A Position based Kinematic Method for the Analysis of Human Gait

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Human joint motion can be kinematically described in three planes, typically the frontal, sagittal, and transverse, and related to experimentally measured data. The selection of reference systems is a prerequisite for accurate kinematic analysis and resulting development of the equations of motion. Moreover, the development of analysis techniques for the minimization of errors, due to skin movement or body deformation, during experiments involving human locomotion is a critically important step, without which accurate results in this type of experiment are an impossibility. The traditional kinematic analysis method is the Angular-based method (ABM), which utilizes the Euler angle or the Bryant angle. However, this analysis method tends to increase cumulative errors due to skin movement. Therefore, the objective of this study was to propose a new kinematic analysis method, Position-based method (PBM), which directly applies position displacement data to represent locomotion. The PBM presented here was designed to minimize cumulative errors via considerations of angle changes and translational motion between markers occurring due to skin movements. In order to verify the efficacy and accuracy of the developed PBM, the mean value of joint dislocation at the knee during one gait cycle and the pattern of three dimensional translation motion of the tibiofemoral joint at the knee, in both flexion and extension, were accessed via ABM and via new method, PBM, with a Local Reference system (LRS) and Segmental Reference system (SRS), and then the data were compared between the two techniques. Our results indicate that the proposed PBM resulted in improved accuracy in terms of motion analysis, as compared to ABM, with the LRS and SRS.

Key Words : Gait Analysis, Skin Movement, Human Motion, Knee Joint, Euler Angle, Bryant Angle, Kinematic

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1. Introduction

Human joints are parts of the human body characterized by fairly complex structure, and joints are subject to frequent injury. In particular, the tibiofemoral joint of the knee, when it engages in the sliding and rolling motion, is compelled to support a force equal to four times the body weight in one's entire lifetime. Therefore it has been estimated that 80% of the population over the age of 55 suffers from diseases such as rheumatoid arthritis or osteoarthritis (Jennifer Manetta et al., 2002). The diagnosis and cure of diseases associated with bones has greatly advanced with the advent of joint imaging techniques, although most of these techniques work only under static conditions (Couteau et al., 2000; Koji et al., 2003). However, as human movement is dynamic, this technique is clearly limited with regard to both application and diagnosis. Therefore, a great deal of research is currently focusing on the analysis and diagnosis of movement, based on ever-more-accurate human body modeling techniques (Moeinzadeh et al., 1983; Abdel-Rahman et al., 1998; Mun et al., 2004). This movement analysis is accomplished in three steps; the selection of a reference system, the application of error reduction methods, and the actual analysis procedure. Therefore, the accurate movement analysis clearly requires the selection of an appropriate reference system, the application of an effective error reduction method, and the employment of a proper analysis method.

Four reference systems are commonly used in the context of gait and movement analysis; the Absolute Reference System, Local Reference System, Segmental Reference System, and the Joint Reference System (Craik and Oatis, 1995). Each reference system has a different character and meaning so the selection of a reference system which is appropriate for the purpose of the analysis becomes the basis of an accurate movement analysis. In particular, the Local Reference Systems (LRS) and Segmental Reference Systems (SRS) consist not only of three points in space but can also be used to analyze relative motion between segments. Also, the SRS contains clear anatomical meanings such as proximal-distal, medial-lateral and anterior-posterior. Therefore, biomechanists and clinicians commonly use these reference systems which are relatively simple for such technicians to understand (Reinschmidy et al., 1997; Hebert et al., 2000; Zhang et al., 2003; Cerveri et al., 2004; Judi Laprade et al., 2005). However, the selection of a proper reference system, which is appropriate to the desired analysis is not, by itself, sufficient to reduce the errors associated with skin movement. Therefore, a host of methods have been developed in previous studies to reduce random and systematic errors due to skin movement (Spoor et al., 1980; Velpaus et al., 1988; Challis et al., 1995; Cappello et al., 1996; Lu et al, 1999; Mun et al, 2001; 2003). However, the use of an error reduction method is an irreducibly limited method for the reduction of skin marker errors, because movement analysis must be based on a wide variety of phenomenological aspects of the human body. Therefore, these residual errors, which are applied to Euler angle and Bryant angle methods, become a factor generating significant cumulative errors in most traditional analysis procedures.

In the clinical analysis of human motion, the general methodology for the kinematic data recovery is to use Euler angle and Bryant angle orientation coordinates (Neil et al., 1996). Fundamentally, the method to use Euler angle and Bryant angle involve the calculation the position and orientation of a body segment throughout a successive axial rotation expressed as a value such as x'-y'-z'. Therefore, the successive rotation effect expressed by the Euler angle method under dynamic conditions would be identical to that of a static condition to be standard in the motion analysis of a rigid body as in a robot. However, because the skin movement artifact is of remarkable magnitude: a few millimeters, up to 40 mm during a typical gait cycle (Cappozzo et al., 1996), skin marker error including each axis is generate a cumulative effect. Also, because typical human motion analysis utilizes the inverse dynamic method, using a force plate and 3-D trajectory marker data, this analysis is proceeds according to the following sequence: ankle joint, knee joint and hip joint. In result, the rotation matrix included each joint throughout the analysis procedure is used several times, therefore this cumulative error effect increases continuously through the sequence.

