Functionally oriented and clinically feasible quantitative gait analysis method

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Abstract—*A protocol for clinical gait analysis is described, and data from 30 normal adult female subjects are presented. Extensive application to pathologic subjects has proven to be feasible and sufficiently accurate. The method is based on a particular location and attachment of retro-reflective markers on the body and on a particular arrangement of four TV cameras. A motion analyser measures the 3D coordinates of each marker. A modelling approach, based on individual anthropometric measurements, and a functional approach, based on kinematical considerations, are used to estimate the location of hip, knee, and ankle joint centers and the orientation of the flexion-extension axis of the knee. 3D relative and absolute movements of pelvis and lower limbs are obtained and shown to be consistent with functional anatomy.*

Keywords—*Clinical gait analysis, Locomotory biomechanics, Walking measurements*

Med. Biol. Eng. Comput., 1998, **36**, 179–185

1 Introduction

Clinical usefulness of quantitative gait analysis has been increasingly reported (Sutherland, 1978; DeLuca, 1991; Lee *et al.*, 1992; Creena *et al.*, 1992, Gage, 1993; Camoriano *et al.*, 1995, Frigo *et al.*, 1996*a*; Romanò *et al.*, 1996). Opto-electronic systems are commonly used for these applications, and most of the methods proposed for the calculation of kinematics and kinetics refer to markers attached to the skin (Harris and Wertsch, 1994). Provided the measurement system is sufficiently accurate (and this is guaranteed for many commercially available systems) and appropriate biomechanical models are used, biomechanical functions can be described comprehensively. However, in clinical applications, reliability and accuracy are greatly affected by how much the acquisition system interferes with the patient, and by difficulties arising from bone deformities, uncontrolled movements, use of orthoses and assistive devices. The 'Fatiguability' of the subject must also be considered as a limiting factor for complex protocols to be applicable.

Taking into account these considerations, a protocol for clinical applications of gait analysis has been developed at the Bioengineering Centre in Milan (Pedotti and Frigo, 1992; Rabuffetti *et al.*, 1992). This protocol has been used in a clinical setting at both the Bioengineering Centre in Milan (laboratory S.A.F.Lo.-Servizio di Analisi della Funzionalità Locomotoria) and at the Spaulding Rehabilitation Hospital in Boston. In this work, the methodology and the 3D kinematic data obtained in a normal adult population are presented and discussed.

2 Methods

2.1 *Laboratory configuration and hardware*

The general layout of the laboratory is illustrated in Fig. 1. An optoelectronic system* was used to measure the 3D coordinates of 15 mm hemispherical retro-reflective markers attached to the subjects' body, at a sampling rate of 100 Hz (Ferrigno and Pedotti, 1985; Ferrigno *et al.*, 1990; Borghese and Ferrigno, 1990; Pedotti and Ferrigno, 1995). Four video cameras were used, with two cameras placed posterolaterally on each side of the subject.

The working volume (2 m × 3 m × 1 m) was calibrated using a precision grid. The resultant accuracy was assessed, prior to each subject session, by measuring the distance between two markers connected by a stick that was moved in the whole calibrated volume. This measure was known *a priori* to be 200 mm. The difference between the *a priori* and the measured distance was within 0.2 mm. The markers' coordinates were lowpass filtered in the frequency domain by an algorithm that estimates the optimal cut-off frequency and minimises the residual noise in the signal (D'Amico and Ferrigno, 1990). The resultant cut-off frequency was in the range of 3–7 Hz.

Ground reaction forces were simultaneously collected at 100 Hz sampling rate, using two force plates† staggered along the walkway. The stride temporal phases were obtained from force platform data by analysing the ground reaction vertical component (Stanhope *et al.*, 1990; Rabuffetti and Frigo, 1995).

2.2 *Protocol and subjects*

Preparation of the subjects included the attachment of markers in the following positions (Fig. 2): lower prominence

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First received 26 February 1996 and in final form 10 June 1997

