2009 IEEE International Conference on Control and Automation
Christchurch, New Zealand, December 9-11, 2009

Reliability of Measuring Human Segment Three-dimensional Orientation Using Wearable Sensor System

Kun Liu, Tao Liu, Kyoko Shibata, Yoshio Inoue

Abstract – **A new method for analyzing human segment three-dimensional orientation is presented. A wearable sensor system based on triaxial accelerometers alone was developed, and later was tested on a thigh to calculate the pitch and yaw angles of the thigh with the accelerations captured by the system. Wearing the developed system, eight volunteer subjects walked in the work space of a high-accuracy camera system at three self-selected slow, normal and fast walking speeds. Calculated pitch and yaw angles of the thigh in the local frame were compared with those measured from high accuracy camera system. The mean differences (e_{mean}) of the calculated angles range from 0.4478° to 1.1351° in the Yr direction and from 0.5667° to 0.8973° in the Xr direction at three different speeds. The correlation coefficients (R) were above 0.8113 in the Yr direction and above 0.9031 in the Xr direction across all conditions. The result shows that the wearable sensor system gives a reasonably accurate measurement of the pitch and yaw angles of the lower limb segment. Therefore, the original method using only one kind of sensor (accelerometer), since it's accurate, inexpensive and simple, is suitable for the evaluation of human segmental orientation, for the purpose of assessing spatio-temporal gait parameters and monitoring the gait function of patients.**

Keywords: **Segmental orientation, Triaxial accelerometer, Gait analysis**

I. INTRODUCTION

In the medical field, parameters of human motion, especially the orientations of lower limb segments are very important for clinicians to determine suitable treatments for patients [1, 2]. Gait analysis has become an effective tool for quantifying surgical intervention effects and evaluating patients' conditions [3]. Therefore, it is essential to detect the orientations of lower limb segments in biomechanical applications.

In the laboratory, an optical motion capture system for kinematic data combined with force platforms for kinetic data constitute a complete gait analysis system, but it has limitations on the volume, weight, price and moving area for the observed subject. It takes a large space and almost impossible to be used outside the laboratory and it can only capture a few gait cycles. Therefore, it can be only used as an accurate reference system but not a daily wearable and portable surgical device.

With the development of the micro-machined sensors, parameters of human motion can be measured outside a specialized laboratory with ambulatory sensor-based systems. Various kinematic sensor techniques and sensor-based wearable systems have been developed for studying gait analysis [4-9], which are low-power, low-price or low-volume, and viable in capturing kinematic and kinetic data of human limbs.

As far as the measurement of certain lower limb segment orientation, joint angle has been estimated using accelerometers and/or gyroscopes [10, 11]. However, when accelerometers were used to measure the accelerations of human lower limbs, the measured accelerations along each sensitive axis were vector sum which composed of gravitational and linear acceleration components and noise [12]. The actual resultant acceleration cannot be simply integrated to predict the rotational angular velocity and displacement of the segment, since it contains not only translational and rotational acceleration components, but also a gravitational acceleration component and noise [13]. If the angular displacement of the rotational segment is calculated using the equation $\theta=(180/\pi\cdot arcsin(a_z\ /g)$, the measured subject must hold still, or the linear acceleration component must be neglected. Hooman Dejnabadi et al. [14] obtained the angular displacement of the rotational segment by numerically integrating the angle velocity captured by a gyroscope, but integrated result was distorted by offsets or drifts. H. J. Luinge et al. [15] presented another method, but the gyroscope offset must be continuously recalibrated and the orientation must be continuously corrected. H. Dejnabadi et al. [16] gave a method to solve the contradiction, getting the angular displacement with both accelerometer and gyroscope, and switching between the two sensors according to the wave frequency of the body segment. However, it was

Manuscript received April 29, 2009. This work was supported by Mechanical and Dynamic Lab in Kochi University of Technology Japan.

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hard to give an accurate switching frequency. Karol J. et al. [17] presented a methodology for joint angle measurement, but it combined three kinds of sensors (gyroscope, accelerometer and magnetometer), and the study was limited to a static system (no global rotational or linear acceleration existed). Therefore, a complementary approach should capture and analysis data with less number and complexity of sensors, and should calculate more parameters of the lower limb motion rather than directly measuring the required data by various sensors.

