31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Patient-Specific Walking Pattern Simulation in a Gait Trajectory Guiding Device

Muhammad Kamrul Hasan, Jang-Ho Park, Seung-Hun Park, Sun-Hee Hwang, Gon Khang

***Abstract*— Repetitive training is of much importance for restoring full-fledged gait ability. At present, task-specific repetitive approach has been proved to be the most effective motor learning concept. In this regard, a gait trajectory guiding device with partial body weight support system can be a solution for gait rehabilitation. This paper presents a complete gait study with an objective to implement the motion of a natural walking pattern in the automated foot-boards of a gait trajectory guiding device. In our developed motion algorithm of foot-boards we have concentrated on adaptation of patient-specific true walking trajectory, determination of variable velocity pattern along different degrees of freedom and time-division for simulating different phases of a complete gait cycle. Gait database, collected from disparate sources and previous gait-studies have been used for kinetic and kinematic analysis of human walking. We have modeled those data based on the previous researches done in this area and adopt them for our motion algorithm. A precise velocity pattern and time-division have been described along different axes so that patient's biofeedback and postural stability in different walking phases can be recorded accordingly and motion-correction of the foot-boards can be done in consecutive cycles through iterative learning control algorithm with the help of motion sensors.**

I. Introduction

MODERN concepts of motor learning have drastically modified the framework of rehabilitation from a conventional neuro-developmental therapy to a more dynamic, task-oriented approach [1]. It is hypothesized that the technique works in part by stimulating remaining force, position, and touch sensors in the legs during stepping in a repetitive manner and that residual circuits in the nervous system learn from this sensor input to generate motor output appropriate for stepping [2]. Several fundamental studies demonstrated that the modulation of many spinal reflexes is also task and phase dependent and contributes to the control of walking and balance [3]. Apart from the post-stroke or SCI patients with hemi paretic lower extremities, task specific repetitive gait training can also be effective for adapting with prostheses for the patients who have gone through surgical procedure for including artificial limbs. Aiming for a device for task-oriented gait rehabilitation training, we have developed a gait trajectory guiding system based on automated foot-boards as shown in Fig.1 [4], [5]. Two hypotheses have led us to this work. First, if we can serve a patient suspended from a body weight support harness with a real and personalized gait-like movement by placing his/her feet on guided foot boards, patient's other joints like ankle, knee, hip will also follow an ideal gait pattern in different phases of a complete walking cycle. Second, allowing the legs to move freely in different degrees of freedom and without any restriction (*e.g.* robotic orthosis) will be more effective for optimal muscular activation during the guided training which will aid the patient to restore an accurate gait afterwards.

The foot-boards' movement has been programmed in such a way, so that it will provide the patients a natural walking experience, while the pressure sensors mounted on the foot-boards will keep track of patient's training performance and postural stability in consecutive sessions. This device acts as an automated tool for the therapists for rehabilitating gait disabled people through repetitive training with consistency and efficiency.

Fig. 1. Prototype of our gait trajectory guiding device: NiceWalker.

The major concern in designing such a device is, the foot-boards have to have the provisions of multidimensional movement with adequate degrees of freedom and variable velocity for supporting a total simulation of natural gait pattern of a normal subject. For doing that, at first we have

Manuscript received April 7, 2009. This work was supported by Korea Institute of Industrial Technology Evaluation and Planning under the Core Technology Research for Silver Medical Devices Project (ITEP-10029934).

M. K. Hasan, J H Park and S H Park are with U-Health Lab, Department of Biomedical Engineering, Kyung Hee University, South Korea (corresponding author: Seung-Hun Park. phone: +82-31-202-2686; fax: +82-31-205-9062; e-mail: parksh@khu.ac.kr).

S H Hwang and G Khang are with Rehabilitation Engineering Lab, Department of Biomedical Engineering, Kyung Hee University, South Korea.

to settle an optimum motion algorithm for the foot-boards for catering the patients of different ages, heights and genders with a walking experience as near to real as possible. The focus of this paper is adaptation of human walking pattern for motion trajectory of such a device considering all the gait parameters and developing an algorithm for a patient specific gait simulation in the foot-boards.