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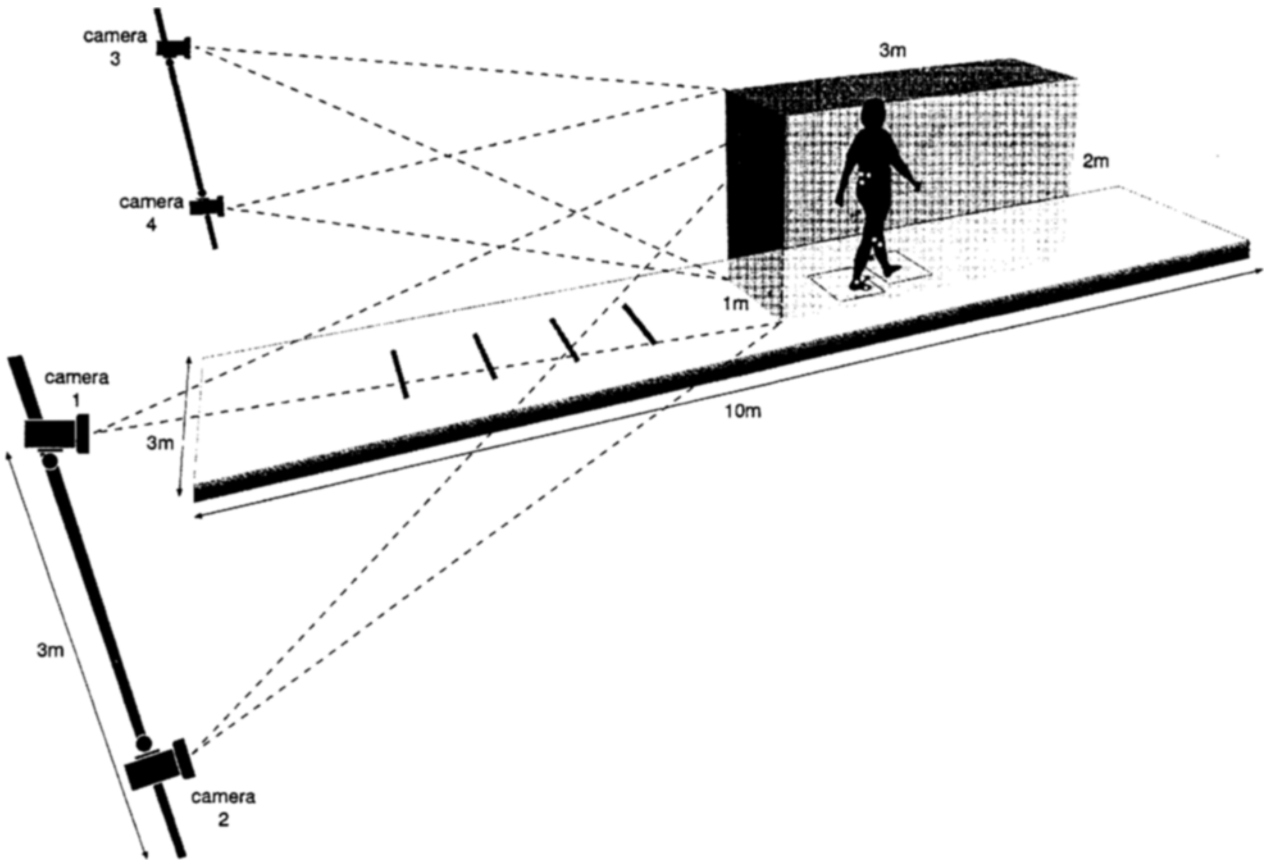

Fig. 1 *General layout of the gait laboratory: four video cameras were used with two cameras placed posterolaterally on each side of the subject; two force plates are staggered along the walkway*

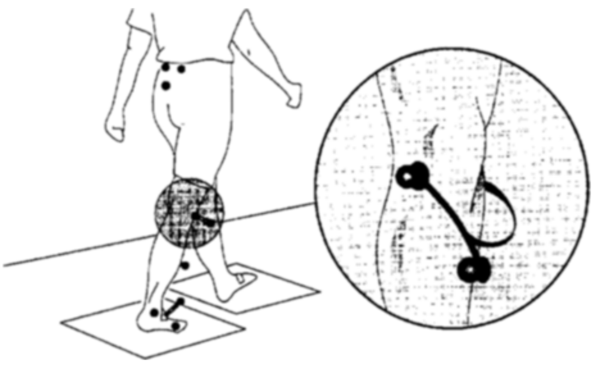

Fig. 2 *Locations of the markers on a subject, placed over the following bony landmarks: lower prominence of the sacrum, posterior superior iliac spines, lateral femoral condyles, lateral malleoli, fifth metatarsal heads; three additional markers attached to the ends of rigid wands are fixed over bilateral lateral femoral condyles, anterior tibial shaft, and forefoot*

of the sacrum (swimsuits were worn and fixed to the skin over the sacrum by double-sided adhesive tape), posterior superior iliac spines, lateral femoral condyles, lateral malleoli, and fifth metatarsal heads. Three additional markers on each lower limbs were attached to the ends of wands rigidly fixed over:

(*a*) the lateral femoral condyles (actually their structure included an attachment to the medial condyles as well, as depicted in Fig. 2 inset)
(*b*) the anterior tibial shaft
(*c*) the forefoot

They are 'the extended markers'.