In this paper, a novel method was presented and expatiated to estimate orientation of the lower limb segment. To validate the feasibility of the method, a wearable detection system based on only three accelerometers for ambulatory recordation of lower limb segment orientation was developed. Then, to capture the gait motion parameters using both the developed device and a high accuracy camera system, the device was worn on thighs of volunteer subjects, and the subjects walked in the working space of the camera system. The angular displacements of the thighs in the local fame were calculated, and the reliability of the developed device was analyzed compared with the camera system.

II. METHODS AND MATERIALS

A. Calculation of the angle displacement and angle velocity in 2-D frame with a fixed origin

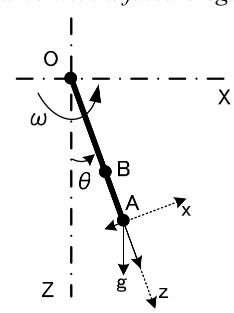

Fig.1 Illustration of a rigid body swinging in *xoz* reference frame

When a rigid body rotates in 2-D reference frame *xoz* about a fixed rotation point, at two points A and B, the actual acceleration are composed with rotational acceleration and gravitational acceleration.

At the point A,

$$a_{Ax} = \ddot{x} + g\sin\theta = r_A\ddot{\theta} - g\sin\theta \tag{1}$$

$$a_{Az} = \ddot{z} + g\cos\theta = -r_A\dot{\theta}^2 + g\cos\theta \tag{2}$$

At the point B,

$$a_{Bx} = \ddot{x} + g\sin\theta = r_B\ddot{\theta} - g\sin\theta \tag{3}$$

$$a_{Bz} = \ddot{z} + g\cos\theta = -r_B\dot{\theta}^2 + g\cos\theta \tag{4}$$

Based on $(1)\cdot r_B - (3)\cdot r_A$ and $(2)\cdot r_B - (4)\cdot r_A$, the following equations were obtained:

$$r_B a_{Ax} - r_A a_{Bx} = (r_B - r_A) g\sin\theta \tag{5}$$

$$r_B a_{Az} - r_A a_{Bz} = (r_B - r_A) g\cos\theta \tag{6}$$

Then the angular displacement was obtained as follow.

$$\theta = \sin^{-1}\left(\frac{a_{Ax}r_B - a_{Bx}r_A}{g(r_A - r_B)}\right) \text{ or } \theta = \cos^{-1}\left(\frac{a_{Az}r_B - a_{Bz}r_A}{g(r_B - r_A)}\right) \tag{7}$$

Based on (1)-(3) and (2)-(4), the angular acceleration and angular velocity of the rotational rigid body were shown as follows:

$$\dot{\omega} = \ddot{\theta} = \frac{a_{Ax} - a_{Bx}}{r_A - r_B} \tag{8}$$

$$\omega^2 = \dot{\theta}^2 = \frac{a_{Az} - a_{Bz}}{r_B - r_A} \tag{9}$$

As was shown above, it was obvious that angular displacement, angular velocity and angular acceleration of a rotational rigid body can be obtained only using the actual accelerations at two points, which can be measured only using accelerometers instead of combination with accelerometers, gyroscopes and other kind of sensors. Then, the method can be extended to measure 3-D orientation of thigh to do gait analysis.

B. Explanation of the novel approach for the thigh orientation

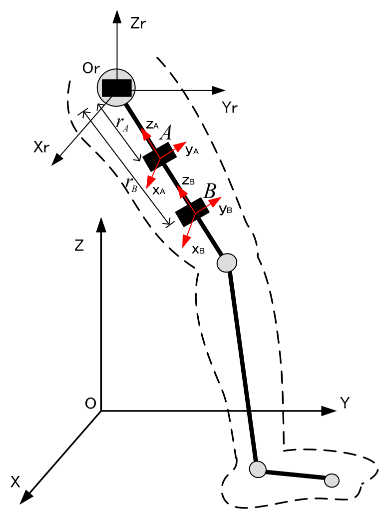

Fig.2 Schematic of the lower limb with accelerometer-based wearable sensor system on the thigh during ordinary steps

To identify the human segment three-dimensional orientation, the relative attitude of the thigh compared with the trunk in three-dimensional space was analyzed, i.e. the orientation of the thigh in the local coordinate system was calculated. As Fig.2 shows, the following systems of coordinates were introduced to analyze the orientation of the thigh:

1, O-XYZ is global frame with the axes X, Y, and Z pointing outward, upward, and forward.

