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Contributions of the Lower Extremity Joint on the Support Moment in Normal Walking and in Unexpected Step-down Walking

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Relative contributions of lower extremity joints on the support moment were investigated in this study. Three-dimensional gait analyses were performed in normal walking and in unexpected step-down walking. For both gait studies, inverse dynamics were performed to obtain each joint moment of the lower extremity, which was applied to the forward dynamics simulation to determine the contributions on the support moment at different phases of walking. The forward dynamic simulation results showed that, in normal walking, the ankle plantar flexors contributed significantly during single-limb-support. However, the ankle plantar flexors, knee extensors and hip extensors worked together during double-limb-support. In unexpected step-down walking, the important contributors on the support of the body during single-limb-support were not only ankle plantar flexors but also knee extensors. This study, analyzing the relative contributions of the lower limb joint moments for the body support, would be helpful to understand different unexpected walking conditions and compensatory mechanisms for various pathological gaits.

Key Words : Gait Analysis, Gait Simulation, Support Moment, Normal Walking, Unexpected Step-Down Waling

1. Introduction

There are two important tasks of the lower extremity in normal gait : the generation of the forward movement and the stable support of the upper body (Winter, 1991). Simon et al. (1978) performed gait analyses in normal walking and investigated the role of the posterior calf muscles. Later, Perry (1990) compared the gait of normal subjects to those with no significant calf muscle function, and reported that the normal calf muscle function helped the center of gravity (COG) of the body located more anterior to the center of pressure (COP). This indicated that the generation of forward movement velocity in walking might result from the active control by the plantar-flexion. Sutherland et al. (1980) also added another clinical evidence that a controlled fall by the passive mechanism might contribute to the generation of forward velocity in walking.

Especially, the function of the body support has been recently received much attention. The concept of the support moment has been used to determine the relative contribution of the lower extremity joint moments to prevent the collapse

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of the lower limb. The support moment was defined as the sum of all joint moments in the lower extremity (Winter, 1990). By its definition, positive values were regarded as extensor moments for preventing collapse and negative values as flexor moments for facilitating collapse (Winter, 1990). Actually, we should be able to determine the relative contribution of the individual muscles and the passive elements to the body support. However, it is very complicated to determine individual muscle forces due to the redundancy of the musculoskeletal system.

The gait analysis based on the support moment has some limitations to apply compensatory pathological gait. The support moment assumed that all kinds of flexor moments are contributors to collapse. This assumption would be broken for patients who have knee hyperextension. Another limitation of the support moment is that it does not distinguish which portion of the given joint moments is used for the support of the body and for the generation of forward progression. The last limitation of the support moment is that the contribution of each joint to support of the body is supposed that it is directly proportional to the amount of its moment. This supposition is only valid because of the multi-link characteristics of the human body. Zajac and Gordon (1989) reported that the torque at a single joint would generate accelerations at all joints and the magnitude of these accelerations could be varied with respect to the configuration of the body segments as well as the magnitude of the torque. Thus, the analysis on the body support during locomotion is generally limited to the net joint moments.

Of a special interest in the support moment is Kepple's works. Kepple et al.(1989) calculated the relative contributions of the lower extremity joint moments to forward progression and support during gait. They found that the ankle plantar flexors with a significant assist from the knee extensors produced the forward progression and the plantar flexors during single-limb-support and a combination of ankle plantar flexors, knee extensors and hip extensors during double-limb-support produced the support largely (Kepple, 1997). Those results gave convincing answers to analyze the relative contributions of the net joint moments to forward progression and support.

Falling is a serious problem among the elderly population, frequently resulting in physical injuries. The unexpected walking is one of the most probable cause of falling in the elderly. Recently a few studies have been performed to understand the process of the recovery balance mechanism against unexpected situations. In static position, ankle strategy, hip strategy and combined strategy were used to maintain the balance of the human body (Runge, 1999). However, the postural recovery mechanism based on the support moment in unexpected walking has not been clearly defined yet.

In this paper, we attempted to analyze the relative contributions of the lower extremity joint moments for the support in normal walking and in unexpected step-down walking.

