Ambulatory Estimation of 3D Lower Limb Gait Posture in Anatomical Coordinate Frame using Wearable Sensor System

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Analysis of the lower limb posture is important in the quantitative assessment of lower limb joint disorders and in the design of better prosthetic devices. To estimate the three dimensional (3D) kinematics of lower limb joints, the most widely used method was based on the modeling of each lower limb joint as a sequence of three hinges, which depended on the sequence order of the Euler angles, and integration of the angular velocity would result to distortion by offset and angle drift. In this paper, a method to estimate 3D kinematics of the knee joint for the lower limb gait posture in anatomical coordinate frame was proposed and tested.

An original approach for ambulatory estimation of knee joint kinematics in anatomical coordinate system was presented, which was composed of physical-sensor difference based algorithm and virtual-sensor difference based algorithm. To test the approach, a wearable monitoring system composed of accelerometers and magnetometers was developed and evaluated on lower limb. The flexion/extension (f/e), abduction/adduction (a/a) and inversion/extension (i/e) rotation angles of the knee joint in the anatomical joint coordinate system were estimated. The three knee joint angles within the anatomical coordinate system were independent of the orders which must be considered when Euler angles were used. Besides, since there were no physical sensors implanted in the knee joint based on the virtual-sensor difference based algorithm, it was feasible to analyze knee joint kinematics with less numbers and kinds of sensors than ever before. Compared with result from the reference system, the developed wearable sensor system was available to do gait analysis with fewer sensors and high degree of accuracy.

The wearable sensor system was composed of two pieces of MAG3 (analog inertial sensor) and two MM-2860 (accelerometer) and tested on the thigh and shank to estimate kinetics of knee joints for the lower limb gait posture analysis. Five subjects were asked to walk in the effective area of the referenced camera system. Each parameter obtained from the wearable sensor system and the referenced system was compared and the result showed that the method is simple and accurate for lower limb gait posture analysis.

The method is not limited by the sequence order of the Euler angles, that is, rotations about each axis in the anatomical coordinate system are independent. Since there was no integration of angular acceleration or angular velocity, the result was not distorted by offset and drift. Therefore, the method is suitable for the quantitative assessment of the anatomical gait posture of the patients and gives useful evaluation information for the clinicians. And it can also be used in other conditions, such as measuring the posture of rigid segments or upper limb.
AMBULATORY ESTIMATION OF 3D LOWER LIMB GAIT POSTURE IN ANATOMICAL COORDINATE FRAME USING WEARABLE SENSOR SYSTEM

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ABSTRACT
Knee joint kinematics analysis using an optimal sensor set and a reliable algorithm would be useful in the gait posture analysis. An original approach for ambulatory estimation of knee joint kinematics in anatomical coordinate system was presented, which was composed of physical-sensor difference based algorithm and virtual-sensor difference based algorithm. To test the approach, a wearable monitoring system composed of accelerometers and magnetometers was developed and evaluated on lower limb. The flexion/extension (f/e), abduction/adduction (a/a) and inversion/extension (i/e) rotation angles of the knee joint in the anatomical joint coordinate system were estimated. In this method, since there was no integration of angular acceleration or angular velocity, the result was not distorted by offset and drift. And the three knee joint angles within the anatomical coordinate system were independent of the orders which must be considered when Euler angles were used. Besides, since there were no physical sensors implanted in the knee joint based on the virtual-sensor difference based algorithm, it was feasible to analyze knee joint kinematics with less numbers and kinds of sensors than ever before. Compared with result from the reference system, the developed wearable sensor system was available to do gait analysis with fewer sensors and high degree of accuracy.

1. INTRODUCTION
Ambulatory estimation of lower limb posture is very important in the diagnosis of patients with stroke, Parkinson or knee osteoarthritis disease [1][2], and useful to evaluate the rehabilitation of patients. Based on the kinematic and kinetic data of the lower limb, proper treatment for the patients can be chosen by the clinicians [3]. A complete understanding of joint kinematics is the key point for the lower limb posture analysis. In the lab, the typical system for gait analysis is the optical motion system for kinematic data combined with force platforms for kinetic data [4]. However, since the system is bulky and expensive, it is not applicable for the out-lab ambulatory estimation of lower limb posture in the daily life.

With the development of more and more advanced inertial sensors, many algorithms for calculating the parameters of human kinematics were put into practice, and more and more wearable sensor systems based on inertial sensors were developed for gait analysis and clinical application[5-7], kinematics data for gait analysis such as knee joint angles, thigh and shank orientations were obtained [8-10].