The following anthropometric measurements were taken for each subject: standing height; body weight; thigh length (distance between the greater trochanter and the lateral femoral condyle); lower leg length (distance between the lateral femoral condyle and the lateral malleolus); foot length (distance between the heel and the great toe); pelvic width (distance between the right and left superior iliac crests); pelvic height (distance between the superior iliac crest and the ischium when seated on a rigid chair); vertical distance between the posterior iliac spine and superior iliac crest (VD); intracondylar distance at the knee; intramalleolar distance at the ankle; and the distance between the first and fifth metatarsal heads. The thickness of the skin (ST) was assumed to have a standard value of 5 mm, but there is a possibility of using different estimates.

Informed consent was obtained from each subject, and the methods were approved by the Spaulding Rehabilitation Hospital Institutional Review Board. Each subject was asked to walk barefoot across the entire length of the walkway at a normal, comfortable speed. Bilateral force plate and kinematic data from five walking trials for each subject were collected. Standing still upright was also analysed and entered in the procedure for the estimation of anatomical landmarks (see below).

For comparison with the data in the literature, data from a population of 30 normal female subjects, without history of gait deviation, musculoskeletal or neurological dysfunction, were analysed and are presented here; demographic data, gait velocity and stride length are reported in Table 1. In addition, more than 300 patients with different forms of locomotive diseases have been analysed with the same protocol. They included 25 hemiplegic patients; 52 children with spina bifida; 15 children with cerebral palsy; 92 rheumatologic and osteo-arthritic patients, 35 subjects with residua of congenital hip

Table 1 Subject demographics of the population involved in the study

	Age, years	Height, m	Mass, kg	Velocity, m s^{-1}	Stride length, m
mean	30	1.62	58	0.79	0.78
SD	6	0.06	6	0.06	0.05

dysplasia; 35 amputees with above and below knee prostheses; 24 subjects with foot deformities; 12 subjects with knee ligament diseases; 22 subjects with spinal deformities; and various other different forms of locomotive diseases.

General considerations about the application of our protocol to these pathologies are reported below.

3 Kinematic analysis

3.1 *Pelvis frame and hip joint centre*

Using the markers located on the pelvis (posterior superior iliac spine and lower prominence of the sacrum) a Cartesian system of axes was defined as follows (Fig. 2): the medial/lateral axis (z axis) parallel to the line joining the two posterior superior iliac spines; the anterior/posterior axis (x axis) perpendicular to the plane on which the three pelvis markers lie; the vertical axis (y axis) perpendicular to the previous two axes. The origin was located midway between the two posterior superior iliac spines (Fig. 3*a*).

