Article can be accessed online at http://www.publishingindia.com

A Control Framework of Lower Extremity Rehabilitation Exoskeleton based on Neuro-Muscular-Skeletal Model

Lei Shi*, Zhen Liua and Chao Zhang**

Abstract

A control system framework of lower extremity rehabilitation exoskeleton robot is presented. It is based on the Neuro-Musculo-Skeletal biological model. Its core composition moudle, the motion intent parser part, mainly comprises of three distinct parts.The first part is signal acquisition of surface electromyography (sEMG) that is the summation of motor unit action potential (MUAP) starting from central nervous system (CNS). sEMG can be used to decode action intent of operator to make the patient actively participate in specific training. As another composition part, a muscle dynamics model that is comprised of activation and contraction dynamic model is developed. It is mainly used to calculate muscle force. The last part is the skeletal dynamic model that is simplified as a linked segment mechanics. Combined with muscle dynamic model, the joint torque exerted by internal muscles can be exported, which can be ued to do a exoskeleton controller design. The developed control framework can make exoskeleton offer assistance to operators during rehabilitation by guiding motions on correct training rehabilitation trajectories, or give force support to be able to perform certain motions. Though the presentation is orientated towards the lower extremity exoskeleton, itis generic and can be applied to almost any part of the human body.

Keywords: Rehabilitation Exoskeleton, Surface Electromyography (sEMG), Neuro-Musculo-Skeletal model, Muscle Dynamics Model, Skeletal Model

1. Introduction

With the rapid arrival of the aging society and the increase of physical movement disorder patients caused by various disease, the demand for occupational therapists has increased drastically. The traditionalrehabilitation mainly relies on therapist's one-by-one rehabilitation therapy, obviously whichis gradually not in conformity with the needs. In parallel to this situation, researchers have been using robotic technologies to develop many kinds of assistive and rehabilitative devices for people with disabilities or to develop medical devices used by caregivers. Our research team's core work is just to use robot technology to develop a intelligent rehabilitative tranning system that can be used to do lower limb gait rehabilitation for patients following a disease or a neurological condition. For this purpose, we develop a exoskeleton device that can be worn around human lower limb to offer assistance to patients during rehabilitation of the locomotor system. Its recovery principle is that robot drives patients to simulate normal subjects walking to complete the rehabilitation training mission under the control of control system

To date, this kind of rehabilitation has been developed for many rehabilitation purposes by other researchers and many of the clinical trials also verify they are valid. Among them, the lokomat exoskeleton is an example of the early gait trainer (Jezernik, Colombo, Keller, Frueh, & Morari, 2003, Riener, 2012). Evidence based data shows that lokomat therapy can improve gait symmetry, walking ability, increases muscle strength and so on compared to conventional physical therapy in stroke patients (Husemann, Müller, Krewer, Heller, & Koenig, 2007;

* Graduate School of Engineering, Nagasaki Institute of Applied Science, Aba-machi, Nagasaki, Japan. E-mail: xh_lambert@126.com

** Department of Advanced Technology and Science for Sustainable Development, Nagasaki University, Bunnkyomachi, Japan.

Westlake & Patten, 2009). Moreover, there are many other exoskeletons besides LOKOMAT. They were generally divided into two categories: treadmill gait trainer and over-ground gait trainer. Other treadmill exoskeletons besides LOKOMAT have LOPES (van Asseldonk & van der Kooij, 2012; Veneman et al., 2007), ALEX (Banala, Kim, Agrawal, & Scholz, 2009), ANdROS (Unluhisarcikli, Pietrusinski, Weinberg, Bonato, & Mavroidis, 2011) and so on. The developed over-ground gait trainers have Hybrid Assistive Limb (HAL) (Sankai, 2011) from University of Tsukuba, EXPOS from Sogang University (Kong & Jeon, 2006) and Vanderbilt exoskeleton (Farris, Quintero, & Goldfarb, 2011). For these exoskeleton devices, regardless of their different mechanical types, some common considerations must be paid on the design of their control system.