In the diagnosis of joint disorders resulting from injury or disease, in the quantitative assessment of treatment, and in the general study of locomotion, the 3D knee joint kinematics and the thigh and shank orientation are crucial. J. Faver et al. [11][12] developed an ambulatory system to measure the 3D knee angles by filtering and integrating the gyroscope signals from the thigh and shank, the f/e and a/a rotational angles were obtained. However, the data derived by integration of angular acceleration or angular velocity was distorted by offset and angle drift, and a proper calibration was not given. H. Dejnabadi et al. [4][15] gave a new approach to estimate sagittal kinematics of lower limbs without accumulation of errors. Virtual accelerometers were fixed in the knee joint center and ankle to measure the joint rotational angles using external skin-mounted accelerometer and gyroscope, but only the f/e joint angle in the sagittal plane was estimated. K. O’Donovan [13] presented a technique which used a combination of rate gyroscope, accelerometer and magnetometer to measure 3D inter-segment joint angles, but the investigation of the performance of the technique was limited to a static system, and there was no evaluation for dynamic system. R. Takeda [14] proposed a novel method for measuring human gait posture using tri-axial accelerometers and gyroscopes, in which the optimization algorithm used for estimating gravitational acceleration gave an optimal lower limb gait posture. However, since the algorithm involved searching for large number of combinations, it was not suitable for small computing devices. Willemensen et al. [16] gave a method using only accelerometers to estimate the rotational angles of lower
extremities without integration, but the movements of the lower limb joints were only analyzed in the sagittal plane and thus two-dimensional.

We have recently presented a double tri-axial accelerometers based approach to measure the lower limb orientation angles [7]. In the paper, only one kind of motion sensor (accelerometers) was used to estimate the angles for lower limbs orientation in 3D space. The algorithm was based on the difference between two groups of accelerations without integration, which was simple and time-saving for micro board. However, the method was only evaluated in a restrained condition that the lower limb segments were assumed to be rigid segments and the subjects walked in a straight forward way with very little trunk sway, skin artifacts and no significant i/e rotation of the leg, thus not a true 3D analysis of the joint kinematics.

For a further application of the double-sensor difference based algorithm to estimate 3D lower limb joint kinematics for gait analysis, in this paper, an original approach based on accelerometers and magnetometers for ambulatory estimation and analysis of 3D knee joint kinematics was presented. The f/e angle, a/a angle and i/e rotation angle in the anatomical joint coordinate system were estimated. A wearable sensor system composed out of two MAG’s (inertial measurement unit composed out of an accelerometer, a magnetometers and a gyroscope) and two MM2860s (accelerometer) was developed, then tested on the lower limb. Without integration of angular acceleration or angular velocity for 3D lower limb joint kinematic analysis, the calculated result was not distorted by offset and drift. And the knee joint kinematic parameters within the anatomical coordinate system were independent of the order of the joint rotations.

2. METHODS

Estimation of 3D kinematics of inter segment joint is based on the modeling of the lower limb joint as a ball and socket joint permitting three degrees of freedom (DOF) rotations: f/e, a/a, and i/e rotation. To analysis the knee joint angles for lower limb posture, a novel method composed of physical-sensor difference based algorithm and virtual-sensor difference based algorithm is explicated.

The acceleration measured on a lower limb segment is a resultant acceleration signal containing gravitational acceleration, translational acceleration, rotational acceleration and noise, which cannot be separated directly. Besides, since the sensors are attached on the lower limb, the skin motion artifact due to impact loading and muscle activation can readily contaminate the measured signal. However, if two accelerometers are attached at two different positions (different distance to the rotational joint) with every two corresponding axes in the same direction, the gravitational acceleration, translational acceleration, skin motion artifact and other noise acting on the two sensors are the same except the rotational accelerations. To exploit the difference between the rotational accelerations, a simple algorithm based on the difference between double physical sensors, named physical-sensor difference based algorithm here, is explicated to estimate the rotational angles of lower limb segments.

\[
a_{ij} = g - \mathbf{R}_{ij} - \mathbf{r}_{ij} \omega_i \times \omega_j
\]

where \(a_{ij}\) is equivalent acceleration at point \(P_{ij}\), \(i\) is segment index, \(j\) is point index (\(j=0\): origin of segment), \(g\) is gravitational acceleration, \(\mathbf{R}_{ij}\) is the position of point \(P_{ij}\) relative to the global coordinate system (O-XYZ).

