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Measurement of Lower Extremity Kinematics During Level Walking

M. P. Kadaba, H. K. Ramakrishnan, and M. E. Wootten

Orthopaedic Engineering and Research Center, Helen Hayes Hospital, West Haverstraw, New York, U.S.A.

Summary: A simple external marker system and algorithms for computing lower extremity joint angle motion during level walking were developed and implemented on a computer-aided video motion analysis system (VICON). The concept of embedded axes and Euler rotation angles was used to define the three-dimensional joint angle motion based on a set of body surface markers. Gait analysis was peformed on 40 normal young adults three times on three different test days at least 1 week apart using the marker system. Angular motion of the hip, knee, and ankle joints and of the pelvis were obtained throughout a gait cycle utilizing the three-dimensional trajectories of markers. The effect of uncertainties in defining the embedded axis on joint angles was demonstrated using sensitivity analysis. The errors in the estimation of joint angle motion were quantified with respect to the degree of error in the construction of embedded axes. The limitations of the model and the marker system in evaluating pathologic gait are discussed. The relatively small number of body surface markers used in the system render it easy to implement for use in routine clinical gait evaluations. Additionally, data presented in this paper should be a useful reference for describing and comparing pathologic gait patterns. **Key Words:** Gait analysis—Joint angles—Gait parameters—Biomechanical model—Sensitivity analysis.

Quantitative gait analysis is an important clinical tool for quantifying normal and pathological patterns of locomotion, and has been shown to be useful for prescription of treatment as well as in the evaluation of the results of such treatment (1,6,16,17). Typically, data acquired during a clinical gait analysis include relative positions and orientations of body segments, foot–floor reaction forces, temporal–distance parameters, and phasic activity of muscles of the lower extremities. Several practical methods in current use provide relative orientation of segments either directly or as a derived parameter from measurements of relative position of segments. For example, electrogoniometers (5,10–12,24) have been used to record instantaneously the three-dimensional joint rotation of lower extremity. Accelerometers have also been used for indirect measurement of angular displacements of limbs (8,14,20). Interrupted light photography has been used to derive sagittal plane motion patterns (15,18) by monitoring reflective markers placed on key anatomical locations. Cine film photography (15,23) has been utilized to quantify the motion patterns in three dimensions. Modern computer-aided systems such as VICON (4) and SELSPOT (2) provide accurate three-dimensional spatial positions of reflective skin (surface) markers placed on key anatomical sites on the lower extremities. From these positional data, the relative angular rotation of the individual body segments are derived using analytical techniques based on a biomechanical model of the lower extremity.

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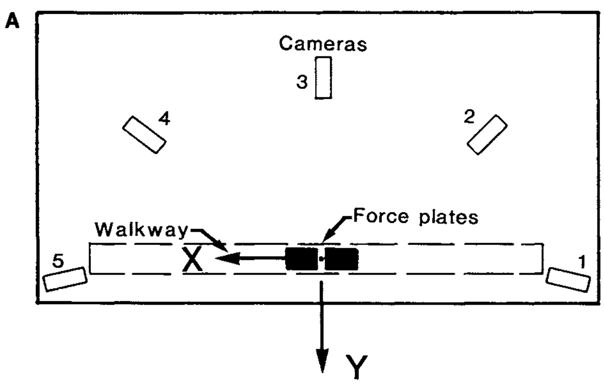
Address correspondence and reprint requests to Dr. M. P. Kadaba at Orthopaedic Engineering and Research Center, Helen Hayes Hospital, Rt. 9W, West Haverstraw, NY 10993, U.S.A.

Sutherland et al. (22) and Murray et al. (15) utilized the coordinates of key anatomical points, obtained from a cine film system, to compute joint angle motion using planar definitions. A nonorthogonal joint coordinate system with the associated Cardan angles was proposed by Grood and Suntay (7) and Suntay et al. (21) for describing the motion of knee joint. Euler angle definitions were used by Chao (5) for the measurement of knee joint motion using a triaxial goniometer. Tylkowski et al. (25) also utilized Euler angle definitions to compute hip joint motion from trajectories of body surface markers derived from cine film. Cappozzo (4) developed a system to compute joint angle motion based on the concept of Cardan angles. Antonsson (2), using the concept of a screw axis (helical axis) of motion, devised a method to compute limb rotations from limb orientation data recorded using an optoelectronic system. The concept of helical axis was also utilized by Shiavi et al. (19) in the measurement and analysis of knee joint motion using a six degrees of freedom goniometer.