Thus, in order to ensure an accurate motion analysis, a previous study underlined the importance of selecting a proper reference system (Blacharski et al., 1975 ; Pennock and Clark, 1990 ; Fujie et al., 1996), as well as error reduction method which compensates for skin movement artifacts (Spoor et al., 1980 ; Velpaus, 1988 ; Challis et al., 1995 ; Cappello et al., 1996 ; Lu et al., 1999). In the analysis procedure, the Euler angle and Bryant angle methods were used for the recovery of kinematic data (Apkarian et al., 1989 ; Koopman et al., 1995 ; Biryukova et al., 2000 ; Rivest., 2004). However, if the previous ABM is applied during this analysis procedure as the final step in motion analysis, several problems impose themselves : not only the aforementioned cumulative error problem occurs but it also becomes impossible to maximize the effects of the error reduction method and the reference system selection applied during preprocessing. Therefore, we felt that there was a clear need for a study to solve the cumulative error problem inherent to the use of the Euler angle and Bryant angle methods.

The objectives of this study were to : (1) propose a new Position Based Kinematic Method which considered both translational and angular motion between markers representing identical body segments during gait, and to verify the efficacy of this method, compared to the previously used Angular Based Kinematic Method, specifically with regard to the regional effects of the displacement of human locomotion and (2) to compare the analysis errors via the application of each method (ABM, PBM) based on the LRS and SRS, both of which are commonly used in gait analysis.

2. Analysis

2.1 Gait experiment

Ten healthy male subjects, none of whom exhibited any lower extremity pathology, were enrolled into this study. The gait date was normalized at 100% per gait cycle (stance, swing), and ten gait trials were conducted with each subject. Gait speed was defined as an average forward velocity of 1.2(0.9) m/s, and the 3-D trajectories of the markers were used as inputs for the knee joint kinematic analysis. Video data was collected with a 6-camera Vicon 460 (Oxford Metrics Limited, Oxford, UK) system, equipped with an opto-electric motion analysis system. During walking, data was collected at a frequency of 60 Hz for the camera motion system. The data was then filtered with a fourth-order Butterworth, zero-lag, low-pass filter with a 7 Hz cut-off frequency. This cut-off frequency was determined via residual analysis, as was described by winter (winter, 1990). This cut-off frequency was determined because 99.7% of the signal power was contained in the lower seven harmonics below 6 Hz and the signal power including the frequency above 7 Hz was regarded as noises (winter, 1990). In the experiment, the location of the markers which comprised the LRS were chosen in locations in which relatively less skin movement occurs, from an anatomical standpoint. Also, markers which represented anatomical landmarks were used to define the SRS. That is to say, the landmarks on the shank could include the medial and lateral malleolus and the lateral femoral epicondyle. On the thigh, the landmarks could include the medial and lateral epicondyle and the greater trochanter. Fig. 1 shows the locations of the placed markers in this experiment and Fig. 2 shows the overall gait analysis system.

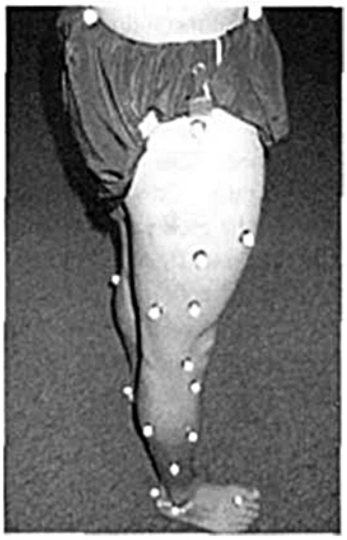

Fig. 1 The location of the attached markers

Table 1 Subject Character, mean values (±SD)

Sex	Age (years)	Height (cm)	Weight (kg)
male	24.3 (1.4)	174.3 (1.5)	66.7 (6.1)

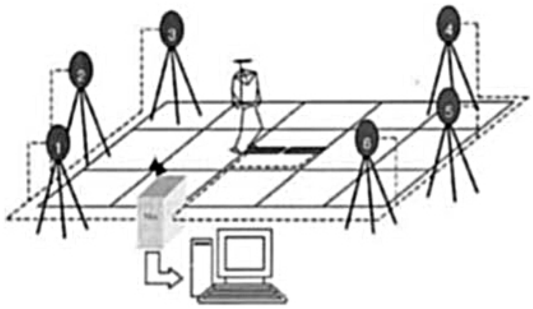

Fig. 2 Configuration of the gait analysis system

2.2 Angular-based method (ABM)

The general angular based kinematic configuration is shown in Fig. 3. The position of an arbitrary point, **P**, on the shank segment can be located using a relative coordinate technique, as the following:

$$\mathbf{r}^P = \mathbf{r}_1 + \mathbf{A}_1\mathbf{s}'_{12} - \mathbf{A}_2\mathbf{s}'_{21} + \mathbf{A}_2\mathbf{s}'_{2P} \tag{1}$$

where $\mathbf{r}_1$ is a vector from the global coordinate reference frame to the thigh reference frame, $\mathbf{A}_1$ and $\mathbf{A}_2$ are the transformation matrices from the each segment's local coordinate system to the global coordinate system, and $\mathbf{s}'_{12}$, $\mathbf{s}'_{21}$ and $\mathbf{s}'_{2P}$ are constant vectors given in the segment's local coordinate system.