II. Gait Analysis For trajectory Determination

A. *Step Symmetry*

After analyzing the video gait data from [6] and coordinate data from [7], we have found that the horizontal distances between two feet over time show an almost sinusoidal shape which depicts that the horizontal trajectory of the foot-boards should be set in a way so that their mutual distances follow sinusoidly changed values. In Fig.2, the vertical bars represent the distances between feet over time for a sample data while the 2 period moving average trend line indicates its sinusoidal nature.

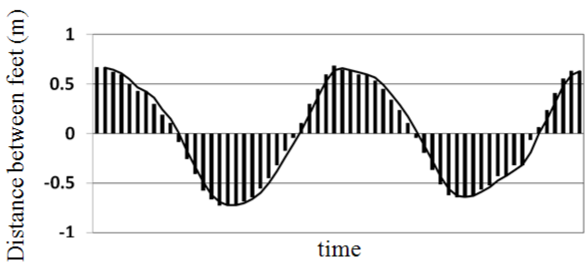

Fig. 2. The horizontal distance between two feet over time for a sample data. Here total 106 frames were considered when the sampling frequency was 69.9 frames/sec.

From this observation it can be inferred that, as both the forwarding and backwarding foot-boards have to travel the same distance at the same time duration, for keeping step symmetry, the variable horizontal velocity pattern should be the same for both foot-boards.

B. *Complete Gait Cycle*

A gait cycle can be divided into two phases: Stance Phase and Swing Phase where the stance phase comprises of 2 double support phases and 1 single support phase [8].

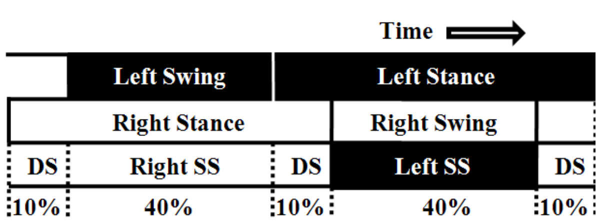

Fig. 3. Different phases of a complete gait cycle (DS = Double Support, SS = Single Support).

As from step symmetry information it has been decided that the forwarding and the backwarding foot-boards will travel the same length of path, for same duration with same velocity pattern, a complete gait cycle has to be divided equally between the two foot-boards. From this deduction we have decided to simulate *Double Support Phase + Swing Phase* in the forwarding plate and *Double Support Phase + Single Support Phase* in the backwarding plate.

C. *Stride Parameters*

Stride Length: There is no significant difference between the corresponding step and stride lengths in repeated trials or between successive step and stride lengths in the same trial [9]. Walking is possible at a wide variety of combinations of *step length sl* and *step frequency sf* $(velocity\ v = sl \times sf)$. A person when asked to walk at a particular velocity is most likely to choose parameters which minimize energy expenditure [11] and the step and stride lengths are related systematically and significantly with height [9], [10]. These observations are expressed in the experimentally derived equations [11] called normalizing formulae, which show a linear relationship between sl and *sf*, where sl and *body_height* are measured in meter, and *sf* in steps/min:

$$\frac{sl}{sf \times body\_height} = 0.004 \tag{1}$$

As, $sl = v/sf$

$$sl = \sqrt{0.004 \times v \times body\_height} \tag{2}$$

Here actually the *body_height* normalizes the equation. It indirectly represents the length of the legs, which has an effect on the preferred step length. Based on (1), if a velocity is defined, a "natural" step length and step frequency can be calculated; in the case where a velocity and a step length are specified a more angular motion might result if the step length deviates significantly from the "natural" one. The parameters have to be checked at a certain threshold to ensure that they are within anatomical limits defined by locomotion attribute e.g. sf_{max}= 182 step/min or sl_{max}= 1.08m [12].

Stride Width and Out-toeing: Stride width is a measure of the transverse distance between points on the central long axes of the feet during foot-to-floor contact. The out-toeing indicates the angle formed by the long axis of the foot with the plane of progression. From the gait experiments of [9][10], it has been found that the average stride width is 8±3cm and the average out toeing angle is 6.76 ± 2.5^0. The tolerance limits here are quite small, that's why we can set the mean values for these two stride parameters which will fit patients of different ages, heights and genders.