2. Methods

2.1 3D motion analysis

A 27 year-old male subject (height: 170 cm, weight: 68 kg), with no gait problems in gait, participated in the three-dimensional motion analysis. Six infrared cameras were used to capture sixteen reflective markers for the gait analysis of the lower extremity.

Fig. 1 is the picture of sixteen reflective markers for motion analysis, based on the Davis protocol (Davis, 1991). Four makers were used to describe pelvic motion such that LASI and RASI were placed directly over the left, right anterior superior iliac spine, and LPSI and RPSI were placed directly over the left, right posterior superior iliac spine. Eight markers were placed on the leg. LKNE and RKNE were placed on the lateral epicondyle of the left, right knee. LTHI, RHTI were placed on the lower lateral surface of the thigh. LANK and RANK were placed on the lateral malleolus along an imaginary line the passes through the transmalleolar axis. Similarly to the thigh markers, LTIB and RTIB located along the tibia over the lower shank to determine the alignment of the ankle flexion axis. Four markers were placed on the foot. LTOE and RTOE located on

Table 1 The definition of the marker set used in this study

Segment	Name	Position
Pelvis	LASI	Left ASIS
	RASI	Right ASIS
	LPSI	Left PSIS
	RPSI	Right PSIS
Leg	LKNE	Left knee
	RKNE	Right knee
	LTHI	Left thigh
	RTHI	Right thigh
	LANK	Left ankle
	RANK	Right ankle
	LTIB	Left tibia
	RTIB	Right tibia
Foot	LTOE	Left toe
	RTOE	Right toe
	LHEE	Left heel
	RHEE	Right heel

Fig. 1 Marker set for the 3D motion analysis based on the Davis protocol (Davis, 1991)

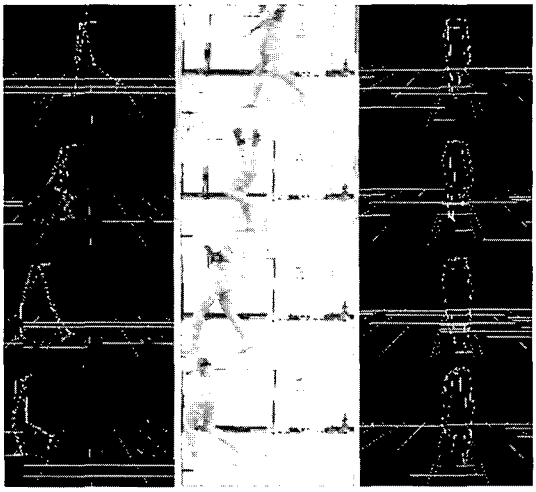

Fig. 2 Stick figures and photos at different gait phases in unexpected step-down walking, obtained by the 3D motion analysis

the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot. LHEE and RHEE located on the calcaneous at the same height above the plantar surface of the foot as the toe markers. The subject walked along the 10 m-walkway comfortably after enough practice.

For unexpected step-down walking, a movable platform was designed to provide a vertical perturbation during gait using a hydraulic pump and an AC servo motor, as shown in Fig. 2. Joint motions of the hip, knee and ankle in normal walking were analyzed by 3D motion analysis system (Vicon 612, Vicon Motion Systems Inc., USA).

Ground reaction forces were measured by four force plates (Advanced Mechanical Technology Inc., USA ; Kistler Instruments Ltd, Switzerland) during walking. All data were collected synchronously during experiments.

2.2 Computer simulation

Computer simulations were performed by the functional virtual prototyping tools MSC. ADAMS (MSC software corporation, USA) and Biomechanics Modeler BRG. LifeMod (Biomechanics Research Group Inc, USA).

A 3D virtual skeletal model for the present computer simulation was composed of seven segments : two feet, two shanks, two thighs and a pelvis. It was made based on the anthropometric data of the subject contained in the SLF file. The SLF file contained body measurement parameters and motion capture data. Gait simulations were performed by translational and rotational motion capture data.