With the advent of computer-aided video motion analysis systems, clinical gait laboratories are proliferating rapidly. In spite of the advantages of computer-aided video motion analysis over cine film systems, problems with tracking closely spaced markers make measurement of joint angle motion labor intensive. Therefore, for routine clinical use, the external marker system must be simple and yet rigorous enough to define the relative motion of the rigid body segments in three dimensions. Despite the vast literature related to lower extremity kinematics, a detailed description of the external marker system for computing the motion at the pelvis, hip, knee, and ankle joints during gait is not available. The definition of the axes or planes about which the limb rotations take place as well as the methods to construct these axes and planes based on body surface markers are also lacking. In this paper, we present a simple marker system that can be easily implemented for routine clinical gait evaluations. We describe in detail the definition of axes and planes as well as the techniques for constructing them. We present the results of a sensitivity analysis designed to demonstrate the limitations associated with the joint angle measurement system.

DEFINITION OF PARAMETERS

In gait analysis, human body segments are modeled as rigid bodies and the relative rotation is as-



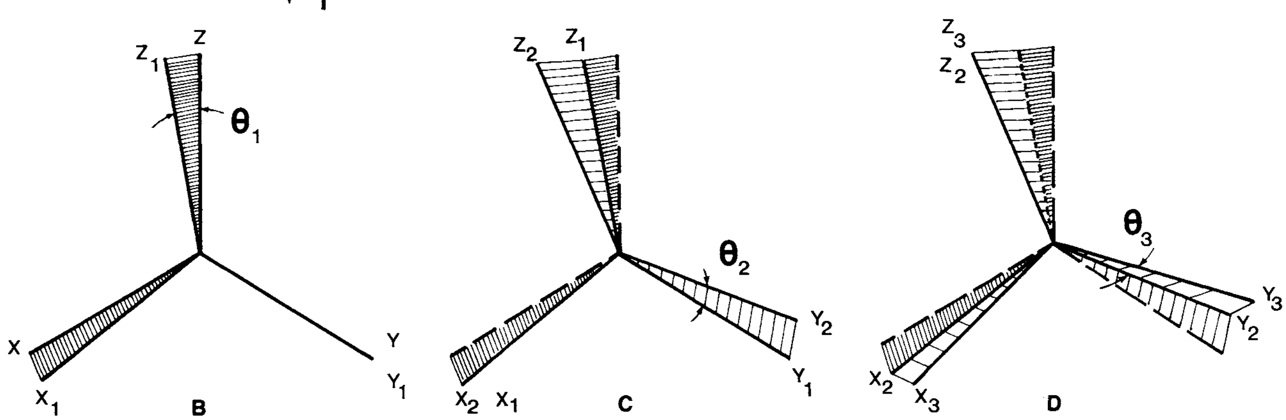


FIG. 1. **(A)** Camera configuration and absolute reference system in a horizontal plane. Absolute *Z* direction (not shown) is perpendicular to both *X* and *Y* directions pointing away from the paper. **(B)** Rotation about *Y* axis. θ_1, pelvic tilt—hip flexion/extension—knee flexion/extension—ankle plantar/dorsiflexion. **(C)** Rotation about X_1 axis. θ_2, pelvic obliquity—hip AB/adduction—knee varus/valgus. **(D)** Rotation about Z_2 axis. θ_3, pelvic rotation—hip rotation—knee rotation—foot rotation.

sumed to take place about a fixed point in the proximal segment, which is considered to be the center of the joint. Euler angles have been successfully applied to describe relative rotations of one segment with respect to another reference segment in a three-dimensional space (5). These angles are defined as a set of three finite rotations assumed to take place in sequence to achieve the final orientation from a reference orientation. A better method for describing joint angle motion would be the orthopaedic angles as defined by Lewis and Lew (13). Essentially, orthopaedic angles are the same as Euler angles but they are defined according to the clinical terms such as flexion, abduction, etc.

In order to calculate the relative Euler angles, it is necessary to define a set of orthogonal embedded axes both in the moving segment as well as in the reference segment. In the absolute orthogonal reference system (X, Y, and Z in Fig. 1A), defined here, the X axis is along the walkway, the Z axis is the vertical pointing upwards, and the Y axis is perpendicular to both X and Z directions, forming a right-handed cartesian coordinate system. For the pelvis, the reference axes are the absolute coordinate axes. For the thigh segment, the reference axes are the pelvic-embedded axes. For the shank, the references are thigh-embedded axes, and for the foot, the references are the shank-embedded axes. The orthopedic angles describing the lower extremity limb rotations are defined as follows: When a particular segment rotates in the right-handed direction through an angle θ_1 about the reference Y axis, the resulting angles with reference to a four-segment lower extremity model are pelvic tilt (upward), hip extension, knee flexion, and ankle plantarflexion. If a left-handed rotation takes place, then the resultant angles are pelvic tilt (downward), hip flexion, knee extension, and ankle dorsiflexion. At this point, the new orientation of the embedded axes of the moving segment is denoted by X_1, Y_1, and Z_1 (Fig. 1B). When the segment rotates in the right-handed direction through an angle θ_2 about the rotated X_1 axis, the rotations are defined as pelvic obliquity, hip ab-/adduction, and knee varus/valgus. This rotation is not considered for the ankle and the reasons will be described later. The new orientation of the axes of the moving segment is now denoted by X_2, Y_2, and Z_2 (Fig. 1C). When the segment further rotates through an angle θ_3 about the new Z_2 axis to achieve its final position, the angular displacements are now defined as pelvic rotation, hip rotation, knee rotation, and ankle rotation. This final orientation of the embedded axes is X_3, Y_3, and Z_3 (Fig. 1D).