The most common method used in the calculation of the angular orientation of these segments is the acquisition of the transformation matrix via the evaluation of the angular displacement of a given body segment from the displacement data measured using three non-colinear sensors. The sensor points on the each body segment (denoted as points $\mathbf{P}_1$, $\mathbf{P}_2$, and $\mathbf{P}_3$ on the thigh segment) are then used in the definition of the transformation matrix, $\mathbf{A}_1$, which relates the body coordinate frame to the global frame. This transformation can expressed as the following:

$$\mathbf{A}_1 = \begin{bmatrix} a_{11} & a_{12} & a_{13} \\ a_{21} & a_{22} & a_{23} \\ a_{31} & a_{32} & a_{33} \end{bmatrix} = [\hat{\mathbf{f}}_1\ \hat{\mathbf{g}}_1\ \hat{\mathbf{h}}_1] \tag{2}$$

where

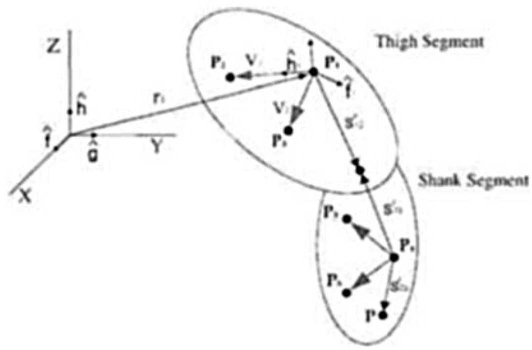

Fig. 3 ABM configuration

$$\hat{\mathbf{h}}_1 = \frac{\mathbf{v}_1}{\|\mathbf{v}_1\|}$$
$$\hat{\mathbf{f}}_1 = \frac{\mathbf{x}_1}{\|\mathbf{x}_1\|} \tag{3}$$
$$\hat{\mathbf{g}}_1 = \hat{\mathbf{h}}_1 \times \hat{\mathbf{f}}_1$$

The transformation matrix can be equivalently expressed in terms of the Euler angles, as follows:

$$\mathbf{A}_1 = \begin{bmatrix} c\psi c\phi - s\psi c\theta s\phi & -c\psi s\phi - s\psi c\theta s\phi & s\psi s\theta \\ s\psi c\phi - c\psi c\theta s\phi & -s\psi s\phi - c\psi c\theta c\phi & -c\psi s\theta \\ s\theta s\phi & s\theta c\phi & c\theta \end{bmatrix} \tag{4}$$

where **s=cos** and **s=sin**. Thus, the Euler angles for each segment can be determined using the following equations:

$$-\tan\psi = \frac{\sin\psi}{-\cos\psi} = \frac{a_{13}}{a_{23}} \tag{5}$$

$$\cos\theta = a_{33} \tag{6}$$

$$\tan\phi = \frac{\sin\phi}{\cos\phi} = \frac{a_{31}}{a_{32}} \tag{7}$$

Whereas this method can be used to determine the orientation of each segment well, it fails to consider some additional information which is available form the sensors. The proposed Position-based method makes use of all the sensor data, in order to simultaneously determine segment orientation, and estimate localized sensor movement.

2.3 Proposed position-based method (PBM)

The proposed position-based kinematics coordinate system was formulated according to a Cartesian coordinate system which was fixed to the body and moves along with it, as was shown in the Fig. 3. The vector used to locate **P** in the XYZ frame is provided by Eq. (8) as the following:

$$\mathbf{r}^P=\mathbf{r}_1+\mathbf{v}_2+\mathbf{A}_{13}\mathbf{s}'_{12}-\mathbf{A}_2\mathbf{s}'_{21}-\mathbf{v}_3+\mathbf{v}_4+\mathbf{A}_{2p}\mathbf{s}'_{2p} \quad (8)$$

where $\mathbf{s}'_{12}$, $\mathbf{s}'_{21}$ and $\mathbf{s}'_{2p}$ are constant vectors with respect to the body-defined $x'_1y'_1z'_1$, $x'_4y'_4z'_4$, and $x'_6y'_6z'_6$ coordinate reference frames, respectively, and the vector $\mathbf{v}_i$ is determined from the measured point data, $\mathbf{v}_1=\mathbf{P}_1-\mathbf{P}_2$, etc.