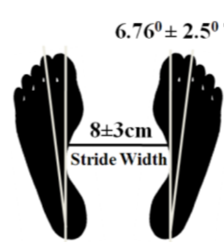

Fig. 4. In-toeing, Out-toeing and Stride width.

D. *Double Support Phase*

The Double Support Phase ranges from Heel Strike or Initial Contact of one foot to Toe Off of another, that's to say Double Support Phase has to be simulated in both the foot-boards at the same time for the same duration.

According to the observation of Fig. 5 based on coordinate data of [7], it is obvious that, the horizontal displacement for both foot-boards during the Double Support Phase is quite insignificant and therefore can be neglected. However during double support phase both the foot-boards have motion in vertical direction.

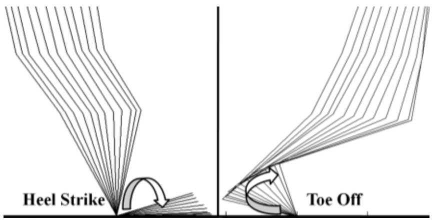

Fig. 5. Insignificant horizontal displacement during double support phase.

Experimental data [11] suggest that in human walking there is an approximately linear relationship between the step frequency and the duration of the double support state as a percentage of a cycle, i.e. the duration of the double support state decreases with increasing step frequency [12]. Based on results from different experiments, Inman et al. have suggested an empirical formula for describing the time for double support phase t_{ds} in terms of step frequency sf and cycle time t_{cycle}

$$t_{ds} = [(-0.16 \times sf) + 29.08] \times \frac{t_{cycle}}{100} \qquad (3)$$

Since sf is known as one of the locomotion parameters, and because of

$$t_{cycle} = 2 \times t_{step} = 2/sf \qquad (4)$$

From (3) and (4) we can deduce that

$$t_{ds} = -0.0032 + \frac{0.5816}{sf} \qquad (5)$$

E. Foot Board Angle and Elevation

The angle and vertical elevation of the foot boards over time are two must-to-know information for developing the motion algorithm of foot-boards. From the raw coordinate data provided by [7] we have determined this information.

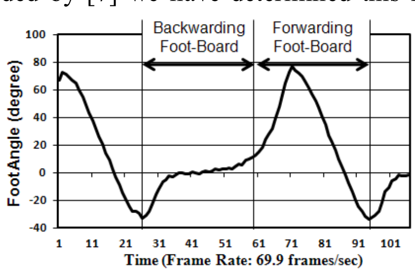

Fig. 6. Sample data of Foot Angels for a complete gait cycle.

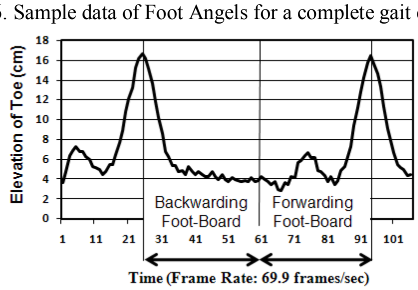

Fig. 7. Sample data of Elevation of Toe/Front portion of a Foot-Board

III. ALGORITHM FOR MOTION PATTERN OF FOOT-BOARDS

A. Foot-Boards Movement Trajectory

For a sagittal plane, from the gait study of Section II, it can be concluded that each foot trajectory can be denoted by a vector $T_f = [X_f(t), Z_f(t), \theta_f(t)]$ where $(X_f(t), Z_f(t))$ is the coordinate of the ankle position and $\theta_f(t)$ denotes the angle of the foot. For simplicity in robotic movement design as shown in Fig. 8 we have adopted the trajectory vector as $T_F = [X_F(t), Z_{1F}(t), Z_{2F}(t)]$ which supports all the components of T_f.

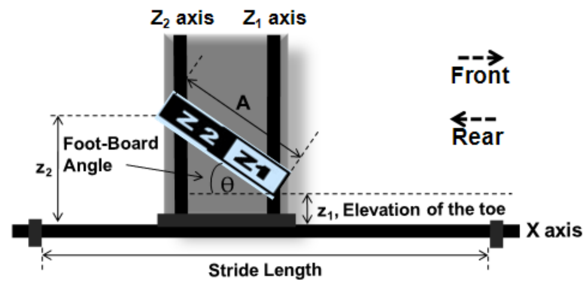

Fig. 8. Axis-configuration of the Foot-Boards.

