33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

A detailed 3D ankle-foot model for simulate dynamics of lower limb orthosis

Jun Inoue, *Student Member*, *IEEE*, Wenwei Yu ,*Member, IEEE*,Kang Zhi Liu, *Member, IEEE,* Kazuya Kawamura ,and Masakatsu G. Fujie, *Senior Member*, *IEEE*

Abstract— **The objective of this study is to develop a 3D ankle-foot model containing toe expression for designing an AFO (ankle-foot orthosis) with a training function. Two experiments were conducted to (1) show the influence of toes by comparing walking with and without an AFO, and (2) clarify the functions of toes during walking by correlating the activity of the major muscles controlling the ankle and the toes to the sole pressure data during walking. By analyzing the results of these two experiments, the necessary components and conditions of a detailed 3D foot-ankle model for developing an AFO with a training effect were clarified. A model was built and examined with empirical facts, and data were collected from the AFO simulation.**

I. Introduction

Recently, the number of people with lower-leg disabilities has been increasing rapidly due to the high incidence of strokes and other diseases [1]. As part of the walking assistance and rehabilitation for these disabled people, an AFO (ankle-foot orthosis) plays an important role. Using the AFO can enable patients to resume their normal social activities at an earlier time. However, if suitable training is not performed, their muscular power is seriously weakened, their ROM (range of motion) of joints diminishes and they forget the start timing for moving their muscles. Since not all people have the most beneficial rehabilitation, once the patient starts using the AFO, he or she could depend on it for the rest of his/her lifetime. This is not the optimal situation for most patients.

Consequently, we think an AFO should fix a minimum number of joints, because this will help prevent the patient from relying on the AFO.

Some new orthoses are being developed in big projects [2][3]. These new orthoses enable more natural walking and natural use of muscles. But, these projects last approximately ten years and incur a very large cost, and these two factors pose a barrier for creating new, more desirable orthoses. So the influence of the AFO on muscles needs to be clarified before the beginning of development.

The objective of this study is to develop a three-dimensional foot-ankle model containing toe expression for designing an AFO.

Some research on the 3D kinematics of a foot model has been conducted. Delp et al. created OpenSim and simulated the whole body [4]. Kim et al. created a 9-segment foot model for normal walking [5]. Takashima et al. performed a dynamic model analysis of the human foot, but the models need actual measurements for each condition [6]. Our proposed model can evaluate muscle activity by using a statistics method. This model is indispensable for making an improved foot orthosis and its application to other fields is expected.

Manuscript received April 15, 2011. This work is supported in part by grant from the Moritani Scholarship Foundation and by Global COE (Centers of Excellence) Program "Global Robot Academia," in part by a Grant for Scientific Research (A) (90198664) from the Ministry of Education, Culture, Sports, Science and Technology of Japan, and in part by a Waseda University Grant for Special Research Project (Project number: 2011A-607).

J.Inoue, is with the Members of the Graduate School of Science and Engineering, Waseda University, Japan. (59-309, 3-4-1, Okubo, Shinjuku Ward, Tokyo, Japan, e-mail: j.inoue@ruri.waseda.jp, tel: +81-3-5286-3412, fax: +81-3-5291-8269)

Wenwei Yu, Kang Zhi Liu are with the Members of the Faculty Chiba University, Chiba, Japan.

M. G. Fujie, Kazuya Kawamura are with the Members of the Faculty of Science and Engineering, Graduate School of Science and Engineering, Waseda University, Japan.

II. Sole pressure measurement walking with and without an AFO

The first of two experiments investigated the differences during walking with and without wearing an AFO. The two subjects were a non-disabled male and a disabled male with both lower legs paralyzed due to a spinal cord injury. The disabled male has disturbance sensation under the knee and equinovarus feet. Both subjects were in their twenties. Sole pressure was measured using F-Scan (Tekscan) during walking.

Both subjects walked on a treadmill without assistance, such as a cane. The able subject's walking speed was set to 3 km/h, and the disabled subject's walking speed was set to 1.5 km/h for safety reasons, because he usually walked with a cane. To measure the sole pressure in normal walking (without the AFO) of the able-bodied subject, a pressure sensor sheet was placed between the foot and the shoe. When the disabled subject wore the AFO, the sheet was inserted between the foot and the AFO.

