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The comparison of joint kinematic error using the absolute and relative coordinate systems for human gait †

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Abstract

Minimizing artifacts from skin movement is vital for acquiring more accurate kinematic data in human movement analysis. There are several stages that cause skin movement artifacts and these stages depend on the selection of the reference system, the error reduction method and the coordinate system in clinical gait analysis. Due to residual errors, which are applied to the Euler and Bryant angle methods in each stage, significant cumulative errors are generated in the motion analysis procedure. Thus, there is currently a great deal of research focusing on reducing kinematic errors through error reduction methods and kinematic error estimations in relation to the reference system. However, there have been no studies that have systematically examined the effects of the selected coordinate system on the estimation of kinematic errors, because most of these previous studies have been mainly concerned with the analysis of human movement using only the human models that are provided in the commercial 3D motion capture systems.

Therefore, we have estimated the differences between the results of human movement analyses using an absolute coordinate system and a relative coordinate system during a gait, in order to establish which system provides a more accurate kinematic analysis at the ankle joint. Six normal adult subjects with no neurological or orthopedic conditions, lower extremity injuries, or recent history of lower extremity surgery were used in this study. The analysis was conducted at a walking speed of 1.35m/s. For the clinical estimation, we used a cardinal plane based on the segmental reference system and the differences were plotted on the planes. From this analysis, when a relative coordinate system was in the gait analysis, the average kinematic error occurring during the gait was determined to be 13.58mm, which was significantly higher than the error generated with an absolute coordinate system. Therefore, although the relative coordinate system can also be used to calculate the ankle joint center during the clinical gait analysis, the absolute coordinate system should be employed in order to obtain more accurate joint kinematic data. In addition, the results from this study can be used as a basis to select an appropriate coordinate system with regards to the diagnostic accuracy level required for various kinds of gait disorders.

Keywords: Gait analysis; Joint kinematic error; Absolute coordinate system; Relative coordinate system; Segmental reference system

1. Introduction

3D human movement analysis has long been considered a useful method for ergonomic, sports biomechanical, and gait analysis. In particular, clinical gait analysis has been adapted for the quantitative diagnosis of pathological gait, which is generally associated with a variety of functional deformities, muscle weakness, sensory loss, and impaired motor control. In the diagnosis of pathological gait, several factors are normally used in clinical gait analysis such as the relative angles of the joints, forces, moments and power. Thus, accurately measuring kinematic and kinetic factors at each joint is essential for the analysis of pathological

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gait mechanisms.

In general, the segments of the human body are assumed to behave as rigid bodies, but the soft tissue interposed between the skin and the bone consists of elastic and damping components that are typically exposed to inertial movements and to shape changes due to muscle activity. Therefore, extracting input data with a minimal amount of error is the most important factor to consider when selecting the appropriate method, which will ultimately be the one that minimizes errors associated with skin movement during the four movement analysis stages: the selection of a reference system, the application of error reduction methods, the selection of a coordinate system, such as an absolute or a relative coordinate, and the analysis procedure.

Skin movement artifacts pose a significant problem in this regard, since they tend to generate analytical errors in the movement analyses when 3D motion capture systems are used. A variety of previous studies have been performed with the goal of reducing errors caused by skin movement artifacts. However, skin movement errors cannot be minimized when only this reference system is utilized. Therefore, there is a need to develop novel methods for minimizing skin movement errors during data acquisition and analysis. A variety of studies have been conducted to establish such a system [1-7]. However, these error reduction methods are not reliably reproducible, because movement analysis is 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, generate significant cumulative errors in most traditional analysis procedures [8]. In addition, these previous studies conducted human movement analysis using only the human model provided in the commercial 3D motion capture systems. Therefore, they have overlooked the estimation of the kinematic error, which occurs by the selection of the coordinate system [9-12]. In other words, there have been no studies in which different coordinate systems were applied to human movement analysis to evaluate joint kinematic errors.