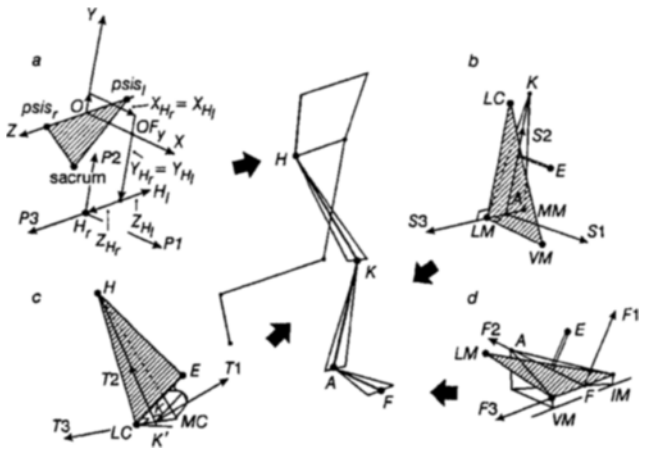

Fig. 3 *(a) Pelvis anatomical frame and hip joint coordinate system: X, Y and Z= pelvis anatomical axes; H_r= right hip joint centre; H_l= left hip joint centre; OF_y= measured y distance between the posterior iliac spines and the upper iliac crest; P1, P2, and P3 = resultant hip anatomical axes; (b) knee anatomical reference frame: LC = lateral femoral condyle; MC = medial femoral condyle; E = extended knee marker; H = hip joint centre; T1, T2, and T3 = resultant anatomical reference axes, with the origin K = knee joint centre; (c) ankle anatomical frame: LM = lateral malleolus; MM = medial malleolus; LC = lateral femoral condyle; VM = fifth metatarsal head; K = knee joint center; E = shank extended marker; S1, S2, and S3 = anatomical reference axes, with the origin A = ankle joint centre: (d) foot anatomical reference frame: LM = lateral malleolus; VM = fifth metatarsal head; IM = first metatarsal head; E = foot extended marker; A = ankle joint centre; F1, F2, and F3 = local anatomical axes, with F = forefoot midpoint*

The local coordinates of the hip joint centres (RH = right hip centre and LH = left) were estimated bilaterally, in this reference frame:

$$RH_X = LH_X = R_X \cdot PH + ST \qquad (1)$$

$$RH_Y = LH_Y = -R_y \cdot PH + VD \qquad (2)$$

$$RH_Z = -LH_Z = R_z \cdot (PW/2 - ST) \qquad (3)$$

R_x, R_y are the ratios between the X (or Y) hip coordinates and the pelvis height PH. R_z is the ratio between the Z hip coordinates and the pelvis width PW. R_y and R_z were predicted through regression equations obtained by analysing frontal radiographs of 175 subjects of both sexes, without any morphological abnormality of the pelvis, ranging from 12 to 76 years of age (PERRAZZO and UGLIETTI, 1992). On averatge $R_y = 0.69$ ± 0.03 SD), $R_z = 0.63(\pm 0.07$ SD) for the males ($N = 74$), and $R_y = 0.71(\pm 0.03$ SD), $R_z = 0.63$ (± 0.04 SD) for the females ($N = 101$). A rough estimation for R_x yielded a value of 0.46, but this was without statistical meaning, being based on a few images and on a measurement of a skeleton model (in practice sagittal radiographs of pelvis are rarely available).

Thus, an additional procedure was implemented to correct the anterior/posterior coordinate estimation. The algorithm was aimed at minimising and the standard deviation (SD) of the distance between the hip centre positions ($\vec{H}$) and the knee marker positions ($\vec{K}$) all along the stride cycle:

$$SD(dx) = \sqrt{\sum_{n=1}^{N}\left(|\vec{H}_n + dx \cdot \vec{i}_n - \vec{K}_n| - \sum_{n=1}^{N}|\vec{H}_n + dx \cdot \vec{i}_n - \vec{K}_n|/N\right)^2/(N-1)} \qquad (4)$$

where dx is the unknown antero-posterior displacement; $\vec{i}$ is the x unit vector; n is the nth sample and N is the number of available samples. The solution of the minimisation algorithm is the value $d\tilde{x}$ that makes:

$$SD(d\tilde{x}) \leq SD(dx), \forall dx \qquad (5)$$

The final estimation of the hip centre position was thus obtained as

$$\vec{H}'_n = \vec{H}_n + dx \cdot \vec{i}_n \qquad (6)$$

At the end of the minimisation procedure, the target function *SD* was always considerably reduced, and in many cases achieved a value of less than 3 mm.

A hip joint orthogonal frame was defined with its origin in the hip joint centre (*H*) and the three axes parallel to the previously defined pelvis frame: the anterior/posterior axis parallel to the x axis (*P*1), the longitudinal axis parallel to the y axis (*P*2), and the medial/lateral axis parallel to the z axis (*P*3).

3.2 *Thigh frame and knee joint centre*

The geometry of the knee wand was such that the line between the two attachment points (medial and lateral femoral condyles) was perpendicular to the line between the extended knee marker and the lateral femoral condyle marker; during walking, artificial movements of the wand occurred primarily in the vertical plane, and so the rotation of the thigh around its longitudinal axis was relatively unaffected by these artifacts. The medial/lateral knee axis was thus defined as the perpendicular to both the longitudinal axis of the thigh and the line of the knee wand.

As the intracondylar distance was measured, the 3D coordinates of the knee joint centre (*K*) were obtained by moving from the lateral femoral marker along the medial/lateral axis for half of that distance. The anterior/posterior axis ($T1$) was computed as the perpendicular to the plane identified by the hip and knee joint centres and lateral femoral condyle, and directed anteriorly. The longitudinal axis ($T2$) was the line from the knee to the hip joint centres and was directed cranially. The medial/lateral axis ($T3$) was perpendicular to both axes and directed to the right.

