Proceedings of the 4th World Congress on Electrical Engineering and Computer Systems and Sciences (EECSS'18)
Madrid, Spain – August 21 – 23, 2018
Paper No. EEE 110
DOI: 10.11159/eee18.110

Development of Smart Wearable Exoskeleton Robot

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Abstract - Wearable lower extremity exoskeletons are assistive tools developed for the purposes of labor-saving in effort, lowering bone joint loads and reducing wear and wear on joints during walk or rehabilitation exercises. There are two applications, with one being for medical rehabilitation so patients who need motion exercises can avoid the waiting time in hospitals and also reduce the manpower strain of physical therapists. The other application is in autonomous walking control in which the electromyography (EMG) of human body is used to control activation and shutdown of assistive tools. These two applications both use embedded systems as a control center and g-sensors for gesture recognition to control the balance of exoskeletons. Motion detection by EMG allows computer systems to learn the information on how patients would like to control the exoskeletons. The autonomous walking mode uses zero moment point and ANFIS to stabilize the body from tilting during walking motion. D-H method is used to define the robotic linkage coordinates, followed by the use of homogeneous matrices to derive forward kinematics. Algebra and geometry are then used for inverse kinematics, with PID control making the motions smoother.

Keywords**:** Exoskeleton, Gait Analysis, Assistive Walking, G-Sensor, Electromyography Signals, Zero Moment Point, Forward And Inverse Kinematics.

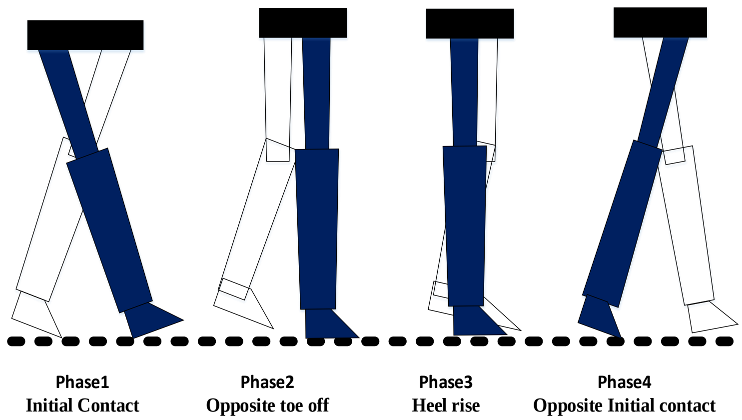
1. Introduction

In recent years, the global aging issues continue to emerge and the demand for rehabilitation that goes with elderly care just increases day by day. Under these circumstances, a smart rehabilitation equipment that assists people in various rehabilitation exercises will certainly ease doctors' burden and provide good quality in medical care.

Balance and stability need to be considered in the studies related to lower extremity exoskeletons. In addition, assistance and control based on the principle of zero moment point [3] need to be added to the center of gravity. Therefore, the compensation generated by the upper torso during walking for instantaneous balance control [5] is used to conduct ANFIS computation.

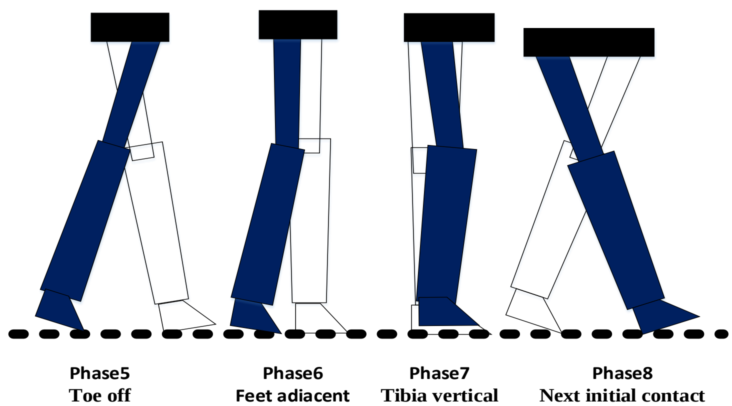
This paper mainly presents the use of g-sensors and electromyography [9], DC motors, encoders and Maxon original driver, with an embedded system from Raspberry Pi of the UK that pairs up with Broadcom's ARMv8 architecture to process various signals. The angle signals from the g-sensors, the EMG of thigh quadriceps and the signals from the rotary encoders of motors are acquired and analyzed real-time. Then, forward and inverse kinematics and the principle of zero moment point are used to derive an operation mode that allows stable walking. In a human-machine interface, the modes of rehabilitation or autonomous walking can be selected, making the equipment smart and multi-functional.