As a kind of wearable robot (Pons, 2008), the distinctive, specific and singular aspect of exoskeleton is its kinematic chain maps on to the human limb anatomy. Thus its controller design must be imposed strict requirements as regards safety, effectiveness and dependability. It must be designed as person-oriented device and is under the control of operators at all times. For control of the exoskeleton rehabilitation robot, of course, a large number of control system have been proposed in earlier studies using various approaches, such as machine learning, decoders, pattern recognition, and proportional control (Park & Khatib, 2006; Schultz & Kuiken, 2011). Except for proportional control, these control methods have two inherent drawbacks: (1) they only allow the subject to perform predetermined movements and (2) they limit the user's ability to control the magnitude of torque production. Alternatively, proportional myoelectric controllers use the subject's muscle activation to control the magnitude of joint torque for the powered device, which may be more beneficial in lower-limb control (Ferris & Lewis, 2009). But most of the previous work proposes complex mechanisms or systems of sensors. Meanwhile, many researchers also use the EMG directly to generate machine control commands for robot (Lee & Lee, 2005). However, most of the previous works decode only finite lower limb postures from surface electromyography (sEMG) signals, which can cause many problems regarding smoothness of motion, especially in the cases where the robot performs everyday life tasks. Therefore, effective controller entails the necessity for continuous and smooth control. Besides, studies have shown that active involvement for operators in the production of a motor pattern results in greater motor learning and retention than passive movement (Kaelin-Lang, Sawaki, & Cohen, 2004; Lotze, Braun, Birbaumer, Anders, & Cohen, 2003; Schouenborg, 2004). So in our system design, in order to make the patient actively participate in the task specific training, sEMG is adopted to decode intent of operator.

In the work, we construct a hybrid control scheme that combines the model-based control system and sensor-based control system. For the design of sensor-based control system, you can complete the design by referring to the design methods of control system of traditional robots. Therefore, the work mainly talk about the other sub-control system of model-based control system. In the system, Neuro-Musculo-Skeletal model is adopted. The sub-control system is an intuitive interactive interface between exoskeleton and operator. Compared to the traditional control by way of an external device, for example, a keypad or a wheel and so on that has to be manipulated, intuitive interface can reduce operator's mental load, that is,the operator can focus on fulfilling a task with the exoskeleton rather than focus on mere control of the device (Fleischer, 2007).

This paper is organized as follows: Section II provides a brief description of physiology of Neuro-Musculo-Skeletal and human motion control; Section III focus on the description about the control framework based on neuro-musculo-skeletal model developed in the work; Section IV gives a closing remarks and future work

2. Physiology of Neuro-musculo-skeletal and Human Motion Control

Movements of the body are brought about by the harmonious contraction and relaxation of selected muscles. Rehabilitation referred in this article is mainly to rebuild human limbs motor function such as walking gait, so as to maximize the patient's quality of life (QOL). (Shi & Liu, 2014).

2. 1 Generation, Transmission and Transformation of Nerve Impulses

In Neuro-Musculo-Skeletal model, the connection of the nervous and muscular system is the so called -motor neurons originating from the spinal cord or brain stem. Neurons create an electrical impulse (nerve impulses or an action potential (AP)) that transmitted across neuromuscular junctions to the motor endplates sitting on top of the muscle fibers. At this time, the muscle is activated to contract by the nerve signal and human movement is finally driven by skeletal muscle contraction force.

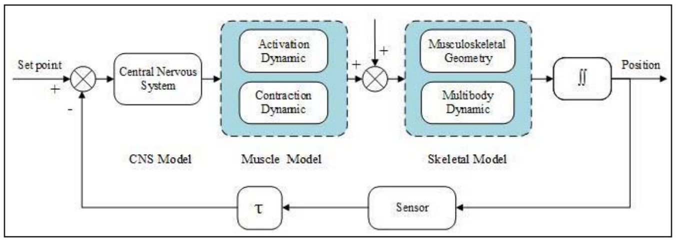
The-motor neuron, together with the axon, motor endplate, and the muscle fibers they connect to make up the fundamental block of motor control and are called a motor unit (Merletti & Parker, 2004). In general, motor units are fired in a random pattern and are not synchronized when a motor unit action potential (MUAP) is evoked. Studies on single motor units revealed that one stimulation pulse creates a single twitch response from the muscle. With increasing frequency of those pulses, the twitches start to merge and the force production of the muscle becomes continuous and increases. When the stimulation frequency is further increased, the twitches come closer to a permanent maximum contraction of the muscle at which point no further force can be generated. If this contraction is performed voluntary (no reflex, no spasm) it is called maximum voluntary contraction.