Two independent algorithms were applied to extract information from the kinematics and to improve the estimation of the joint centre and flexion/extension axis.

First, an optimisation algorithm was applied to minimise the SD of the distance between the knee joint centre and the ankle marker. As well as the hip joint correction, only the anterior/posterior coordinate of the knee was made to change.

After this repositioning, the distance between the knee centre and the ankle marker had, in many cases, an SD of less than 2.5 mm.

The second functional correction was based on a definition of the knee flexion/extension plane. The ankle marker was referred to the thigh reference frame, and its trajectory, described all along the recorded movement, was projected on the plane perpendicular to the thigh longitudinal axis (Fig. 4). The best regression line was computed, which minimises the RMS distance from the trajectory points in a range of knee flexion between 10° and 30°, where the trajectory looked to be straight. The knee flexion axis was defined as the perpendicular to that line and was assumed as the thigh medial/lateral axis. Consequently, the anterior/posterior axis was redefined as the perpendicular to this axis and to the thigh longitudinal axis.

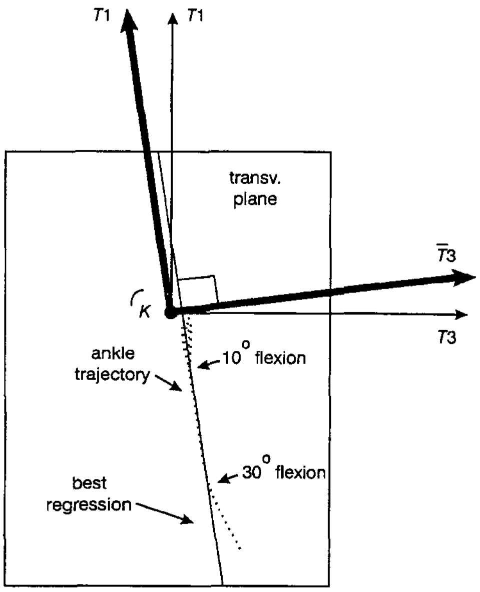

Fig. 4 *Anatomical-functional frame at the knee; definition of the flexion/extension knee axis based on functional concepts; T1 = anterior-posterior thigh axis; T3 = medial-lateral axis; the ankle joint trajectory is projected on the plane perpendicular to the thigh longitudinal axis; a departure from a straight line reflects adduction/abduction movemement; the best regression line of this trajectory is computed within a joint angle excursion of 10°–30°, where it is almost linear; a functional flexion-extension axis is defined as the perpendicular to this regression line; this is at a certain angle with the previously defined lateral axis; this offset is used to correct the orientation of the T1 and T3 axes, and to make the lateral axis coincidental with the functional axis*

3.3 *Shank frame and ankle joint centre*

The centre of the ankle joint was estimated when the subject was standing in a upright position, by first defining a plane on which the lateral malleolus, the lateral femoral condyle and the fifth metatarsal head lie. The relative location of the ankle joint centre was then estimated by translating the lateral malleolus marker by half of the measured intramalleolar distance perpendicularly to this predefined plane (Fig. 3*c*).

The anatomical reference frame of the shank was defined by an anterior/posterior axis ($S1$) perpendicular to the plane identified by the knee joint centre, the ankle joint centre and the lateral malleolus; by a longitudinal axis ($S2$) joining the ankle joint centre to the knee joint centre; and by a transverse axis ($S3$) perpendicular to the other two axes. The origin was assumed to be the ankle joint centre (A).

3.4 *Foot frame*

The distal extremity of the foot segment was defined by a point in the middle of the line connecting the first and the fifth metatarsal heads (forefoot mid-point). Its identification was based on the measurement of the distance between the first and the fifth metatarsal heads. This assumes that in an upright standing posture:

(*a*) these two points could reach the ground, and thus they were at the same level (as shown later, this assumption consists of defining a foot medial/lateral axis horizontal in a standing position).

(*b*) the line which connects these two points is perpendicular to the line which connects the fifth metatarsal head and the ankle joint centre (the fact that the first metatarsal head could be in a slightly different position than assumed here is in practice not significant, as only the apparent foot length would be affected).