MARKER SYSTEM AND EMBEDDED AXES

The marker system described here was designed with a minimum of markers to simplify the identification of marker trajectories. The position of markers (2 cm in diameter, weighing 4.4 g, developed in this study) is shown in Fig. 2 and was selected to satisfy the rigid body assumption as well as other practical requirements described by Cappazzo (4). Two markers are placed on the right and left anterior superior iliac spines (ASIS). One other marker is placed on a stick 10 cm long extending from the top of the sacrum (L4–L5) and in the spinal plane. It is stabilized by a flexible triangular plate attached to the body with an elastic belt. Four other markers are placed on the following locations of the particular limb under consideration: greater trochanter, directly lateral to the estimated average axis of rotation of the knee joint, lateral malleolus, and space between the second and third metatarsal heads.

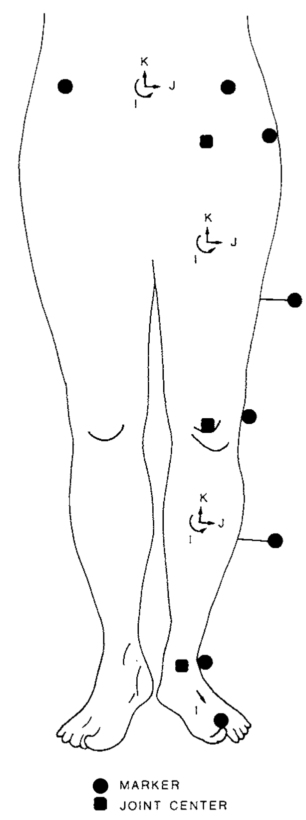


FIG. 2. Marker configuration and embedded coordinate systems.

One cuff is positioned on the midthigh and another on the midshank sufficiently distal to the hip and knee joints to avoid interference during walking. Wands, 7 cm long, with markers at the tip are attached to these cuffs. The cuffs are aligned laterally with the long axis of bones to reflect the neutral rotation angles while standing in a normal position. The axes of the wands are also aligned such that they are in line with the flexion–extension axis of the corresponding distal segment.

An empirical relation, based on a pelvic radiograph study (J. Gage, S. Tashman, personal communication, 1985), is used to estimate the location of the hip joint center relative to the ASIS location and pelvic orientations. In this method, the X,Y,Z coordinate distances of the hip center from the ASIS marker are calculated as a function of the leg length. The location of the hip joint center can also be computed using the distance between the two ASISs as the independent variable (3). The knee center is assumed to lie in the plane defined by the knee marker, thigh-wand marker, and hip joint center, halfway between the femoral condyles. In a similar way, the ankle center is assumed to fall in the plane defined by the ankle marker, the knee center, and the shank-wand marker, and located halfway between the malleoli.

The three-dimensional coordinates of the following points in the absolute reference system are used to calculate the embedded coordinate systems: sacral wand tip, right and left ASIS markers, hip center, knee center, ankle center, thigh-wand tip, shank-wand tip, and foot marker. The embedded coordinates are represented by three orthogonal unit vectors **I**, **J**, and **K** along the embedded X, Y, and Z axes, respectively. For the pelvic coordinates, **J** is the unit vector along the line from the right ASIS to the left ASIS marker. The unit vector **I** is perpendicular to **J**, pointing forward, and is in the plane defined by both ASIS and sacral markers. The third unit vector **K** is perpendicular to both **I** and **J**, defining a right-handed cartesian coordinate system.

For the thigh, the unit vector **K** is in the direction from knee center to hip center. The second unit vector **J** is in the plane defined by the hip center, knee center, and the thigh-wand marker in an orientation perpendicular to the unit vector **K** and points to the subject's left side. The third vector **I** is calculated from the cross product of **J** and **K**. The construction of the shank unit vectors is identical to the thigh unit vectors with the knee center, ankle center, and shank wand, replacing the hip center, knee center, and thigh wand. Since only two markers are used on the foot, only two angular motions can be derived for the ankle. Therefore, only one unit vector is required to compute the foot orientation. This is calculated from the line segment joining the ankle center and the marker at the foot (between the second and third metatarsal heads).