A gait cycle is the time period of movements in which one foot heel contacts the ground to when that same foot again contacts the ground. This cycle can be divided into two phases. The first is the stance phase as shown in Fig.1 and the second is the swing phase as shown in Fig.2. In Phase 1, there are four step changes, from the first, initial contact, to the second, response, followed by the third, mid-stance and then the fourth, stance end and at the same time entering Phase 2. Entering the 5th movement, pre-swing, the other foot moves and leaves the ground to perform the initial swing (6th movement). It then accelerates into mid-swing (7th movement) and then decelerates to enter terminal swing (8th movement), thus completing a gait cycle. The principle of zero moment point [1] is then added to this gait cycle to complete the steady-state walk.



Stance phase

Fig. 1: Gait definition of the first phase-Stance Phase.



Swing phase

Fig. 2: Gait definition of the second phase-Swing phase.

2. Method

In this study, both the mechanical design and the assembly of finished product are original. The system uses webpage and C++ control and is written originally. The EMG of human body [9] is extracted for real-time monitoring, together with the computation using the principle of zero moment point and forward and inverse kinematics, to make this assistive technology smarter.

2.1. Exoskeleton Structure

The exoskeleton structure made for this study, as shown in Fig.3, consists of hip supports, thigh supports, calf supports and soleplates. The hips and thigh form one degree of freedom, the thighs and calves form another and then the calves and feet form another. Therefore, the exoskeleton structure in this study has 6 degrees of freedom, with each being an independent motor joint. The 6 rotary joints act as pitch joints to control the gait size. With g-sensors, a reducer with a reduction ratio of 1:1400, motors and motor encoders, the design of the two joints of feet is based on zero moment point to make a steady-state trajectory, thus increasing the area of contact between the bottom and the ground. This assistive tool offers a large torque output to help patients.

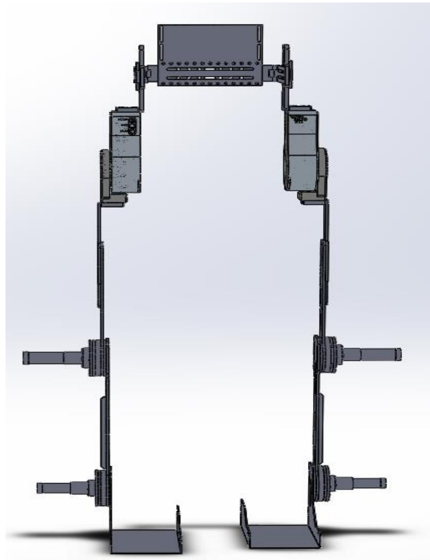


Fig. 3: 3D Schematic Showing an Exoskeleton Structure.

2.2. Process Flow Showing The Movements Of A Wearable Walking Exoskeleton

A flow of the wearable exoskeleton assistive tool is shown in Fig.4. In this study, it includes a medical rehabilitation mode and an EMG control mode, offering assistance to meet different needs. The medical rehabilitation mode allows medical staff to program effective parameter settings. During the process, the g-sensors and motor encoder send out real-time signals through Raspberry Pi embedded system to determine angles, further sending operating signals to the motor. In the EMG control mode, the status of signals from the g-sensors and motor encoder are determined by the Raspberry Pi embedded system and then the EMG signals activate the threshold settings [9], making the motor operates.

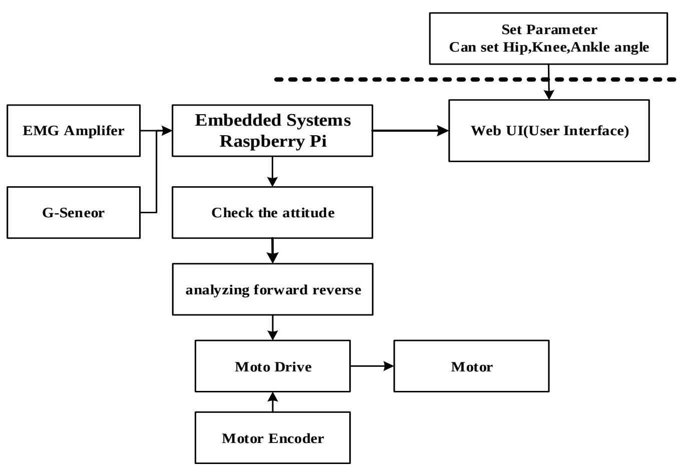


Fig. 4: System flow chart.

