Original Article

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Effect of heel pressure pad attached to ankle-foot orthosis on the energy conversion efficiency in post-stroke hemiplegic gait

Keisuke Kon, PO, PhD[1)*], Yasuyuki Hayakawa, PO, PhD[1)], Shingo Shimizu, PT, PO[1)], Takeshi Tsuruga, ENG, PhD[1)], Shin Murahara, PO[1)], Hirokazu Haruna, PT, PhD[2)], Takumi Ino, PT[2)], Jun Inagaki, ENG, PhD[3)], Sumiko Yamamoto, ENG, PhD[4)]

[1)] Department of Prosthetics and Orthotics, Faculty of Health Sciences, Hokkaido University of Science: 7-15-4-1 Maeda, Teine, Sapporo, Hokkaido 006-8585, Japan

[2)] Department of Physical Therapy, Faculty of Health Sciences, Hokkaido University of Science, Japan

[3)] Department of Information Engineering, Faculty of Engineering, Hokkaido University of Science, Japan

[4)] Department of Health and Welfare Studies, International University of Health and Welfare, Japan

Abstract. [Purpose] This study aimed to analyze the effect of heel pads in ankle-foot orthoses on dynamic motion aspects of gait in stroke patients from the viewpoint of energy conversion efficiency. [Subjects] Fourteen chronic stroke patients who were ambulatory and had lower extremity motor function categorized as Brunnstrom stage IV participated in the study. [Methods] A three-dimensional motion analysis system was used to assess the effect of heel pad intervention on dynamic motion gait parameters using a single-system A-B-A design. [Results] The results showed that a heel pad attached to the ankle-foot orthosis caused significant retention of the center-of-pressure at the heel during the heel rocker function and significant increase in the dorsiflexion moment and the height of the center of gravity. [Conclusion] The present study showed that a heel pad attached to the calcaneal region of an ankle-foot orthosis caused slight retention of the center-of-pressure at the heel during the heel rocker function along with center of gravity elevation in the stance phase and improved the energy conversion efficiency, especially on the non-paretic side.

Key words: Heel pad, Three-dimensional motion analysis, Hemiplegic gait

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INTRODUCTION

Falls during daily activities are a concern in stroke patients who have central nervous system damage. Falling stems from the inability to lift the ankle (dorsiflexion) triggered by the failure of central nervous system mechanisms caused by stroke.

Previously, Duncan[1)] reported that a stimulus input in the plantar calcaneal region in an individual with central nervous system damage induced a dorsiflexion reflex as a neurophysiological action. This mechanism is known as "heel-gait-cast-treatment[2)]". Later research included studies such as one by Charles et al.[3)] on electromyographic analysis of the effects of a calcaneal region stimulus in individuals with post-stroke hemiplegia on activation of dorsiflexion muscle activity.

It has also been reported that in stroke patients, posture and gait[4, 5)] were affected by stimulating[6, 7)] the peroneal nerves within an ankle foot orthosis (AFO). Additionally, there have been some reports in which dynamic motion data were changed by neurological intervention in this manner. However, there are few studies on the effect of AFO on gait with a focus on energy, which is an indication of the total efficiency. Haruna and his co-workers[8)] reported that the continuous use of a Gait Solution AFO developed by Yamamoto[9, 10)] increased the rate of mutual exchange between kinetic energy and potential energy in the gait of a hemiplegic patient. This change showed that use of the Gait Solution AFO improved the energy conversion efficiency of the body's center of gravity using the so-called inverted pendulum. We investigated the effects of 27 different shapes and sizes of heel pads on gait using a plantar flexion resistive AFO in healthy subjects[11)]. This study showed that heel pads fixed to an AFO retained the center of pressure (COP) at the heel and increased the lever arm of the ground reaction force vector to the ankle related to the dorsiflexion moment. These effects facilitated the inverted pendulum movement and raised the COG. However, there is a relative lack of research describing the effect of a heel pad on gait in hemiplegic patients. Further studies are therefore required to elucidate

*Corresponding author. Keisuke Kon (E-mail: kon@hus.ac.jp)