Since the orthopedic angles specify the relative orientation of the distal moving segment with respect to the proximal reference frames, the corresponding rotational matrix can be derived in terms of these angles. Let the unit vectors of the proximal reference frame in the absolute reference system be represented by **I**, **J**, and **K**, and the unit vectors in the distal-embedded system of the moving segment be $\mathbf{I}_3$, $\mathbf{J}_3$, and $\mathbf{K}_3$. Then the following relationship can be easily derived based on orthopedic angles θ_1, θ_2, and θ_3 defined previously for the pelvis, hip, and knee:

$$\begin{bmatrix} \mathbf{I}_3 \\ \mathbf{J}_3 \\ \mathbf{K}_3 \end{bmatrix} = \begin{bmatrix} C1 * C3 + S1 * S2 * S3 & C2 * S3 & -S1 * C3 + C1 * S2 * S3 \\ -C1 * S3 + S1 * S2 * C3 & C2 * C3 & S1 * S3 + C1 * S2 * C3 \\ S1 * C2 & -S2 & C1 * C2 \end{bmatrix} \begin{bmatrix} \mathbf{I} \\ \mathbf{J} \\ \mathbf{K} \end{bmatrix} \quad (1)$$

Here C1 refers to the cosine of angle θ_1 and S1 refers to the sine of angle θ_1, and similar notations apply to other terms.

From this, the rotational angles can be calculated as shown below:

$$\begin{aligned} \theta_2 &= \arcsin(-\mathbf{K}_3 \cdot \mathbf{J}) \\ \theta_1 &= \arcsin[(\mathbf{K}_3 \cdot \mathbf{I})/\cos(\theta_2)] \\ \theta_3 &= \arcsin[(\mathbf{I}_3 \cdot \mathbf{J})/\cos(\theta_2)] \end{aligned} \quad (2)$$

For the ankle joint, the direction cosine matrix relating the foot frame and shank frame may be derived based on two orthopedic angles θ_1 and θ_3 as

$$\begin{bmatrix} \mathbf{I}_3 \\ \mathbf{J}_3 \\ \mathbf{K}_3 \end{bmatrix} = \begin{bmatrix} C1 * C3 & S3 & -S1 * C3 \\ -C1 * S3 & C3 & S1 * S3 \\ S1 & 0 & C1 \end{bmatrix} \begin{bmatrix} \mathbf{I} \\ \mathbf{J} \\ \mathbf{K} \end{bmatrix} \quad (3)$$

The rotational angles of the foot can now be calculated as

$$\theta_3 = \arcsin(\mathbf{I}_3 \cdot \mathbf{J})$$
$$\theta_1 = \arcsin(\mathbf{K}_3 \cdot \mathbf{I}). \quad (4)$$

In deriving the above equations, an assumption is made regarding the sequence of rotations in three dimensions. At each of the joints, flexion–extension is assumed as the first rotation since the major motion occurs in this plane. Ab-/adduction is assumed to take place next in sequence about a rotated axis. Finally, internal–external rotation is assumed to take place next about the third rotated axis.

METHODS AND MATERIALS

Motion analysis was performed using a computer-aided video motion analysis system with five infrared cameras (VICON) under the control of a computer (DEC PDP 11/34). The results of three-dimensional accuracy and resolution (static and dynamic) of the system showed that the system has a composite accuracy of ±3 mm and a resolution of ±2 mm in each of the three coordinate directions (9). Foot contact patterns were recorded using pressure-sensitive foot switches (developed at Rancho Los Amigos Hospital) attached to the heel, first and fifth metatarsals, and great toe of each foot.

A group of 40 normal healthy subjects (age range of 18–40 years, 28 males and 12 females) with no previous history of musculoskeletal problems was evaluated. The subjects were evaluated on three different test days at least 1 week apart in order to assess the repeatability of motion data (27). Prior to recording the gait parameters, the height, weight, lower limb length, knee width, and ankle width of each subject were measured. After a brief orientation session, the subjects were asked to walk at their natural speed along the walkway to assess the individual's free walking speed. Subsequent to the practice session, four sets of gait data were collected over a 3 m portion of the 9 m walkway. One more set of data corresponding to the standing position (static data) were also recorded, in order to correct for any misalignment of the wand markers. These procedures were repeated for each of the lower limbs.

Data Analysis

Gait parameters (velocity, cadence, single stance time, etc.) were calculated for each run using foot switch data. The beginning and end of gait cycles were obtained from foot switch signals. A five point window (Hanning) with weighing coefficients 1, 3, 4, 3, and 1 was used for smoothing raw three-dimensional marker trajectories before computing the joint angle motion. The gait cycles were extended or compressed in time to yield a normalized gait cycle of 64 equally spaced data points. All gait cycles were expressed as a function of a unit (100%) cycle length irrespective of the actual time for a stride. Three out of four cycles of data from each test session were selected and the mean and standard deviation for each joint angle pattern were computed for each subject. Since the subjects were evaluated on three different days, a total of nine data sets for a particular subject were averaged, yielding a representative pattern of motion data for that individual. Both right and left limb data were grouped separately. Further, the mean and standard deviations at each point of the gait cycle were determined by averaging the mean joint angle data of all of the subjects.