This coordinate system selection scheme takes into account the effect of localized sensor movements such as translational and angular errors due to the movement of the skin during the gait experiment. Consider the single segment depicted in Fig. 4. The unit vector, $\hat{\mathbf{h}}_1$, lies along the segment $\overline{\mathbf{P}_1\mathbf{P}_2}$. While the segment is permitted to change in length, the coordinate frames located at $x'_1y'_1z'_1$ and $x'_2y'_2z'_2$ exhibit identical local-to-global transformation matrices, $\mathbf{A}_1$. The change in length for the $\overline{\mathbf{P}_1\mathbf{P}_2}$ segment is a measure of both skin movement and experimental error which takes into account only the information provided by the two sensors.

Additional errors for the segment can be determined using the data provided by the third sensor. During the initial calibration phase of the gait experiment, the angle, θ_0, is determined between the $\hat{\mathbf{g}}_1$ axis and the $\overline{\mathbf{P}_1\mathbf{P}_3}$ segment, and the angle, θ_i is estimated to be identical to angle, θ_0, under the dynamic conditions. Again, $\hat{\mathbf{g}}_1$ is a unit vector, whereas the $\overline{\mathbf{P}_1\mathbf{P}_3}$ segment is permitted to change in length. Similarly to the $\overline{\mathbf{P}_1\mathbf{P}_2}$ segment, changes in length are related to both experimental errors and skin movement. The coordinate transformation, $\mathbf{A}_{13}$, from the $x'_3y'_3z'_3$ coordinate reference frame to the global frame is as follows:

$$\mathbf{A}_{13}=\mathbf{A}_1\mathbf{A}'_{13} \quad (9)$$

where

$$\mathbf{A}'_{13}=\begin{bmatrix}1 & 0 & 0\\ 0 & \cos(\theta_i-\theta_0) & -\sin(\theta_i-\theta_0)\\ 0 & \sin(\theta_i-\theta_0) & \cos(\theta_i-\theta_0)\end{bmatrix} \quad (10)$$

and

$$\cos\theta_i=\frac{\hat{\mathbf{g}}_1\cdot\mathbf{v}_{12}}{\|\mathbf{v}_{12}\|} \quad (11)$$

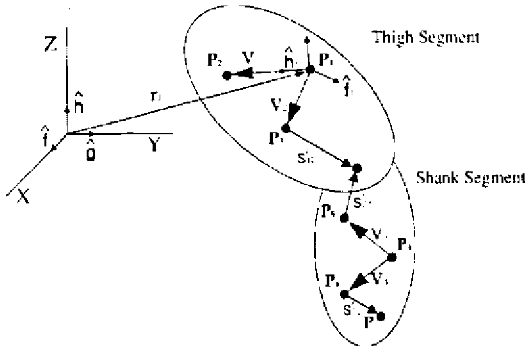

Fig. 4 PBM configuration

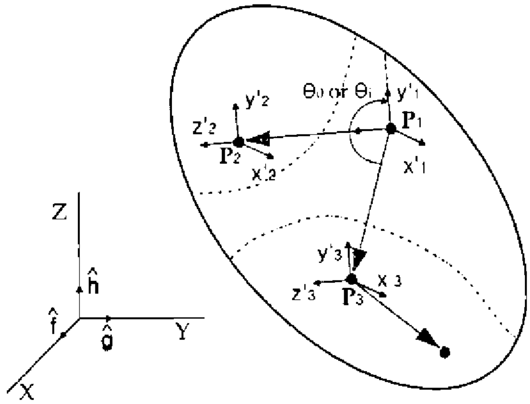

Fig. 5 Configuration of the thigh segment

Eq. (11) is evaluated form the time-dependent walking frames. In cases in which no angular error changes occur between the initial calibration measurement and the gait measurement, then all coordinate reference frames on the segment, $x'_1y'_1z'_1$, $x'_2y'_2z'_2$, and $x'_3y'_3z'_3$ will exhibit identical rotational transformations. Otherwise, the angular error constitutes a metric for the localized movements of the sensors.

2.4 Validation of the PBM

In order to verify the developed method, (1) the extent of the joint dislocation at the knee was compared between the ABM and proposed PBM and (2) the degree to which 3-D translation motion occurred in the knee joint was compared with the previous study, using bone pin marker set. The joint dislocation, which was shown to occur in cases in which the joint was not constrained, was calculated as the distance between