In my control scheme of the exoskeleton, the surface electromyography (sEMG) have been used as nerve control signal to identify voluntary movement from a patient. sEMG signal of a muscle is the summation of MUAPs evoked at the same time and can be directly measured invasively with surface electrodes located on top of the skin. Fig. 2 shows the main hardware composition that completes the process from the sEMG signal generation to the controlled object, whose details can refer to (Shi, Liu, & Wang, 2013)

The time between the emission and detection of the sEMG signal can be neglected in the context of this work. But there is also a time between emission and force production. This time, called the electromechanical delay, is reported to be about 50-80ms (Cavanagh & Komi, 1979; Zhou, Lawson, Morrison, & Fairweather, 1995), mainly due to low muscle fiber conduction velocity and the chemical processes which lead to contraction. It allows the signal evaluation process to start before the force production begins, reducing the latency of control systems coupled to EMG signals. Effects of muscle fatigue are not taken into account in this work. An analysis of these effects can be found in (Petersen, Hansen, Aagaard, & Madsen, 2007).

So the control scheme is the ability to detect the user's intention prior to the actual contraction of the muscle. However, one of problems by using sEMG is the system will not work properly if it is applied to the user with muscle disorder. This is also why a hybrid control scheme is used. Fig 1 is a conceptual scheme of the neuro-musculo-skeletal system from the perspective of control. The scheme shows a feedback control system.

Figure 1. A Conceptual Scheme of the Neuromusculoskeletal System.



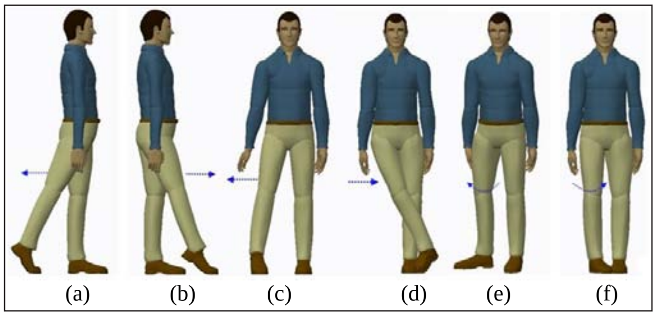
During the whole process from the creatation of nerve impulses that is based on feedback signal from sensor such as eyes to movement generation, the essential characteristic of human biological system is that it is converged in nature. About four levels of integration are included in the neuro-musculo-skeletal system (De Luca, LeFever, McCue, & Xenakis, 1982; Winter, 1980, 1984). But they are mainly divided into three parts from the perspective of biomechnical modeling and control in the work, that is, the central nervous system (CNS) model, muscle dynamics (activation and contraction dynamic) and skeletal dynamic model.

2. 2 Musculoskeletal Model

The construction of musculo-skeletal model requires an understanding of the anatomy of musculo skeletal systems. According to (Knudson, 2007), anatomy is the study of the structure of the human body and provides essential labels for musculo skeletal structures and joint motions relevant to human movement.

In anatomy, each bone is a complex living organ that provides attachment points for muscles to allow movements at the joints. For human joints, according to predominant tissues that supports the articular elements together, joints typically have three major types, that is, fibrous, cartilaginous, or synovial (Liu Kun, 2010) joints. Among of them, the synovial joint is the main joint type associated with lower limb movement. Mobility varies considerably and a number of subcategories are defined based on the specific architecture and topology of the surfaces involved (e. g. planar, saddle, ball and socket). For different categories of joints, corresponding movements are permitted, e. g. flexion and extension, medial and lateral rotation and so on. For example, hip joint that is the link between the pelvis/trunk and the lower limbs is a ball-and-socket joint and has several kinds of movement, as shown in Fig 2.

Figure. 2. Possible hip Jointmovements and their Corresponding Types Are: (a) Extension (b) Flexion (c) Abduction (d) Adduction (e) External Rotation (f) Internal Rotation. These Movement Occurrs in three Main Planes, i. e., Sagittalplane (Movements of a and b), Coronalplane (Movements of c and d) and Transverse Plane (Movements of e and f), which are Introduced when Researching Motion Related to human. For Details Such as Definitions of These Movement Types and Planes, you can Refer to the Relevant Books (Knudson, 2007).