2, O_r-$X_rY_rZ_r$ is local coordinate system with the origin O_r fixed at the hip joint. Suppose it maintains constant orientation in space with the axes parallel to the axes of the global system. In the joint rotation convention for a hip joint

[19], the X_r axis describes the flexion-extension motion; the Y_r axis is used to measure abduction and adduction.

3, A-$x_A y_A z_A$ and B-$x_B y_B z_B$ are the sensor coordinate frame systems for two accelerometers which are attached to the thigh at position A and B. The x_A axis is perpendicular to the sagittal plane of the femur; the z_A axis is the long axis of the thigh segment (along the femur); the y_A axis is orthogonal to both axis x_A and z_A. The axes in the sensor frame B-$x_B y_B z_B$ direct the same orientation with the axes of A-$x_A y_A z_A$ correspondingly

To analysis the lower limb segmental orietation, the flexion-extension angle θ_X and abduction-adduction angle θ_Y are used to describe the orientation of the thigh in the local coordinate frame.

When the subjects perform straight-line walking trials, the origin O_r (the hip joint) of the local coordinate system is not fixed in the global frame. In this paper, only the translational acceleration on the hip joint was considered. The actual resultant accelerations on the thigh were composed of translational acceleration, rotational acceleration and gravitational acceleration. The composition of the acceleration ($\mathbf{a}_A$) at point $\boldsymbol{A}$ which can be measured by an accelerometer is formulated as follows:

$$\mathbf{a}_A = \dot{\boldsymbol{\omega}} \times \mathbf{r}_A + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \mathbf{r}_A) + \mathbf{R}_{\theta_Y} \cdot \mathbf{R}_{\theta_X} \cdot (\mathbf{g} + \mathbf{a}_{O_r}) \tag{10}$$

where $\mathbf{r}_A$ is the position vector of point $\boldsymbol{A}$ in the local coordination system, $\boldsymbol{\omega}$ is the angular velocity ($\boldsymbol{\omega} = \dot{\boldsymbol{\theta}}$, $\dot{\boldsymbol{\omega}} = \ddot{\boldsymbol{\theta}}$), $\mathbf{g}$ is the gravitational acceleration vector, $\mathbf{a}_{Or}$ is the translational acceleration of the hip joint in the global coordinate system, $(\mathbf{g} + \mathbf{a}_{O_r})$ is the actual acceleration measured with an accelerometer at the hip joint which will be indicated with $\mathbf{A}_{Or}$ in the following equations. $\mathbf{R}_{\theta_Y}$ and $\mathbf{R}_{\theta_X}$ are rotation matrices by angles θ_X and θ_Y from the global coordination system to the local coordinate system.

As Fig.2 shows, two triaxial accelerometers were fixed at two positions $\boldsymbol{A}$ and $\boldsymbol{B}$ on the thigh with corresponding axes in the same direction, therefore, two sets of actual accelerations $\mathbf{a}_A$ and $\mathbf{a}_B$ at positions $\boldsymbol{A}$ and $\boldsymbol{B}$ were measured. The rotational radiuses about the local origin O_r were measured as r_A and r_B, the following the equation can be obtained with the rotational radiuses and the measured accelerations:

$$\begin{aligned} &\mathbf{a}_A \cdot r_B - \mathbf{a}_B \cdot r_A \\ &= \left[\dot{\boldsymbol{\omega}} \times \mathbf{r}_A + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \mathbf{r}_A) + \mathbf{R}_{\theta_Y} \cdot \mathbf{R}_{\theta_X} \cdot \mathbf{A}_{O_r}\right] \cdot r_B \\ &\quad - \left[\dot{\boldsymbol{\omega}} \times \mathbf{r}_B + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \mathbf{r}_B) + \mathbf{R}_{\theta_Y} \cdot \mathbf{R}_{\theta_X} \cdot \mathbf{A}_{O_r}\right] \cdot r_A \end{aligned} \tag{11}$$