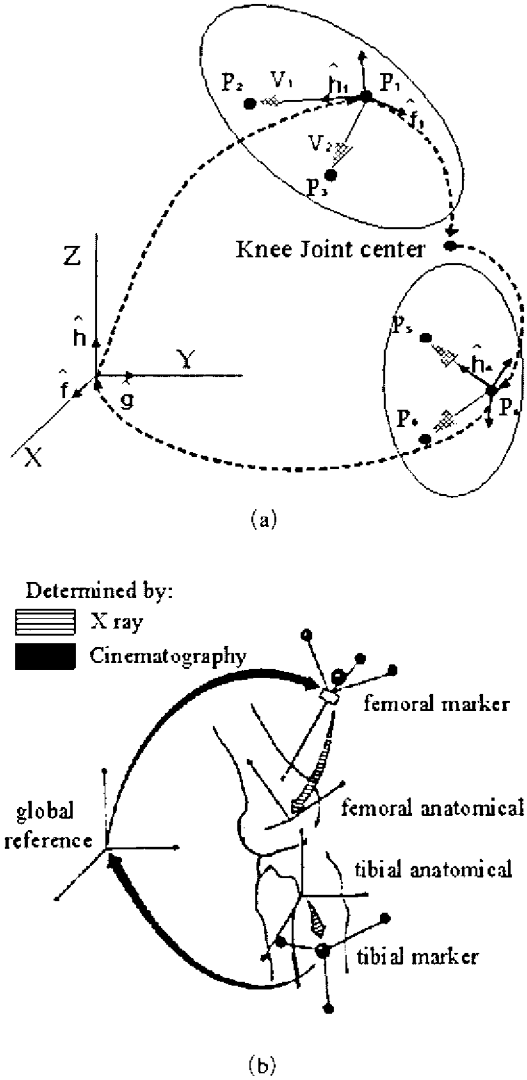

Fig. 6 (a) Configuration of the analysis method using a relative coordinate system and (b) a configuration of the analysis method using the Bone pin marker set (Scanned picture from Lafortune et al., 1992)

the two end points, with the proximal end point of the distal segment designated as the joint center (Kepple et al., 1994; Lu et al., 1999). This study employed a relative coordinate system, and the knee joint was consisted of the markers from the thigh and shank within the static calibration step. Also from this standpoint, the summation of the joint dislocation at the knee was calculated during the gait cycle (Fig. 6(a)). In the absence of skin movement, the origin of the thigh with regard to the origin of the shank exhibit only anatomical displacements due to the sliding and rolling motion inherent to the geometry of the knee joint. However, this joint displacement can result in an apparent non-anatomical displacement due to skin movement and body deformations.

Secondly, in order to estimate the accuracy of the results of our analysis, we compared the ABM and PBM with results of the previous study based on bone pin marker set (Lafortune et al., 1992; Reinschmidt et al., 1997). The bone pin marker set means intra-cortical pins fixed directly into bones and this method that the markers of the end of pins is composed knee joint and analyze the joint motion is very insertable (Fig. 6(b)). The positions of triads in the global frames of reference were determined through cinematography while their positions in the respective anatomical frames of reference were obtained through X-ray. As, in this method, no errors due to skin movement occurred, it is possible to estimate the effects of skin movement with regard to the ABM and PBM. This study compared 3-D translation motion of the knee joint, in terms of anterior/posterior, lateral/medial, and compression/distraction motion during walking.

3. Results

Figure 7(a) represents the angle variation between markers, designated as $\theta_i - \theta_0$, used in Eq. (10) for both the thigh and shank segments. (b) shows the mean values of the maximum and minimum of the angle variation $\theta_i - \theta_0$ values. In detail, the angle variation for the thigh was measured at 4.7(1.4) degrees and the angle variation for the shank was measured at approximately 2.6(0.6) degrees. This result was consistent with the results of a previous study, in which the value assigned to skin displacement, as tracked by shank markers was found to be smaller than that measured by thigh markers (Alberto Leardini et al., 2005).

Figure 8 shows a tibiofemoral joint dislocation at the knee joint. (a) shows the joint dislocation values obtained using each method (ABM, PBM)

