Quantitative Evaluation of Unrestrained Human Gait on Change in Walking Velocity

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Abstract— **In human gait motion analysis, which is one useful method for efficient physical rehabilitation to define various quantitative evaluation indices, ground reaction force, joint angle and joint loads are measured during gait. To obtain these data as unrestrained gait measurement, a novel gait motion analysis system using mobile force plates and attitude sensors has been developed. On the other hand, a human maintains a high correlation among the motion of all joints during gait. The analysis of the correlation in the recorded joint motion extracts a few simultaneously activating segmental coordination patterns, and the structure of the intersegmental coordination is attracting attention to an expected relationship with a control strategy. However, when the evaluation method using singular value decomposition has been applied to joint angles of the lower limb as representative kinematic parameters, joint moments related to the rotational motion of the joints have not yet been considered. In this paper, joint moments as kinetic parameters applied on the lower limb during gait of a normal subject and a trans-femoral amputee are analyzed under change in walking velocity by the wearable gait motion analysis system, and the effectiveness for quantitatively evaluate the rotational motion pattern in the joints of the lower limb by using joint moments is validated.**

I. Introduction

Human gait motion involves the biomechanical system, the surrounding environments and the central nervous system. Recently, patients with gait problems such as lower-limb amputation from illness, industrial or traffic accidents have increased, and physical rehabilitation has also been emphasized because of our aging society [1]-[7]. It has been reported that human gait motion is constructed by multiple simple coordination patterns, and it is thought that easier gait motion training is given by showing each extracted pattern [8].

In a previous study [8], a new quantitative evaluation method was proposed to obtain principal intersegmental coordination and its activity pattern by calculating joint angular velocity and superficial myoelectric potential in motion, arranging their time changes along time or muscle in an observation matrix. An evaluation index was defined based on such a matrix and its singular value decomposition. However, when the evaluation method using singular value decomposition has been applied to joint angles of the lower limb as representative kinematic parameters, joint moments related to the rotational motion of the joints have not been considered. However, their use conditions and measurable amount of steps are limited because they must be installed on a purpose-built structure. Conventional devices must be converted into wearable ones to resolve those problems. Therefore, we have developed a wearable gait motion analysis system using mobile force plates and attitude sensors for unrestrained gait measurement [9].

In this paper, joint moments as kinetic parameters applied on the lower limb during gait, in addition to joint angles as kinematic parameters, are analyzed during the gait of a normal subject and a trans-femoral amputee under change in walking velocity by the wearable gait motion analysis system. This system consists of mobile force plates like sandals and attitude sensors made from acceleration, gyro, and geomagnetism sensors. Measured experimental data are recorded on a laptop PC via data logger by wireless LAN. In gait experiments, ground reaction force, center of pressure, joint angle, joint moment and limb postures can be calculated by the developed system. In the end, we validated the effectiveness of quantitative evaluation based on intercomparison and consideration of all activities concerning the rotational motion pattern in the joints of the lower limb by using joint moments.

II. Experimental Methodology

A. Subject and Experimental Facility

In this study, five healthy male subjects and one male unilateral trans-femoral amputee with a prosthetic limb participated. The latter has worn a trans-femoral prosthesis of normal socket-type for at least 30 years. The total mass includes body mass plus the mass of the prosthetic limb. The experiments took place at Doshisha University, Kyoto, Japan. Human research ethical approval was received from Doshisha University and written consent was obtained from this subject.

B. Experimental Facility

The experimental facility for measuring ground reaction forces and joint moments applied to a limb or prosthetic limb of a subject is the developed wearable gait motion analysis

*Manuscript received April 2, 2014.

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system shown in Fig. 1. This system consists of both mobile force plates using a thin-type three-axis force sensor and attitude sensor structured by acceleration, gyro, and geomagnetism sensors. Moreover, the mobile force plate is fixed beneath a pair of sandals. This system has less cost and fewer constraints. The experimental data measured by mobile force plate and attitude sensor are transferred to PC via data logger by wireless LAN and recorded on the laptop PC. In gait experiments, ground reaction force, center of pressure, joint angle, joint moment and limb postures can be calculated by the outputs of the mobile force plate and attitude sensor.