The main coordinate systems applied to human movement analyses are the absolute coordinate system (ACA) and the relative coordinate system (RCS). The ACA is a method of human movement analysis that is based on a global reference system, and the RCA is a human movement analysis method that is based on a relative coordinate system. From the perspective of dynamic systems, the absolute coordinate system is advantageous in that it can be used to simplify the resultant equations, which leads to more efficient and more accurate numerical solutions. The principal advantage of the relative coordinate system is that it can be used for an intuitive description of the physical system in the context of a multi-body dynamic system [13].

However, each of these coordinate systems only provides the same results when applied to a rigid body system that is in a static and dynamic state. Therefore, each coordinate system, when applied to the human body, contains a variety of sources of error, including skin movement errors and six degrees of freedom in the motion of the joint. Thus, when the trajectories of joint centers are calculated by ACS and RCS there is a lack of consistency between the two linked body segments. In addition, there is a possibility that the variability of the joint trajectories will be larger when using RCS, because of the cumulative error that is caused by skin movement artifacts. We believe that the quantitative evaluation of skin movement error via an applied coordinate system is necessary for clinical gait analysis.

Thus, the principal objective of this study is to give an estimation of the kinematic errors applied to ACS and RCS for more accurate kinematic data recovery at the ankle joint during clinical gait analysis, which can provide a basis for selecting an appropriate coordinate system with regards to the diagnostic accuracy level that is required in various kinds of gait disorders.

2. Methodology

2.1 Gait analysis

A 6-camera Vicon 460 (Oxford Metrics Limited, Oxford, UK) system equipped with an opto-electric motion analysis system was used to collect video data. During walking, the sample frequency of the camera motion system was 60Hz. The data was then filtered by using a fourth-order Butterworth, zero-lag, low-pass filter with a cut-off frequency of 7Hz [14]. This cut-off frequency was selected because 99.7% of the signal power was contained in the lower seven harmonics when the frequency was below 6Hz and the signal power at frequencies above 7Hz was regarded as noise [14]. Table 1 provides a description of the equipment used in this study.

The study group was comprised of six healthy subjects, all over the age of 20 (6 men; mean age ± stan-

Table 1. Specification of gait analysis apparatus.

Equipment	Maker	Model	Specification
Camera	VICON	MCam2	1. User selected frame rates: up to 1,000 fps 2. Pixel of digital CMOS sensor: 1,280 x 1,024 3. Resolution: 1,280 × 1,024 pixels
Force Plate	AMTI^{INC}	OR6-7	1. Available force: 4,450~17,800 (N)

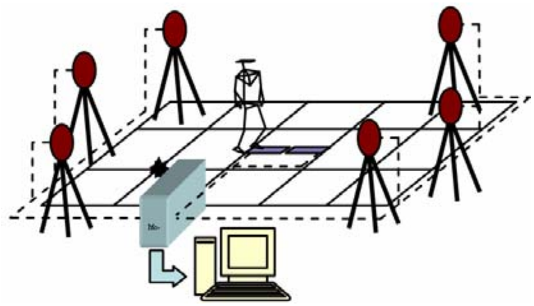


Fig. 1. Schematic view of the gait analysis apparatus.

dard deviation [SD], 25.4±2.9year; mean height, 173.1±4.6cm; mean mass, 71.8±7.5kg). 16 reflective markers were attached to anatomical landmarks on each subject and the subjects walked on a 9m walkway including a centrally placed, embedded force platform (sampling frequency of 1000Hz, Kistler, and type 9287). The average walking velocity was 1.35 m/s (±0.12m/s) and each subject performed five walking trials.

2.2 Absolute and Relative coordinate system

The vectors used to locate P in the XYZ frame for the absolute and relative coordinate system are provided in Eq. (1) and Eq. (2), respectively. r^{p_ab} is a vector from the shank reference frame to the ankle position and r^{p_re} is a vector from the thigh reference frame to the ankle position through the knee joint center and the shank local reference frame in the global XYZ reference system.

$$r^{p\_ab} = r_4 + A_2 s'_{2p} \tag{1}$$

$$r^{p\_re} = r_1 + A_1 s'_{12} - A_2 s'_{21} + A_2 s'_{2p} \tag{2}$$

where $\mathbf{r_1}$ and $\mathbf{r_4}$ are vectors from the global reference frame to the thigh and shank reference frames, $\mathbf{A_1}$ and