Fig. 3 The skeletal model based on the subject's anthropometric data

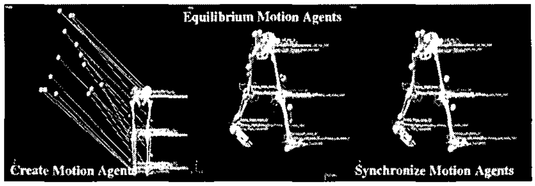

Fig. 4 Equilibrium analysis and synchronization body marker location with data location

Fig. 3 shows the skeletal model generated anatomically by using the 3D modeling program, LifeMOD. As the motion of upper limbs, in general, does not affect human gait, only the lower limb model was used.

In normal walking, there was a very small difference between the COG of head-arms-trunk (HAT) and the pelvic center in normal walking (Whittle, 1997). Therefore, in this study, mass and the moment of inertia of the pelvis were merged with those of the upper body. As a result, the pelvic center was the substitute for the COG of HAT.

Three-axis kinematics joints at the hip, knee and ankle were generated for each limb. Hip and ankle joints are considered as spherical joints and the knee was assumed as a hinge joint. Motion capture data obtained by the 3D motion analysis system were imported to the generated skeletal model. In order to drive the skeletal model to capture the simple joint angle histories for each joint, motion agents were added to the model. The motion agents have the effect of guiding the model to track the segment motion contained in the motion input file.

Fig. 4 shows the positions of reflexive markers and motion agents, fitting the model to the motion capture data, and synchronizing with the motion agents. The synchronization process reduced the slight differences between the model and the motion capture data.

The model for this study calculated the vertical acceleration of the upper body, which was produced by the support moment estimated at each joint. The mathematical theory for the modeling was outlined by Zajac and Gordon (1989), who explained that the moments produced by muscle forces around a joint would generate accelerations at all joints of the body. This formulation can be written by Eq. (1)

$$q=M^{-1}T+M^{-1}C+M^{-1}G+M^{-1}F \qquad (1)$$

In this equation, $\boldsymbol{q}$ is the matrix for joint accelerations, $\boldsymbol{M}^{-1}$ the inverse of the inertia matrix. $\boldsymbol{T}$ the matrix for joint moments, $\boldsymbol{C}$ the matrix for Coriolis terms, $\boldsymbol{G}$ the matrix of gravitational terms, and $\boldsymbol{F}$ the matrix for external forces.

Accelerations that was calculated exclusively by joint moments can be obtained by setting the matrices $\boldsymbol{C}$, $\boldsymbol{G}$, and $\boldsymbol{F}$ to zero in Eq. (1)

$$q=M^{-1}T \qquad (2)$$

Contact ellipsoids were generated at the calcaneous and at five metatarsal heads, and then the corresponding contact forces between the contact ellipsoids and the floor were defined. Inverse dynamics simulations were performed to obtain each joint moment in the lower limbs. Then, the forward dynamics simulations were performed to determine relative contributions of each joint moment on the support moment.

In order to calculate joint moments for the support in walking, gait phases were divided into right mid-stance, double limb support, and left mid-stance (Fig. 5). We used the Kepple's method for the simulation. After arranging the model, for each data frame the gravity value and all except one of the joint moments were set to zero. The solver module of the ADAMS software was then used to calculate the vertical acceleration of the pelvic center produced by that joint moment during the 0.001s simulation. We supposed that this short simulation interval assured that joint positions remained virtually unchanged, and that joint velocities remained near zero.

3. Results

Fig. 5 and Fig. 6 show the simulation results using the forward dynamics for normal and unexpected step-down walking respectively. It is noted that the magnitude of the ground reaction force vector in unexpected step-down walking is significantly greater in right mid-stance than in left mid-stance. The left knee flexion angle at right mid-stance is also greater than that at left mid-stance.

Fig. 7 shows the vertical accelerations of the pelvic center generated by each joint moment in different phases of normal walking. At right mid-stance, right ankle moment was the most primary contributor for the vertical accelerations. Left hip moment was the second contributor. At double-limb-support, right ankle contributed even more significantly and left ankle moment also increased a little. In left mid-stance, left ankle moment was the primary contributor but right ankle moment contributed less.

