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A DUAL MODE EMG-CONTROLLED ROBOTIC ORTHOSIS

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ABSTRACT

This paper presents the design and development of an EMG-controlled robotic orthosis aimed to assist stroke patients in rehabilitation process, and possibly to mitigate the adverse effects caused by stroke. The robot could operate in dual control modes, (i) by using EMG signals to control the joint direction and (ii) by using a reference gait trajectory as a control signal. Mathematical model of the robot is derived via Lagrange equations. The dual mode control system is implemented successfully and tested in both simulation and experiments.

Keywords: electromyography, robotic orthosis, control.

INTRODUCTION

When stroke patients suffer from paralysis, continued rehabilitation and therapy will be able to help them to recover but the process could be prolonged. If the paralysis happens to the lower limbs, rehabilitation might be needed to restore the basic gait function [1]. The rehabilitation is suggested to be carried out within the first six months after stroke where the most recovery occurs [2]. The sooner the rehabilitation starts, the more efficient the brain pick up those normal movements again at the mean time recognizing the sensory and motor pathways of the brain [3].

Patients who suffer from lower limbs paralysis are not able to complete the gait rehabilitation by themselves. A personal physiotherapist is needed in order to support them throughout the rehabilitation. The employment of physiotherapists as a sole trainer in the rehabilitation exercise requires huge stamina and intensive focus on the patients and this often leads to fatigue of the trainer. The increasing involvements of mechatronics and robotics technology in biomedical engineering applications over the last two decades has opened up a great avenue in rehabilitation process to reduce the burden of a physiotherapist.

Lokomat and AutoAmbulator are the two common stationary rehabilitation systems available commercially. These devices basically support the stroke patients to walk on a treadmill. AutoAmbulator provides robotic arms attached to the patients for controlling their lower limbs [4]. As for the Lokomat, the device is an exoskeleton incorporated with a body weight support system, along with linear actuators to control the joint angles at the knee and hip [5]. These devices share the same disadvantage. The movements are limited to sagittal plane only, therefore does not provide significant balance training. There is also a limitation to the walking trajectory. The patients are guided to perform a pre-set walking trajectory through certain control strategies [6]. Therefore patients do not perform walking volitionally.

There is an alternative approach other than treadmill-centred technology, which is to use foot plates to guide the feet in order to reproduce gait trajectories. An example of such approach is the Gait Trainer. The device uses a crank and a gear system as guidance to the feet, simulating the stance and swing phases of walking [7]. The approach can also provide more degrees of support to stroke patients. Meanwhile the Haptic Walker is designed for more random movements of the feet to adapt walking on different surfaces [8]. Other systems such as GaitMaster and G-EO Systems even allow the simulation of stair ascent and descent [9-10].

Robotic devices with over-ground walking system are mobile and can be carried around. This has improved the mobility and convenience in usage. For example, KineAssist has a mobile base and assists movements for pelvis and torso, while allowing the assistance of therapists by leaving the patients' legs freely [11]. Meanwhile WalkTrainer utilizes an actuator to track the movements of the patient and is capable of controlling the motion of pelvis in six degrees of freedom with its parallel robotic structure [12]. However, the balance training is difficult due to the hip movement being restricted. The Hybrid Assistive Leg (HAL) is a wearable system comprised of exoskeleton driven by electric motors [13-15]. The exoskeleton can be controlled by the EMG signals of the muscles when the user trying to move the limbs. The power units are used to amplify the user motion and the suit is powered by batteries.

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This paper presents a design, development and control analysis of a dual mode two degree-of-freedom (DOF) robotic orthosis. The robot can be operated in two modes of control, namely (i) reference trajectory control and (ii) EMG control. It aims to assist stroke patients in rehabilitation process, and possibly to mitigate the adverse effects caused by stroke.