Now from experimental data of [8], at heel strike or initial contact, the feet are in 25^0 dorsiflexion with the toes up, followed by a total contact with the ground at the end of the loading response- the toes drop towards neutral alignment and maintain this position throughout mid stance. With heel rise in terminal stance, the foot dorsiflexes up to 20^0. This motion continually increases throughout pre-swing to a final position of 70^0 extension. Based on these findings, we can model our data in the following manner for angles of the foot-boards over time:

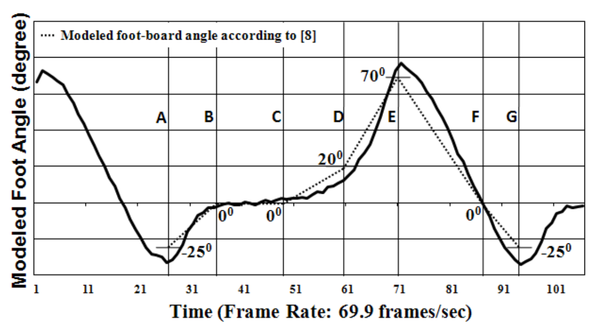

Fig. 9. Data modeling for foot-board angles over time (Time durations of AB, BC, CD, DE, EF, and FG are described in Table I, Table II and Table III.

B. Velocity and Acceleration

From (3) we can find out the amount of time for double support phase and as stated above, for this time there will be no horizontal displacement in both the foot-boards. Fig. 10 shows a sample horizontal velocity pattern based on the data collected from [7] which illustrates that, in a normal walking the horizontal velocity during swing phase follows an elliptical shape-

$$V_{horz} = \sqrt{b^2(1 - t^2/a^2)} \qquad (6)$$

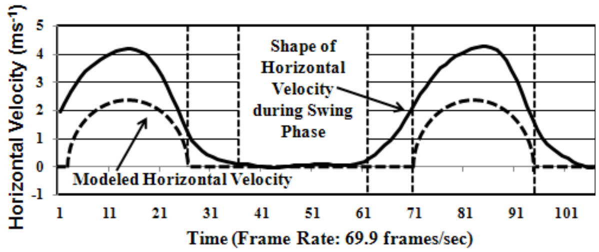

Fig. 10. Sample data of Horizontal velocity of foot during different phases of walking. Modeled Horizontal Velocity was calculated considering v= 5km/h and body_height= 1.8m.

Now from, $\frac{dx}{dt} = V_{horz} = \sqrt{b^2\left(1-\frac{t^2}{a^2}\right)}$

After integration we can determine that,

$$x = \frac{b}{a}\left(\frac{t\sqrt{a^2-t^2}}{2} + \frac{a^2}{2}\sin^{-1}\frac{t}{a}\right) \tag{7}$$

From user input of step frequency or average velocity and height of particular subject, using (1) the value of x and step duration ($t_{max} = x/average\ velocity$) can be calculated. Plugging the value of x= *step length*, $t = t_{step} - t_{ds}$ and $a = t_{step} - t_{ds}$ in (7), we can determine the elliptical shape of the horizontal velocity pattern and acceleration from (6) for the forwarding foot-board during swing phase. According to the decision in *Step Symmetry* subsection of Section II, same velocity pattern will be implemented in the backwarding foot-board during its Single Support Phase.

Fig. 10 shows that, the modeled velocity during swing phase follows an elliptical shape. But one thing to be noted here, in case of the original velocity of the sample data, this graph is also showing some horizontal velocity before toe-off and after the heel-strike. Those velocities are may be from the horizontal component of the upward velocity of feet along Z1 or Z2 direction. These horizontal components have also contributed to the sample data velocity measured during the swing phase. That's why the modeled velocity in Fig. 10 seems to be smaller than the original one.

C. Time Division

In a gait trajectory guiding device apart from being coherent with the patient specific perfect gait pattern, precise time division of the total movement is necessary for keeping track of patient's biofeedback in different phases of the cycle. Although the Ground Contact Force (GCF) signals do not directly provide feedback signals for the control of assistive devices, they do provide a foundation for detecting human motion phases and enable assistive devices to adaptively change the algorithms for each motion phase for better estimation of the feedback signals [13]. It will also be used for reducing the tracking error by a trial-and-error procedure: after each repetition of the gait cycle the feed-forward control signal can be improved by some learning rule. The photoelectric motion sensors installed along every axis of the movement and hall sensors inside the Servo Motor will act in this regard.

