (deg)		
	tilt	ob	rot	fe	aa	ie	fe	aa	ie
1	0.1	0.1	0.1	0.1	0.2	0.3	0.2	1.7	0.9
2	0.2	0.1	0.1	0.1	0.3	0.7	0.2	1.9	1.0
3	0.2	0.3	0.2	0.1	0.3	0.6	0.5	2.9	2.0
4	0.3	0.7	0.6	0.2	0.4	0.4	0.4	2.2	0.9
5	0.2	0.2	0.3	0.3	0.5	0.9	0.4	1.9	1.0

respectively). This variability is a consequence of the larger error associated with the identification of the anatomical medio-lateral axis of the femoral AF.

The effectiveness of the UP-CAST method is evidenced by the low error propagation to joint kinematics. As evidenced by the data reported in Table 4, pelvis, hip and knee angles showed a reduced ratio of inter-operator and inter-trial error if compared with that of the Vicon Clinical Manager protocol (Davis et al., 1991) as assessed in two different laboratories (Charlton et al., 2004; Schwartz et al., 2004). Note that these assessments differed in the number of subjects, operators, and trials per subject. Within the UP-CAST method, the angles more affected by calibration error propagation were knee ab-adduction and internal–external rotation with $\bar{\sigma}_{\phi}^{subj}$ errors never higher than 3 deg. These angles were confirmed to be the most sensitive to all experimental errors, as kinematic cross talk (Della Croce et al., 2005) and soft tissue artifact (Leardini et al., 2005).

Two further strength points of the UP-CAST method regard its practical applicability. First, the time required for landmark identification is drastically reduced. For the pelvis, distal femur, tibia and fibula the identification of the landmarks necessary to perform a conventional calibration

Table 4
$\bar{\sigma}_{\phi}^{\text{subj,oper}}/\bar{\sigma}_{\phi}^{\text{subj,trials}}$ for the proposed UP-CAST method and for the Vicon Clinical Manager (VCM) assessed in two laboratories, using different numbers for subjects, operators, and trials per subject. Most of the numbers were extracted from graphs. (ob: obliquity; rot: rotation; fe: flexion extension; aa: abduction–adduction; ie: internal–external rotation)

	Pelvis			Hip			Knee		
	tilt	ob	rot	fe	aa	ie	fe	aa	ie
UP-CAST	1.0	1.0	1.0	1.0	1.0	1.0	1.0	1.4	1.1
VCM (Charlton et al., 2004)	–	–	–	1.4	1.3	1.5	1.6	1.6	1.8
VCM (Schwartz et al., 2004)	2.8	3.7	7.2	4.2	3.8	3.0	~2	5.0	~3

(Cappozzo et al., 1995; Davis et al., 1991) could last 10–15 min, while only 5–6 min are required to calibrate the selected areas using the UP-CAST procedure. Moreover, a larger number of landmarks can be made available without increasing the calibration time. Potentially, the method allows for the identification of anatomical axes defined using specific geometric characteristics of the bone surface and not only punctiform landmarks. This is the case, for example, with the spherical head of the femur or with the geometrical axis of the femoral condyle defined as a line connecting the centers of the posterior circular surfaces of the medial and lateral condyles (McPherson et al., 2005). Second, the UP-CAST method may be applied successfully by non-skilled operators, since it does not strictly require professional anatomy-specific knowledge. This represents a key point while doing movement analysis, because it eases the anatomical calibration without compromising with precision.

While using the UP-CAST calibration, if a subject-specific digital model of the bone is not available, as is generally the case, attention has to be paid to the limitations associated with the morphological difference between the selected template and the bone under analysis. Although the sensitivity to template differences of the repeatability of ALs identification is as yet not been assessed, it may be hypothesized that better results could be achieved if a bone database is made available, thus allowing the selection of a specific template that best matches the subject's morphology. This issue becomes more critical when the estimate of a subject-specific bone and the relevant accuracy are pursued, rather than only the repeatability of ALs location. In this respect, it is interesting to observe that the dispersion with which the ALs of a given bone may be determined using the UP-CAST method (Table 2) is similar to the relevant biological variability as assessed in White et al. (1989) and in Kepple et al. (1998). This circumstance highlights the importance of applying the UP-CAST procedure using the best-possible template bone.

The extension of the present results to overweight subjects requires further investigations. A correlation analysis between body skin-fold and mass indices and joint kinematics repeatability evidenced high correlations ($r = 0.8$ on average) between iliac crest thickness and pelvic rotations as well as between body mass index and knee rotations, only two of these correlations being significant. The low number of volunteers and the range of their BMI does not allow a generalization of the present results to subjects with higher BMI. It should be emphasized, however, that when applying movement analysis to overweight subjects, there may be more disruptive error sources than those associated with anatomical calibration.