SYSTEM ARCHITECTURE

Figure-1 illustrates the system architecture of the design. The main unit is a mechanical exoskeleton frame designed to strap onto the user. Four DC motors (Maxon DC brush motor model 148867) controlled by separate motor driver serve to power the hip and knee joints on both legs. A microcontroller (Arduino Mega) board is responsible for receiving motion sensor signals and sending control commands to the motor drivers. A wireless EMG unit (Shimmer ExG) acquires muscle strength signal from the user and send it to a laptop via its built-in Bluetooth. The laptop subsequently communicates with the microcontroller board after processing the EMG signals in Matlab.

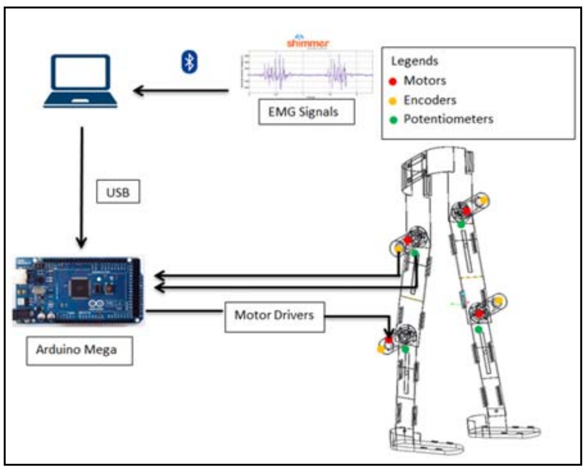

Figure-1. Schematic diagram of the robotic orthosis.

The mechanical frame is made of two aluminum plates for each link with a slot mechanism connected to it. This design facilitates the users to adjust the length and width of the link according to their height. Furthermore the design does not require additional material and no wielding is needed. To fasten the exoskeleton frame to the user, Velcro strap has been chosen to be the most suitable material as it has varies adjustable length and it can withstand high tension.

The microcontroller board (Arduino Mega 2560) is used to drive and control four DC motors through the motor driver. Discrete digital signals are used to control the direction whereas digital Pulse Width Modulation (PWM) signals are used to control the current fed to the motor.

Encoders within the motor driver are used to measure the angular position and velocity of the motor shaft. To double confirm the angular position of the link, a 100 kΩ linear potentiometer is used. It is attached to the stationary link right below the motor and connected to the rotating link by a latch made of a thin aluminum plate.

The photograph of the completed prototype is shown in Figure-2.

DYNAMIC MODEL

The exoskeleton could be represented by a uniform two-link double pendulum model connected in series where the motors are attached to the front of each link. The dynamic model could be derived by using the Lagrange equation as follow:

$$\frac{\partial}{\partial t}\left(\frac{\partial T}{\partial \dot{q}_i}\right)-\frac{\partial T}{\partial q_i}+\frac{\partial V}{\partial q_i}+\frac{\partial F}{\partial \dot{q}_i}=Q_i \qquad i=1,2,\ldots,N \tag{1}$$

where

q_i = Generalised Displacement Coordinates
T = Kinetic Energy of the system
V = Potential Energy of the system
F = Rayleigh Dissipation Function
Q_i = Generalized Force associated with coordinate q_i

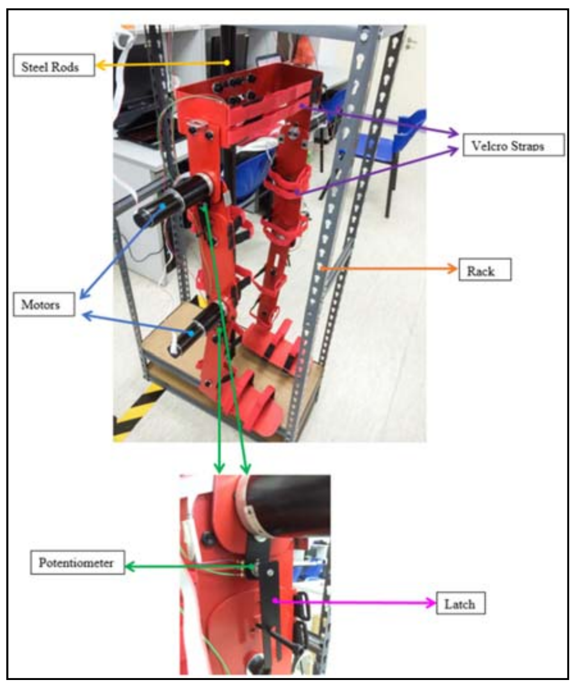

Figure-2. Photograph of the completed prototype.