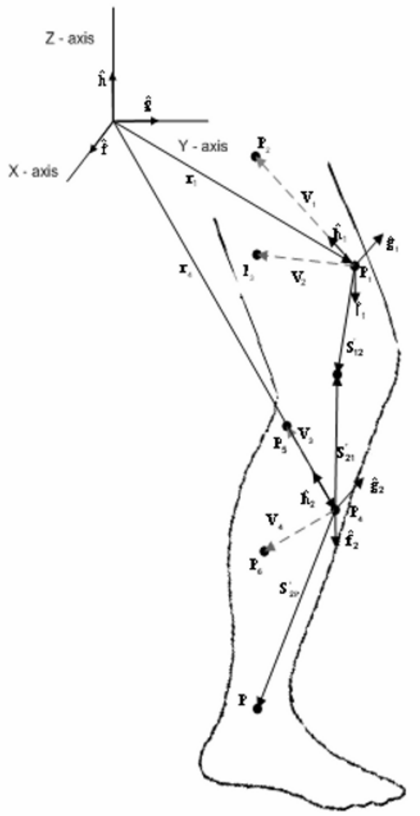


Fig. 2. Configuration of the Lower body (thigh & shank) of the Human.

$\mathbf{A_2}$ are the transformation matrices from each segment's local coordinate system to the global reference frame, and $\mathbf{s'_{12}}$, $\mathbf{s'_{21}}$ and $\mathbf{s'_{2p}}$ are constant vectors given in the segment's local coordinate system.

The most frequently used method to calculate 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 with three non-collinear sensors. The sensor points on each of the body segments (denoted as points $\mathbf{P_1}$, $\mathbf{P_2}$ and $\mathbf{P_3}$ on the thigh segment, and points $\mathbf{P_4}$, $\mathbf{P_5}$ and $\mathbf{P_6}$ on the shank segment) are then subjected to the transformation matrixes, $\mathbf{A_1}$ and $\mathbf{A_2}$, which relates the body reference frames to the global frame. These transformations can be expressed as follows:

$$A_1 = \begin{bmatrix} a_{11} & a_{12} & a_{13} \\ a_{21} & a_{22} & a_{23} \\ a_{31} & a_{32} & a_{33} \end{bmatrix} = \begin{bmatrix} \hat{f}_1 & \hat{g}_1 & \hat{h}_1 \end{bmatrix} \tag{3}$$

Where,

$$\begin{aligned} & \hat{h}_1 = \frac{v_1}{\|v_1\|} \\ v_1 = P_2 - P_1 & \\ v_2 = P_3 - P_1 \quad & \hat{f}_1 = \frac{x_1}{\|x_1\|} \\ x_1 = v_1 \times v_2 & \\ & \hat{g}_1 = \hat{h}_1 \times \hat{f}_1 \end{aligned} \tag{4}$$

and

$$A_2 = \begin{bmatrix} a'_{11} & a'_{12} & a'_{13} \\ a'_{21} & a'_{22} & a'_{23} \\ a'_{31} & a'_{32} & a'_{33} \end{bmatrix} = \begin{bmatrix} \hat{f}_2 & \hat{g}_2 & \hat{h}_2 \end{bmatrix} \tag{5}$$

Where,

$$\begin{aligned} & \hat{h}_2 = \frac{v_3}{\|v_3\|} \\ v_3 = P_5 - P_4 & \\ v_4 = P_6 - P_4 \quad & \hat{f}_2 = \frac{x_2}{\|x_2\|} \\ x_2 = v_3 \times v_4 & \\ & \hat{g}_2 = \hat{h}_2 \times \hat{f}_2 \end{aligned} \tag{6}$$