In the definition of coordinate system, forces and moments regarding x, y and z-axis directions are defined as F_x, F_y, F_z and M_x, M_y, M_z when the positive rotation of each axis is clockwise and coordinate systems are right-handed. Specifically, x, y and z-axes correspond to the anatomical medio-lateral (lateral is positive), anterior-posterior (anterior is positive) and vertical (upward is positive) directions. It is noted that the original point of the total coordinate system is a one-sided foot with the original point of the coordinate system in the heel-side mobile force plate.

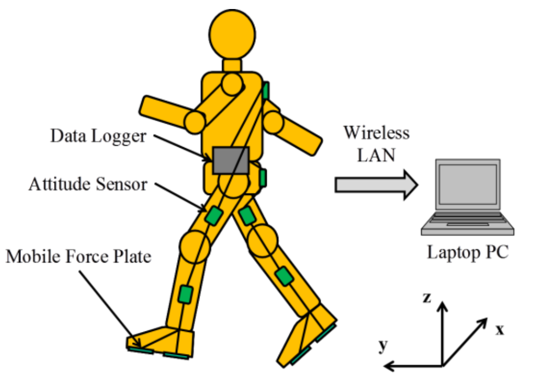

Fig. 1. Constitution of developed wearable gait motion analysis system.

C. Experiment Description

The subject performs each activity at a different rate of velocity on a treadmill (Fig.2). Concrete descriptions of each activity and the experimental fields are given. Actually, about 20 [min] of practice is performed before the experiments. Ground reaction forces and kinematic parameters are measured for at least 10 steps of each activity because of the unrestrained gait measurement. Sampling frequency and cut-off frequency of the low-pass filter in the experimental facility are 100 [Hz] and 10 [Hz].

Fig. 2. Treadmill

D. Data Analysis

Obtained patterns of ground reaction forces and kinematic parameters for each gait cycle of the various activities are analyzed when the first and last steps recorded for each trial are eliminated to avoid the initiation and termination of walking. One gait cycle is divided into four phases (double support phase 1,2 ,single support phase, swing phase) from behavior of the ground reaction force, and the joint moment and joint angle are calculated.

$$\boldsymbol{F} = \begin{pmatrix} F_x \\ F_y \\ F_z \end{pmatrix} = \begin{pmatrix} F'_{xtoe} + F'_{xheel} \\ F'_{ytoe} + F'_{yheel} \\ F'_{ztoe} + F'_{zheel} \end{pmatrix} , \quad (1)$$

Then, ankle joint moment about x-axis M_{xankle} is as follows. $g, m_{foot}, l_{y1}, l_{y4}, l_{y7}, l_{z1}, l_{z4}, I_{x1}, \ddot{\theta}_{x1}$ are acceleration of gravity, mass of foot, each moment arm on the lower limb, moment of inertia about x-axis of ankle joint, and angular acceleration about x-axis of ankle joint.

$$\begin{aligned} M_{xankle} &= l_{y1}F'_{ztoe} - l_{z1}F'_{ytoe} + l_{y4}F'_{zheel} - l_{z4}F'_{yheel} \\ &\quad - l_{y7}\, m_{foot}g + M'_{xtoe} + M'_{xheel} + I_{x1}\ddot{\theta}_{x1} \end{aligned} , \quad (2)$$

E. Evaluation Method of Intersegmental Coordination

In the present study, a specific evaluation method based on the principle of singular value decomposition proposed by Tsuchiya, Funato and Aoi [8] is used for quantitatively evaluating intersegmental coordination as typified by rotational motion about each joint of the human lower limb during unrestrained human gait as follows.