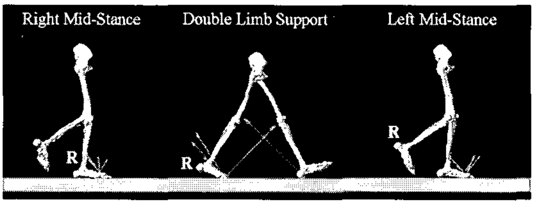

Fig. 5 Simulation results from the forward dynamics in normal walking

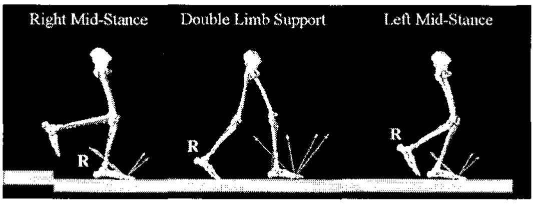

Fig. 6 Simulation results from the forward dynamics in unexpected step-down walking

In normal walking or postural balance, many investigators reported that the ankle joint moment plays the most important role to support the whole body. The present simulation result shows a good agreement with the Kepple's work (Kepple, 1997). Right ankle moment was larger than left ankle moment in our simulation results. This might be the reason that the subject has the right limb for his dominance.

Fig. 8 shows the vertical accelerations of the pelvic center in unexpected step-down walking. In unexpected step-down walking, right ankle joint also was the most primary contributor for the vertical accelerations. At right mid-stance, the magnitude of the vertical acceleration contributed by the right ankle joint alone in unexpected step-down walking was about five times larger than that in normal walking. In addition, the vertical acceleration of the right knee in unexpected step-down walking was significantly larger than that in normal walking. The roll-over mechanism of the foot in normal walking did not occur in unexpected step-down walking. In unexpected step-down walking, forefoot contacted to the

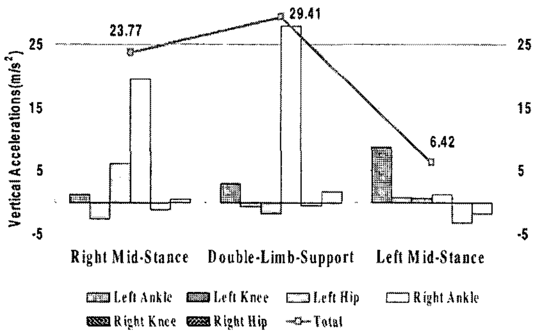

Fig. 7 Vertical accelerations of the pelvic center in normal walking

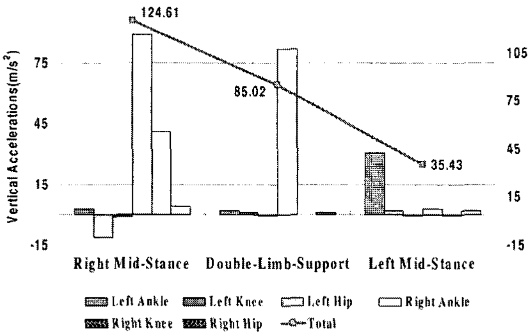

Fig. 8 Vertical accelerations of the pelvic center in unexpected step-down walking

ground at first and then progressed to the next phase, double-limb-support. Total accelerations in unexpected step-down walking were much larger than in normal walking. At left mid-stance phase, the subject kept the balance for repositioning of the over-advanced COG through that the subject stepped left foot forcefully. This caused that left ankle moment at left mid-stance was larger than in normal walking.

4. Conclusion

This study has attempted to analyze the relative effects between the lower extremity joint moments to support the body in normal walking and in unexpected step-down walking.

In normal walking, the ankle plantar flexors worked significantly during single-limb-support, but the ankle plantar flexors, knee extensors and hip extensors worked together during double-limb-support. In unexpected step-down walking, the important contributors during single-limb-support are not only ankle plantar flexors but also knee extensors.

This study, analyzing the relative contributions of the lower limb joint moments for the body support will be helpful to understand many unexpected walking and compensatory mechanisms for various pathological gaits.

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