The local anatomical frame was defined as follows: the anterior/posterior axis ($F1$) perpendicular to the plane which includes the centre of the ankle joint, the forefoot midpoint, and the fifth metatarsal head; the longitudiinal axis ($F2$) connecting the forefoot midpoint and the ankle joint centre; and the lateral axis ($F3$) perpendicular to the other axes. The origin F was placed in the forefoot midpoint (Fig. 3*d*).

3.5 *Relative joint angles*

The joint angles, which represent the relative orientation of the different local reference frames, were computed according to the Euler definition (GROOD *et al.*, (1983). For the pelvis, the angles were computed with respect to the absolute laboratory frame, where X is the progression direction, Y is the vertical axis and Z is the transverse axis.

The three Euler angles for each segment or joint describe joint flexion-extension, or pelvis tilt, as the rotation around the transversal proximal axis; joint internal-external rotation, or pelvis horizontal rotation, as the rotation around the distal longitudinal axis; joint adduction-abduction, or pelvis obliquity, as the rotation around the floating axis, the axis perpendicular to the distal longitudinal and to the proximal transverse axes.

4 Results

Consistent with the aims of this work, the feasibility of the analysis in a clinical environment has to be considered a result. The methods presented here have been applied to more than 300 patients (adults and children) with a broad spectrum of neurologic and orthopaedic diagnoses, many of whom have been unable to walk without an assistive device or the assistance of another person.

Preparation of the subjects required approximately 15 min. The time for the acquisition of ten trials depended mainly on the ability of the subject to achieve a well reproducible steady-state ambulation, which facilitated identifying a starting point for arriving correctly on the force plate with one foot. Walking velocity was, of course, another factor, as well as the 'fatiguability' of the patients, who could require frequent rest periods. In no case did the acquisition time exceed 90 min.

Visualisation of markers throughout the gait cycle was without difficulty. In particular, free arm swing during walking (a component of comfortable walking) did not interfere with the markers' visualisation when subjects were walking fast (normal subjects), when they were walking slowly with their arms hanging in a fixed position, nor when they were handing an assistive device. Only in a few cases with a relatively high internal rotation of the foot and knee joint surface, the markers located on the fifth metatarsal head and the lateral femoral condyle, respectively, had to be raised from the skin surface by a 3–4 mm high support to make them more easily detectable by the video system. (The height of the support was, of course, taken into account when the knee joint centre and the forefoot mid-point were computed.)

In many cases, electromyography through surface electrodes was used to analyse the activity of the main lower limb muscles. A portable telemetric device was attached anteriorly to the subject by belts, and a maximum of eight pre-amplified electrodes, located on the lower limb, were connected to it by wires. This additional arrangement did not interfere with the kinematic protocol.

Data from our normal control population are reported in Table 1 (demographics and global gait parameters) and Fig. 5 (body segments and joint kinematics). In each graph in Fig. 5, the curves from each trial and for each subject (total for 300 trials) were time normalised for the gait cycle and averaged. Zero value refers to when the local frames of the distal and proximal segment are aligned.

The percentage of stance phase duration, marked on the graphs, was on average 61.77 (mean) $\pm$ 1.63 SD). The mean SD (MSD) along the stride cycle and the MSD divided by the peak-to-peak amplitude of the average curve (normalised MSD) are displayed in Table 2. The MSD ranged from 2.36° (ankle dorsi-plantar flexion) to 5.82° (hip internal-external rotation), and the NMSD from 6% (knee flexion-extension) to 70% (pelvis anterior-posterior tilt).

5 Discussion

The time courses of the joint angular motion are in agreement with those provided in the literature (MURRAY *et al.*,