2.3. The Use Of Human-Machine Interface

The controls needed in this study are done through a touch screen on the tool, as shown in Fig.5. This system has two functional options which are control and smart modes, as shown in Fig.6. In the control mode, the angle changes of motors and the number of repetitions for the rehabilitation actions can be set in the system parameter settings. This enables medical staff to just stand by and monitor the process. In the smart mode, the EMG signals of human body drive the tool to move. Upon completing the parameter settings and menu options, the assistive tool is activated. The monitoring window can provide the EMG signals of human body, rotary angles of motors and the current posture to let patients and the people doing the monitoring to clear understand both the body and the tool status.

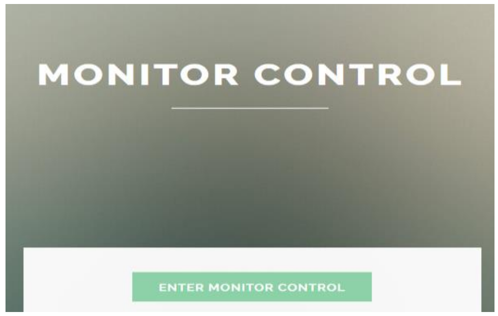


Fig. 5: Human-machine interface.

Fig. 6: Menu options.

2.4. Wisdom-Assisted Human Body Motion Detection And Analysis

Human body angle detection will use the three-axis accelerometer and three-axis gyroscope to detect the human body's action state, the sensor detects the signal provided to the embedded system to calculate the angle value and applied to the parameters in the program writing, the parameters provided to the motor One of the driving conditions, and with the motor speed control in different angles to give different speeds, while the user's monitoring screen can be observed in the human body dynamic changes, to provide users and medical personnel real-time status monitoring.

Acceleration accelerometer for measuring the acceleration of the chip, with the acceleration in the aids and space generated by the acceleration of gravity in the free-fall accelerometer and the acceleration of gravity will cancel each other offset Accelerometer three-axis output will be zero, but the accelerometer when the angular velocity is calculated, an error occurs due to the acceleration motion. Therefore, when the accelerometer is used to monitor the state, it will be used simultaneously with the gyroscope. The accelerometer will not accumulate errors for a long time and will be sensitive and accurate for a short time, so as to achieve accurate state detection.

Because of through this action detection, with the myoelectric signal analysis, auxiliary with positive and negative movement, ANFIS, and the theory of stable walking, it makes the first step in the wisdom of the human body EMG signal source difference analysis through angle detection, coupled with the theory of kinematics and gait stability, intelligent accessory walking movement.

2.5. Electromyography Analysis

Surface electromyography (SEMG) is used to extract signals. Electrode pads are placed along the muscle fibers of thigh quadriceps. Before use, the excessive body hair on the skin need to be removed. Rubbing alcohol needs to be applied to clean the skin surface to ensure that the skin and the pads are free of any interference. After applying the pads, the signals are amplified by the front-end amplifier. High- and low-frequency noises are removed after they go through both high-pass filter (1KHz) and low-pass filter (20Hz), followed by the ADC conversion in which 10KHz sampling rate is performed for the analog and digital signals before the conversion to the embedded system. The EMG signals can be presented and recorded on a webpage after being processed and then go through C++ programming for further signal processing, as shown in Fig.7.

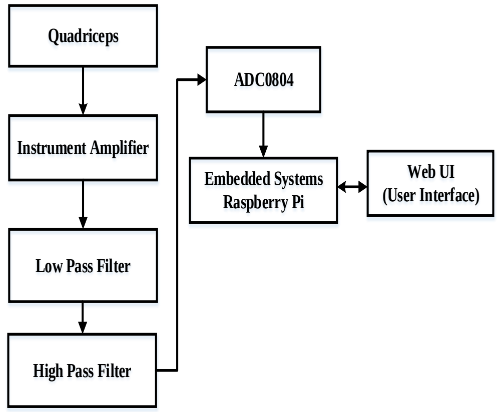


Fig. 7: Acquisition process of EMG signals.