The transformation matrices can be expressed in terms of the Euler angles as follows:

$$A_1 = \begin{bmatrix} c\varphi_1 c\psi_1 - s\varphi_1 c\theta_1 s\psi_1 & -c\varphi_1 s\psi_1 - s\varphi_1 c\theta_1 c\psi_1 & s\varphi_1 s\theta_1 \\ s\varphi_1 c\psi_1 - c\varphi_1 c\theta_1 s\psi_1 & -s\varphi_1 s\psi_1 - c\varphi_1 c\theta_1 c\psi_1 & -c\varphi_1 s\theta_1 \\ s\theta_1 c\psi_1 & s\theta_1 c\psi_1 & c\theta_1 \end{bmatrix} \tag{7}$$

$$A_2 = \begin{bmatrix} c\varphi_2 c\psi_2 - s\varphi_2 c\theta_2 s\psi_2 & -c\varphi_2 s\psi_2 - s\varphi_2 c\theta_2 c\psi_2 & s\varphi_1 s\theta_1 \\ s\varphi_2 c\psi_2 - c\varphi_2 c\theta_2 s\psi_2 & -s\varphi_2 s\psi_2 - c\varphi_2 c\theta_2 c\psi_2 & -c\varphi_2 s\theta_2 \\ s\theta_2 c\psi_2 & s\theta_2 c\psi_2 & c\theta_2 \end{bmatrix} \tag{8}$$

Where, $c = cos$ and $s = sin$. Thus, the Euler angles for each segment can be determined from the following equations:

$$\cos\theta_1 = a_{33}, \quad \cos\theta_2 = a'_{33} \tag{9}$$

$$-\tan\phi_1 = \frac{\sin\phi_1}{-\cos\phi_1} = \frac{a_{13}}{a_{23}}, \quad -\tan\phi_2 = \frac{\sin\phi_2}{-\cos\phi_2} = \frac{a'_{13}}{a'_{23}} \tag{10}$$

$$\tan\psi_1 = \frac{\sin\psi_1}{\cos\psi_1} = \frac{a_{31}}{a_{32}}, \quad \tan\psi_2 = \frac{\sin\psi_2}{\cos\psi_2} = \frac{a'_{31}}{a'_{32}} \tag{11}$$

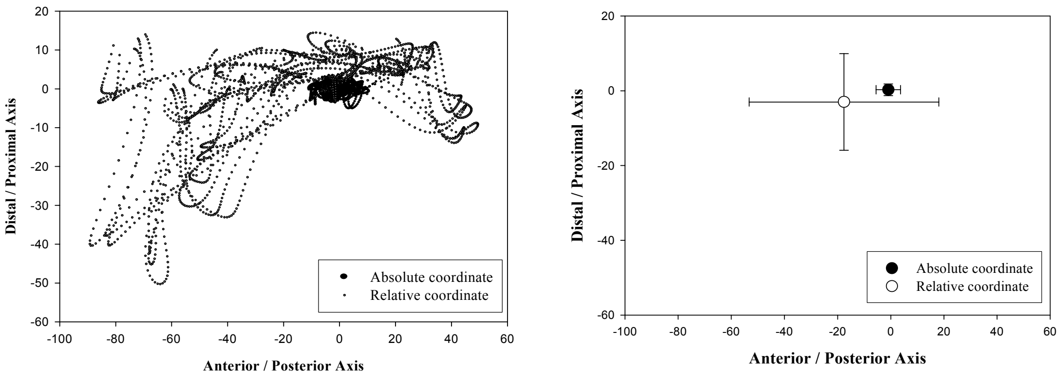
Finally, in order to estimate the accuracy of the analysis results of each coordinate system, we compared the variability of the ankle translational movement using ACS and RCS on the three cardinal planes, the sagittal, frontal and transverse planes, because the exaggerated variability of the ankle position on each plane denotes a high error ratio.