The sole pressure distribution obtained from the measurements of walking of the disabled subject is shown in Fig. 1. The white lines show the locus of the center of the load. When a disabled person walks with an AFO, the displacement of the load is limited to a comparatively narrow range, as can be seen from this result.

To investigate the cause of this distribution, the data was divided and analyzed in the toe, the fore foot and the hind foot

(Fig. 2). This graph shows that the load is applied to the toes when not wearing the AFO, but the load is hardly applied to the toes when wearing the AFO.

These results show that there is almost no torque applied to the metacarpophalangeal (MP) joint when walking with the AFO, but in many research the MP joint is very important for walking.

It is required therefore, to make an AFO that stimulates the muscles moving the MP joint when walking in order to increase the future possibility that a person with a disabled leg can walk without the AFO.

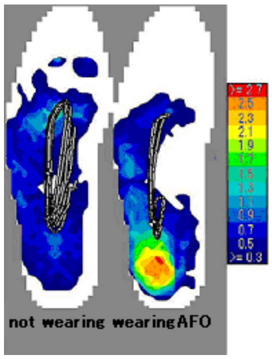

Fig. 1: Load distribution of the disabled subject

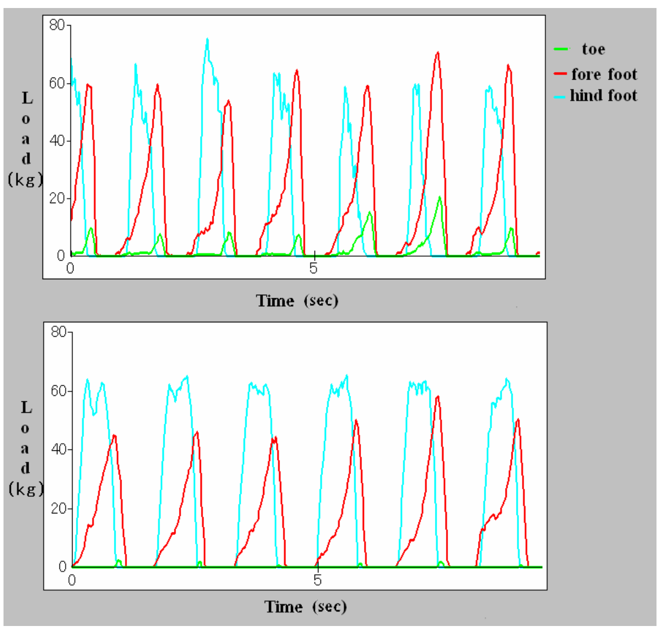

Fig. 2: Load transition of the disabled subject
(upper: without AFO lower: wearing AFO)

III. Measurement of the Muscle Activity and Motion of the Foot During Walking

In the measurement of muscle activity and foot motion, we considered the requirements that must be included in the walking model and obtained data from walking by the able subject. The data were also used to determine the relationship between muscle activity and joint angle or load on each part. Since it was only necessary to see the correspondence between the motion and the muscles on an able-person's foot, unlike in the previous experiment, only the able subject was chosen. In addition to the data of sole pressure, an electromyogram (EMG) of the muscles related to the angle of each portion of the foot was recorded. To measure one muscle EMG is very difficult. But the muscle in a near position is work similar. The angular velocity was taken from motion capture (VICON) during walking. The EMG of three muscles, tibialis anterior (TA), extensor digitorum longus (EDL), and peroneus longus (PL), were also measured at the same time because these three muscles can oppose foot drop and clubfoot, which are characteristic of leg paralysis.

In this experiment, 14 motion capture markers were used. One marker was on the knee, and 13 markers were on the foot. Because normal-size markers are too big for typical foot sizes, smaller markers with 0.16 inch diameter were used to avoid overlapping of the markers for the visual motion capture system and therefore losing accuracy or tracking of the movement.

The data of the stance phase was divided between a control term (0–30% GC (gait cycle)) and a propulsive term (30–50% GC). The control term is the time needed to regain the trunk balance lost in the swing phase, and the propulsive term is the time when an impelling force is generated.