In this paper, the signals of the EMG control mode need to be monitored in real-time. Therefore, the time domain processing being used has to take the linear envelope method as the most basic processing. The linear envelope method performs full-wave rectification of the original signals and then adopts program coding to allow the low-pass filter to display real-time status.

$$\mathrm{EMG(Full-wave-rectifier)} = |\mathrm{EMG}| \quad (1)$$

After obtaining the signals from the low-pass filter, we next use root mean square to obtain the average of EMG signals.

$$\mathrm{RMS} = [\frac{1}{T}\int_{T}^{1+T} EMG^2(t)dt]^2 \quad (2)$$

2.6. Stable walking

To maintain body balance and not tilting with the equipment on during the gait, the center of gravity needs to be controlled. We would take the upper limbs for the purpose of real-time balance control during the gait. Fig.8 shows the distribution diagram of the reaction force between feet and ground during walking stride. The reaction force is set at point P and we also define the concentration force at N and torque as M. If P is on a flat plane and the resultant moment of force of the ground and point P is zero, then point P can be defined as the zero moment point. During the walking stride, or the phase with one leg as a support, the reaction point is to be at A based on the theory. Under a standstill with both legs, the reaction point can be within the range of ether point A or B, as shown in Fig.9. Wearing the assistive equipment within this range enables steady walking stride, and based on this theory, the ZMP location (3) of X direction and the ZMP location (4) of X direction can be derived. $X_i, Y_i\ and\ Z_i$ are the coordinates of the i linkage; m_i is the mass, g is the gravitational acceleration, $I_{ix}\ and\ I_{iy}$ are the moment of inertia and $\ddot{\Omega}_{ix}\ and\ \ddot{\Omega}_{iy}$ are the angular velocity.

$$X_{zmp} = \frac{\sum_i m_i[(\ddot{Z}_i - g)X_i - Z_i\ddot{X}_i] - \sum_i I_{iy}\ddot{\Omega}_{iy}}{\sum_i m_i(\ddot{Z}_i + g)} \quad (3)$$

$$Y_{zmp} = \frac{\sum_i m_i[(\ddot{Z}_i - g)Y_i - Z_i\ddot{Y}_i] + \sum_i I_{ix}\ddot{\Omega}_{ix}}{\sum_i m_i(\ddot{Z}_i + g)} \quad (4)$$

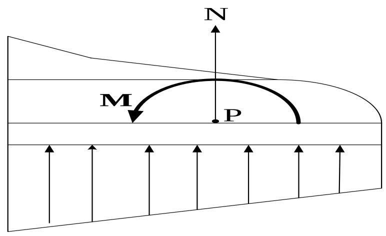


Fig. 8: XMP point.

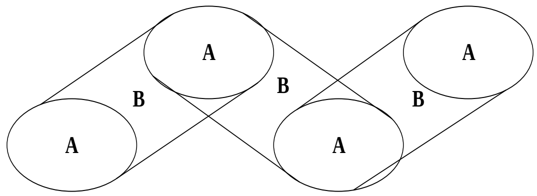


Fig. 9: Gait range.

2.7. Forward and Inverse Kinematics Of The Assistive Equipment

Forward and inverse kinematics are used in this study [4], together with the principle of ZMP to plan a stable gait path. A D-H parameter table is defined for each joint of the equipment, followed by the use of homogeneous matrices to derive forward kinematics and then the use of algebra and geometry for inverse kinematics.

Linkage mechanism of the equipment needs to be provided in order to derive forward kinematics, and the placement of motors is symmetrical right and left. Therefore, to define 6 linkage coordinates, we can first generate a parameter table for the right side and then define the left side based on the rule of symmetry. The D-H table defines that: α_i is the distance from Z_n axis to Z_{n+1} axis on X_{i+1} axis. α_i represents the angle from Z_n axis to Z_{n+1} axis on X_{i+1} axis. d_i represents the distance from X_i axis to X_{i+1} axis on Z_{i+1} axis. θ_i represents the angle from X_i axis X_{i+1} axis on Z_{i+1} axis. By defining the coordinates of joints, we then derive the forward kinematics (M) of the unilateral hip joint coordinates (3) corresponding to the ground coordinates (0).

Table 1: D-H linkage parameters.

i	α_i	a_i	d_i	θ_i
1	α_1	0	0	θ_1
2	α_2	0	0	θ_2
3	α_3	0	0	θ_3

$$^0M_3 = {}^0M_1\, {}^1M_2\, {}^2M_3 \tag{5}$$

2.8. The Use of ANFIS

Applied to the computation of ZMP to make the movement action smoother and closer to the actual human body movement, ANFIS is a flow that processes theory and human logical inference. This method can deal with inaccurate or fuzzy systems and composite algorithm is used which adjusts all parameters in the system. As ZMP requires real-time balance and the movements need to mimic the actual walking actions, we bring in angles and the readings from motor encoder into the ANFIS models for computation.