Table 1. Subject characteristics

Evaluation parameters	A1	A2	A3	A4	A5	A6	A7	A8	A9	A10	A11	A12	A13	A14
Age (Y)	65	44	36	70	59	49	52	63	64	66	68	69	57	58
Gender (M: male, F: female)	M	F	F	M	M	F	M	M	M	M	M	F	M	F
Paretic side	Left	Left	Right	Left	Right	Left	Left	Right	Right	Right	Right	Right	Left	Right
Brunnstrom stage (Lower limb)	IV	IV	IV	IV	IV	IV	IV	IV	IV	IV	IV	IV	IV	IV
ROM (Paretic ankle) :Plantarflexion–Dorsiflexsion	40–5	40–5	45–20	45–20	40–20	40–0	45–15	45–20	40–7	40–10	40–15	35–20	45–0	40–10
Foot clonus	±	±	+	+	±	+	±	±	+	±	+	±	±	+
Type of everyday use orthosis	PLS	PLS	PLS	PLS	JAFO	PLS	JAFO	PLS	PLS	PLS	PLS	PLS	JAFO	PLS
Average walking speed (m/min)	50.4	33.6	42.0	48.0	45.3	44.4	41.8	52.3	42.7	45.6	49.5	39.5	40.5	38.8

PLS: plastic leaf spring AFO, JAFO: Joint AFO

the effect of a heel pad on hemiplegic gait.

The simplest method of calculating energy by using positional information on COG is to use the energy conversion efficiency calculation reported by Cavana et al[12]. Cavana[13] integrated the force component (acceleration component) obtained from a force plate to calculate the kinetic energy and potential energy, and they calculated the energy conversion efficiency from a physics standpoint. The method proposed by Cavana et al[14–19]. was applied to this study. This enabled us to focus on the conversion of potential energy and kinetic energy obtained from the movement of the COG. This study aimed to analyze the effect of heel pads in ankle-foot orthoses on dynamic motion aspects of gait in stroke patients from the viewpoint of energy conversion efficiency.

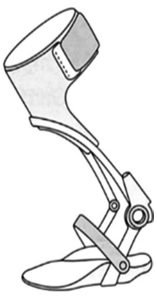

Fig. 1. Joint AFO used in this study

SUBJECTS AND METHODS

Fourteen chronic stroke patients (9 males and 5 females; age, 59±10 years) who were ambulatory and had lower extremity motor function categorized as Brunnstrom[20] stage IV participated in the study. Their characteristics are shown in Table 1. Selection criteria included subjects who could follow verbal instruction, did not have any orthopedic disease, used an AFO routinely on a daily basis, and were able to walk independently. The selected subjects in this study were able to walk forward by means of a heel strike when using an AFO. There was no restriction regarding cane use, or lack thereof, in the interest of ensuring an adequate number of subjects.

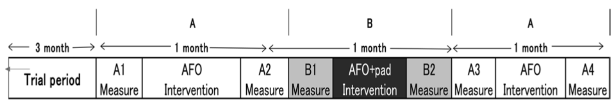

Fig. 2. Study protocol

This study was conducted after obtaining approval from the Ethics Committee of Hokkaido University of Science. The content of the study was explained to cooperating research facilities and the participants in order to obtain their consent, and written informed consent has been obtained from each subject.

Fig. 3. Heel pad

The AFO[9, 10] used for the study was a joint AFO (JAFO, Gait-solution-design, Pacific Supply, Japan), as shown in Fig. 1, and measurements were begun after the angle and resistance of the orthose The AFO[9, 10] used for the study was a joint AFO (JAFO, Gait-solution-design), as shown in Fig. 1, and measurements were begun after the angle and resistance of the orthoses were adjusted to match each subject's condition over a three-month trial period.

The study design was an A-B-A single-system design (Fig. 2). As a baseline, only JAFOs tailored for each subject were used for one trial month (A1 to A2). Next, the JAFOs were fitted with 3-mm-thick rubber heel pads with 60 degrees of hardness at the plantar calcaneal region of the JAFOs, as shown in Fig. 3, in order to examine the effect of the press-in force on the calcaneal region; these heel pads were used for one month (B1 to B2). Finally, the heel pads were removed, and only the JAFOs were used for one month (A3 to A4).