3. Control Framework Based on Neuro-Musculo-Skeletal Model

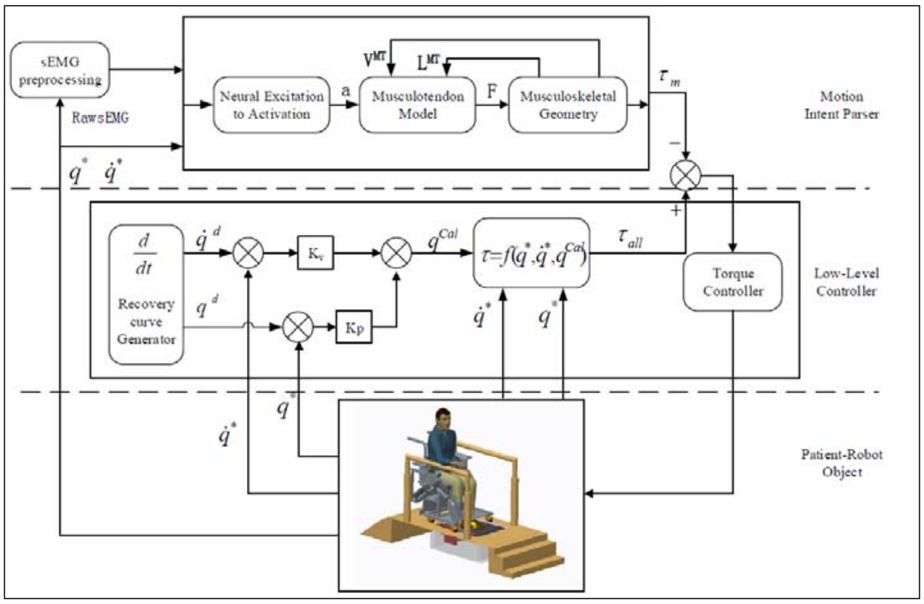
The work aims to develop an innovative neuromuscular control theoretic formulation to control a exoskeleton rehabilitation robot for the lower limb. The approach adopted here in developing the interface of human-robot sensorimotor control is based on an inverse model (dynamic and kinematics) coupled with nonlinear feedback. As the nervous system plans and regulates movement, it does so by taking into account the mechanical properties of the muscles, the mass and inertial properties of the body segments, and the external forces arising from contact with the environment. The overall system can be represented schematically as in Fig. 3., which is subdivided into three distinct parts: the motion intent parser, lower-level controller and controlled object. Among of them, the motion intent parser that computes the suitable support torque is the most important composition part. So the following mainly give a description for it.

The part of the motion intent parser mainly comprises of three distinct parts, which are signal acquisition of surface electromyography (sEMG), muscle dynamic model and skeletal dynamic model.

3. 1 Signal Acquisition of sEMG

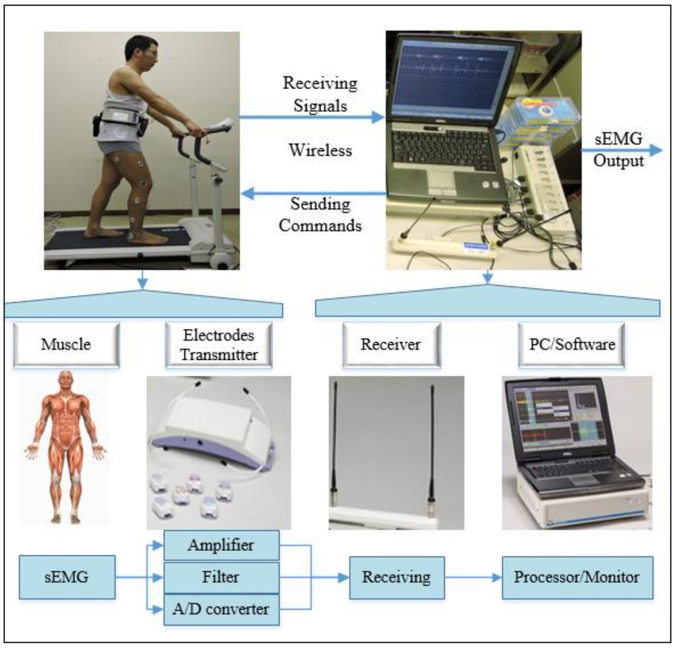
sEMG is the summation of motor unit action potential (MUAP) starting from central nervous system (CNS).