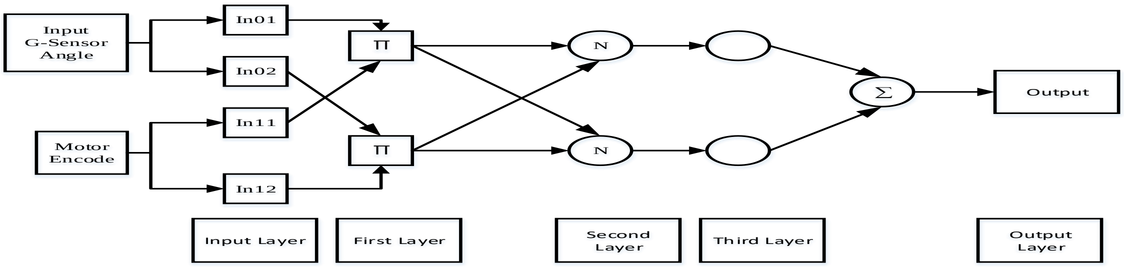


Fig. 10: Schematics showing ANFIS.

3. Simulation Results and Discussion

In this experiment, the values derived from the theory of zero moment point while the actual human body is in a steady standstill posture can generate steady walking stances within the ZMP theory, as shown in Fig.11. However, there would be instability and potential fall-over if we deliberately stay away from that position, as shown in Fig.12. The experiment finds that if the center of gravity of the gait is not in the path derived from theory in advance, the instantaneous change of X- and Y-axis can be huge, according to the charts. The amplitude larger than 2000 indicates that the center of gravity is lost and it can easily cause fall. Among these data, we only need to determine X- and Y-axis, as these two are the X- and Y-axis directions on actual human feet and the data are the actual acceleration (acceleration = speed / time), minus the gravity, which is the angle change after converting the angles of the g-sensors.

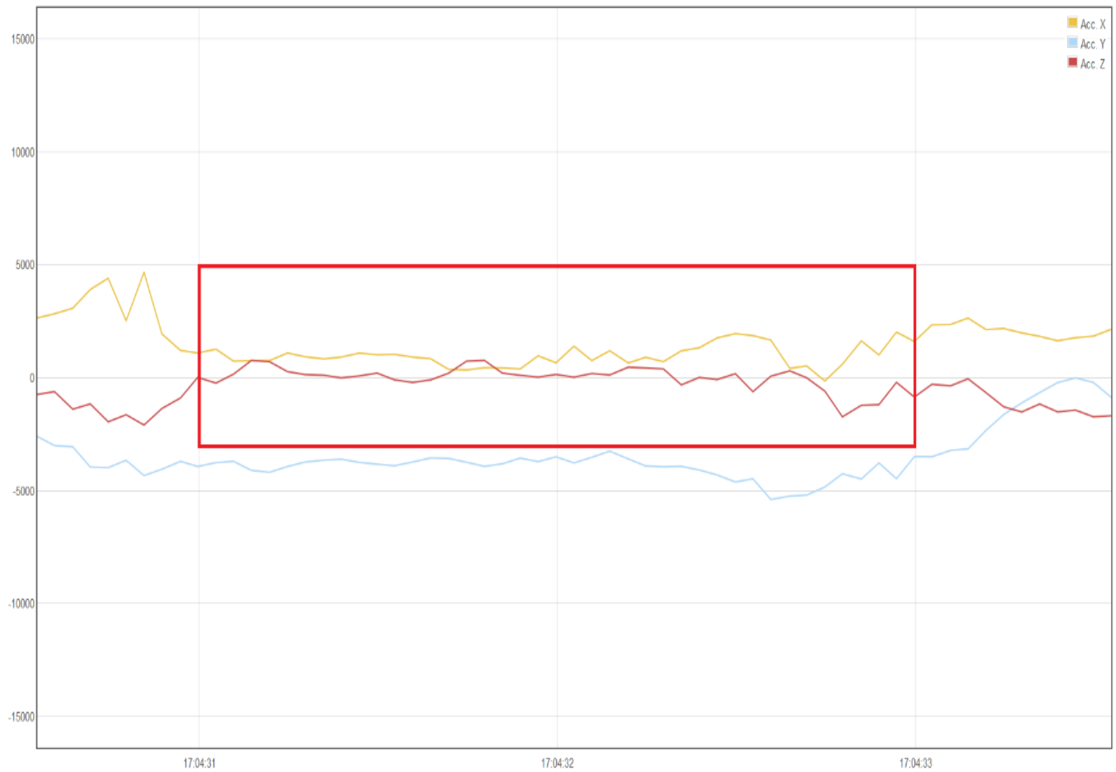


Fig. 11: A complete gait cycle that shows the changes of X and Y and normal walking on ZMP.