Only each joint angle has been applied to the quantitative evaluation in the past study; conversely, each joint moment is also applied in addition to each joint angle. $\theta_{ankle}(t), \theta_{knee}(t), \theta_{hip}(t), M_{ankle}(t), M_{knee}(t), M_{hip}(t)$ are joint angle and joint moment of ankle, knee and hip as positional information and kinesthetic information in the sense of force in the space, respectively. Here, $\boldsymbol{t}$ is time. An observation matrix is defined as the quantitative evaluation index of matrix **R** that has *m* number of rows and *n(=p+q)* number of columns. *m,n,p,q* are the number of data. $\bar{\theta}, \bar{M}, \bar{\boldsymbol{R}}$ are standardized joint angle, joint moment and observation matrix. Moreover, each physical quantity is defined as $\theta_j(t_i), \bar{\theta}_j(t_i), \bar{M}_j(t_i)\ (i = 1, \ldots, m,\ j = 1, \ldots, r \le n)$. Then, each specific quantitative evaluation index is calculated by using the following formula.

$$\bar{\boldsymbol{R}}(\bar{\theta}, \bar{M}, t) = \begin{pmatrix} \bar{\theta}_1(t_1) & \cdots & \bar{\theta}_p(t_1) & \bar{M}_1(t_1) & \cdots & \bar{M}_q(t_1) \\ \vdots & \ddots & \vdots & \vdots & \ddots & \vdots \\ \bar{\theta}_1(t_m) & \cdots & \bar{\theta}_p(t_m) & \bar{M}_1(t_m) & \cdots & \bar{M}_q(t_m) \end{pmatrix} , \quad (3)$$

Next, if singular value decomposition is performed on $\bar{\boldsymbol{R}}(\bar{\theta}, \bar{M}, t)$ and the equation is arranged, $\bar{\boldsymbol{R}}(\bar{\theta}, \bar{M}, t)$ is expressed as below.

$$\bar{\boldsymbol{R}}(\bar{\theta}, \bar{M}, t) = \sum_{j=1}^{n} \lambda_j \cdot \boldsymbol{v}_j(t) \cdot \boldsymbol{z}_j^{\,T}(\bar{\theta}, \bar{M}) , \quad (4)$$

where $\boldsymbol{v}_j(t), \boldsymbol{z}_j(\bar{\theta}, \bar{M})$ are eigenvectors of $\bar{\boldsymbol{R}}\bar{\boldsymbol{R}}^T, \bar{\boldsymbol{R}}^T\bar{\boldsymbol{R}}$, λ_j is a singular value defined as the contribution ratio of each orthonormal base vector called motion mode, which means human motion pattern, $v_j(t)$ is the time variation pattern of motion mode called temporal coordination, and $z_j(\bar{\theta}, \bar{M})$ is the coordination pattern of motion mode concerning each physical quantity. Each component is serialized by depending on the magnitude of λ_j value, and its contributing rate γ_j is calculated by using the following equation.

$$\gamma_j = \lambda_j^2 \Big/ \sum_{j=1}^{n} \lambda_j^2 \quad . \tag{5}$$

III. Experimental Results and Consideration

A. Contributing Rate

Figure3 shows the contribution of the mode of a normal subject walking and changing speed for each phase. From this, it can be seen that the contribution rate of the first mode is increased as the belt speed increases in double support phases 1 and 2. As the belt speed increases, the proportion of the operation of the kick and depression increases. It can be seen that the contribution rate of the first mode is decreased as the belt speed increases in the single support phase and swing phase. It is considered that the timing of the entire lower extremity becomes faster since the operation of raising a strong kick in double support phase 2 increases before the swing phase as the belt speed increases.

Figure4 shows the contribution of the mode of the trans-femoral amputee walking and changing speed for each phase. Trend contribution rate that decreased in 2 [km/h] from 1 [km/h] and increased in 3 [km/h] was observed. It can be seen that the walking of 3 [km/h] is a more simple walking than 2 [km/h].