Sensitivity Analysis

Accurate definition of the embedded axes is essential to reliable estimation of three-dimensional motion at each joint. In the present Eulerian system, the definition of the flexion–extension axis as well as the rotation axis is crucial. The flexion–extension axis, about which the first rotation in the Euler sequence is assumed to take place, is defined with respect to body surface markers. If the actual flexion–extension motion does not take place about this axis, then the computed joint angles, i.e., flexion/extension, ab-/adduction, and internal/external rotation, would all be in error. To quantify the effects of errors in the definition of the flexion–extension axis, a sensitivity analysis was performed using knee joint angle data from a representative subject. The orientation of the flexion–extension axis in the transverse plane at the knee joint was analytically varied, from +15 to −15° at 5° intervals and the resulting joint angle patterns were recalculated. Similar analyses were performed at the hip and ankle joints; however, only the results for the knee joint will be presented here.

RESULTS

The mean and standard deviation of temporal distance parameters for the group of subjects evaluated in this study are presented in Table 1. When the subjects were grouped according to sex (male,

TABLE 1. *Mean and standard deviation of temporal distance factors*

Parameter	Units	Group I (young adults) Men (N = 28)	Women (N = 12)
Cadence	steps/min	112 ± 9	115 ± 9
Velocity	m/s	1.34 ± 0.22	1.27 ± 0.16
Stride time	s	1.08 ± 0.08	1.05 ± 0.08
Step time	s	0.56 ± 0.02	0.53 ± 0.06
Stride length	m	1.41 ± 0.14	1.30 ± 0.10
Stance phase	% gait cycle	61.0 ± 2.1	60.7 ± 2.6
Double limb support	% gait cycle	10.2 ± 1.5	10 ± 1.4

n = 28; female, n = 12) there were no significant differences in the spatiotemporal parameters between male and female subjects. The overall mean and standard deviation of angular excursions for the subjects along with one standard deviation envelope are shown in Figs. 3–5. The limb rotation angles are the average of nine cycles from each of the 40 subjects (total of 360 gait cycles). Zero percent corresponds to the heel strike and 100% corresponds to the next heel strike of the same limb. The percent standard deviations for the flexion–extension motion at the hip, knee, and ankle were smaller than those for the ab-/adduction or internal and external rotations. The joint angle data also were further divided according to sex. Except for hip ab-/adduction, there were no significant differences between the male and female groups for any of the joint angle patterns.

The effect of errors in defining the embedded axes on the computed angles are shown in Fig. 6 using the knee joint as an example. The knee flexion–extension angle was relatively unaffected while the knee varus/valgus and rotation angles were affected nonuniformly throughout the gait cycle. The results showed that the errors in knee varus/valgus and rotation angles varied with increasing knee flexion angle. The magnitude of the errors in the knee varus/valgus and rotation angles are shown as a function of the knee flexion angle for different magnitudes of error in the definition of embedded axes in Fig. 7A and 7B, respectively. Similar results were obtained at the hip and ankle joints.

DISCUSSION

In this paper, we have presented a detailed description and implementation of a technique for computing lower limb rotations during level walking using a simple marker system. For computing the limb rotation angles, a system of axes was defined based on a set of markers affixed to key anatomical locations. Two factors were considered in choosing the anatomical location. The first was to minimize relative motion between the skin and underlying bony structures, thereby satisfying the rigid body assumption. For the skin-mounted markers as well as the cuff-mounted markers, the rigid body assumption was found to hold (on the average) to within ±3 mm. This did not have a significant effect on the measured joint angle patterns. The second consideration was to minimize the amount of manual intervention needed to sort and track the marker trajectories accurately. In video motion analysis systems, it is common for the trajectories of closely spaced markers to cross each other, thereby making automatic tracking by the computer extremely difficult. Manual intervention is often necessary to identify trajectories of closely spaced markers whose paths intersect. In gait analysis, the trajectories of markers placed on the foot present problems due to their relative proximity to each other. Therefore, in the present system, only two markers were used on the foot to define limiting the measurement of ankle joint motion to flexion–extension and internal–external rotation. Due to the geometry and the size of the foot segment, adding another marker to measure eversion–inversion angle would complicate the data analysis. Further, given the finite accuracy and resolution of the motion analysis system, the estimates of inversion–eversion may not be sufficiently accurate to be of any practical use. By limiting the number of markers on the foot to two, the time required for data analysis is substantially reduced, which renders the system attractive for use in routine clinical gait evaluation.