Then we can get

\[
a_{ai} = g - \mathbf{R}_{ai} - \mathbf{r}_{ai} \omega_i \times \omega_i
\]

\[
a_{ai} = g - \mathbf{R}_{ai} - \mathbf{r}_{ai} \omega_i \times \omega_j
\]

\[
a_{ai} = g - \mathbf{R}_{ai} - \mathbf{r}_{ai} \omega_i \times \omega_i
\]

At the two positions where the two accelerometers are fixed, if the rotational radiuses about the origin \(O\) of the reference coordination system (O-XrYrZr) is measured as \(r_1\) and \(r_2\), the vector of acceleration at the rotation joint can be obtained from Eq.2-Eq.4.

\[
a_{ai} = \frac{r_1 a_{ai} - r_2 a_{ai}}{r_1 - r_2}
\]

At the knee joint, there are three rotational angles in three directions: the f/e angle, a/a angle and i/e rotation angle. To analyze the knee joint rotation angles, a algorithm based on the difference between double virtual sensors implanted in the knee joint, named virtual-sensor difference based, was present in the next part. As the left picture in Fig.2 shows, the lower limb segments, the thigh and shank, are supposed to be rigid segments. Two physical sensors and a virtual sensor in blue are fixed on the upper segment (thigh), and another two physical sensors and a virtual sensor in red are attached on the lower segment (shank). The corresponding axes of the physical sensors and the virtual sensor in the same group are in the same direction. Then the accelerations of the two

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virtual sensors can be calculated from the accelerations measured by every two physical sensors in each group respectively using the physical-sensor difference based algorithm.

When a multisegment rigid body connected with a three degree of freedom (DOF) ball and socket joint is moving in space, as we know, one point should physically have a unique acceleration, therefore, the two virtual sensors in the knee joint must have equal accelerations in the same coordinate frame. At the same position in the knee joint, the two virtual sensors attached in different orientations would measure two groups of accelerations. The difference between the acceleration vectors represents the difference of orientations between the two segments which can illustrate the rotation angles of the knee joint. The relationship of accelerations measured by the two virtual sensors is:

$$ R_{st} = \mathbf{R}^T $$

where \( \mathbf{R} \) is the rotation matrix between the two virtual sensors, which is also between the thigh and shank. It can be expressed with three navigation Euler angles: yaw angle \( \theta_x \), pitch angle \( \theta_y \), roll angle \( \theta_z \) as follow,

$$ \mathbf{R} = \begin{bmatrix} \cos \theta_x & 0 & \sin \theta_x & 0 \\ -\sin \theta_x & 0 & \cos \theta_x & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \begin{bmatrix} \cos \theta_y & 0 & -\sin \theta_y & 0 \\ \sin \theta_y & 0 & \cos \theta_y & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \begin{bmatrix} \cos \theta_z & 0 & -\sin \theta_z & 0 \\ \sin \theta_z & 0 & \cos \theta_z & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} $$

To calculate the navigation angles, another two magnetometers are used to measure the magnetic field data attached on the thigh and shank, with the corresponding axes in the same directions as those of the accelerometers on the thigh and shank respectively. The same way as calculating the navigation angles from accelerations using the virtual accelerometers in the knee joint, two virtual magnetometers attached with different orientations in the knee joint should physically have a unique magnetic field data. The difference between the vectors of magnetic field data represents the difference of orientations between the two magnetometers, which also illustrate the knee joint angles. The relationship of the measured magnetic field data is:

$$ \mathbf{M}_t = \mathbf{R}^T \mathbf{M}_s $$

Based on physical-sensor difference based algorithm and virtual-sensor difference based algorithm, the navigation angles of the knee joint can be calculated from Eq.5-Eq.7, and then the rotation matrix \( \mathbf{R} \) can be obtained.
TABLE III
ANATOMICAL KNEE JOINT COORDINATE SYSTEM

| \( \mathbf{e}_{k1} \) | The axis fixed on the femur and coincided with \( \mathbf{X}_f \)-axis of the femoral coordinate system, indicating f/e rotation and anterior-posterior translation |
| \( \mathbf{e}_{k2} \) | The axis fixed on the tibia and coincided with \( \mathbf{Z}_t \)-axis of the thigh coordinate system, indicating a/a rotation and distal-proximal translation |
| \( \mathbf{e}_{k3} \) | The axis fixed on the tibia and coincided with \( \mathbf{Z}_t \)-axis of the left (or right) tibia/fibula coordinate system, indicating f/e rotation and proximo-distal translation |