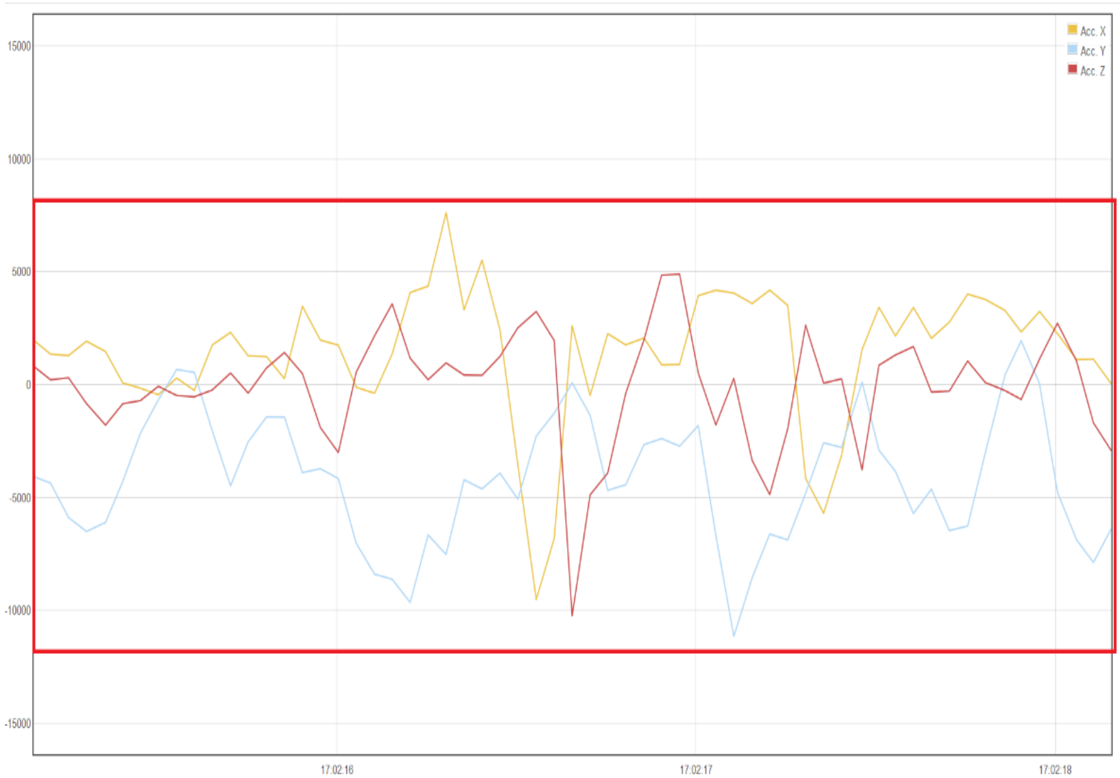


Fig. 12: A gait cycle that shows the changes of X and Y and is not on the points according to the theory of ZMP, and it is completely unstable.

4. Conclusion

This paper proposes that the use of forward and inverse kinematics, together with zero moment point, can offer the best solution for balance during steady walking. The changes in kinematics can make movements closer to the actual human body motions, and ANFIS and PID motor control algorithm can smoothen the motions. Software simulation, the gait verification derived from the actual theory and the data from the analysis of angle information can be used in the gait system. After repeated computation, validation and tests, the application of assistive equipment for patients can be more stable. The functions of equipment proposed in this study can be really helpful to both doctors and patients if they are put to good use, and the workload for medical staff can be reduced, as well. In the future, algorithms that can mimic human brains to make the system even more stable will be developed to allow the equipment to have postures that are even closer to the normal walking so that users and patients can steadily walk every step. The potential use of brainwave in the selection of modes will make the equipment smarter and more convenient.

Acknowledgements

This study is supported by the funding of project number MOST 106-2221-E-167-001 from Ministry of Science and Technology. We hereby offer our sincere gratitude to the Ministry.

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