The correlation coefficients of each muscle and each measured value for every control term and propulsive term are shown in Table 1. The measured values are at the MP joints of each toe: hallux, 2nd toe, and 3rd to 5th toes. The angle of the foot arch is the angle of MP joint-navicular bone-calcaneus bone.

Table 1: Correlation coefficients of each muscle and each measured value

Control term	Load on Hallux	Load on 2^{nd} toe	Load on 3^{rd} to 5^{th} toe	Angle of Hallux
TA EMG	0	-0.47	-0.75	-0.58
EDL EMG	0	-0.32	-0.61	-0.38
PL EMG	0	-0.31	-0.62	-0.51
Propulsive term	Load on Hallux	Load on 2^{nd} toe	Load on 3^{rd} to 5^{th} toe	Angle of Hallux
TA EMG	0.89	0.28	-0.03	-0.46
EDL EMG	0.77	0.30	0.06	-0.32
PL EMG	0.76	-0.27	-0.59	-0.90
Control term	Angle of 2^{nd} toe	Angle of 3^{rd} to 5^{th} toe	Angle of ankle	Angle of foot arch
TA EMG	-0.57	-0.76	0.96	-0.86
EDL EMG	-0.82	-0.96	0.94	-0.97
PL EMG	-0.64	-0.88	0.97	-0.92
Propulsive term	Angle of 2^{nd} toe	Angle of 3^{rd} to 5^{th} toe	Angle of ankle	Angle of foot arch
TA EMG	-0.77	-0.73	-0.34	-0.23
EDL EMG	-0.58	-0.60	-0.42	-0.17
PL EMG	-0.97	-0.99	0.20	-0.73

The TA in the control term has a high correlation with the ankle angle. This is natural if we consider that they are the muscles that manage the dorsiflexion of the ankle joint. However, the correlation of TA and the ankle angle becomes weak for the propulsion term, but correlation with the hallux load becomes high.

While the ankle joint dorsiflexion for a promotion term is passive, it is considered to cause the muscle of the back of the leg to work when force is applied to the hallux and the muscle opposing the TA muscle. Actually, there is some TA training when standing tiptoe.

The control term has a quite high correlation between the EDL and the toe angle, or arch angle. But the propulsive term has a low correlation. Especially, correlation of the arch angle and EDL is generally lost. This result is considered to be dependent on the Windlass mechanism of the plantar aponeurosis. The Windlass mechanism is the phenomenon in which the arch angle is reduced by the plantar aponeurosis attached to the bottom of the toes by pulling the heel to the front in toe dorsiflexion.

In the control terms, there is correlation of the 3rd to the 5th toe angle, ankle angle, etc., and the PL. This is because the PL muscle manages the varus of the foot. It is observed that the action of the varus becomes stronger, if the patient applies a strong load on the outside of the leg, because it trains the Peroneus longus, and the load to the hallux becomes larger, for example, in the case of a sprain in a clinical situation.

IV. Consideration of the requirements for the model

The requirements necessary for a complete model were considered from two different measurements.

First, by analyzing the sole pressure walking with and without the AFO, it was found that the load is hardly applied to the forefoot when the AFO is worn. This implies that the load is not applied to the MP joint, which is the root joint of the toes. This means that after several years of AFO use, the lack of load applied to the MP joint can have a negative influence on the recovery of the patient. Thus, it is necessary to include the MP joint in the simulation model proposed in this paper. Finally, it is important to note the influence that the toes have on walking, as has been reported previously in various studies.[7][8] As shown in the preceding section, the toes are effective in stabilizing the body during walking.

The experiment to find the relationship between an able person's walk and the EMG signals shows that the influence that each toe has on each muscle is completely different. From this result, we conclude that the model needs to have two or more toes. Furthermore, to see the rehabilitation effect, it is necessary to connect the motion of the model to the activity of the muscles. For these reasons, an 8-axis 8-segment model (Fig. 3) was created using 3D multibody simulation software ADAMS (MSC software).8-segment is lower leg, heel, 3-middle foot, 3-toe. We chose three toe among the following reasons. The most load on the big toe and second toe. And antithenar is also important for healthy gait.