In any type of motion analysis system, contacting or noncontacting, a source of error in the estimation of joint angle motion is due to uncertainty in the construction of an embedded coordinate system. In a goniometric system, the alignment of the goniometer determines the orientation of the embedded axis. In the present system, the body surface markers define the embedded axes and therefore their placement is crucial. While the effect of errors in the definition of embedded axes on the flexion–extension angles is small, ab-/adduction and rotation angles are affected significantly. This may be the reason for the large dispersion reported in the literature for the knee varus/valgus and rotation angles and therefore these angles must be interpreted cautiously. While it may be difficult to define the embedded axis exactly, it is at least necessary to be

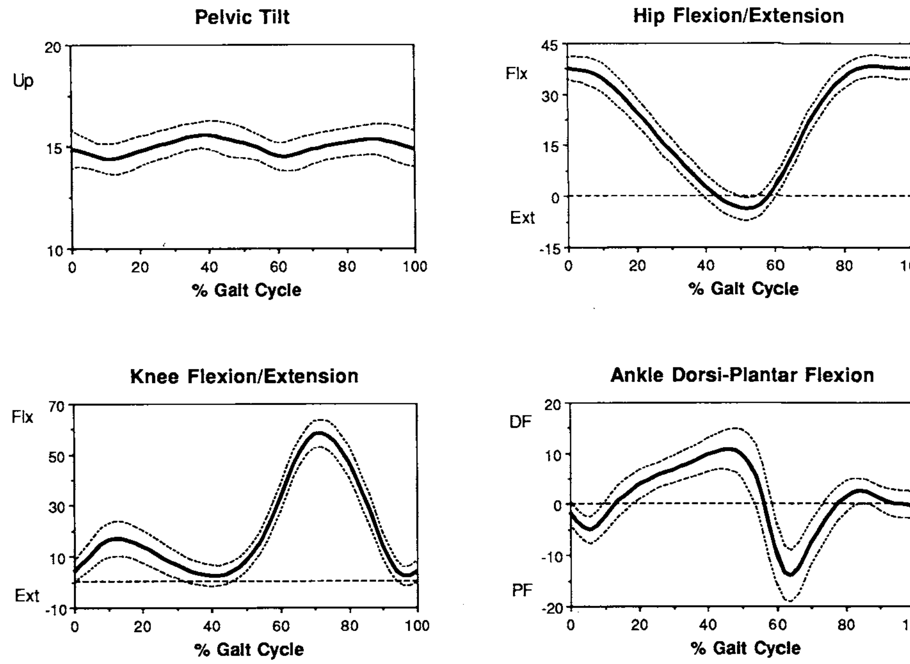


FIG. 3. Mean (thick line) and one standard deviation (dotted lines) of sagittal plane angles of normal adults. All angles are shown in degrees.

consistent in the definition so that it would be possible to compare data between different gait laboratories. For example, for the flexion–extension axis at the knee joint, the line joining the femoral condyles has been previously suggested by Chao et al. (5) and Grood and Suntay (7).

The sensitivity analysis also demonstrated that the error in ab-/adduction and rotation angles increased with increasing flexion angle at hip, knee, and ankle joints. In view of this, joint angle patterns of patients with flexion contractures (e.g., cerebral palsy patients) may be susceptible to errors throughout the gait cycle. Therefore, in such cases, the ab/adduction and rotation angles must be interpreted with caution.

Another source of error is due to uncertainty in defining the neutral axis or plane for the transverse plane rotations. Previously, it was suggested that a reference data set with the subject standing still (static) be used to obtain the position of the neutral

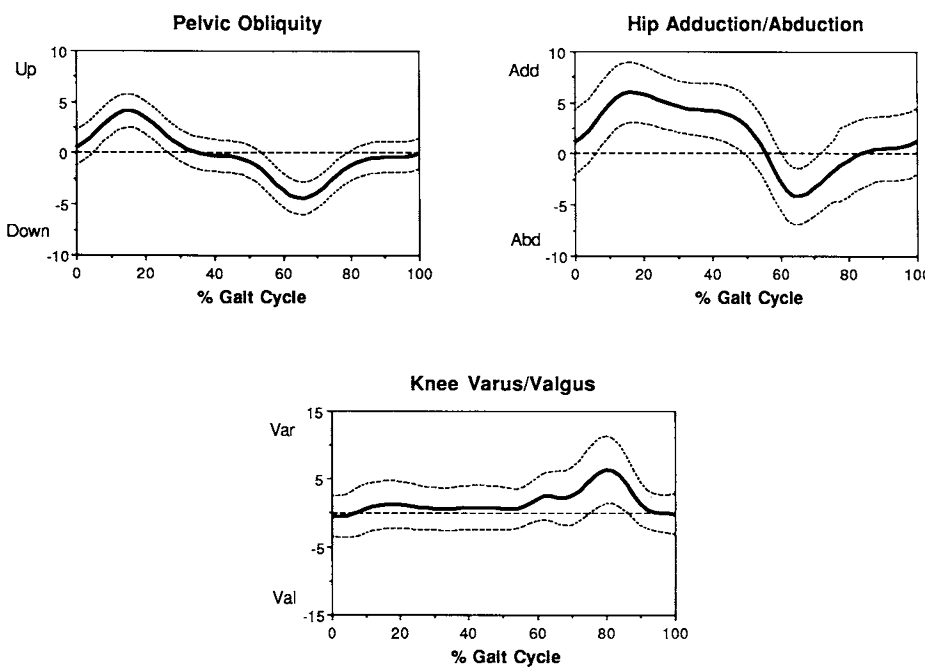


FIG. 4. Mean (thick line) and one standard deviation (dotted lines) of frontal plane angles of normal adults. All angles are shown in degrees.