Figure 3. The Framework of Overall Control System



So that it can be used to decode action intent of operator to make the patient actively participate in specific training . Fig 4 is a aEMG signal acquisition system adopted in our experiment. It is a multi-channel wireless telemetry system that is a product of NIHON KOHDEN corporation in japan. Before starting to obtain signal, a lot of preparations must be done, which is very important if a good quality signal is to be obtained. In my experiment,it includes (1) shaving the excess hairto obtain even lower skin resistance; (2) Using alcohol to removal of dirt, oil, and dead skinin order to reduce any skin resistance and allow electrodes to be attached without coming loose; (3) a part of subjects whose skin surface are dry use electrodegel (Elefix Z-181BE made in NIHON KOHDEN corporation

Figure 4. The Key Elements of the System from SEMGto Robot



in japan) rubbed into the skin, which can dramatically improve the quality of the recorded signal; (4) In the decision of the specific site for the electrode, in order to assure repeatability for different subjects, various bony landmarks are used as a reference. After finding the position for one subject, marks are made so as to assure that the electrodes are over the same muscle fibers in different trials; (5) The inter-electrode distance is a important parameters and you should make sure that this distance is consistent throughout all subjects and trials. In my experiment, this step is skipped because the electrodes has fixed electrode geometries.

3. 2 Muscle Dynamic Model

Act as the input, sEMG signals get into the muscl edynamics and estimate the current joint moment contribution of the operator based on the resulting muscular forces. The module consists of activation dynamic and contraction dynamic model. Activation dynamic corresponds to the transformation of sEMG to activation of contractile and a modified non-linear first-order dynamic model (Thelen, 2003; Winters, 1995) is used in the system, as shown in Equ.

$$\frac{da}{at} = \frac{u-a}{\tau_a(a,u)} \tag{1}$$

Where, u is excitation, a is activation, is a variable time constant that varies with activation level and whether the muscle activation level is increasing or decreasing.

$$\tau_a(a,u) = \begin{cases} \tau_{act}(0.5+1.5a) \\ \dfrac{\tau_{deact}}{0.5+1.5a} \end{cases} \tag{2}$$

Where, is the activation time constant and is the deactivation time constant. Typical values for activation and deactivation time constants are 10 ms and 40 ms, respectively.

The raw sEMG signals are acquired and need to do some preprocessing before calculating a (t). The preprocessing mainly includes rectification, filter, smoothing (Shi et al., 2013). And during the process of modeling, all parameters are derived through scaling values found in the literature. And in order to do analysis for different persons and muscles conveniently,no dimensional method is implemented, that is, all parameters used in the calculation is normalization value. For instance, all forces and length quantities are normalized to maximum isometric muscle force (F_0^M) and optimal muscle fiber length (F_0^M), respectively. That is to say, $F^M_{norm} = F^M / F_0^M$ and $L^M_{norm} = L^M / L_0^M$. The details can refer to (Shi & Liu, 2014).

Once we have obtained the activation of the muscle, we can compute the resulting force exerted by the muscle using a muscle model. In this work, a type of Hill-type (Hill, 1938) muscle model that is based on the work of (The len, 2003) is used to model muscle contraction dynamics. This type of hill muscle tendon model consists of three components: an active contractile element (CE), a passive parallel element (PE) and a passive series element (SE) (Rosen, Fuchs, & Arcan, 1999).

3. 3 Skeletal Dynamic Model

The skeletal dynamic model is simplified as a Linked Segment Digital Human Model (LS-DHM) in the work. For for the sake of generality, the template model LS-DHM that comprises 19 segments and 46 DOFs is developed firstly, and thenit will be configured to create a special and simplified analyzing model based on a specific study, for example, the study of lower limb walking movement.

Combined with muscle dynamic model, the joint torque exerted by internal muscles can be exported by Equ. Using an inverse dynamic model of LS-DHM derived from equations of motion, a exoskeleton controller can be designed (Shi & Liu, 2014).

$$\boldsymbol{M}_k(t) = \sum_{i=1}^{Ne} \boldsymbol{F}_e^i(t) \cdot \boldsymbol{d}_e^i(t) - \sum_{j=1}^{Nf} \boldsymbol{F}_f^j(t) \cdot \boldsymbol{d}_f^j(t) \tag{3}$$

Where, is the joint identifier. For lower limb, could be hip, knee, ankle and other joint indetifier is the number of extensor muscles for joint, is the number of flexor muscles for joint; is the force produced by extensor muscle at time; is force produced by flexor muscle at time; andare moment arm of extensor and flexor muscle at time, respectively.