The following table summarizes the time division of horizontal movement of the foot-boards

TABLE I
HORIZONTAL MOVEMENT OF THE FOOT-BOARDS

Backward Foot-board	
t_{ds} (From A to B)	No Displacement
t_{step} - t_{ds} (From B to D)	Displacement with variable velocity -V_{horz}
Forward Foot-board	
t_{ds} (From D to E)	No Displacement
t_{step} - t_{ds} (From E to G)	Displacement with variable velocity V_{horz}

All the vertical movements of both the forwarding and backwarding footboards have to be completed within the time t_{step}. Now following the %duration of different phases of a complete walking cycle suggested in [8], time division for movements along two vertical axes is shown in Tables II and III:

TABLE II
TIME DIVISION FOR MOVEMENT ALONG Z1 AXIS

Backward Foot-board	
t_{ds} (From A to B)	From -25^0 to Ground
t_{step} - t_{ds} (From B to D)	At ground
Forward Foot-board	
t_{ds} (From D to E)	At ground
66% of (t_{step} - t_{ds}) (From E to F)	Elevate 4cm and then descend to ground
33% of (t_{step} - t_{ds}) (From F to G)	From ground to -25^0

TABLE III
TIME DIVISION FOR MOVEMENT IN ALONG AXIS

Backward Foot-board	
t_{ds} + 50% of (t_{step} - t_{ds}) (From A to C)	At ground
50% of (t_{step} - t_{ds}) (From C to D)	From ground to 20^0
Forward Foot-board	
t_{ds} (From D to E)	From +20^0 to +70^0
66% of (t_{step} - t_{ds}) (From E to F)	From +70^0 to ground
33% of (t_{step} - t_{ds}) (From F to G)	At ground

From the above information we can easily derive the vertical velocity along Z1 and Z2 direction over time as shown in Fig. 11 (Z1 and Z2 were described in Fig. 8).

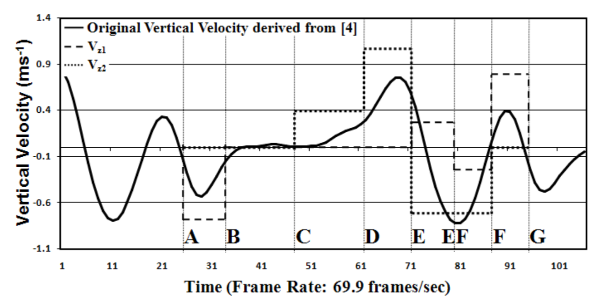

Fig. 11. Vertical velocity of foot-boards along Z1 and Z2 axes in comparison with original vertical velocity of foot. Modeled vertical velocity V_{z1} and V_{z2} were calculated considering v= 5km/h and body_height= 1.8m.

On the basis of the time divisions for horizontal and vertical movements along X, Z1 and Z2 axes shown in Table I, II and III, we have simulated our proposed person specific gait algorithm in MATLAB. Fig. 12 depicts motion of the foot-boards during backward and forward movements. In this figure we have also shown the co-ordinate points for ankle, knee, hip and base rib joints derived from sample data of a normal subject, as according to our first hypothesis we are expecting that the foot-boards movement will also help the patients to follow an ideal trajectory and angle for these joints.