3. Result

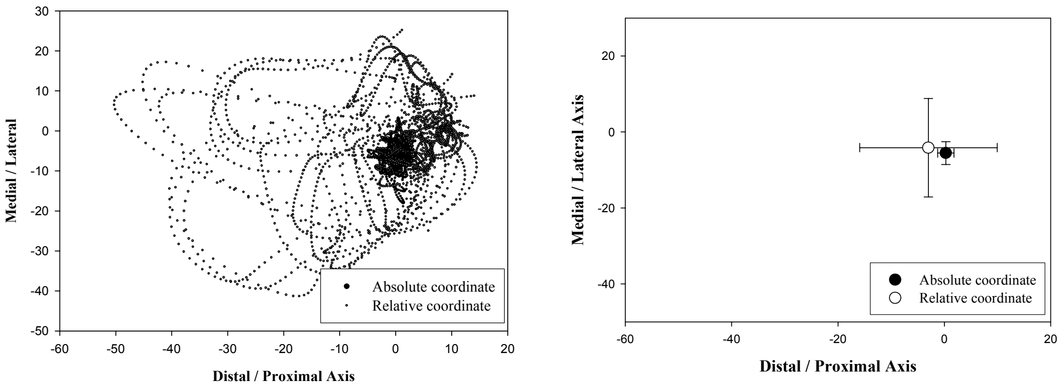
The variability of the ankle translational movement in each anatomical axis when the ACS and RCS were used for movement analysis is provided in Fig. 3. As mentioned previously, all of the analysis results are shown on the cardinal planes based on a segmental reference system to clinically estimate the results of analysis. In each figure, the graph on the left shows the ankle movement pattern of six subjects and the graph on the right shows the average and standard deviation of the ankle joint movement in the segmental reference frame of the shank segment.

As shown in the graphs on the right side, the variability of the ankle translational movement when using RCS is larger than that generated with ACS. In particular, the ankle joint movement indicates that there was a great degree of variability, approximately 60mm in the anterior/posterior direction and over 20mm in the other directions. However, the variability of the ankle joint movement when using ACS was at maximum 8mm, which was significantly less than the movement variability of the RCS.

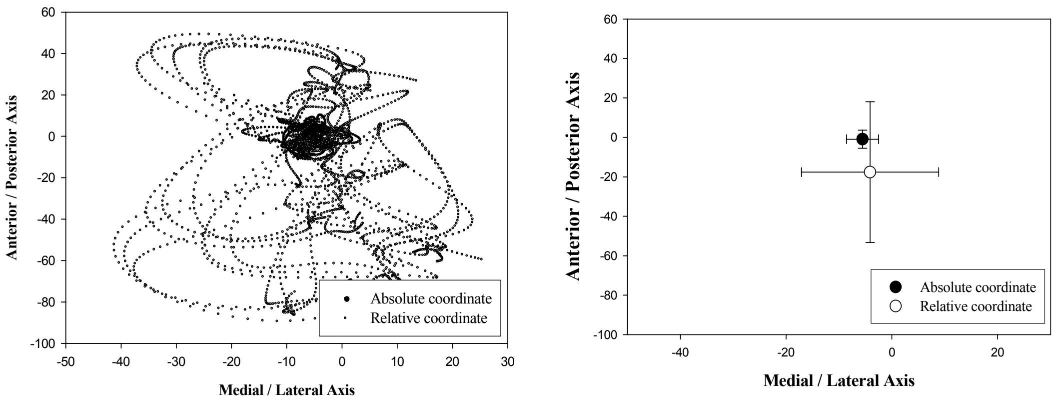
The mean values of the 3-D translational ankle joint motion determined from the ACS and RCS in the segmental reference system in each subject are shown in Table 2. The average value of the ankle joint translational motion measured when using ACS was 3.31mm in the anterior/posterior direction and the value obtained when using RCS was 20.82mm. In the cases of the distal/proximal and medial/lateral directions, the average values of the ankle joint translational motion determined in the ACS were less than those observed when using RCS.



(a) Sagittal plane



(b) Frontal plane



(c) Transverse plane

Fig. 3. The ankle joint movement pattern on the each cardinal plane (Sagittal / Frontal / Transverse planes).

Table 2. The average values of 3D translational.