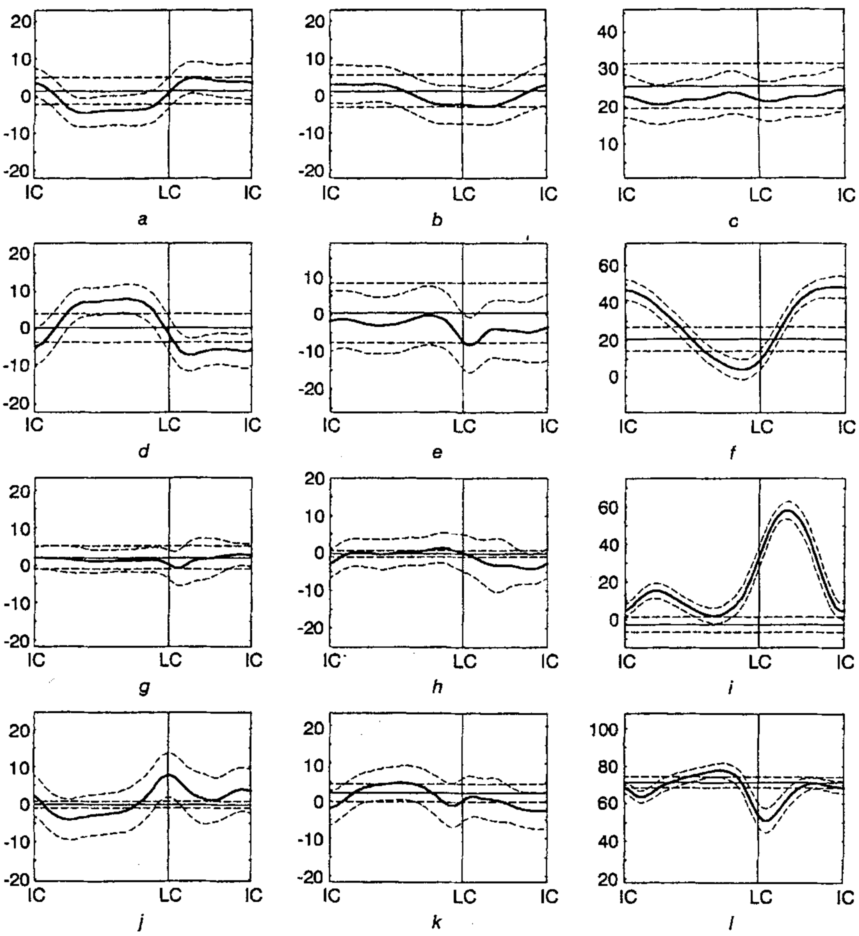

Fig. 5 *Pelvis, hip, knee and ankle kinematic data (angles in degrees); the x axes represents percent gait cycle; IC = initial contact, LC = last contact; solid line = mean; dashed line ± 1 SD; horizontal reference lines = mean ± 1 SD of angles measured in standing posture; labels refer to the meaning of the positive and negative values, respectively; (a) pelvis obliquity (b) pelvis horizontal rotation (c) pelvis anterior-posterior tilt (d) hip adduction/abduction (e) hip int./ext. rotation (f) hip flexion/extension (g) knee adduction/abduction (h) knee int./ext. rotation (i) knee flexian/extension (j) ankle adduction/abduction (k) ankle pronation/supination (l) ankle dorsi/plantar flexion*

Table 2 *Kinematics: overall and individual parameters*

Segment joint	Angular variable	MSD, ° mean (range)	NMSD, % mean (range)	Minimum peak, ° mean (SD)	Maximum peak, ° mean (SD)
Pelvis	anterior/posterior tilt	2.55 (0.68–8.65)	70 (13–265)	10.61 (5.19)	25.68 (5.82)
	obliquity	3.11 (1.22–8.82)	34 (10–108)	−5.17 (4.04)	5.84 (4.25)
	horizontal rotation	4.23 (1.35–8.12)	65 (18–176)	−4.31 (4.87)	5.43 (5.08)
Hip	flexion/extension	3.12 (1.66–7.10)	7 (324)	3.66 (5.48)	49.63
	adduction/abduction	3.25 (0.72–7.57)	21 (5–50)	−8.52 (4.26)	9.15 (3.81)
	internal/external rotation	5.82 (2.50–10.16)	67 (17–213)	−10.47 (7.37)	2.59 (7.55)
Knee	flexion/extension	3.21 (1.53–8.34)	6 (3–18)	0.79 (3.27)	58.81 (3.73)
	adduction/abduction	2.54 (1.05–4.91)	66 (15–149)	−2.20 (4.02)	5.10 (3.88)
	internal/external rotation	3.76 (2.00–7.69)	49 (20–80)	−7.08 (6.14)	3.51 (3.64)
Ankle	dorsi/plantar flexion	2.36 (1.36–4.61)	9 (4–18)	50.21 (6.39)	78.58 (3.40)
	adduction/abduction	3.81 (1.68–7.49)	33 (13–73)	−5.45 (5.18)	9.03 (6.47)
	pronation/supination	3.32 (1.33–7.12)	37 (13–82)	−4.34 (4.88)	6.76 (5.24)