The equation of motion is formulated by using Lagrangian dynamics (Greenwood, 1997), as shown in Equ.

$$\boldsymbol{M}(\boldsymbol{q})\ddot{\boldsymbol{q}} + \boldsymbol{C}(\boldsymbol{q},\dot{\boldsymbol{q}})\dot{\boldsymbol{q}} + \boldsymbol{G}(\boldsymbol{q}) = \boldsymbol{Q}_{nc} \tag{4}$$

Where, are vectors of displacement, velocity and acceleration; is a pose-dependent inertia matrix comprising body anthropometry, and comes from taking the second derivative the kinetic energy. The next factor represents what are called the coriol is and centrifugal or coupling effects of the manipulator system on joint torques, which is a vector. Note it includes both angular position and velocity terms. The matrix includes forces based on the influence of gravity is non-conservative force (internal dissipative forces, any external forces).

4. Closing Remarks and Future Work

The work mainly develops a control system framework based on Neuro-Musculo-Skeletal model. The control framework can make exoskeleton offer assistance to patients during rehabilitation by guiding motions on correct training, rehabilitation trajectories, or give force support to be able to perform certain motions. sEMG is adopted as an indicator of subject's voluntary intention in the system, so it is an intuitive interactive interface between the exoskeleton and operator. Compared to the traditional control by way of an external device, for example, a keypad or a wheel and so on that has to be manipulated, intuitive interface can reduce operator's mental load, that is, the operator can focus on fulfilling a task with the exoskeleton rather than focus on mere control of the device.

Though the presentation of the framework of control system is orientated towards the lower extremity exoskeleton rehabilitation robot, the method is generic and can be applied to almost any part of the human body. In the future, we'll apply the interface in prototype machine so as to validate and make it better meet the real system requirements.

References

Banala, S. K., Kim, S. H., Agrawal, S. K., & Scholz, J. P. (2009). *Robot assisted gait training with active leg exoskeleton (ALEX)*. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 17(1), 2-8.

Cavanagh, P., & Komi, P. (1979). Electromechanical delay in human skeletal muscle under concentric and eccentric contractions. *European Journal of Applied Physiology and Occupational Physiology*, 42 (3), 159-163.

De Luca, C., LeFever, R., McCue, M., & Xenakis, A. (1982). Control scheme governing concurrently active human motor units during voluntary contractions. *The Journal of Physiology*, 329 (1), 129-142.

Farris, R. J., Quintero, H. A., & Goldfarb, M. (2011). *Preliminary evaluation of a powered lower limb orthosis to aid walking in paraplegic individuals*. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 19 (6), 652-659.

Ferris, D. P., & Lewis, C. L. (2009). *Robotic lower limb exoskeletons using proportional myoelectric control.* Conference Proceedings on IEEE Engineering in Medicine and Biology Society, 2119-2124.

Fleischer, C. (2007). *Controlling exoskeletons with EMG signals and a biomechanical body model.* Scuola Superiore Sant'Anna.

Greenwood, D. T. (1997). *Classical dynamics*. Courier Corporation.

Hill, A. (1938). *The heat of shortening and the dynamic constants of muscle*. Proceedings of the Royal Society of London B: Biological Sciences, 126 (843), 136-195.

Husemann, B., Müller, F., Krewer, C., Heller, S., & Koenig, E. (2007). Effects of locomotion training with assistance of a robot-driven gait orthosis in hemiparetic patients after stroke a randomized controlled pilot study. *Stroke*, 38 (2), 349-354.

Jezernik, S., Colombo, G., Keller, T., Frueh, H., & Morari, M. (2003). Robotic orthosis lokomat: A rehabilitation and research tool. *Neuromodulation: Technology at the Neural Interface*, 6 (2), 108-115.

Kaelin-Lang, A., Sawaki, L., & Cohen, L. G. (2004). Role of voluntary drive in encoding an elementary motor memory. *Journal of Neurophysiol*ogy, February, 93 (2), 1099-1103.

Knudson, D. (2007). *Fundamentals of biomechanics* (2nd ed.). Springer Science & Business Media.

Kong, K., & Jeon, D. (2006). *Design and control of an exoskeleton for the elderly and patients.* IEEE/ASME Transactions on *Mechatronics*, 11 (4), 428-432.