Subject (mm)	Absolute coordinate system			Relative coordinate system		
	Anterior / Posterior	Distal / Proximal	Medial / Lateral	Anterior / Posterior	Distal / Proximal	Medial / Lateral
1	3.15	1.36	1.85	24.96	15.11	14.34
2	5.70	1.34	2.25	24.88	9.14	9.61
3	2.93	1.74	2.92	22.63	18.62	10.88
4	3.38	1.37	4.74	21.82	12.66	8.32
5	1.44	1.39	2.22	16.27	6.42	8.11
6	3.25	1.36	1.69	14.33	7.39	13.30
Average (SD)	3.31 (1.4)	1.43 (0.2)	2.61 (1.1)	20.82 (4.5)	11.56 (4.8)	10.76 (2.6)

*SD: Standard Deviation

4. Conclusions

Soft tissue artifacts are commonly considered the most troublesome source of error in human movement analyses. Although many researchers have attempted to reduce skin movement errors by selecting a proper reference system and a proper error reduction method, the cumulative error in the analysis procedures appears impossible to circumvent. Therefore, the primary objective of this study was to provide a coordinate system capable of minimizing cumulative errors by determining a proper reference system and a proper error reduction method. We estimated the kinematic error of the ankle joint when the ACS and RCS were used during the gait and determined which coordinate system minimized the residual error.

To compare the experimental results from 3D ankle kinematic error in both the ACS and RCS, a lower extremity model including the thigh, shank and foot segments was constructed in the initial step of this study. The results were then compared and analyzed in three cardinal planes in the segmental reference system. In the cardinal plane, the anatomical axes, the anterior/posterior, proximal/distal, and medial/lateral, encompassed the anatomical landmarks of the human body, and the ankle kinematic errors were estimated in the reference frame of the anatomical axes of the shank segment.

The experimental results of this study are summarized as follows:

(1) The ankle kinematic errors in the anterior/posterior and medial/lateral directions were determined to be at maximum 5.7mm and 4.74mm, respectively, when the ACS was used in step 3, the selection of the coordinate system. However, when the RCS was used, the errors in the anterior/posterior and medial/lateral directions were at maximum 24.96mm and 14.34mm, respectively. The kinematic errors in the distal/proximal direction were 11.74 and 8.62mm when the ACS and RCS were used, respectively. These results indicate that when the RCS was used the magnitude of error in the analysis occurred because of repetitive transformation in the thigh segment, knee joint, and shank segment.

(2) It was also apparent that when using the RCS the kinematic error was induced by knee joint translations on each anatomical axis, such as the anterior/posterior, medial/lateral, and distal/proximal axes. As a result, the cumulative error was caused not only by skin movement artifacts of the thigh and shank segments, but also by knee joint movement when RCS was applied to human movement analysis.

Human movement analysis has been used for a wide variety of reasons, and in the analysis an inverse dynamics method has been employed in which known joint variables, including displacement, velocity, and acceleration, are used to calculate the forces and moments that are applied to the joints during movement [15]. Therefore, the accuracy of the computational model is profoundly dependent on the accuracy of the input data [16]. To maximize the accuracy of the input data, a coordinate system that minimizes error should be selected. This is especially true in the case of clinical gait analysis, since the accuracy of the analysis results is exceedingly important to clinical decision-making for the diagnosis of patients. Therefore, in this study, we assessed the effects of different coordinate systems by comparing the kinematic analysis results of the ankle joint: the four steps of human movement analysis, the selection of a reference system, the application of error reduction methods, the selection of a coordinate system such as absolute and relative coordinate systems, and the analysis procedure. From this comparative analysis, it has become apparent that the magnitude of the kinematic errors caused by the selection of the coordinate system at the ankle joint depends on the whether the ACS or RCS are applied to gait analysis. In addition, we found that the joint kinematic accuracy obtained with the ACS is far higher than that attained by using RCS in the kinematic recovery of the ankle joint center. However, ACS and RCS should both be used in clinical gait analysis to achieve the level of diagnostic accuracy that is required in various kinds of different gait disorders, even though the joint kinematic accuracy using ACS is much higher than that of RCS. The reason is that the analysis of each pathological gait demands different analytical factors and diagnostic accuracy level. The results of this study are limited in that they were not obtained under practical clinical conditions. Therefore, in future studies, it will be necessary to conduct kinematic and gait analysis in each coordinate system under practical clinical conditions.

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