To analyze knee joint kinematics in the anatomical coordinate system, the final step of data analysis is to translate the navigation angles into knee joint angles in the anatomical knee joint coordinate system, the f/e angle, a/a angle and i/e rotation angle, which were quantified by Good et al. [17] and have been widely used to analyze kinematics of lower limb joint motion. The anatomical landmarks on the bones, lower limb segments fixed coordinate systems and anatomical joint coordinate system for the knee joint kinematic analysis are defined in Fig.3 and indicated in Table 1-3. Suppose three unit vectors, \( \mathbf{A}_X, \mathbf{A}_Y, \mathbf{A}_Z \), are obtained from the measured signals about each axis in the tibia/fibula coordinate system (\( \mathbf{O}_{t} - \mathbf{X}_t \mathbf{Y}_t \mathbf{Z}_t \)), then another three unit vectors, \( \mathbf{B}_{Xt}, \mathbf{B}_{Yt}, \mathbf{B}_{Zt} \) in the femur fixed coordinate system (\( \mathbf{O}_{f} - \mathbf{X}_f \mathbf{Y}_f \mathbf{Z}_f \)) are calculated using the rotation matrix \( \mathbf{R} \) expressed with navigation angles,

\[
\begin{align*}
\mathbf{B}_{Xt} &= \mathbf{RA}_{Xt} \\
\mathbf{B}_{Yt} &= \mathbf{RA}_{Yt} \\
\mathbf{B}_{Zt} &= \mathbf{RA}_{Zt}
\end{align*}
\]

Finally, the floating axis, the f/e angle \( \theta_{f/e} \), a/a angle \( \theta_{a/a} \) and i/e rotation angle \( \theta_{i/e} \) of the knee joint in the anatomical knee joint coordinate system can be calculated as follows,

\[
\mathbf{e}_{k2} = \frac{\mathbf{A}_z \times \mathbf{B}_{Xf}}{|\mathbf{A}_z \times \mathbf{B}_{Xf}|}
\]

\[
\theta_{f/e} = -\sin^{-1}\left(\frac{\mathbf{e}_{k2} \cdot \mathbf{B}_{Zf}}{|\mathbf{e}_{k2}||\mathbf{B}_{Zf}|}\right)
\]

\[
\theta_{a/a} = \cos^{-1}\left(\frac{\mathbf{B}_{Xf} \cdot \mathbf{A}_z}{|\mathbf{B}_{Xf}||\mathbf{A}_z|} - \frac{\pi}{2}\right)
\]

\[
\theta_{i/e} = -\sin^{-1}\left(\frac{\mathbf{e}_{k2} \cdot \mathbf{A}_f}{|\mathbf{e}_{k2}||\mathbf{A}_f|}\right)
\]

3. EXPERIMENT

In order to evaluate the presented approach, a wearable sensor system was developed. The proposed wearable sensor system was mainly comprised of one piece of MCU (H8/3694, from Renesas Technology Corp.), two triaxial accelerometers (MM-2860 Sunhayato, Japan), and two MAG’s (MEMSsense, MAG10-1200S050, analog inertial sensor consisting of a triaxial magnetometer, an accelerometer and a gyroscope, ±1200°/s, ±10g, ±1.9 Gauss). The MCU was used to capture accelerations and magnetic field data from the sensors, store data in the EEPROM real time and communicate with a PC after each test. The sampling frequency was 100 Hz and the A/D had a 12-bit resolution.

As shown in Fig.4, in the wearable sensor system, each group of sensors on the thigh or shank contained one piece of MM-2860 and one piece of MAG’s, and every two corresponding axes of the sensors in the same group were in the same direction. The two MM-2860s attached at positions A and D on the thigh and shank were used to measure accelerations, and the two MAG’s fixed at positions B and C were used to capture accelerations and magnetic field data simultaneously. The four groups of accelerations measured by MM-2860s and MAG’s were used to calculate two groups of accelerations in the knee joint using the physical-sensor difference based algorithm, which were regarded as readings of two virtual sensors, and then were used to calculate the navigation angles using the virtual-sensor difference based method in Eq.6. The two groups of magnetic field data captured by the two MAG’s were also used to calculate the navigation angles using the virtual-sensor difference based method in Eq.7. The recorded accelerations about \( Y_t \) and \( Z_t \) axis indicated the anterior-posterior motion, about \( X_t \) axis and \( X_s \) axis indicated vertical motion along the lower limb segments, and about \( Z_s \) axis and \( Z_t \) axis indicated the lateral motion in the sensor coordination system on thigh or shank.