By deriving the system kinetic energy, potential energy and the generalized force, the complete equations of motion of the robotic orthosis are as follows:

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For Link 1:

$$
\begin{aligned}
&\frac{1}{3}m_1 l_1^2 \ddot{\theta}_1 + m_2 l_1^2 \ddot{\theta}_1 + \frac{1}{3}m_2 l_2^2 \ddot{\theta}_1 - \frac{1}{3}m_2 l_2^2 \ddot{\theta}_2 + m_2 l_1 l_2 \ddot{\theta}_1 \cos\theta_2 \\
&- m_2 l_1 l_2 \dot{\theta}_1 \dot{\theta}_2 \sin\theta_2 - \frac{1}{2}m_2 l_1 l_2 \ddot{\theta}_2 \cos\theta_2 + \frac{1}{2}m_2 l_1 l_2 \dot{\theta}_2^2 \sin\theta_2 \\
&+ m_{m2} l_2^2 \ddot{\theta}_1 + \frac{1}{2}m_1 g l_1 \sin\theta_1 + m_2 g l_1 \sin\theta_1 + \frac{1}{2}m_2 g l_2 \sin(\theta_1 - \theta_2) \\
&+ m_{m2} g l_1 \sin\theta_1 = \tau_1 - \tau_2
\end{aligned}
\tag{2}
$$

For Link 2:

$$
\begin{aligned}
&-\frac{1}{3}m_2 l_2^2 \ddot{\theta}_1 + \frac{1}{3}m_2 l_2^2 \ddot{\theta}_2 - \frac{1}{2}m_2 l_1 l_2 \ddot{\theta}_1 \cos\theta_2 \\
&+ \frac{1}{2}m_2 l_1 l_2 \dot{\theta}_1^2 \sin\theta_2 - \frac{1}{2}m_2 g l_2 \sin(\theta_1 - \theta_2) = \tau_2
\end{aligned}
\tag{3}
$$

where m_1, m_2 and m_{m2} are the mass of the first link, second link and the second motor respectively (the first motor is fixed to the frame and does not move). θ_1 and θ_2 are the hip and knee joint angles. l_1 and l_2 are the length of the first and second link respectively. τ_1 and τ_2 are the joint torques.

As mentioned earlier, four DC servomotors with a gear ratio of N=353:1 are used to power the four joints on both legs. The motor reference torques could be computed by referring the motor angular acceleration and motor angular velocity back to the link:

$$
\begin{aligned}
N\tau_{m1ref} &= N^2 J_{m1}\ddot{\theta}_1 + N^2 k_b \dot{\theta}_1 + \tau_1 \\
N\tau_{m2ref} &= N^2 J_{m2}\ddot{\theta}_2 + N^2 k_b \dot{\theta}_2 + \tau_2
\end{aligned}
\tag{4}
$$

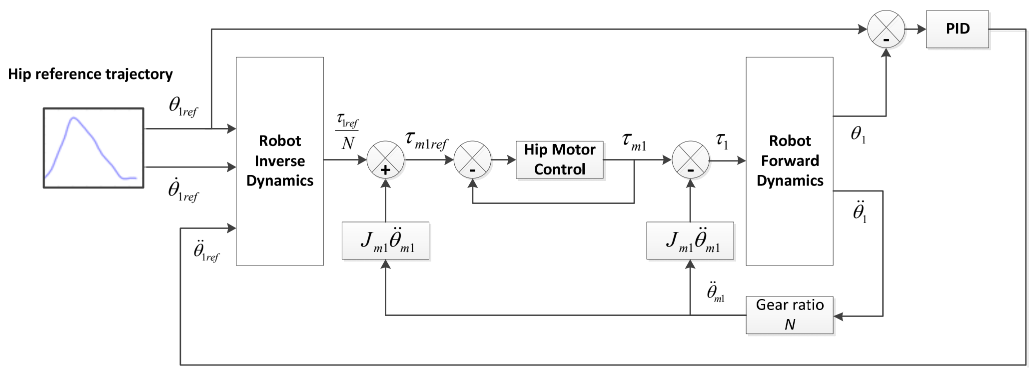

Figure-3. Hip reference trajectory control system block diagram.