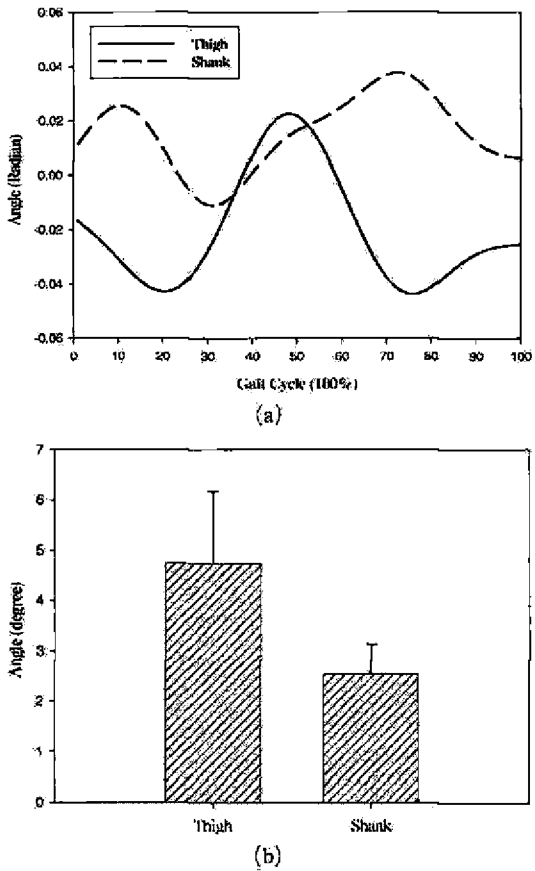

Fig. 7 The pattern of angle variation of on the thigh and shank segment

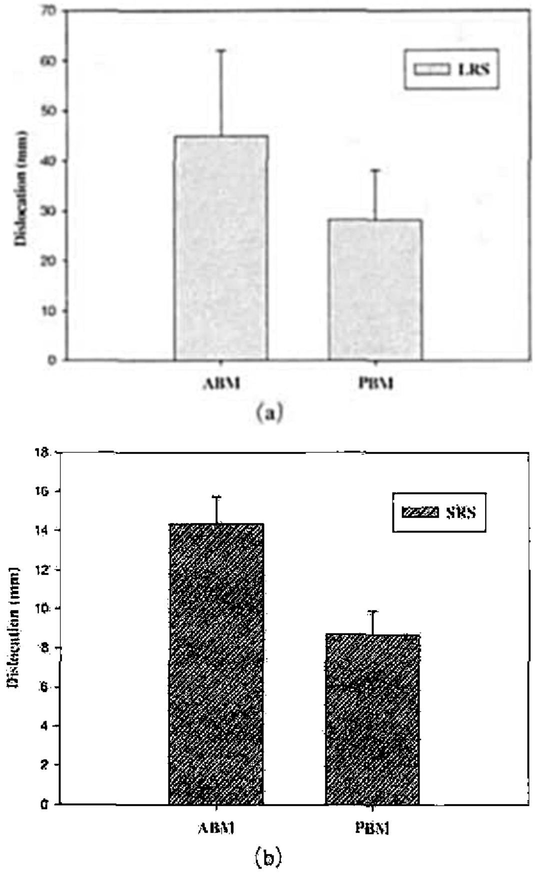

Fig. 8 Joint dislocation at knee for the ABM and PBM based on the LRS (a) and SRS (b)

based on the LRS, and (b) shows the same, but based on the SRS. Also the left side of the figure represents the joint dislocation when using the ABM, and the right side of the figure represents joint dislocation using the PBM. As shown in this figure, the average degree to which joint dislocation occurs at the knee is 45(17.2) mm with the ABM and 28.2(9.9) mm with the PBM based on the LRS. Similarly, the average degree to which joint dislocation occurs at the knee was 14.2(1.4) mm at the ABM and 8.7(1.2) mm at the PBM based on the SRS. As a result, the degree of joint dislocation measured when using the PBM based on the LRS was reduced by 37.3% as compared with that measured with the ABM. When the PBM was used based on the SRS, we noted a 39.2% reduction as compared with that observed when using the ABM.

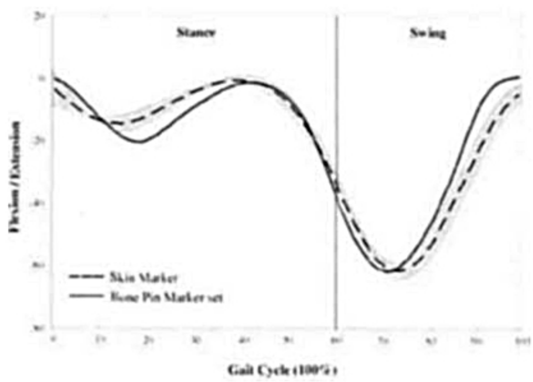

Fig. 9 Angular pattern of knee joint during walking (sagittal plane). The dashed line shows the method using bone pins and the solid line shows the method using skin markers

Figure 9 shows the average pattern of flexion/extension of the tibiofemoral joint during walking. The solid line shows the experiment which used skin markers, and the dotted line is the case in which the cortical pins were directly fixed into the bones. The solid line and dotted line were not completely coincident due to the combined effects of skin movement. However, both of them exhibit a slight flexion followed by an extension during the stance phase, and a more pronounced flexion, also followed by an extension, during the swing phase. In more detail, as the heel struck the ground, the knee exhibited a flexion angle value of 0 degrees for the cortical pins and 8 degrees when using the skin markers. Also, the knee reached initial peak value of 20 degrees in

