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Consistent accuracy in whole-body joint kinetics during gait using wearable inertial motion sensors and in-shoe pressure sensors

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ABSTRACT

To analyze human motion such as daily activities or sports outside of the laboratory, wearable motion analysis systems have been recently developed. In this study, the joint forces and moments in whole-body joints during gait were evaluated using a wearable motion analysis system consisting of an inertial motion measurement system and an in-shoe pressure sensor system. The magnitudes of the joint forces and the moments in nine joints (cervical, thoracic, lumbar, right shoulder, right elbow, right wrist, right hip, right knee, and right ankle) during gait were calculated using the wearable system and the conventional system, respectively, based on a standard inverse dynamics analysis. The averaged magnitudes of the joint forces and moments of five subjects were compared between the wearable and conventional systems in terms of the Pearson's correlation coefficient and the normalized root mean squared error to the maximum value from the conventional system. The results indicated that both the joint forces and joint moments in human whole body joints using wearable inertial motion sensors and in-shoe pressure sensors were feasible for normal motions with a low speed such as walking, although the lower extremity joints showed the strongest correlation and overall the joint moments were associated with relatively smaller correlation coefficients and larger normalized root mean squared errors in comparison with the joint forces. The portability and mobility of this wearable system can provide wide applicability in both clinical and sports motion analyses.

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1. Introduction

Human motion analyses, including joint kinematics and kinetics, have been conventionally performed using integrated systems of optical cameras to capture motion data, and force plates to measure ground reaction forces (GRFs). Although this conventional system has been utilized successfully in many research fields, such as sports and clinics, it is limited to the laboratory work space required by the camera and force plate system [1,2]. To analyze the human motion of daily activities or sports outside of the laboratory, wearable systems consisting of inertial motion sensors and in-shoe pressure sensors have been recently developed [1,3,4]. Their reliabilities were guaranteed by comparing these new systems with the conventional system in terms of the joint moments in three lower limb joints during walking [1], and the joint angles, angular velocities, and moments of body parts during manual material handling tasks [3]. However, there was no study that included a kinetic analysis of a human whole body using the wearable systems. In this study, the joint forces and moments in whole-body joints during gait were evaluated using a wearable motion analysis system consisting of an inertial motion measurement system and an in-shoe pressure sensor system. These results were compared to those measured by the conventional motions capture system with force plates.

2. Methods

Five healthy male subjects (age, 27 ± 1 years; height, 171.4 ± 3.9 cm; weight, 73.3 ± 12.1 kg) participated in this study. This work was approved by our Institutional Review Board and the subjects gave informed consent to the work. The wearable motion capture system (wearable system) consisted of the MVN® motion capture system (Xsens Technologies, Enschede, the Netherlands) with 17 inertial sensors and the Pedar-X® (Novel GmbH, Munich, Germany) in-shoe pressure measurement system. The Pedar-X® in-shoe pressure system

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consisted of 85–99 sensors and its measurable pressure range was 15–600 kPa with a resolution of 2.5 kPa. The conventional motion capture system (conventional system) was composed of the Hawk® system (Motion Analysis, Santa Rosa, CA, USA) with 10 cameras and four MP4060® force plates (Bertec Corporation, Columbus, OH, USA). Seventeen inertial sensors, two in-shoe pressure sensors, and 37 passive photo-reflective markers were attached to the subject's body (Fig. 1a and b). The inertial sensors were fixed to the subject's head, trunk, pelvis, and upper and lower extremities, and the pressure sensors were secured to the subject's bare feet. The passive photo-reflective markers were attached to anatomical landmarks that were pre-determined based on the existing literature [5,6].

To estimate the accuracy of the inertial sensor system, the orientation angle of a rigid body in a three-dimensional (3D) global $X\,Y\,Z$ coordinate system was compared between the inertial sensor system and the optical motion capture system. Three optical markers and three inertial sensors were attached to the plastic rigid frame (Fig. 1c). Optical markers were attached to create a right angle between two vectors between three optical markers, and a local $x\,y\,z$ coordinate system could then be defined. The inertial sensors were placed on the frame where the x' and y' axes of the local $x'\,y'\,z'$ coordinate system in the inertial sensor were parallel to the x and y axes of the local $x\,y\,z$ coordinate system of optical markers, respectively. Then, the initial orientation angles of the x, y, and z axes of the optical markers were same as those of the x', y', and z' axes of each inertial sensor with respect to the global coordinate system. The differences in the orientation angles of the x, y, and z axes of the optical markers and those of the x', y', and z' axes of each inertial sensor were analyzed in terms of the root mean squared error (RMSE) when the frame moved as a single Z-axis rotation and as a free 3-axis rotation in the 3D space.