According to the method used to calculate the angular displacement, by solving the equations set with $\mathbf{a}_A$ and $\mathbf{a}_B$ in the equation (11), the angular displacements about the X_r and Y_r axes in the local frame were calculated and shown as follow:

$$\theta_x = \sin^{-1}\left(\frac{r_B a_{Ay} - r_A a_{By}}{(r_B - r_A)\sqrt{A_{O_r y}^{\;2} + A_{O_r z}^{\;2}}}\right) - \sin^{-1}\left(\frac{A_{O_r y}}{\sqrt{A_{O_r y}^{\;2} + A_{O_r z}^{\;2}}}\right) \tag{12}$$

$$\theta_y = \sin^{-1}\left(\frac{r_B a_{Ax} - r_A a_{Bx}}{(r_B - r_A)\sqrt{A_{O_r x}^{\;2} + A_{O_r y}^{\;2} + A_{O_r z}^{\;2} - \left(\dfrac{r_B a_{Ay} - r_A a_{By}}{r_B - r_A}\right)}}\right) - \sin^{-1}\left(\frac{A_{O_r x}}{\sqrt{A_{O_r x}^{\;2} + A_{O_r y}^{\;2} + A_{O_r z}^{\;2} - \left(\dfrac{r_B a_{Ay} - r_A a_{By}}{r_B - r_A}\right)}}\right) \tag{13}$$

where θ_X is the flexion-extension angular displacement (pitch angle) about the X_r axis; θ_Y is the abduction-adduction angular displacement (yaw angle) about the Y_r axis; ($A_{O_r x}$, $A_{O_r y}$, $A_{O_r z}$) is the measured acceleration ($\mathbf{A}_{Or}$) of the local frame origin (hip joint) in the global frame.

As was shown in (12) and (13), it was obvious that angular displacements of the thigh in 3-D coordinate system can be obtained only using three actual accelerations ($\mathbf{A}_{Or}$, $\mathbf{a}_A$ and $\mathbf{a}_B$) at three points (the hip joint and two arbitrary points on the thigh), which can be measured only with accelerometers instead of combination with accelerometers, gyroscopes and other kind of sensors.

C. Experiment Design

In order to evaluate the feasibility of the presented approach, a wearable and removable elementary system was developed. Three triaxial accelerometer-based chips (MM-2860) and one MCU (H8/3694, from Renesas Technology Corp.) were adopted in the system as fig.3 showed. The MCU was used to capture accelerations from the triaxial accelerometers, store data in the EEPROM real time and communicate with PC after each test.

To capture the accelerations at two points on the thigh, two accelerometers were attached on a board with the corresponding axes in the same direction, and the distance between them was 50mm. Since it was hard to fix the accelerometer on the hip joint exactly and, the hip joint moved together with the trunk at the same speed, the third accelerometer for measuring the translation accelerations of the local frame origin was fixed above the hip joint, on the waist, where more stable accelerations could be captured than on the hip joint. The distance (r_A) from the hip joint to the first accelerometer on the thigh was measured with a ruler in each trial.

Eight subjects (6 males, 2 females, Age: 25±3 yeas, Height: 170±5cm, Mass: 60±11kg) with no history of musculoskeletal pathology and injury were requested to perform three straight-line walking trials at self-selected slow, normal and fast walking speeds. Three groups of three-dimensional accelerations at three positions were measured using the wearable sensor system during each test. At the same time, three marks were fixed on the thigh so that the angular displacements of the thigh in the X_r and Y_r

directions were captured by the high accuracy camera system, which were used as references.

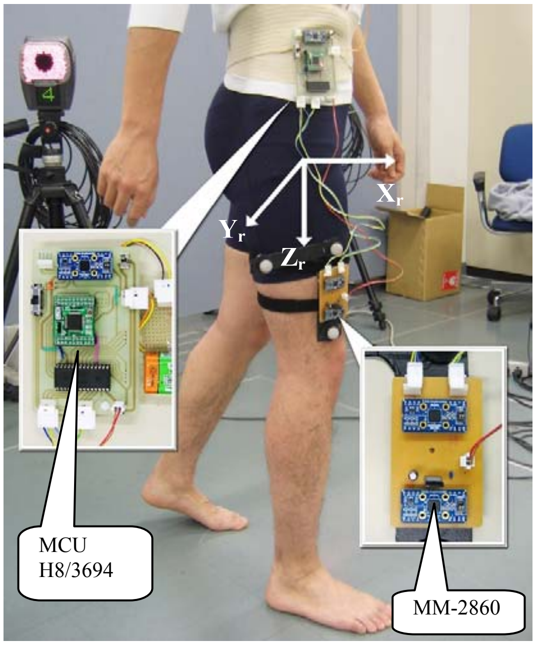

Fig.3 Experiment with the developed wearable sensor system and the high accuracy camera system. In each test, the subject was request to make straight-line walking in the working space of the camera system

III. RESULTS

All signals captured in the experiment were off-line processed by Matlab. A low-pass filter with a cut-off frequency of 20 Hz was used to remove noise from all the initial data.