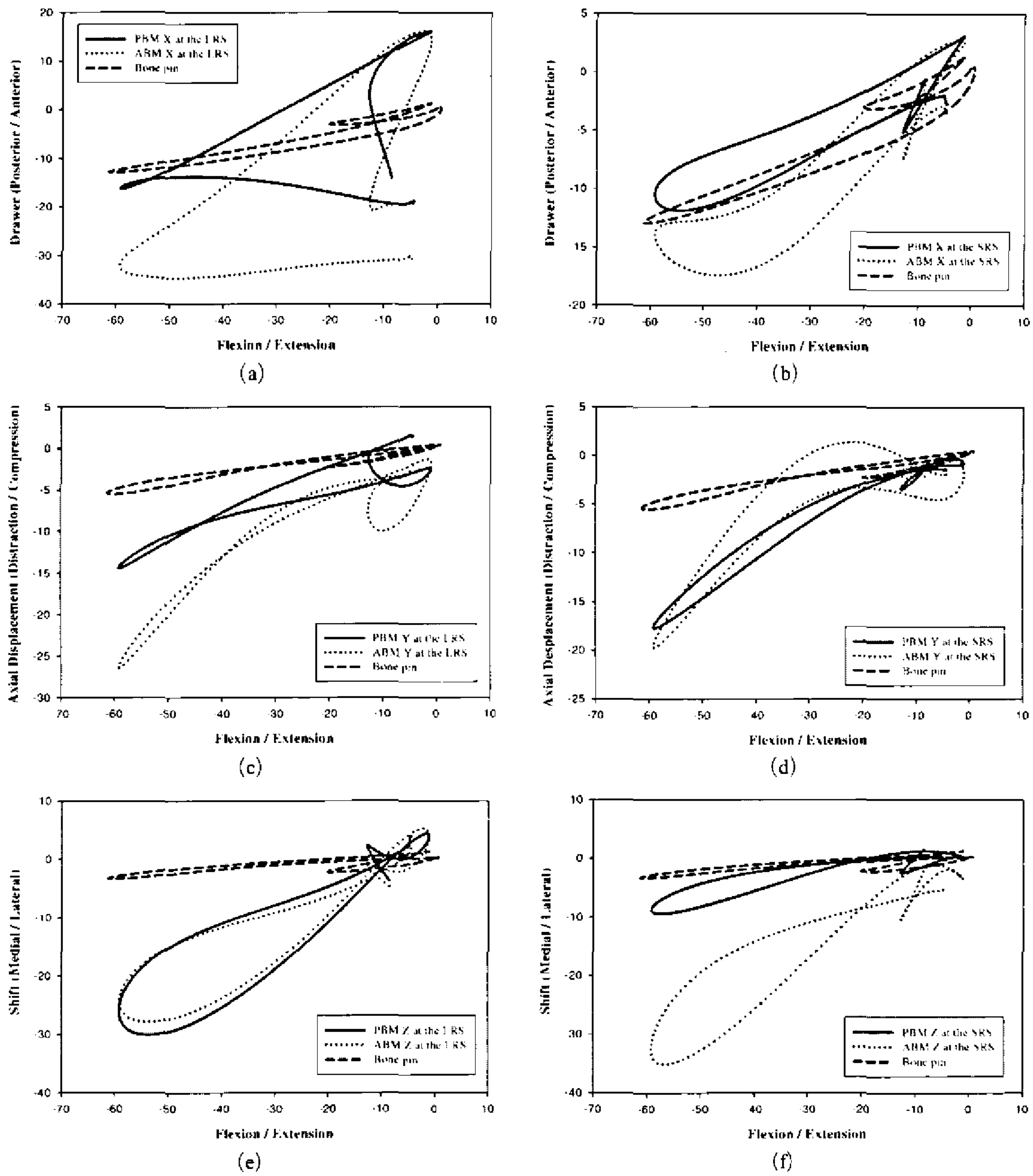

Fig. 10 Pattern of the translation of the tibiofemoral joint as a function of its flexion/extension pattern at the LRS (left) and SRS (right)

approximately 18% of the gait cycle for the cortical pins and 13 degrees, over approximately 12% of gait cycle for this experiment. Thereafter, the knee flexed fully and the values were similar to one another.

Figure 10 represents the 3-D translational motion of the knee joint, according to the flexion/extension angles. 3-D translational motion refers to the linear behavior of the femur with respect to the tibia, expressed in terms of the medial/lateral, anterior/posterior, and compression/distraction planes of movement. As the location of the tibial and femoral anatomical frames of reference were not determined to be of coincident origins, all linear motion was expressed relative to the positions occupied by the bones while walking. The motions of these three directions are regarded as drawer (posterior/anterior), axial displacement (distal/proximal), and shift (lateral/medial), which are represented by the vertical axis in Fig. 10. The left side of Fig. 10 shows each method based on the LRS and the right side of the figure was based on the SRS. As the method using bone pin marker set exhibits a linear relationship, it is quite clear that the 3-D translational motion of the knee joint is comparable to the pattern of flexion/extension angles. The method using skin markers was not found to represent the linear relationship due to skin movement. However, in each of the figures, the PBM exhibited a closer tendency to the method using the bone pin marker set than did the ABM except in figure (e).

Tables 2 and 3 show the maximum, minimum, and mean values of 3-D translational knee joint motion using the ABM and PBM, based on the LRS and SRS. In the bone pin experiments, the ABM based on the LRS involves error values of 40.8, 31.1, and 38.3(mm). In particular, the drawer value was associated with the greatest amount of difference. Also, the PBM based on the LRS involves error values of 25.7, 19.7, and 40.9 (mm), and a reduced error of 22.3% as compared to the ABM, and the drawer value exhibited the most profound reducing tendency, with a value of 15 mm. The ABM based on the SRS involves the error values of 4.1, 12.6, and 28(mm). In particular, the shift value exhibited the greatest degree of difference. Also the PBM based on the SRS was associated with error values of 0.5, 11.5, and 3.71(mm), and also exhibited a reduced error value of 61.1% as compared with the ABM, and the shift value was associated with the most profoundly reduced tendency, with a value of 24 mm.