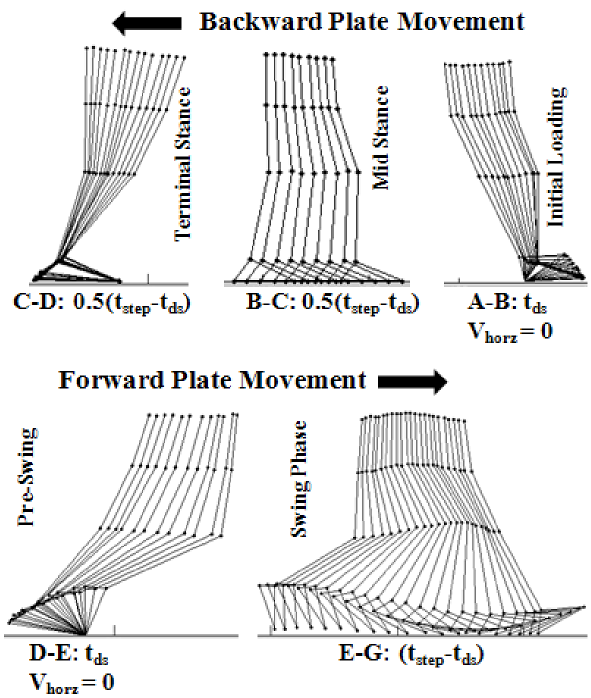

Fig. 12. MATLAB simulation of foot-plate movement for our gait trajectory guiding device.

IV. Conclusion

The main theme of task specific approach is motor learning through repetitive training of a real and ideal walking practice suitable for individual patient's body structure. That's why we have focused our study to find out the best motion algorithm for our gait trajectory guiding device based on true walking pattern so that the patient can experience a real gait like movement and return to a normal life after the training sessions. But in real scenario, applying this motion and velocity pattern precisely in a heavy mechanical system was found to be very complicated. Anyway, the more accuracy we can obtain in simulating gait in our device following the above analysis, the more beneficial it will be for the patients to retain a natural walking ability.

Furthermore, for keeping the foot-boards in its accurate trajectory over time, implementation of a good control algorithm is quite necessary. As we know in different phases of a gait cycle the ground reaction forces of different portions of a foot change, this variation pattern is supposed to be an index of normal gait detection. The time division information will be of much help in this regard.

References

[1] H. Barbeaua, and J. Funga, "The role of rehabilitation in the recovery of walking in the neurological population," Current Opinion in Neurology, vol. 14, pp. 735-740, Dec. 2001.

[2] C. E. Wang, J. E. Bobrow, and D. J. Reinkensmeyer, "Swinging From The Hip: Use of Dynamic Motion Optimization in the Design of Robotic Gait Rehabilitation," in Proc. IEEE International Conference on Robotics and Automation, 2001, pp. 1433-1438.

[3] S. Rossignol, "Neural control of stereotypic limb movements," Handbook of Physiology. Oxford, 1996, pp. 173-216.

[4] M. K. Hasan, S. H. Park, and G. Khang, "Gait Trajectory Guiding Device for Gait Rehabilitation," presented at the Korean Society of Biomechanics Conference with special focus on Rehabilitation and Gerontechnology, October 9-10, 2008, KAIST, South Korea.

[5] M. K. Hasan, S. H. Park, S. J. Seo, D. H. Sohn, S. H. Hwang, and G. Khang, "A Gait Rehabilitation and Training System based on Task Specific Repetitive Approach," presented at The 3rd International Conference on Bioinformatics and Biomedical Engineering, June 11-13, 2009, Beijing, China.

[6] L. Out, J. Harlaar, and C. Doorenbosch, "MoXie Viewer—XML-based sharing of movement analysis results," Gait & Posture, vol. 24, pp. 136-137, Dec. 2006.

[7] D. A. Winter, Biomechanics and Motor Control of Human Movement, Third Edition, John Wiley and Sons, Inc., 2005.

[8] Perry J., Gait analysis: normal and pathological function. Thorofare, NJ: Slack Inc., 1992.

[9] M. P. Murray, A. B. Drought, and R. C. Kory, "Walking Pattern of Normal Men," Journal of Bone and Joint Surgery, vol. 46, pp. 335-360, 1964.

[10] M. P. Murray, R. C. Kory, and S. B. Sepic, "Walking patterns of normal women," Archives of Physical Medicine & Rehabilitation, vol. 51, pp. 637-650, 1970.

[11] V. T. Inman, H. J. Ralston, and F. Todd, Human Walking. Williams & Wilkins, Baltimore, 1981.

[12] A. Bruderlin, and T. W. Calvart, "Goal-Directed, Dynamic Animation of Human Walking," Computer Graphics, vol. 23, Num. 3, July 1989.

[13] K. Kong, and M. Tomizuka, "Smooth and Continuous Human Gait Phase Detection Based on Foot Pressure Patterns," IEEE International Conference on Robotics and Automation, May 19-23, 2008, Pasadena, CA, USA.