1964; 1970; SUTHERLAND and HAGY, 1972; WINTER, 1984; KADABA *et al.*, 1989; 1990; OUNPUU *et al.*, 1991; LAFORTUNE *et al.*, 1992; OBERG *et al.*, 1994) and exhibit relatively low SDs. In particular, the internal/exernal rotations and the adduction/abduction angles historically have shown large oscillations, but show little oscillation in the results presented here and are consistent with the functional anatomy of the joints. Part of the variability is to be ascribed to different individual walking patterns and/or different postural attitudes. They would be reduced if the angles were referred to each subject's standing values. The decision was made to leave this source of variability in the data because, when looking at pathological cases, it is quite likely that abnormal standing posture occurs together with abnormal gait, and both these abnormalities deserve attention.

In comparing the data from this study to previous work, most documentation concerns data in the sagittal plane only and the use of different models and angle definitions. Those works which may be taken as reference, insofar as they use a similar approach, are those of Ounpuu and Kadaba (OUNPUU *et al.*, 1991; KADABA *et al.*, 1989; 1990). Unfortunately, they present only a subset of the joint angles presented here, and the first study of OUNPUU *et al.*, (1991), is concerned primarily with data obtained in a distinct population of children. Nonetheless, the timings and variabilities of the motions which do overlap are similar.

Another important reference is the work of LAFORTUNE *et al.*, (1992), who measured knee motion in the transverse, frontal and sagittal planes by direct intra-cortical pin placement. It is worth noticing that the angles reported (particularly the interal/external and adduction/abduction angles) in four out of five of the subjects presented are very similar to those obtained in our study, both as time course and amplitude are considered. After careful inspection, an offset of about 5° can be appreciated that can be easily explained by a different definition of the longitudinal axis of the thigh. The fifth of the subjects presented in LA FORTUNE *et al.* (1992) has a surprisingly opposite behaviour as far as the adduction/abduction angle is concerned, which was not explained. However, we did not observe any similar discrepancy within our test population.

The concepts of technical and anatomical frames have been primarily theoreticised by CAPPOZZO (1984, 1991) and CAPPOZZO *et al.* (1995) and include the concept of anatomical calibration. Having adopted palpable bony prominences and wands of known geometry to support the technical landmarks, in our procedure the anatomical calibration reduces itself to defining a set of geometrical relationships and anatomical considerations. This approach, which could appear excessively crude at glance, has been complemented by refinements in the joint centre estimation for the hip and knee joints to match kinematical principles. That this functional correction improves the estimation of the hip and knee joint centres was demonstrated by a decrease in the SD of the thigh and shank lengths computed along the stride, which in many cases achieved a value of few millmetres. The orientation of the medial/lateral knee joint axis was also corrected on the basis of shank–thigh relative motion. Although this correction improves the estimation of the knee flexion/extension axis, reducing the cross-talk between flexion/extension and the other two joint angles (sensitivity analysis has been dealt with by RAMAKRISHNAN and KADABA, 1991), it does not add any special requirement in the acquisition phase. Similarly, the ankle and mid-forefoot positions are based on hypotheses that dramatically reduce the need of external measurements and/or special manoeuvres.

That visualisation of the markers with free arm swinging was problem-free was undoubtedly a result of the markers being placed in relatively posterior positions on the subject; in other protocols the problem of markers' obscuration at the hip is taken into consideration seriously (DAVIS III *et al.*, 1991) and sometimes the subjects are asked to walk with crossed arms. This relative posterior placement of the markers is a particular advantage for the many patients with moderately to severely disabled gait who keep their arms fairly close to their sides without much motion during walking. The posterior placement is of a similar advantage when an assistive device or the assistance of another person is necessary. Furthermore, when compensatory trunk movements are of interest, the spinal processes and shoulders can be additionally marked and detected by the same TV cameras. If required, electromyographic apparatus can be applied (provided it is relatively anterior to the patient) without risk of interference with the camera visualisation of the markers.

Acknowledgments—This work has been partially supported by the Italian Ministry of Health Service, by the EU TREAD Project (ERBCHRXCT930205) in the frame of HCM and by the Ellison Foundation of Boston, Massachusetts.

The authors would like to thank William R. Cox for his technical assistance.

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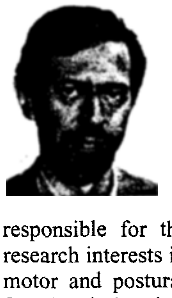
Author's biography



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