Table 2 Comparison between different estimations of the ensemble time-averaged difference of 3-D translational motion of the knee at the LRS

(Unit : mm)	Bone pins	LRS					
		ABM			PBM		
	diff	max	min	diff	max	min	diff
Drawer	14.3	−0.7 (19.3)	−55.8 (22.9)	55.1	2.7 (15.9)	−37.2 (17.3)	39.9
Axial Displacement	7.0	−1.8 (1.2)	−40 (14.3)	38.1	1.4 (0.89)	−25.3 (11.8)	26.7
Shift	5.6	8.4 (3.7)	−35.5 (8.3)	43.9	7.7 (3.7)	−38.8 (9.5)	46.5

Table 3 Comparison between different estimations of the ensemble time-averaged difference of 3-D translational motion of the knee at the SRS

(Unit : mm)	Bone pins	SRS					
		ABM			PBM		
	diff	max	min	diff	max	min	diff
Drawer	14.3	2.3 (1.2)	−16.1 (2.4)	18.4	2.1 (1.5)	−11.8 (0.5)	13.9
Axial Displacement	7.0	1.02 (1)	−18.6 (2.1)	19.6	0.2 (0.6)	−18.3 (1)	18.5
Shift	5.6	8.9 (9.57)	−24.8(12.8)	33.6	2.6 (0.8)	−6.7 (3.4)	9.3

There exists the largest difference of 3-D translation motion in the direction of anterior/posterior between the ABM and PBM from Table 3 since the matrix $\mathbf{A}'_{13}$ is the rotation matrix about X component. In more detail, since the x axis is the medial-lateral direction at the LRS and the matrix $\mathbf{A}'_{13}$ rotates in the medial/lateral direction throughout the matrix $\mathbf{A}_1\mathbf{A}'_{13}$ instead of the previous, angular-based matrix $\mathbf{A}_1$, the drawer values of the PBM and ABM exhibited the most pronounced differences compared to other values (shift, axial displacement) at the LRS. Practically, Fig. 10(a) shows that the patterns of the PBM are closer to those of bone pin experiment than those of the ABM. Also, there exists the largest difference of shift values between the ABM and PBM from Table 4 and Fig. 10(f) since the x axis is in the anterior/posterior direction at the SRS.

4. Discussion

The objective of this study was to minimize the problems posed by skin movement errors, which has been determined to be the most serious issue in motion analysis based on human body modeling. As the previous study for accurate motion analysis was focused narrowly on the selection of a proper reference system and a proper the error reduction method, the cumulative error in the analysis procedures appears to be impossible to overcome. Therefore, in order to resolve the problem of cumulative error, residual errors due to the only use of the Euler angle and Bryant angle methods are eliminated by the use of the error correction matrix, $\mathbf{A}_{13}$. Also, the error values of each method were compared and analyzed, based on the LRS and SRS using anatomical landmarks.

3-D translational knee joint motion due to the use of the ABM and PBM were compared with the experimental results, using bone pin marker set. Because the study using bone pin marker set did not involve errors due to skin movement, the reduction effects of skin marker error of the PBM were estimated via comparisons with the bone pin experiment. In the case of comparison with the ABM, the 3-D translational knee joint motion shown in the PBM represents a more linear pattern than was seen in the experiment using the bone pin marker set, and was determined to have a mean error reduction effect of 41.1%. In particular, the shift displacement occurring in the medial-lateral direction exhibited the largest error reduction effect, with an error rate of 61.1%. This result was acquired via the reduction of the translation and angular errors between markers, using redundant markers and the error correction matrix, $\mathbf{A}_{13}$.

In the case in which the PBM was used for movement analysis, the cumulative error problem was ameliorated to 41.1% However, this result was affected by the applied reference system used. The results of the application of PBM and ABM, which are based on the LRS and SRS using anatomical landmarks, show that the joint dislocation at the knee was reduced by 68.5% in the case of analysis based on the SRS. Because the marker, which were used to construct the SRS, were much closer to the knee joint than in the LRS, the effects due to skin marker errors might be lessened in the analysis procedure.

The result of this study is summarized as follows

(1) The problems due to cumulative errors when the Euler angle or Bryant angle is employed in human motion analysis procedures, can be resolved with the use of redundant markers and the error correction matrix, taking into account the translational and angular motion between markers caused by skin movement.

(2) In clinical motion analysis based on the LRS and SRS, the PBM was used in order to ameliorate the skin movement effects.

Gait and human motion analysis has been applied to a wide variety of purposes and, in analysis, an inverse dynamics method has been employed, in which joint variables including displacement, velocity, and acceleration are known, and the forces and moments at the joints constitute the desired outputs to be computed (Jacob Apkarian et al., 1989). Therefore, the accuracy of the computational model is profoundly dependent on the accuracy of the input data sets. Therefore, this analysis was predicated on the accuracy of the input data. The objective of this study was to propose that the PBM be used to mitigate cumulative errors, and the error rate was analyzed via the application of these methods based on the LRS and SRS. However, the results of this study had the limitation that these results could not be achieved via the performance of practical clinical experiments. Therefore, in future study, this newly-developed method (PBM) requires validation by practical clinical experiments, in order to develop it further, into a useful method for the analysis and practical diagnosis of the patients.

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