Lee, J. W., & Lee, G. K. (2005). Gait angle prediction for lower limb orthotics and prostheses using an EMG signal and neural networks. *International Journal of Control, Automation, and Systems*, 3 (2), 152-158.

Kun, L. (2010). *Research on Wearable Sensor system for 3D lower limb kinematics analysis.* (doctor [D]), Kochi University of Technology, Kochi, Japan.

Lotze, M., Braun, C., Birbaumer, N., Anders, S., & Cohen, L. G. (2003). Motor learning elicited by voluntary drive. *Brain*, 126 (4), 866-872.

Merletti, R., & Parker, P. A. (2004). *Electromyography: Physiology, engineering, and non-invasive applications*. Vol. 11, John Wiley & Sons.

Park, J., & Khatib, O. (2006). A haptic teleoperation approach based on contact force control. *The International Journal of Robotics Research*, 25 (5-6), 575-591.

Petersen, K., Hansen, C. B., Aagaard, P., & Madsen, K. (2007). Muscle mechanical characteristics in fatigue and recovery from a marathon race in highly trained runners. *European Journal of Applied Physiology*, 101 (3), 385-396.

Pons, J. L. (2008). *Wearable robots: Biomechatronic exoskeletons*. Vol. 70, Wiley Online Library.

Riener, R. (2012). Technology of the Robotic Gait Orthosis Lokomat *Neurorehabilitation Technology*, (pp. 221-232), Springer.

Rosen, J., Fuchs, M. B., & Arcan, M. (1999). Performances of Hill-type and neural network muscle models-toward a myosignal-based exoskeleton. *Computers and Biomedical Research*, 32 (5), 415-439.

Sankai, Y. (2011). HAL: Hybrid assistive limb based on cybernics *Robotics Research* (pp. 25-34), Springer.

Schouenborg, J. (2004). Learning in sensorimotor circuits. *Current Opinion in Neurobiology*, 14 (6), 693-697.

Schultz, A. E., & Kuiken, T. A. (2011). Neural interfaces for control of upper limb prostheses: The state of the art and future possibilities. PMR, 3(1), 55-67.

Shi, L., & Liu, Z. (2014). *Design of Soft Human-Robot Interface Based on Neuro-Muscular-Skeletal Model.* IEEE 12th International Conference on paper presented at the Dependable, Autonomic and Secure Computing (DASC).

Shi, L., Liu, Z., & Wang, Q. (2013). *A Novel Method of sEMG Signal Segmentation.* IEEE Ninth International Conference on Mobile Ad-hoc and Sensor Networks (MSN).

Thelen, D. G. (2003). Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. *Journal of Biomechanical Engineering*, 125 (1), 70.

Unluhisarcikli, O., Pietrusinski, M., Weinberg, B., Bonato, P., & Mavroidis, C. (2011). *Design and Control of a Robotic Lower Extremity Exoskeleton for Gait Rehabilitation.* Paper presented at the Intelligent Robots and Systems (IROS), 2011 IEEE/RSJ International Conference on.

van Asseldonk, E. H., & van der Kooij, H. (2012). Robot-Aided Gait Training with LOPES. *Neurorehabilitation Technology* (pp. 379-396), Springer.

Veneman, J. F., Kruidhof, R., Hekman, E. E., Ekkelenkamp, R., Van Asseldonk, E. H., & Van Der Kooij, H. (2007). *Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation*. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 15(3), 379-386.

Westlake, K. P., & Patten, C. (2009). Pilot study of Lokomat versus manual-assisted treadmill training for locomotor recovery post-stroke. *Journal of Neuro-Engineering and Rehabilitation*, 6 (18).

Winter, D. A. (1980). Overall principle of lower limb support during stance phase of gait. *Journal of Biomechanics*, 13 (11), 923-927.

Winter, D. A. (1984). Kinematic and kinetic patterns in human gait: Variability and compensating effects. *Human Movement Science*, 3 (1), 51-76.

Winters, J. M. (1995). An improved muscle-reflex actuator for use in large-scale neuromusculo skeletal models. *Annals of Biomedical Engineering*, 23 (4), 359-374.

Zhou, S., Lawson, D. L., Morrison, W. E., & Fairweather, I. (1995). Electromechanical delay in isometric muscle contractions evoked by voluntary, reflex and electrical stimulation. *European Journal of Applied Physiology and Occupational Physiology*, 70 (2), 138-145.