The UP-CAST method was shown to speed up the anatomical calibration procedure lowering the knowledge required to perform it and, nevertheless, improving the repeatability of joint kinematics. These strength points promote the UP-CAST anatomical calibration as a promising alternative to conventional calibration in the clinical context.

Conflict of interest

None

Acknowledgments

This work was supported by the Italian Ministry of University and Research and by the Istituto Universitario di Scienze Motorie, Roma.

References

Baker, R., 2001. Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait and Posture 13, 1–6.

Camomilla, V., Cereatti, A., Vannozzi, G., Cappozzo, A., 2006. An optimized protocol for hip joint centre determination using the functional method. Journal of Biomechanics 39 (6), 1096–1106.

Cappozzo, A., 1984. Gait analysis methodology. Human Movement Science 3, 27–50.

Cappozzo, A., Catani, F., Della Croce, U., Leardini, A., 1995. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clinical Biomechanics (Bristol, Avon) 10 (4), 171–178.

Cappozzo, A., Cappello, A., Della Croce, U., Pensalfini, F., 1997. Surface-marker cluster design criteria for 3-D bone movement reconstruction. IEEE Transactions on Biomedical Engineering 44 (12), 1165–1174.

Charlton, I.W., Tate, P., Smyth, P., Roren, L., 2004. Repeatability of an optimised lower body model. Gait and Posture 20 (2), 213–221.

Davis, R.B.I., Ounpuu, S., Tyburski, D., Gage, J.R., 1991. A gait data collection and reduction technique. Human Movement Sciences 10, 575–587.

Della Croce, U., Cappozzo, A., Kerrigan, D.C., 1999. Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. Medical and Biological Engineering and Computing 37 (2), 155–161.

Della Croce, U., Leardini, A., Chiari, L., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry. Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics. Gait and Posture 21 (2), 226–237.

Donati, M., Camomilla, V., Vannozzi, G., Cappozzo, A., 2007. Enhanced anatomical calibration in human movement analysis. Gait and Posture 26 (2), 179–185.

Gage, J.R., 2003. Editorial: con: interobserver variability of gait analysis. Journal of Pediatric Orthopaedics 23, 290–291.

Gamage, S.S., Lasenby, J., 2002. New least squares solutions for estimating the average centre of rotation and the axis of rotation. Journal of Biomechanics 35 (1), 87–93.

Grood, E.S., Suntay, W.J., 1983. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. Journal of Biomechanical Engineering 105 (2), 136–144.

Halvorsen, K., 2003. Bias compensated least squares estimate of the center of rotation. Journal of Biomechanics 36 (7), 999–1008.

Hoppe, H., DeRose, T., Duchamp, T., McDonald, J., Stuetzle, W., 1992. Surface reconstruction from unorganized points. In: Proceedings of SIGGRAPH '92. University of Washington, Seattle, pp. 71–78.

Kepple, T.M., Sommer, H.J., Lohmann Siegel, K., Stanhope, S.J., 1998. A three-dimensional musculoskeletal database for the lower extremities. Journal of Biomechanics 31 (1), 77–80.

Leardini, A., Chiari, L., Della Croce, U., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. Gait and Posture 21 (2), 212–225.

McPherson, A., Kärrholm, J., Pinskerova, V., Sosna, A., Martelli, S., 2005. Imaging knee motion using MRI, RSA/CT and 3D digitization. Journal of Biomechanics 38, 263–268.

Noonan, K.J., Halliday, S., Browne, R., O'Brien, S., Kayes, K., Feinberg, J., 2003. Interobserver variability of gait analysis in patients with cerebral palsy. Journal of Pediatric Orthopaedics 23 (3), 279–287.

Ramakrishnan, H.K., Kadaba, M.P., 1991. On the estimation of joint kinematics during gait. Journal of Biomechanics 24, 969–977.

Schwartz, M.H., Trost, J.P., Wervey, R.A., 2004. Measurement and management of errors in quantitative gait data. Gait and Posture 20 (2), 196–203.

Soderkvist, I., Wedin, P.A., 1993. Determining the movements of the skeleton using well-configured markers. Journal of Biomechanics 26 (12), 1473–1477.

Stagni, R., Leardini, A., Cappozzo, A., Benedetti, M.G., Cappello, A., 2000. Effects of hip joint centre mislocation on gait analysis results. Journal of Biomechanics 33 (11), 1479–1487.

Van Sint Jan, S., 2007. Color atlas of skeletal landmark definitions. Guidelines for Reproducible Manual and Virtual Palpations. Churchill Livingstone Title.

White, S.C., Yack, H.J., Winter, D.A., 1989. A three-dimensional musculoskeletal model for gait analysis. Anatomical variability estimates. Journal of Biomechanics 22 (8–9), 885–893.

Wright, J.G., 2003. Editorial: pro: interobserver variability of gait analysis. Journal of Pediatric Orthopaedics 23, 288–289.