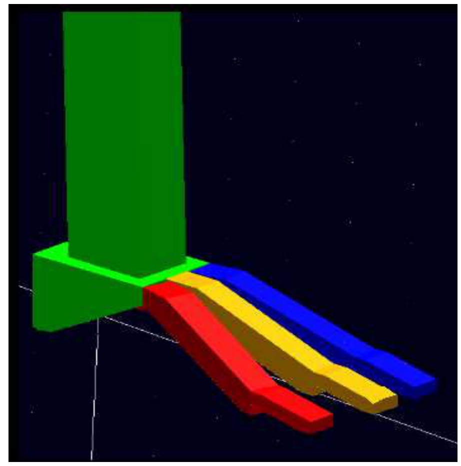

Fig. 3: 8-axis 8-segment foot model

V. Determination of MODEL PARAMETERS

To decide the mass, length, etc., of each component for the simulation model, a right-leg frame model (3B Scientific Co., Ltd) was used. This model is based on the leg of an adult European male. The shoe size is approximately 9 inches. Although it is made of papier-mâché, the model replicates the real parameters of an actual human foot. The papier-mâché model was cut to correspond with each part of the simulation model. The mass of every part and the volume were measured using the Archimedian method. It was hung at two or more places, and the intersection of the perpendicular altitude was determined as the center-of-gravity position. The joint's angles and the rest of anatomical parameters were based on the biomechanics of the human body.

VI. Definition of muscle activity score

To relate the muscle activity to the motion of the simulation model, it was necessary to confirm which movement is related to which muscle from the experimental data. Each data item was changed into a deviation score for each walking cycle by using the measurement of the muscle activity at the time of the able-person's walk and the motion of the foot. Finally, we conducted multiple linear regression analyses and obtained the following equations:

Control term TA

$$a_1=-0.5672+0.057X_2+0.647X_3 \tag{1}$$

Control term EDL

$$b_1=-1.706+0.037X_3 \tag{2}$$

Control term PL

$$c_1=0.619-0.039X_4+0.034X_3 \tag{3}$$

Propulsive term TA

$$a_2=-22.632+0.018X_1+0.340X_3 \tag{4}$$

Propulsive term EDL

$$b_1=+-0.308+0.012X_1 \tag{5}$$

Propulsive term PL

$$c_1=-3.942-0.013X_1-0.017X_5.0.047X_3 \tag{6}$$

where X1: hallux load, X2: 3rd to 5th toe load, X3: ankle angle, X4: 2nd toe angle, and X5: 3rd to 5th toe angle.

We defined the MAS (muscular activity score) as integrated EMG in each walking cycle, since the integration values generally acquired from the EMGs correspond to the muscle power and the straight-line relations at the time of contraction [9]

VII. SIMULATION RESULTS AND CONSIDERATIONS

A. Normal walking

The parameters calculated during the simulation were introduced into each multiple regression equation. The MAS values obtained in the simulation are compared with the actual values shown in Fig.4.

By comparing the real values to the model values, we can conclude that all parameters fit into an error less than 3% with the single exception of the EDL in the propulsive term.

Another consideration is why the MAS of the EDL in the propulsive term differs from the actual value. Table 2 shows the correlation coefficient of each parameter, taken from the experimental value and simulation result and used for the multiple regression equation. Since it is generally accepted that a strong correlation occurs when the correlation coefficient exceeds 0.7, the overall result is very similar to the action shown. But, it is concluded that the reason for the difference is the load of the hallux has a weaker correlation, and the error appears for the propulsive term of the TA muscle.

Table 2: Correlation coefficients of the experimental values and the simulation results

	Correlation coefficient
Load of hallux	0.798
Load of 3-5 finger	0.824
Angle of ankle	0.912
Angle of 2nd finger	0.942
Angle of 3rd finger	0.861