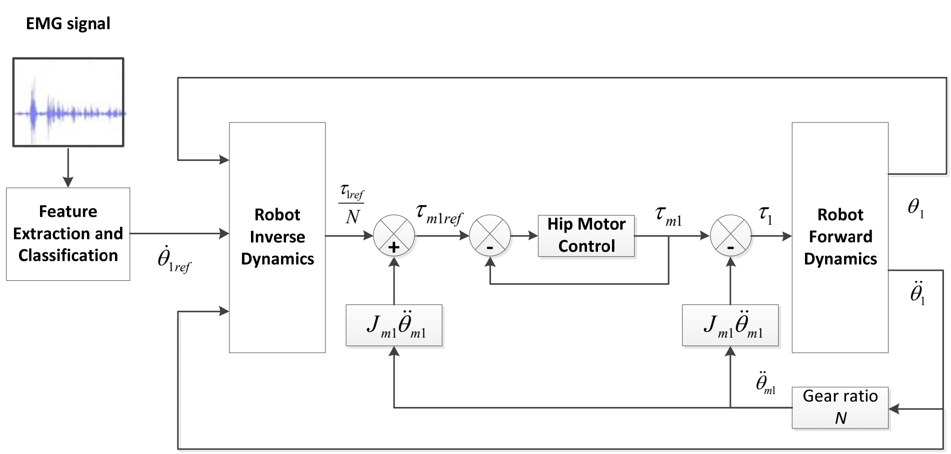

Figure-4. EMG control system (for hip joint) block diagram.

where τ_{m1ref} and τ_{m2ref} are the motor reference torques located at hip joint (first motor) and knee joint (second motor) respectively. N is the motor gear ratio, J_{m1} and J_{m2} are the hip and knee motor inertia, k_b is the motor damping coefficient.

The electrical equations of motion of the motor could be derived from the Kirchhoff voltage law:

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$$V_{m1} = Ri_1 + L\frac{di_1}{dt} + E_1 \tag{5}$$

$$V_{m2} = Ri_2 + L\frac{di_2}{dt} + E_2$$

$$E_1 = k_v\dot{\theta}_{m1};\; E_2 = k_v\dot{\theta}_{m2} \tag{6}$$

$$\tau_{m1} = k_t i_1;\; \tau_{m2} = k_t i_2 \tag{7}$$

where V_{m1} and V_{m2} are the voltage applied to the hip and knee motors, R is the motor winding resistance, i_1 and i_2 are the motor current, L is the motor inductance, E_1 and E_2 are the motor back EMF voltage, k_v is the motor back EMF constant and k_t is the motor torque constant. τ_{m1} and τ_{m2} are the actual motor torques.

CONTROL SYSTEM

The robot can be operated in two modes of control, namely (i) reference trajectory control and (ii) EMG control.

In reference trajectory control, the robot is driven according to a predefined reference trajectory. Specifically, the hip (Joint 1) and knee (Joint 2) angle trajectories are estimated through the data in [15]. Figure-3 shows the block diagram of the hip trajectory control system.

Using hip as an example, the reference trajectory and the reference angular velocity of it are fed into the robot inverse dynamics in order to compute the desired torque. The initial angular acceleration of the hip is assumed to be zero. There is a feedback loop in the motor control to ensure that the value of output torque is as close as possible to the computed torque. Finally, the output torque is fed into the robot forward dynamics in simulation, or in practical implementation, the actual robot. The angular position of the hip is then compared with the reference hip trajectory; the error signal will be fed to a PID controller. This will be treated as the angular acceleration in the following control loop and fed into the robot inverse dynamics subsequently.