The curves in Fig.4 and Fig.5 showed one group of the calculated thigh angular displacements about the X_r and Y_r axes in the local frame, when the subject performed straight-line walking at the self-selected slow and fast speed. The referenced angular displacements were captured by the high accuracy camera system (the calibration error was 0.22% for capturing motion parameters of the moving marks on the thigh).

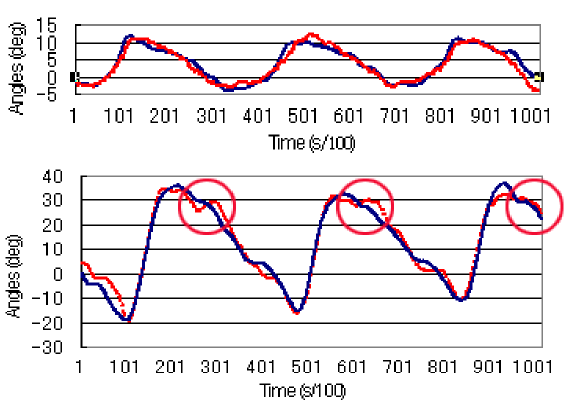

Fig.4 The Calculated (red dotted line) and referenced (blue real line) angular displacements about the X_r and Y_r axes at the self-selected slow speed

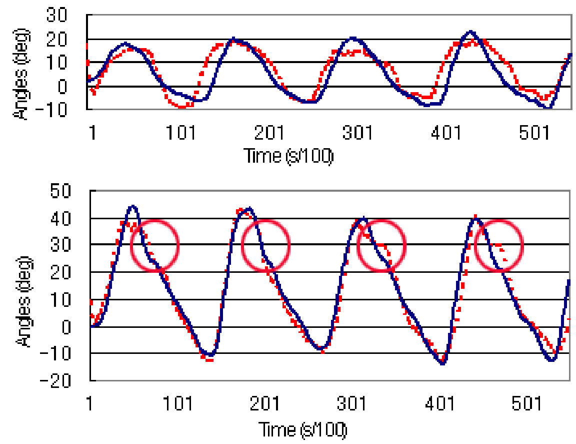

Fig.5 The calculated (red dotted line) and referenced (blue real line) angular displacements about the X_r and Y_r axes at the self-selected high speed

IV. DISCUSSION

Root of the mean of the square differences (RMS) was used to compare the closeness in amplitude of two measurement results captured from the developed wearable system and the referenced camera system. The correlation coefficient (R) was used as a measure of the association between the two systems. The angular displacements in the X_r and Y_r directions from the high accuracy camera system were used as references:

$$RMS = \sqrt{\left(\frac{1}{n}\sum(\theta-\theta_r)^2\right)} \tag{14}$$

$$R = \frac{\left(n\sum\theta\theta_r - \sum\theta\sum\theta_r\right)}{\sqrt{\left(n\sum\theta^2-(\sum\theta)^2\right)\left(n\sum\theta_r^{\ 2}-(\sum\theta_r)^2\right)}} \tag{15}$$

where θ is the angle measured with the developed sensor system, θ_r is the referenced angle given by the camera system, n is the number of the sampling.

In the experiment, the distance (r_A) from the hip joint to the first accelerometer on the thigh depended on where the accelerometer was fixed, and it was different in each test. The range of the distance measured in each test was 200±50mm. The comparison parameters (averages for 8 experiments with 8 subjects at a certain speed of the three) between referenced and calculated angular displacements in the X_r and Y_r directions of each case were indicated in table 1 and table 2, where, e_{mean} is the mean difference, ROM is the range of the the rigid body motion in each direction.

Table 1, Gait velocity conditions (mean±SD) and parameters for analyzing the difference between the referenced and calculated flexion-extension angular displacements captured from the high accuracy camera system and the developed wearable sensor system.

Walking speed(m/s)	$e_{mean}(^\circ)$	$RMS_{\theta y}$	$R_{\theta y}$	ROM(°)
Slow (1.05±0.15)	0.4778	2.4469	0.9321	30°
Preferred (1.40±0.15)	0.7637	3.5379	0.8956	30°
Fast(1.90±0.25)	1.1351	4.1135	0.8113	30°

Table 2, Gait velocity conditions (mean±SD) and parameters for analyzing the difference between the referenced and calculated abduction-adduction angular displacement measured from the high accuracy camera system and the developed wearable sensor system.