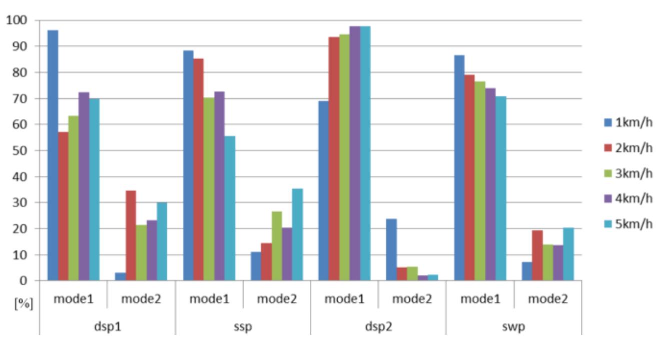

Fig. 3 Contributing rate of normal subject in 1~5 km/h walking

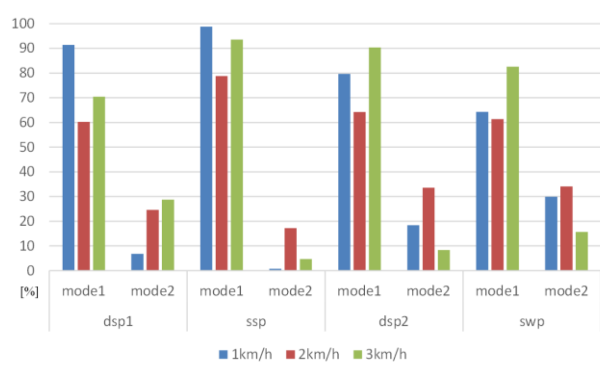

Fig. 4 Contributing rate of trans-femoral amputee in 1~3 km/h walking

B. Spatial Coordination

Figures 5-7 show the spatial coordination of the first mode in each phase for 3 [km/h] normal subject walking. Figures 8-10 show the spatial coordination of the first mode in each phase for 3 [km/h] trans-femoral amputee walking.

It is considered that in double support phase 1, a normal subject produces a braking force by a dorsiflexion moment that is reacted at the ankle joint and an extension moment that is reacted at the knee and hip joints to soften the impact on the heel strike (Fig. 5). On the other hand, the trans-femoral amputee reacts with a dorsiflexion moment at the ankle joint and extension moment at the knee and hip joints (Fig.8). Because it is not possible to kick the ground well in the prosthesis, the trans-femoral amputee has difficulty obtaining a driving force compared to a normal subject. It is considered that the force is transferred to the next phase in order to avoid reducing the driving force obtained as much as possible and kicking the ground in reverse almost without applying a braking force.

In the single support phase, a normal subject extends the knee joint and lower leg while moving the foot forward to land. At this time, the ground reaction force vector moves through the knee backward such that the extension moment works at the knee joint in resistance to this vector. On the other hand, the knee joint of the prosthetic leg is not able to move freely while weight is shifted to the unstable prosthetic leg, and kicking the ground is dangerous. Therefore, it is considered that the trans-femoral amputee provides a propulsive force to kick the ground during the single support phase and accelerate, weight-shifting to the next double support phase.

In double support phase 2, in general to obtain a driving force to kick the ground while moving weight, plantar flexion moment is acted on the ankle of a normal subject (Fig.7). On the other hand, extension moment is acted in the knee joint and dorsiflexion moment is acted in the ankle joint of the trans-femoral amputee (Fig.10). In order to transfer weight to the prosthesis side so as not to reduce the propulsion force obtained in the single support phase, the trans-femoral amputee is left to pull out the healthy foot. It is considered not so much plantar flexion moment that serves to kick the ground as dorsiflexion moment for lifting the toes, which is worked in the ankle.