In the EMG control mode, the EMG signal is used to generate a binary control action to command the hip and knee joints to rotate in the respective direction. The block diagram of the EMG control mode is shown in Figure-4.

RESULTS AND DISCUSSIONS

The simulation model was built in Matlab Simulink. Figure-2 shows the photograph of the prototype as well as its experimental setup. The exoskeleton is assembled and attached securely to the steel rods which are tight fitted into the rack. The feet of the exoskeleton are lifted above the ground while conducting experiments to prevent friction and movement in order to get accurate results. The potentiometers are installed right below the motor and a latch is used to rotate the potentiometer according to the movement of the link below it. The exoskeleton is coated with a layer of insulation and paint to prevent any electric conductivity.

In the EMG control mode, the EMG electrodes were placed on the Rectus Femoris and Biceps Femoris muscles respectively. The user was asked to contract and flex each muscle alternatively so as to generate the respective control command. The raw EMG signal was found to have static offset and it was slightly drifted with time. This was corrected via a high pass filter with zero-phase delay. Figure-5 shows the filtered EMG signal.

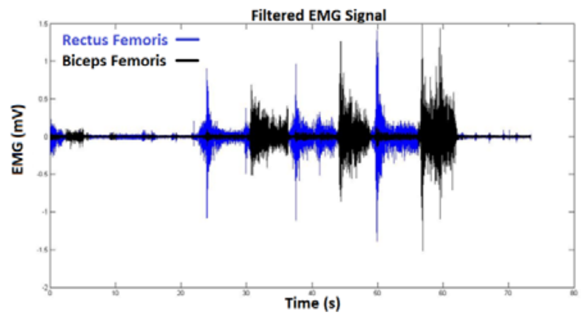

Figure-5. Filtered EMG signal.

Meanwhile the normalized RMS EMG signal was obtained by rectifying the filtered signal; compute its root mean square (RMS) and dividing the RMS value by the maximum voluntary contraction (MVC). These signals were then converted to binary muscle activation commands and sent to the microcontroller to control the motor and link direction.

Figure-6 shows the corresponding experimental test results. In Figure-6(a), red lines are recognized as the forward motion whereas blue lines represent backward motion. These signals were then converted to binary activation signal as indicated in Figure-6(b). When the binary signal is LOW (0), motion command will be sent to the motor and subsequently the link will move backward. Conversely the link will move forward if the signal is HIGH (1). Delay was observed between the first plot (Figure-6(a)) and the second plot (Figure-6(b)). This may be due to the processing of the filtered EMG signals into binary signals for muscle activation.

As a result, the corresponding hip and knee joint angles are plotted in Figure-6(c) and (d) respectively. The rising angle in Figure-6(c) signifies that the hip link is moving forward and vice versa. The angle was peaked at 20^o and -20^o. These values represent the maximum and minimum constraints set upon the hip link in the software for safety purpose. A mechanical limit switch was also installed to ensure the link moves within the constraint angles. The same principle is applied to knee joint as in Figure-6(d).

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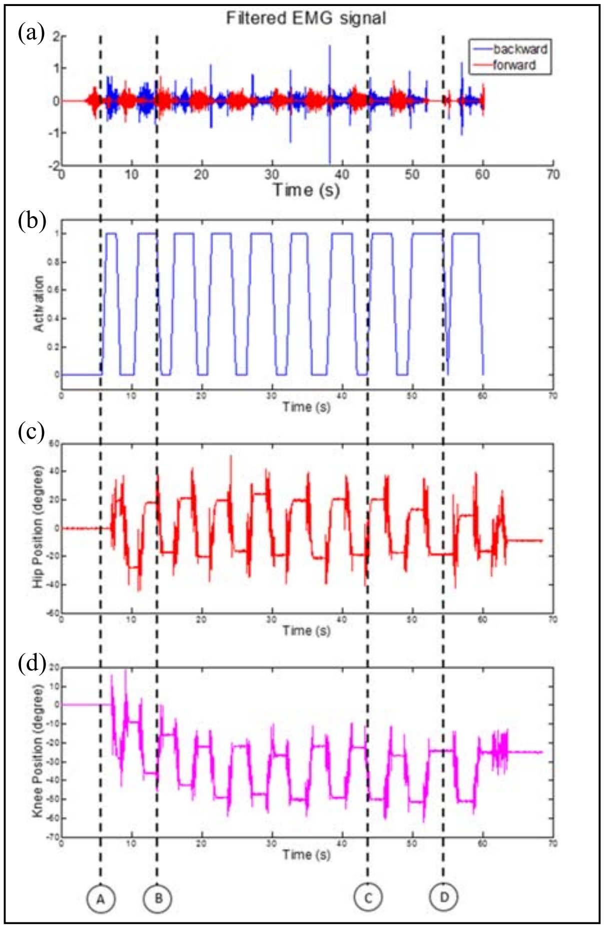