![Fig.4. Illustration for the installation of the sensors on the lower limb to calculate the knee joint angles](image)

Table IV.

<table>
<thead>
<tr>
<th>TABLE IV.</th>
<th>LIST OF THE TERMS USED IN FIG.4</th>
</tr>
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<tbody>
<tr>
<td>( O-ZYX )</td>
<td>Global coordinate system</td>
</tr>
<tr>
<td>( O-ZYZ )</td>
<td>Sensor coordinate system on the thigh</td>
</tr>
<tr>
<td>( O-ZYZ )</td>
<td>Sensor coordinate system on the shank</td>
</tr>
<tr>
<td>( r_t )</td>
<td>Distance from knee joint to the physical sensor B on the thigh</td>
</tr>
<tr>
<td>( r_s )</td>
<td>Distance from knee joint to the physical sensor C on the shank</td>
</tr>
</tbody>
</table>

As indicated in Fig.5, to prevent noise from contaminating the measured accelerations and magnetic field data when...
there were skin motion artifact due to impact loading and muscle activation in the motion, the sensors were fixed on one of the two step slides of a stainless steel telescopic slide rail. The length of the slide rail can be adjusted according to the length of the thigh or shank of different volunteers. The slide rails were connected with a ball and socket joint which allowed three DOF rotations. In initial installation, the slide rails were attached on the lateral surface of the lower limb using elastic straps. The slide rail on the thigh coincided with the line which connected the hip joint and knee joint, and another slide rail on the shank coincided with the line which connected the knee joint and ankle joint.

Fig.5. Experiment using the developed wearable sensor system in the working space of the optical motion system (the camera system).

In the experiment, five volunteers (4 males, 1 females, Age: 25±3 years, Height: 170±5cm, mass: 60±11kg) with no history of musculoskeletal pathology and injury were requested to perform 3 walking trials at self-selected speeds. Four groups of 3D accelerations and two groups of 3D magnetic field data on the thigh and shank were obtained during each motion test. Simultaneously, every three retro-reflective marks were fixed on the shank and thigh. A commercial optical motion analysis system, NAC Hi-Dcam II Digital High Speed Camera Systems (NAC image technology, Japan), was used to track the 3D trajectories of the retro-reflective markers with sampling frequency of 100 Hz, and given the navigation angles of the knee joint as reference. The average calibration error was 0.014% for capturing stationary subject and 0.21% for capturing dynamic subject. Fig.6 shows the procedure of the experiment.

4. RESULT

All signals captured in the experiment were off-line processed. Two groups of accelerations in knee joint regarded as measured by virtual sensors were calculated in each trial and one group was shown in Fig.7. Another group of magnetic field data was shown in Fig.8. The knee joint rotational angles in the anatomical knee joint coordinate system were calculated and shown in Fig.9, compared with the angles obtained from the camera system.

All the related parameters between the referenced and calculated knee joint angles used to evaluate the accuracy of the wearable sensor system compared with the referenced camera system are shown in Table 5, where RMS is the root of the mean of the square differences, R is the correlation coefficient, \( e_{max} \) is the maximum error.

<table>
<thead>
<tr>
<th>Measurement time (sec)</th>
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</thead>
<tbody>
<tr>
<td>Acceleration (m/s^2)</td>
</tr>
<tr>
<td>0</td>
</tr>
<tr>
<td>20</td>
</tr>
</tbody>
</table>

Fig.7 One group of accelerations in the knee joint regarded as measured by virtual accelerometers is shown. The virtual accelerations were calculated from the four groups of accelerations measured by the four physical accelerometers on the thigh and shank. The red line is the acceleration about y axis, black line is the acceleration about x axis, blue line is the acceleration about z axis in the sensor coordinate system.