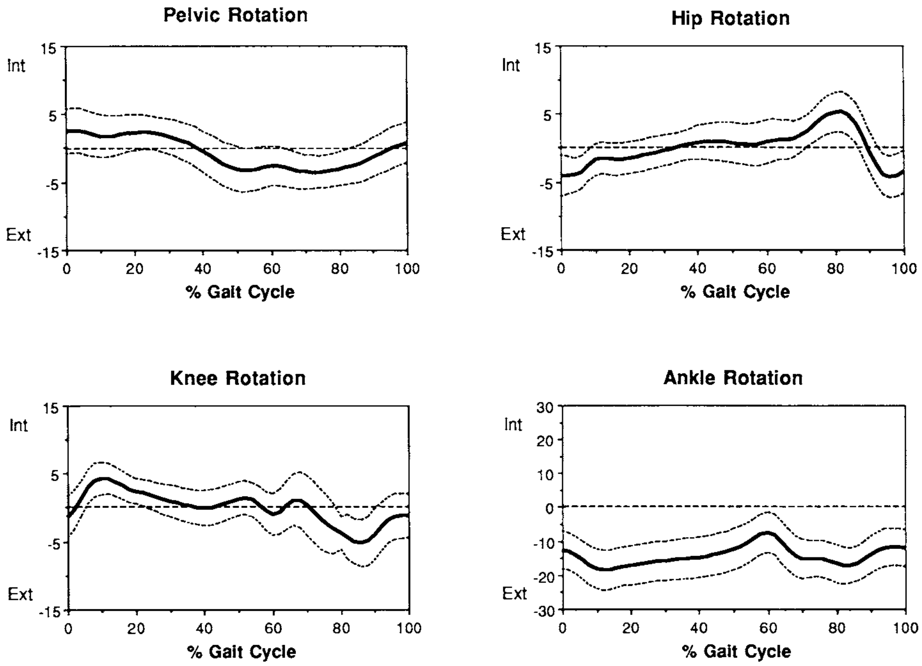


FIG. 5. Mean (thick line) and one standard deviation (dotted lines) of transverse plane angles of normal adults. All angles are shown in degrees.

axis. This procedure was used in this study to obtain a consistent definition of the neutral axis of rotation in the transverse plane. While this procedure yielded reasonable results for normal subjects, it may not be practical in a disabled group, particularly children with cerebral palsy.

The hip joint center estimation is another area that needs further analysis. How well do the empirical equations reflect the location of the true joint center? What happens to the joint angle patterns if there is an error in the location of the hip joint center? To answer some of the questions, the estimated hip center was perturbed in all three directions up to 1 cm and the resulting joint angle patterns were computed. For a 1 cm displacement, a maximum constant offset of 2° in the angle patterns was ob-

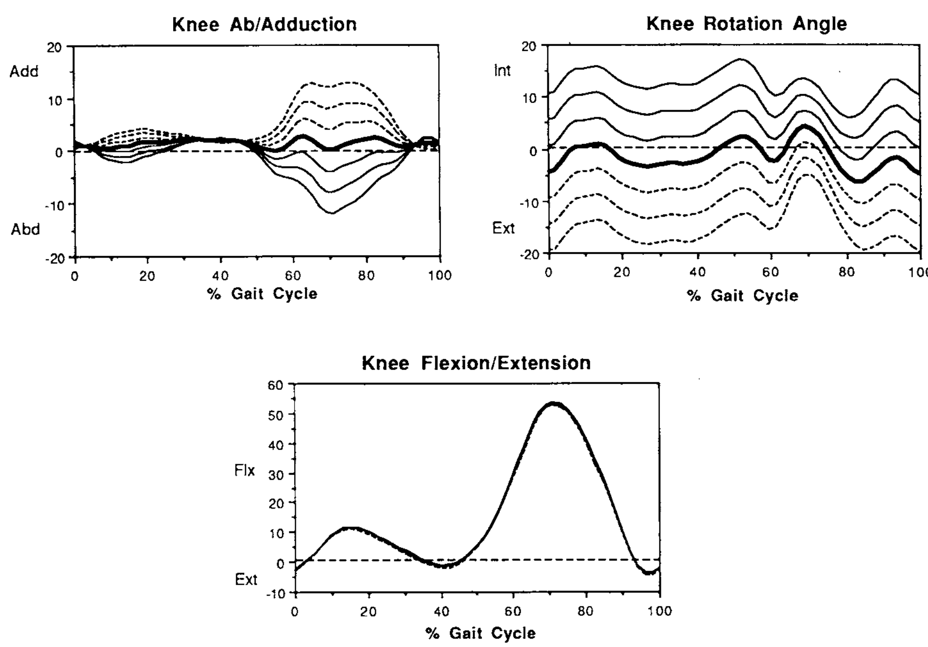


FIG. 6. Errors in the definition of embedded axes on knee angles. All angles are in degrees. Thick line indicates the measured joint angles of a representative subject. The flexion–extension axis is analytically rotated through a range of −15 to +15° from the reference position in steps of 5°. Corresponding knee angles are plotted in thin lines.