Walking speed(m/s)	$e_{\mathrm{mean}}(^{\circ})$	$\mathrm{RMS}_{\theta x}$	$\mathrm{R}_{\theta x}$	$\mathrm{ROM}(^{\circ})$
Slow (1.05±0.15)	0.5667	3.0403	0.9619	40°
Preferred (1.40±0.15)	0.6632	4.1278	0.9371	40°
Fast(1.90±0.25)	0.8973	4.9031	0.9039	45°

As it was shown in table 1 and table 2, in the Y_r direction of the local frame, the rotational angular displacements of the thigh ranged between -30° and 30°, all absolute value of the mean differences ranged from 0.4778° to 1.1351° and all correlation coefficients were greater than 0.8113 across all test conditions. In the X_r direction of the local frame, the rotational angular displacements ranged between -40° and 45°, all absolute value of the mean differences ranged from 0.5667° to 0.8993°, and all correlation coefficients were greater than 0.9039.

As the data showed, when the subjects walked at different speeds, greater speeds resulted to greater e_{mean} and RMS, and smaller correlation coefficient. Since it was hard to firmly fix the wearable sensor system on the thigh like on a rigid body without any relative motion exactly, the errors increased with the increase of the walking speed. Especially when the subjects walked at the high speed, if the circuit board with the two accelerometers on the thigh was fixed loosely, there would be relative motion between the board and the thigh, besides, skin motion artifact and muscle activities also resulted to relative motion which brought noise to the captured accelerations of the femur. Due to the motion of the hip joint, the third accelerometer for measuring the accelerations of the local frame origin was not exactly fixed on the hip joint, and the z axis of the acceleration also could not be vertically placed on the waist exactly as the principle showed, all of which also brought errors to the calculated result. As the walking speed increased, the non-translation acceleration of the local frame origin also increased, which resulted to grater errors. When the three accelerometers were fixed on the human body before every test, there were also some differences in the directions of the corresponding axes, which should be in the same directions exactly.

As Fig.4 and Fig.5 showed, when the subject performed three walking trials at three different speeds, at the terminal swing of each stride, the limb began active deceleration and the yaw and pitch rotational angular displacement of the thigh reached the maximum gradually [18]. When the human body moved ahead continuously, the hip joint also advanced at the same speed accordingly. The relative angular displacement of the thigh in the local frame had decreased before the foot touched the floor. At the same time, since the body weight was accepted by this limb, the reaction force between the foot and the floor increased immediately, i.e. the acceleration in the radial direction increased heavily, therefore the difference between the calculated and referenced result increased much more greatly than other phases action of the stride, as the circled parts of the curves showed in Fig 4 and Fig 5.

Since the subjects in the experiment consisted of a group of healthy young males and females with normal gait, the results should not be generalized to patients, where systematic errors and measuring errors respectively due to axis misalignment and complexity of patient gait are likely to be more significant. Therefore, in the future, more studies are necessary to determine reliability and validity in more diverse groups, especially the clinical populations. And further study for identifying the type and duration of activities should be done.

V. CONCLUSION

There was a clear relationship between the calculated and referenced angular displacements when two triaxial accelerometers were attached at two positions on the thigh, with the corresponding axes in the same direction. The errors varied with the walking speeds of the subjects. In the analysis of the result from the developed wearable sensor system and the high-accuracy camera system, the novel method present in the paper was feasible for ambulatory recording of human body segment 3D accelerations, and calculating the orientation of the lower limb segment when the subject performed normal walking in the daily life.

Results of the study suggested that with simple calculations and fewer kinds and quantities of sensors (only three triaxial accelerometers), orientation of a human segment can be acquired. The technology will be useful to provid simple, rapid and objective information for the researcher or clinician about key gait-related variables.

Since there were no differentiation and integration in the approach, the result was not distorted by offsets or drifts. And the system can obtain continuous orientation angles of the thigh without switching between different sensors at different wave frequencies of the body segment. In addition, since the two accelerometers were fixed in the same condition at two different positions on the thigh, the difference of the measured accelerations was just resulted by the rotational accelerations which related to the orientation angles, therefore, the result based on simple operation of the difference between the accelerations was not affected by the noise acted on the two sensors simultaneously.

In the future, it is feasible to integrate only two triaxial accelerometers on one micro-electronics chip to do human gait analysis during daily activities of patients or persons who need health evaluation and an integrated wearable sensor can be developed with high precision.

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