<table>
<thead>
<tr>
<th>Measurement time (sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Magnetic field (Gauss)</td>
</tr>
<tr>
<td>0</td>
</tr>
<tr>
<td>1.5</td>
</tr>
</tbody>
</table>

Fig.8 One group of magnetic field data in the knee joint regarded as measured by virtual magnetometers, which were in fact measured by two physical magnetometers on the thigh and shank. The red line is the reading about y axis, black line is the reading about x axis, blue line is the reading about z axis in the sensor coordinate system.
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As curves in Fig.8-9 and the result in Table 5 show, the results obtained from the wearable sensors are close to those from the optical motion analysis system, with small RMS and large correlation coefficient. Therefore, it was feasible to use the method and the wearable sensor system to analyze the lower limb joint kinematics, such as knee joint kinematics in this paper. In the experiment, the gravitational acceleration, translational acceleration and noise, which were simultaneously acting on every two sensors in each sensor group, were eliminated based on the physical-sensor difference based algorithm, i.e., the accelerations of the virtual sensors in the knee joint were calculated only based on the differences between the rotational accelerations at every two positions on the thighs or shanks. However, since the rails with the sensors were attached on the lower limb segments with bandage, skin motion artifact due to impact loading and muscle activation would lead to relative motion between the board and the lower limb segments, which would contaminate the measured rotational accelerations, and then bring errors to the calculated rotation angles of the knee joint. Besides, since the distances from the sensors to the knee joint were just measured at the beginning of each test, unfixness of the sensors or the rails in each trial would also bring errors to the actual distances used for the calculation of rotation angles.

Since the seventies of last century, the advantages and disadvantages of use of kinematic sensors were discussed by Padgaonkar and Morris et al. [18] [19], and recently, Daniele et al. [20] demonstrated that architectures with sole accelerometers did not allow an accurate reconstruction of joint kinematics, then gyroscopes were introduced in the algorithm in their further research [21]. Although the insensitivity of gyroscopes improved the trajectory reconstruction, it was necessary the design and the introduction of a real-time algorithm for the drift compensation. In our method, it was the first to consider the knee joint kinematics in anatomical coordinate system using accelerometers and magnetometers without gyroscopes. Since there was no integration of angular acceleration or angular velocity for the calculation of the knee joint angles, the results were not distorted by offset and drift. And that using double physical-sensors in the same condition decreases the affection of noise. It was more simple and practical to use virtual sensors than implant physical sensors in the knee joint to calculate the rotation angles in the anatomical knee joint coordinate system.

In the study of kinematics and dynamics of the human lower limb, many methods have been developed, first only in the sagittal plane [15] and then in the 3D space [21]. The method most widely used to compute 3D kinematics is based on the modeling of each lower limb joint as a sequence of three hinges. The main disadvantage of this method is that the results depend both on the sequence order of the f/e, a/a, i/e rotation and the definition of the axes about which the rotations are expressed: a fixed laboratory frame or mobile axes. In this paper, the knee joint kinematic parameters within the anatomical coordinate system are independent of the
sequence order of the knee joint rotation angles. Consideration of the errors obtained in all of the experiments with the developed wearable sensor system compared with the camera system showed that the max errors of the \(\gamma/e\) angle were lower than 4.13° and the correlation coefficient was higher than 0.91 during a 7s acquisition, the max errors of the \(a/a\) angle were lower than 3.86° and the correlation coefficient is higher than 0.93, and the max errors of \(\gamma/e\) rotation angle were lower than 3.65° and the correlation coefficient was higher than 0.93, showing that the wearable device well satisfied the knee joint kinematics analysis for the clinical lower limb motion evaluation.

Since the newly launched analog inertial sensors MAC\(^1\) were used in the experiment, which were capable of sensing rotation, acceleration and magnetic field about three orthogonal axes and packaged in a single SMT (0.70 \(\times\) 0.70 \(\times\) 0.40 inches), the wearable sensor system on the lower limb segment was minified and convenient to wear on for patients. Especially compared with the inadequate in costs and encumbrance of the optoelectronic equipments, the wearable sensor system could introduce adequate and necessary quantitative analysis of joint kinematics. Another advantage of this method is that the developed device is not model-dependent which is very practical for time-limited clinical applications with many patients or space-unlimited continuous clinical evaluation for a patient to wear on in his daily life.

The combination of physical-sensor difference based algorithm and virtual-sensor difference based algorithm in this paper is original for the analysis of 3D knee joint kinematics in the anatomical coordinate system. With the continuously decreasing of costs and miniaturization of the inertial sensors, we are working for realization of sensor assemblies in a single chip to measure two groups of 3D angular accelerations and one group of 3D magnetic field data simultaneously for joint kinematic analysis, then promote it to develop wearable systems for clinical applications, such as ambulatory measurement and analysis of lower limb gait in the daily life for patients or health persons.

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REFERENCES


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