Walking motion data were captured during a full gait cycle from a right heel strike to the next right heel strike with subject-preferred speed using the wearable and conventional systems simultaneously. In the conventional system, the positions of 37 anatomical landmarks were recorded directly using photo-reflective markers. Complete GRFs were also measured. In the wearable system, the orientation angles of each body segment, the vertical component of GRF calculated by integrating pressure values, and the center of pressure (CoP) data were recorded. The positions of the anatomical landmarks were then calculated from the orientation angles of the body segments using the implemented software in the MVN®. The 3D GRF was then artificially restored based on a previous study [7] as the 3D GRF was directed from the CoP point to the center of gravity of each subject's whole body as follows:

$$\vec{\mathrm{GRF}} = \left(\frac{\mathrm{GRF}_z}{v_z}\right)\vec{v}$$

where $\vec{\mathrm{GRF}}$ is the GRF vector, GRF_z is the vertical component of GRF, $\vec{v}$ represents an unit directional vector of the 3D GRF, and v_z is the vertical component of the directional vector of GRF.

A dynamic model of the human whole body, consisting of 16 segments (head, thorax, lumbar, pelvis, upper arms, forearms, hands, thighs, shanks, and feet) and 45 degrees of freedom linkages for 15 joints (cervical, thoracic, lumbar, shoulders, elbows, wrists, hips, knees, and ankles), was used as in our previous study [8]. Joint centers were defined based on the previous studies [9,10]. The reference frames of each segment were commonly set [5,6]. The joint angles were calculated as the Euler angles of the distal segment reference frame relative to the proximal segment reference frame. The magnitudes of the joint forces and moments were then calculated based on a standard inverse dynamics analysis using the motion data and GRF with MATLAB® (MathWorks™, Natick, MA, USA) [8,11]. Joint kinematic values, such as the angular velocity and acceleration of the ith segment, were calculated from motion capture data by the finite difference technique. The joint force and moment acting on the ith body segment were then calculated starting from the distal segment with the GRF and moment [8,11] using the following equilibrium equations:

$$\vec{F}_i^e = m_i(\vec{a}_i - \vec{g}) - \vec{F}_{i-1}^e$$
$$\vec{M}_i^e = I_i\vec{\alpha}_i + \vec{\omega}_i \times (I_i\vec{\omega}_i) - \vec{l}_i \times \vec{F}_i^e - \vec{l}_{i-1} \times \vec{F}_{i-1}^e - \vec{M}_{i-1}^e$$

where m_i is the ith segmental mass, $\vec{a}_i$ is the translational acceleration vector of the ith segment's center of gravity, $\vec{g}$ is the

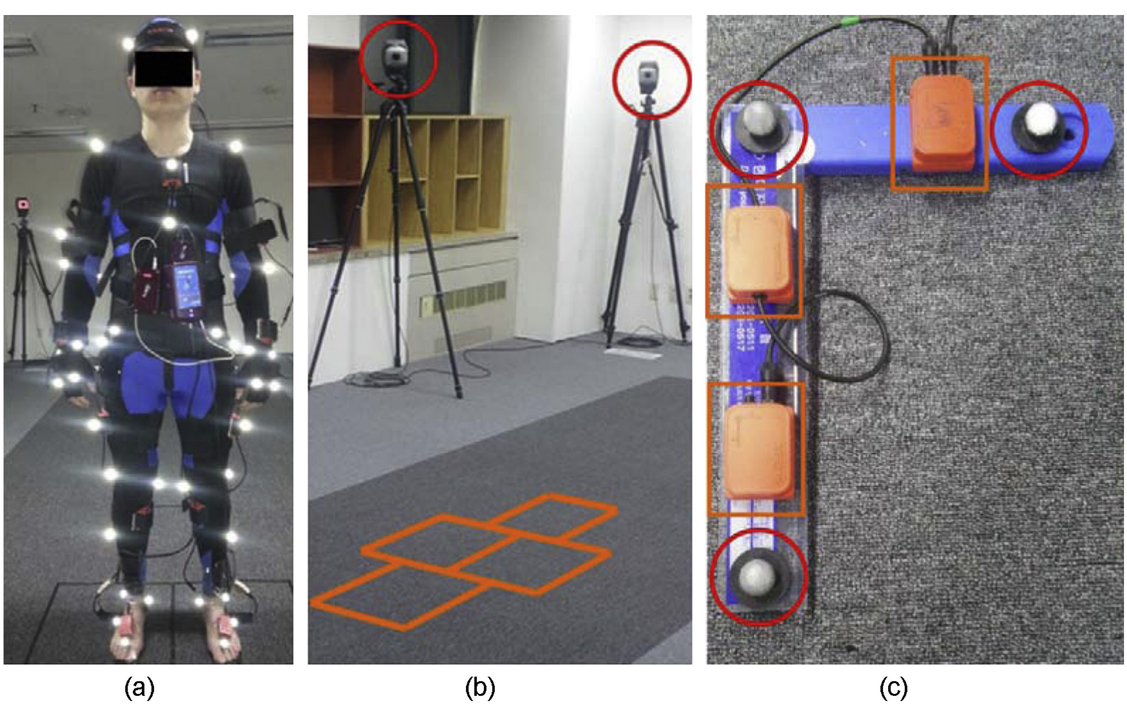

Fig. 1. (a) A subject equipped with inertial measurement system, in-shoe pressure system, and 37 photo-reflective markers, (b) the optical motion capture camera system and force plates, (c) a plastic rigid frame equipped with three optical markers and three inertial sensors.

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