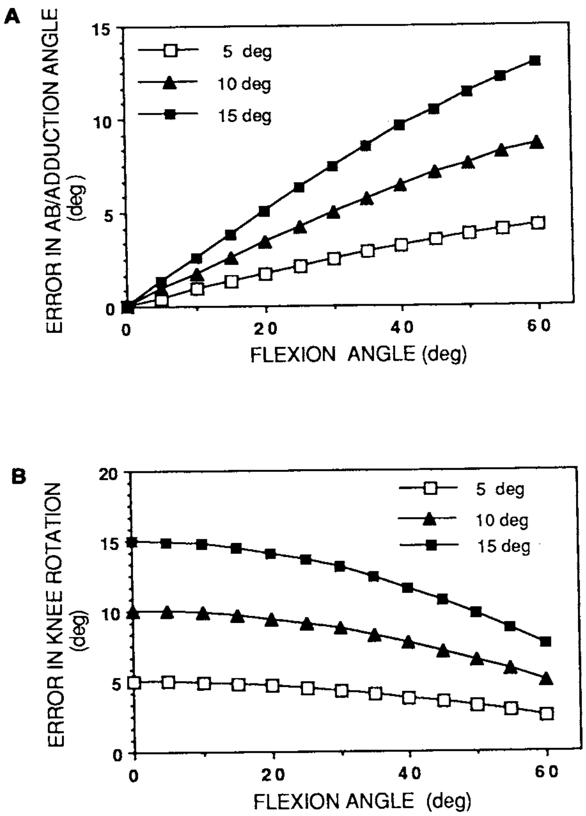


FIG. 7. Error in knee varus/valgus angle (**A**) and rotation angle (**B**) as a function of knee flexion angles for errors in the definition of knee flexion–extension axis.

tained. The ranges of limb rotations, however, were not affected.

A summary of the results (range of motion) from the present study along with the results from other laboratories are compared in Table 2, where the number of subjects is denoted by N. The age range of subjects in all of these studies was approximately similar. Results from this study are similar to results reported by Sutherland et al. (23) at all of the joints except the rotation angle of the pelvis. Specifically, flexion/extension at hip, knee, and ankle joints was quite similar. The difference in the range of pelvic rotation may be due to the different definitions used in measuring this angle. Sutherland et al. (23) defined pelvic rotation based on the coordinates of the tip and base of the sacral stick in a horizontal plane while the same angle is defined as a third rotation in the Euler sequence in our study. The range of motion for the knee flexion–extension angle in this study was lower than those measured using goniometers (5) and the reasons for this are not clear. There were no other remarkable differences in joint angles measured between this study and others listed in Table 2.

In summary, we have described a system of measuring three-dimensional angular motion of the pelvis, thigh, shank, and foot based on a four-segment rigid body model of the lower extremity. Embedded coordinates were assigned to these segments based

TABLE 2. *Comparison of joint angle (degrees) data (mean total range of motion) with previous work*

	Present study $N = 40^a$	Sutherland (23), $N = 15$	Winter (26), $N = 16$	Isacson et al. (10), $N = 20$	Chao et al. (5), $N = 110$	Johnston and Smidt (11), $N = 33$	Murray et al. (15), $N = 60$
Age of subject group (years)	18–40	19–40	—	25–35	19–32	23–55	20–55
Measurement technique	Vicon	Cine film	Video	Goniometer	Goniometer	Goniometer	Interrupted light
Pelvis							
Tilt	2.8	2	—	—	—	—	—
Obliquity	8.4	9	—	—	—	—	—
Rotation	9.2	15	—	—	—	—	8
Hip							
Flexion	43.2	43	43	30.2	—	52	42
Adduction	11.6	14	—	13.6	—	13	—
Rotation	13	9	—	9.9	—	12	—
Knee							
Flexion	56.7	58	64	60.6	68.0	—	60
Varus	13.4	—	—	9.0	10	—	—
Rotation	16.0	12	—	12.9	13	—	—
Ankle							
Flexion	25.5	28	28	19.4	—	—	28
Rotation	15.7	17	—	12.9			

[a] 40 subjects evaluated three times/day on three different test days.

on a set of surface markers and the relative rotations between segments were determined using orthopedic Euler angle definitions. The errors introduced by inaccuracies in the definition of the embedded coordinate system (flexion–extension axis) and alignment were quanitified. A group of 40 normal subjects was evaluated and the results were presented as a normative data base that can be used for comparison purposes. It is hoped that the joint angle measurement technique presented in this paper will provide a uniform method for data acquisition so that it will be possible to compare and/or share gait data between clinical centers.

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