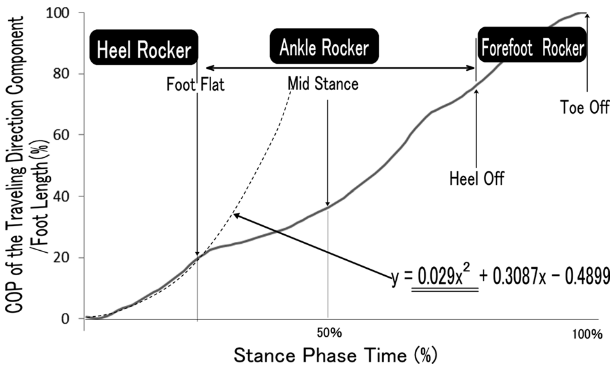

Fig. 4. Calculation of the second-order coefficient of COP

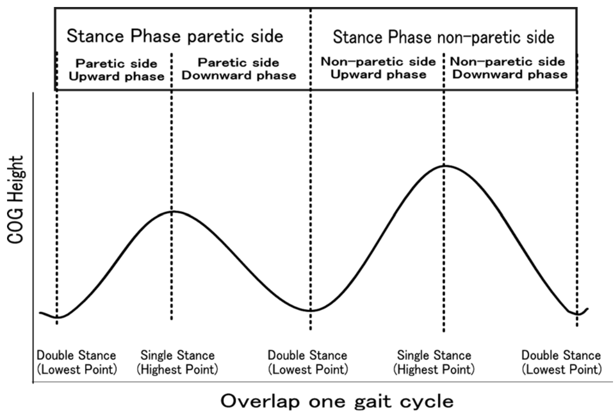

Fig. 5. Separate phases of mechanical energy of COG

Measurements were taken at A1 to A4, and were implemented on the same days for A2 and B1 and for B2 and A3, falling under the timings for the heel pad intervention. This was to examine the following: stability of the baseline by comparing A1 to A2; immediate effects of the heel pads by comparing A2 to B1; impact of the long-term effects of the heel pad intervention by comparing A2 to B2; and the carry-over effects by comparing B2 to A4.

Gait was measured using a three-dimensional (3D) motion analysis system VICON-MX (with six cameras, manufactured by Vicon, Oxford, UK) and five force plates (AMTI). The VICON-MX cameras and force plates were electronically synchronized. The sampling frequency for each measuring instrument was 240 Hz for infrared cameras and 960 Hz for floor reaction force-based measurements. The measurement path surface was 9.5 m, with three force plates arranged in the middle on the left side and two on the right side. For 3D motion measurement, reflective markers with a 14-mm diameter were applied to 21 points on the body. The application positions were the forehead, the right shoulder blade, both acromions, the lateral epicondyles of both humeri, both radial styloid processes, both anterior superior iliac spines, both posterior superior iliac spines, both hip joints (the distal 1/3 point connecting the anterior superior iliac spine and great trochanter), both knee joints (the anteroposterior radius 1/2 point, 1/3 point, and midpoint), both ankle joints (the lateral malleolus center for the healthy side and the orthosis axis center for the orthosis side), both calcaneal regions and both fifth metatarsal heads. The point of interest during measurement on the walking path was data acquisition for 10 trials where one foot was placed on a force plate.

The data obtained from the 3D motion analysis system were used to create middle markers between markers with the Vicon's Body Builder (Example: Midpoints between the knee joint and ankle joint markers).

Thereafter, the walking speed, stance-phase maximum height of the COG, dorsiflexion moment (maximum value of the loading response), and floor reaction force action point (COP) were found from the "Plug-In-Gait model". These parameters were normalized by body height and weight.

The COP retention rate: the COP of the traveling direction component during heel rocker was extracted, and a secondary function was calculated as per the dotted line in Fig. 4. This quadratic function represents the progression of COP. For instance, a greater quadratic function shows that the plantar COP is advancing quickly forward, which is an index for examining the effects of the heel pad.

The energy conversion efficiency: Using the research methods of Cavagna et al.[12–19] makes it possible to calculate the energy conversion efficiency from the COG obtained from the 3D motion data. This is done by applying a reverse pendulum theory using the force of gravity to calculate the energy conversion efficiency during walking from the kinetic and potential energies during walking.