Figure-6. (a) The filtered EMG signal, (b) corresponding binary activation commands, (c) angular position of hip joint and (d) angular position of knee joint.

Four motion cycles were selected to further illustrate the relationship and the connection between the EMG signals, binary activation commands and the corresponding hip and knee joint angles. These cycles are plotted in dotted lines and denoted as region A, B, C and D in Figure-6. Region A shows that when the binary activation signal is rising and it represents forward mode. After a delay of 1 second, the hip link started to move forward whereas the knee link moved backward concurrently. The delay was observed to reduce at Region B and C. Region D shows a special occasion where there is a delay of about 500 milliseconds in showing the binary muscle activation signal. However this did not affect the actual performance. Results indicated that the hip and knee links were already entered the backward mode before the activation signal turned to LOW (0). Therefore the delay could be attributed to huge processing load in Matlab and the communication between Matlab and the microcontroller board Arduino Mega 2560. However this delay affects the graph plotting only and has no effect on the actual performance.

In the reference trajectory control mode, simulation study was performed in Matlab Simulink. Figure-7 shows the simulation results of hip and knee joint angles. The red lines indicate the predefined reference trajectory as control signal while the blue lines show the response of the hip and knee joint respectively.

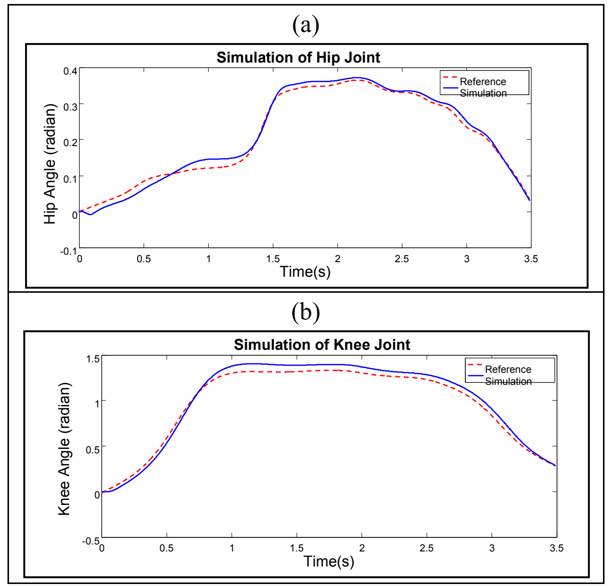

Figure-7. Simulation results of (a) hip joint and (b) knee joint.

Figure-7 shows that the control system is able to track the trajectory with satisfactory results. The deviation of the actual response and the reference trajectory is small. The small deviation could be attributed to the transient response of the PID controller as well as the tuning of the control parameters.

Meanwhile a preliminary experimental test was also conducted in the reference trajectory control mode. In the first test, the reference trajectory was sub-divided into 28 reference set points over a period of 3.5 s. This implies that the link has to reach the set point within 125 ms. however the microcontroller board could handle the control signal at every 20 ms at best and hence the actual response lagged the reference trajectory significantly. Nevertheless, the actual response was able to follow the gait profile correctly but in a slower pace. The performance was improved significantly in the second test where only 14 reference set points were used and the duration of the control cycle changed to 250 ms. Overall, the error could be attributed to the limitation in hardware processing speed as well as inaccurate sensor data acquisition.