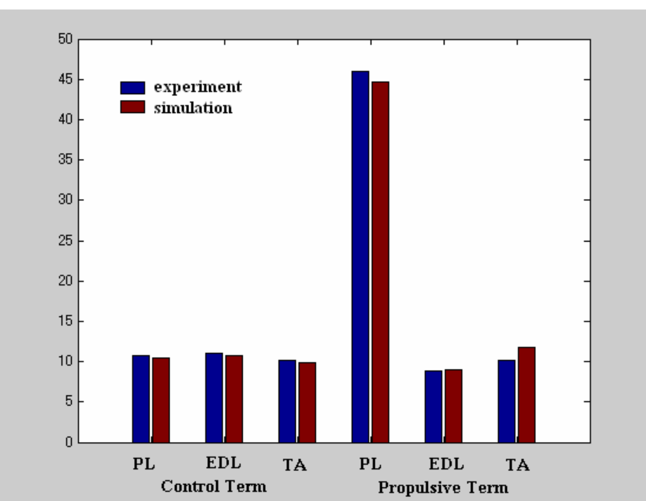

Fig.4: MAS of each muscle

Table 3: Simulation results of the MAS of the PL

	No disturbance	disturbance 50N	disturbance 100N
Control term	11.15	11.68	11.69
Propulsive term	39.89	42.15	49.09

B. Disturbances and environmental changes

Three conditions that are not normal walk were simulated.

1. Arch angle at the time of the increment of the load.

Usually, if the load applied to the knee from the upper part of the body is increased, the load applied to the vertical arch of the leg also increases, so the vertical arch angle becomes larger. Fig. 5 shows the transition of the arch angle when the load is increased to 20 kg in comparison to the arch angle of a normal knee during walking..

Fig. 5: Simulation results of the vertical arch angle

2. Reaction of the PL to the inversion disturbance.

Table 3 expresses the MAS of the PL in the walking simulation, in which the inversion disturbance was added at the end of the control term and the walking was normal, The inversion disturbance is the same power as that causing a sprain. Tokuoumaru reported that strengthening the PL has influence on a sprain [10].

3. The reaction of the TA when the load is increased during dorsiflexion.

Nakao reported that the level of activity of the TA increases when the foot is in dorsiflexion and the load is increased. Fig. 6 shows the level of activity of the TA in normal walking and that applying a 20 kg load. The level of activity is same until about one third of the graph; however, it greatly changes from the middle of the simulated time when it carries a load. This is consistent with the result reported by Nakao [11].

From these three results, it can be said that the simulation responds to environmental changes or disturbances.

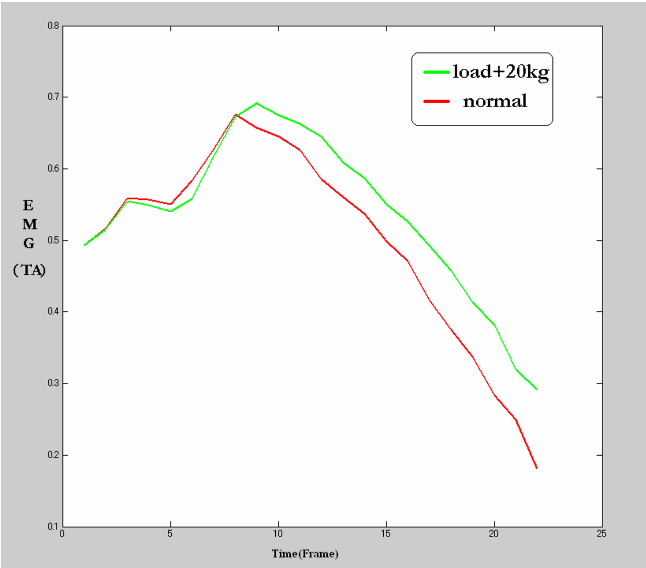

Fig. 6: Simulation result of the EMG of the TA

C. Simulation of wearing the AFO

Next, walking with the AFO was simulated. The AFO model consists of a plank for the sole to prevent rotation of the ankle. Table 4 represents the MAS for normal walking and walking with the AFO.

The same tendency from normal walking to walking with the AFO is seen in the other parameters except for the control term in the TA and the propulsive term in the EDL. These two are influenced by the restriction imposed by the AFO. During the control term, normally the TA supports the foot's weight, but when wearing the AFO, the weight is supported by the AFO instead of the TA. This means that other elements strongly influence the TA. In the propulsive term of normal walking, the toes should bend and all the weight should be concentrated on them, but when the AFO is worn, this weight is distributed to the entire AFO. Because the power to the calf and the belt ware cannot be measured, more experiments and parameters are needed in the model.