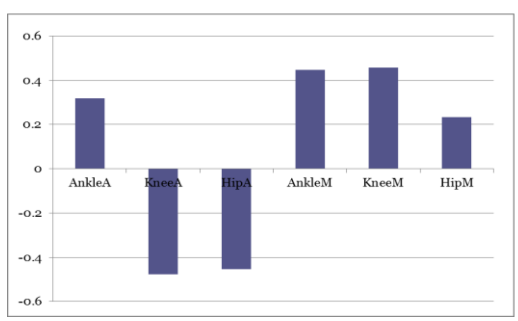

Fig. 5 Spatial coordination of normal subject at double support phase 1

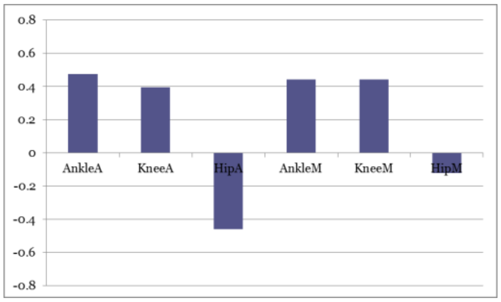

Fig. 6 Spatial coordination of normal subject at single support phase

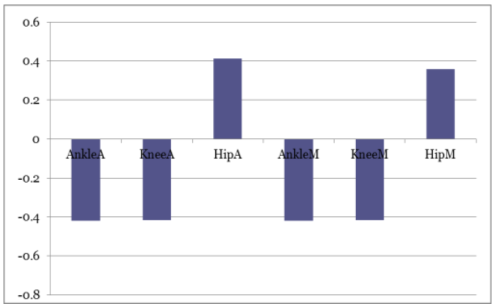

Fig. 7 Spatial coordination of normal subject at double support phase 2

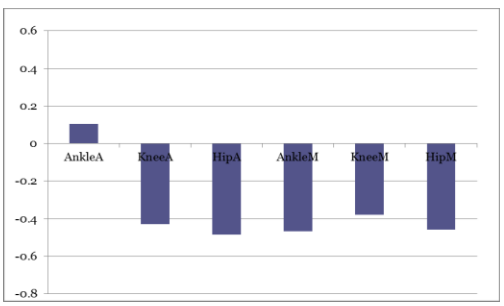

Fig. 8 Spatial coordination of trans-femoral amputee at double support phase 1

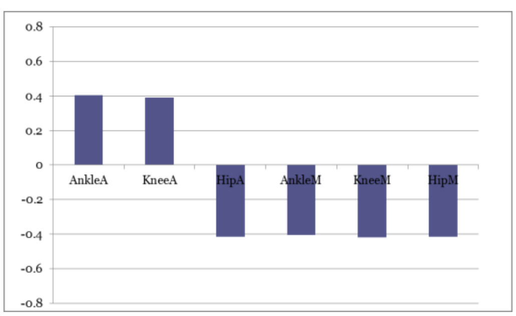

Fig. 9 Spatial coordination of trans-femoral amputee at single support phase

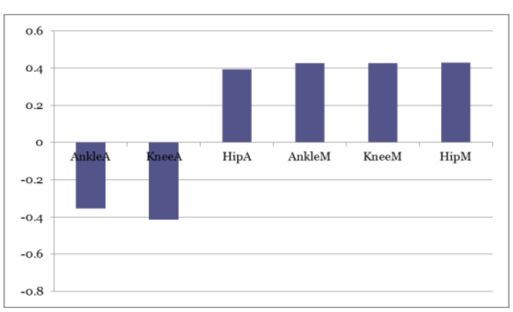

Fig. 10 Spatial coordination of trans-femoral amputee at double support phase 2

IV. CONCLUSION

In this paper, particular attention was focused on joint moments as representative kinetic parameters of the human lower limb during gait in addition to joint angles. Specifically, the rotational motion pattern in the joints of the lower limb considered as the attitude control mechanism was quantitatively evaluated from a viewpoint of kinetics and kinematics by their intercomparison, and their characteristics were investigated. Then, joint angles and joint moments applied on the human lower limb performing different walking velocities were calculated by a novel gait motion analysis system using mobile force plates and attitude sensors for unrestrained gait measurement. The motion pattern had a high correlation in the intersegmental coordination, which is extracted among such physical quantities by using singular value decomposition. In the end, we validated the effectiveness of the quantitative evaluation based on the biomechanical consideration of each joint moment, as well as the kinetics and kinematics.

ACKNOWLEDGMENT

This study was partially supported by a Grant-in-Aid for Scientific Research (B)(25289053), Japan Society for the Promotion of Science.

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