Gait was divided into the paretic stance and the non-paretic stance using COG displacement in the vertical direction, as per Fig. 5. Then, the reverse pendulum motion was separated into a rising section and a falling section from the information on the lowest point and the highest point for the respective COG heights, and the energy conversion efficiency was calculated in each section as follows. To calculate the kinetic energy and potential energy during ambulation, the kinetic energy (Ek.y) in the direction of progress of the body COG was calculated by the formula, $Ek.y=1/2mVy^2$, the kinetic energy (Ek.z) in the vertical direction by $Ek.z=1/2mVz^2$, the potential energy (Ep) by Ep=mgz, the total kinetic energy(Ek) by Ek=Ek.y+Ek.z, and the total energy (Etot) by Etot=Ek+Ep, where m is the body mass, g is the acceleration due to gravity (9.8 m/s^2), Vz is the vertical velocity of COG, and Vy is the velocity in the direction of COG progress.

The external work for accelerating the COG in the direction of progress was calculated using Formula 1, the external work of raising the COG in the vertical direction against gravity by Formula 2, and the external work generated by the muscles to move the COG was calculated using Formula 3.

$$Wy=\Sigma\Delta Ek.y \quad (1),$$

$$Wz=\Sigma\Delta\ (Ek.z+Ep) \quad (2),$$

$$Wext=\Sigma\Delta Etot \quad (3),$$

where Δ is the amount of energy increase or decrease, and Σ is the summation of the data of Δ. Conversion efficiency

Table 2. Evaluation parameters showing the changes from baseline

Normalized parameters increase rate (%)	No Pad A2		Pad B1		Pad B2		No pad A4	
Retention rate of COP	0.3±2.2		Δ 10.6±8.5	**	Δ 23.5±15	**	Δ 12.1±12	**
Ankle dorsiflexion moment	Δ 1.0±1.2		13.4±16		52.9±15	**	27.3±37	
Paretic-side COG	0.1±1.0		6.8±2.0	*	9.1±2.8	**	8.5±3.4	**
Non-paretic-side COG	Δ 0.2±1.2		4.3±3.2		7.4±2.8	**	7.1±2.3	**
Paretic-side upward Energy conversion efficiency	2.0±2.0		Δ 12.0±15		26.0±12	**	20.0±13	**
Paretic-side downward Energy conversion efficiency	1.2±2.0		3.4±6.7		Δ 1.9±8.0		8.2±3.9	
Non-paretic-side upward Energy conversion efficiency	4.1±2.9		8.8±12		21.4±9.8	**	22.2±12	**
Non-paretic-side downward Energy conversion efficiency	1.4±4.2		19.4±8.8	**	16.8±8.8	**	21.9±8.8	**

Values are group mean ± deviation. **: significant difference base line A1. **: $p<0.01$, *: $p<0.05$
Δ: minus

Table 3. Matrix of correlation between parameters

	Retention rate of COP	Ankle dorsiflexion moment	Paretic-side COG	Non-paretic-side COG
Retention rate of COP	-	0.62	0.56	0.39
Ankle dorsiflexion moment	**	-	0.64	0.44
Paretic-side COG	**	**	-	0.72
Non-paretic-side COG	**	**	**	-

Upper part shows correlation coefficients.
Lower part shows a significant difference between parameters. **: $p<0.01$

(%recovery) was calculated using Formula 4.

$$\%recovery=(|Wy|+|Wz|-Wext)/(|Wy|+|Wz|) \times 100 \quad (4)$$

Analysis methods: For each assessment parameter, data after heel pad intervention were standardized using the mean and standard deviation of the baseline (A1). This made it possible to nullify the effects between subjects and the differences in units between assessment parameters to look at the relative growth rate versus the baseline. In the statistical analysis, data were analyzed using the statistical software, Excel-Toukei 2008 (Social Survey Research Information Co., Ltd.). Friedman test and Steel-Dwass-test for multiple comparisons were performed to compare the standardized numerical values. Spearman test was also performed to test the correlation among the values. Statistical significance was established at a probability value of 0.05.