Table 4: Simulation results of change of MAS for normal walking and walking with an AFO(normal→with AFO)

Muscular activity score (experiment)	Term Muscle	Muscular activity score (simulation)
10.77→18.01	Control term PL	10.46→17.77
10.97→13.23	Control term EDL	10.73→17.69
10.19→8.54	Control term TA	9.94→12.82
45.96→7.92	Propulsive term PL	44.63→4.19
8.82→7.46	Propulsive term EDL	9.03→16.92
10.21→5.00	Propulsive term TA	11.82→6.15

Ⅷ. CONCLUSION

In this study, we first measured the sole pressure in two different situations, wearing and not wearing an AFO. We found that the load is not applied to the MP joint when the AFO is worn.

Next, we measured the sole pressure, the motion and the EMG of the foot, and the muscles involved during normal walking. From these data we considered the parameters necessary for the simulation model.

The mass, volume, center of gravity, and the moment of inertia of each portion were measured from a human foot model. We then defined the MAS by putting together these measurement data and confirmed the validity of the simulation model in normal walking.

However, it was found that some muscles do not behave as predicted by the AFO simulation, and so we investigated the causes.

In the near future we will increase the number of parameters used in our model until we find an exact model that corresponds to an actual foot, and then we will apply the model to various situations. We will make new AFO model in simulation and attach to the foot model. And compare normal walk simulation and walk with AFO simulation.

REFERENCES

[1]"Disabled person white paper" Ministry of Health, Labor and Welfare http://www8.cao.go.jp/shougai/whitepaper/index-w.html last access date 2010/03/09

[2] Tanida Sousuke, Kikuchi Takehito, Furusho Junji, Ozawa Takuya, Ostuki Kikuko, Fujikawa Takamitsu, Morimoto Shoji, Hashimoto Yasunori、Yasuda Takashi, "The development and clinical evaluation of an intelligent ankle foot orthosis using compact type MR fluid brakes" Journal of the Society of Biomechanisms 34(2), 124-131, 2010-05-01

[3] S. Yamamoto, A. Hagiwara, T. Mizobe, O. Yokoyama, and T. Yasui, "Development of an ankle-foot orthosis with an oil damper," Prosthet. Orthot. Int., vol. 29, pp. 209–19, Dec. 2005.

[4] Delp SL, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. 2007;54:1940–1950.

[5]S.J.Hwang,H.S.Choi,Y.H.Kim,"Motion Analysis Based on a Multi-Segment Foot Model in Normal Walking "Proceedings of the 26th AnnualInternational Conference of the IEEE EMBS San Francisco, CA, USA • September 1-5, 2004

[6] T.Takashima,H.Fujimoto,W.Kakihata,A, Takanishi " Model Analysis of the Human Foot Include Subtalar Joint,Taroclural Joint, Taroclural Joint, and Foot Arch" Transactions of the Japan Society of Mechanical Engineers. C, Vol.69, No.688, 111-116, 2003

[7]H.Takemura,H.Iwama,J.Ueda,Y.Matumoto,T.Ogasawara "A Study of the toe Function for Human Walking." The JSME Symposium on Welfare Engineering 2003(3) pp.97-100 20031107

[8]Alexandria K.B, Richard W, Mark C, Thomas M" The windlass mechanism during normalwalking and passive first metatarsalphalangeal joint extension" Clinical Biomechanics 13(3):190-194

[9] Y.Utiyama,T.Kobayasi K.Mase,"The guide to a measuring method" Kyo-do Isyo Publishing company

[10]K.Tokuoumaru,M.Takahasi,K.Seki, "The effects of exercise of the peroneus longus and tibialis posterior muscles on heel raise" Japanese Journal of Physical Fitness and Sports Medicine Vol. 58 (2009) , No. 3 p.387-394

[11]H.Nakao,M.Hasimoto,Y.Miyamoto,Y.Masuda,H.Takato,H.Isaji"Changes in Muscle Activities of the Lower Leg and the Foot when the Foot Arch Bears a Load" journal of physical therapy science Vol. 24 (2009) , No. 3 pp423-426