CONCLUSIONS

In conclusion, the prototype of a dual mode EMG controlled robotic orthosis has been designed, fabricated

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and tested. The dual mode control system, which is able to operate with EMG signals or reference trajectory, has been implemented successfully. The performance of the control system could be further improved with an upgrade in the microcontroller hardware. Higher processing power is needed in order to read high frequency encoder data instead of using potentiometer as the joint motion sensor.

REFERENCES

[1] Hollands K.L., Pelton T.A., Tyson S.F., Hollands M.A. and van Vliet P.M. 2011. Interventions for coordination of walking following stroke: Systematic review. Gait & Posture. 35(3): 349-359.

[2] Aziz N.A., Leonardi-Bee J., Phillips M.F., Gladman J., Legg L.A. and Walker M. 2008. Therapy-based rehabilitation services for patients living at home more than one year after stroke. Cochrane Database of Systematic Reviews. 16(2): CD005952.

[3] National Stroke Association of Malaysia (NASAM). Physiotherapy. http://www.nasam.org/english/rehab-physio.php.

[4] Fisher S, Lucas L and Thrasher TA. 2011. Robot-assisted gait training for patients with hemiparesis due to stroke. Topics in Stroke Rehabilitation. 18(3): 269-276.

[5] Jezernik S., Colombo G., Keller T., Frueh H. and Morari M. 2003. Robotic orthosis Lokomat: A rehabilitation and research tool. Neuromodulation: Technology at the Neural Interface. 6: 108–115.

[6] Vallery H., Duschau-Wicke A. and Riener R. 2009. Generalized elasticities improve patient-cooperative control of rehabilitation robots. In: 11th IEEE International Conference on Rehabilitation Robotics (ICORR), Kyoto, Japan.

[7] Hesse S. and Uhlenbrock D. 2000. A mechanized gait trainer for restoration of gait. Journal of Rehabilitation Research and Development. 37(6): 535-541.

[8] Schmidt H, Hesse S, Bernhardt R. and Krüger J. 2005. HapticWalker – a novel haptic foot device. ACM Transactions on Applied Perception. 2(2): 166-180.

[9] Yano H., Tamefusa S., Tanaka N., Saitou H. and Iwata H. 2010. Gait rehabilitation system for stair climbing and descending. In: 2010 IEEE Haptics Symposium, Waltham, USA. pp. 393-400.

[10] Hesse S., Waldner A. and Tomelleri C. 2010. Research Innovative gait robot for the repetitive practice of floor walking and stair climbing up and down in stroke patients. Journal of Neuro Engineering and Rehabilitation. 7(30): 1-10.

[11] Peshkin M., Brown D.A., Santos-Munné J.J., Makhlin A., Lewis E., Colgate E.J., Patton J. and Schwandt D. 2005. KineAssist: a robotic overground gait and balance training device. In: 9th International Conference on Rehabilitation Robotics. pp. 241-246.

[12] Bouri M., Stauffer Y., Schmitt C., Allemand Y., Gnemmi S., Clavel R., Metrailler P. and Brodard R. 2006. The WalkTrainer: a robotic system for walking rehabilitation. In: IEEE International Conference on Robotics and Biomimetics, Kunming, China. pp. 1616-1621.

[13] Okamura J., Tanaka H. and Sankai Y. 1999. EMG-based prototype powered assistive system for walking aid. In: Proceedings of Asian Symposium on Industrial Automation and Robotics, Bangkok, Thailand. pp. 229-234.

[14] Lee S. and Sankai Y. 2002. Power assist control for walking aid with HAL-3 based on EMG and impedance adjustment around knee joint. In: Proceedings of IEEE/RSJ International Conference on Intelligent Robots and Systems, Lausanne, Switzerland. pp. 1499-1504.

[15] Suzuki K., Mito G., Kawamoto H., Hasegawa Y. and Sankai Y. 2007. Intention-based walking support for paraplegia patients with Robot Suit HAL. Advanced Robotics. 21(12): 1441-1469.