RESULTS

Evaluation parameters standardized by baseline A1 are shown in Table 2. Comparisons of A1 to A2: no heel pad group as a control, B1: immediate intervention effects of the heel pad, B2: Long intervention effects of the heel pad, and A4: carry-over effects after removing the heel pad are also shown in Table 2.

The parameter of retention rate of the COP indicated retention of the COP at the heel as immediate effects (Table 2-B1). In contrast, there were no immediate effects in the paretic-side COG (Table 2-B1). However, long-term effects were observed in the paretic-side COG (Table 2-B2). In addition, a significant correlation was shown between retention rates of the COP and paretic-side COG (Table 3).

Paretic-side upward energy conversion efficiencies at B2 were also significantly increased with long-term effects in the paretic-side COG (Table 2-B2). All parameters of long-term effects (Table 2-B2) and carry-over effects (Table 2-A4) in energy conversion efficiency were significantly increased except the paretic-side downward.

DISCUSSION

The heel pads fixed the COP at the heel rocker section during walking, and they increased the ankle joint dorsiflexion moment.

In this regard, it appears the heel pads fixed the COP at the heel rocker section during walking, and they increased the ankle joint dorsiflexion moment. In this regard, it appears that the phenomenon of COP retention (Fig. 6-2) maintained the distance between the floor reaction force acting line and the ankle joint center (Fig. 6-3). The increase in the ankle joint dorsiflexion moment also caused the paretic-side COG height to rise. This is likely because there is a large ankle joint dorsiflexion moment in the paretic-side heel rocker period, and this produces forward rotation of the paretic lower extremity around the heel (Fig. 6-4), indirectly causing the paretic-side COG height to rise (Fig. 6-5, 6).

In general, the COG height in normal walking peaks at mid-stance, and it has left-right symmetry. During post-stroke walking, however, often a decrease in the support

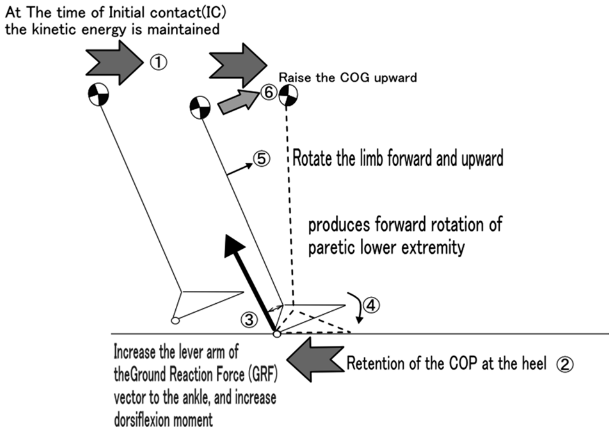

Fig. 6. Mechanism of COG rise

of the paretic side or other factors causes the paretic-side COG height to have a lower maximum value than the non-paretic side[20–22], resulting in a loss of symmetry (Fig. 5). The heel pads, however, caused the maximum value of the paretic-side COG height to rise, which reduced the left–right difference in COG height.

We do, however, feel that despite the lack of an increase in the dorsiflexion moment, the appearance of raising the paretic-side COG as an immediate effect (Table 2-B1) was caused by the higher energy conversion efficiency in the non-paretic-side descent in the direction of travel (Table 2-A4). This indicates that when the energy conversion efficiency on the non-paretic side increases, the paretic foot can land in a state where kinetic energy is maintained, as in Fig. 6-1, and an effect of retaining COP is also obtained, thus yielding a dual effect on increasing the COG height in the paretic stance. That is, forward inertia is produced at the instant of landing. It therefore appears that even a slight moment in the dorsiflexion direction contributes to raising the COG.

We also obtained the long term effect on all parameters except for the paretic-side downward energy conversion efficiency. These results coincided with the results of previous studies[21–23], which showed that the continuous use of an AFO increased the kinetic energy of non-paretic side descent.

The present study was an experimental investigation of the effect of heel pad intervention on dynamic motion assessment parameters using a single-system A-B-A design in 14 stroke patients.

The results showed that a heel pad attached to the calcaneal region of an AFO caused slight retention of the COP at the heel during the heel rocker function along with COG elevation and raised energy conversion efficiency, especially on the non-paretic side.

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