Verification of Mathematical Model for Upper Limb Spasticity with Clinical Data

J Yee¹, C Y Low¹*, P Ong¹, W S Soh¹, F A Hanapiah², N A Che Zakaria³, S von Enzberg³, L Asmar⁴ and R Dumitrescu⁴

¹ Faculty of Mechanical and Manufacturing Engineering, Universiti Tun Hussein Onn Malaysia, 86400 Parit Raja, Johor, Malaysia.
² Faculty of Medicine, Universiti Teknologi MARA, 47000 Sungai Buloh, Selangor, Malaysia.
³ Faculty of Mechanical Engineering, Universiti Teknologi MARA, 40450 Shah Alam, Selangor, Malaysia.
⁴ Fraunhofer Institute for Mechatronic Systems Design, 33102 Paderborn, Germany.

E-mail: cylow@uthm.edu.my

Abstract. Healthcare providers in the field of physical and rehabilitation medicine play a vital role to help patients suffering spasticity readapting themselves to their normal daily activities. Mathematical modeling of spasticity has the potential to avoid the issue of variability in the assessment of spasticity using the Modified Ashworth scale (MAS). In this work, an existing mathematical model for upper limb spasticity is verified using clinical data sets of upper limb spasticity collected in Malaysia at the level of MAS 1+. The data set consists of torque values measured at each elbow angle as the elbow extends from a full flexion position to a full extension position during slow and fast stretch of the forearm. The aim is to find out the capability of the mathematical model and lay a foundation for the future work on data-driven modeling of upper limb spasticity based on the Modified Ashworth Scale.

1. Introduction

Spasticity is a velocity-dependent reflex contraction of muscle which will result in the inability in muscle control of human body[1]. Upper limb spasticity (ULS) is a more specific term for the manifestation of spasticity on one’s upper limb or arm. ULS will affect the normal movement of arm of the sufferers, as the muscles of one’s arm will be stiff and difficult to control. ULS is usually seen in patients or victims who experienced stroke, cerebral palsy (CP) or traumatic brain injury due to the damage on the central nervous system[2]. A patient could slowly restore the normal functionality of spastic arm by undergoing proper rehabilitation. However, rehabilitation physicians and therapists have to be well-trained before they are able to provide top-notch healthcare services to the ULS sufferers in order to avoid inflicting injury to the patients[3].

In the current practice, the trainees and students in the field of physical and rehabilitation medicine may at first learn and feel the signs of ULS through role-play sessions with the other peers before proceeding to clinical training to reduce the possible injury as a result of low proficiency. This is due

¹ To whom any correspondence should be addressed.
to the higher awareness in societal responsibility for a lower tolerance of margin for medical errors[4]. Thus, new trainees in the field are not expected to have direct physical encounter with a patient before undergoing clinical training. However, without direct feeling and sensation of a spastic arm, the trainees would not be able to fully grasp the signs and essence of ULS.

Simulation for the medical education appears to be one of the workable solutions to be better equipped in understanding ULS besides directly engaging with patients. The types of simulation that can be employed in medical education range from low-technology simulation to a realistic patient simulator[5], with different complexity, fidelity, and cost. Simulation for ULS is still a less explored area in the research study. In the field of rehabilitation, there are recently a few haptic devices developed for training purpose. The haptic devices used for ULS rehabilitation training are commonly known as upper limb disorder simulator or ULS part-task trainer and can imitate the signs of spasticity in upper limb. However, these devices have their own limitation[6].

Several ULS part-task trainers currently existing in the academic circle include a haptic device for elbow joint spasticity simulation of a child’s arm[7], the Haptic Elbow Spasticity Simulator (HESS)[1,8–10], and ULS part-task trainer named BITA[2,11–16]. In order to produce a ULS simulator with a higher fidelity, the HESS research team has developed a set of mathematical equations to describe the relationship between the movement of elbow angle and the resistance produced by the spastic arm[9]. This paper is about verification of the mathematical model based on the clinical data collected from Malaysian patients with ULS. A further analysis on the real and simulated clinical data is conducted to compare and verify the functionality of the developed mathematical equation.

2. Current program
If you don’t wish to use the Word template provided, please set the margins of your Word document as follows The This paper will be mainly built upon the foundation of the previously developed ULS part-task trainer named BITA. Currently, BITA is using a primitive program in generating the corresponding torque for the different angle of elbow according to the speed of the stretches, namely slow stretch and fast stretch. When the robotic arm is passively extended slowly which is still lower than a preset angular velocity threshold, the robotic arm would behave according to the torque value assigned to the respective angle in the “slow stretch torque value set” . However, if the robotic arm is passively extended beyond a range of degree within a preset time value, the robotic arm enters the fast stretch mode, The function will behave as shown in Figure 1.
3. Methodology

3.1 Clinical data collection

Sessions of clinical data collection have been conducted in two Malaysian hospitals/specialist centre. The clinical data collection is approved by the ethics committee of the Ministry of Health Malaysia and Universiti Teknologi MARA (UiTM). The clinical data collection system was designed and assembled by the research team of UiTM with the clinical data collection was supervised and conducted by a Consultant Rehabilitation Physician from the Faculty of Medicine of UiTM.

Prior to the clinical data collection phase, the measurement system and equipment were developed. The clinical data collection system consists of two sensors and the whole data acquisition system. The main equipment involved in this measurement system includes a goniometer, USB data-acquisition (DAQ) interface, muscle strength meter (MSM), and electrode amplifier. The specification of the equipment used is summarised in Table 1.

| Equipment                  | Ref. | Brand                   | Specifications                                      |
|----------------------------|------|-------------------------|-----------------------------------------------------|
| Goniometer (Vernier GNM-BTA) | [17] | Vernier                 | Range 0-340° (±170°)                                 |
|                            |      |                         | Accuracy ±3°, ±1° with calibration                   |
|                            |      |                         | Resolution 0.06°                                    |
|                            |      | 13-bit(SensorDAQ)       | Stored calibration values: Slope: 90°/V             |
|                            |      |                         |                                                    |
|                            |      |                         | Intercept: -45°                                     |
| USB DAQ interface (Vernier SensorDAQ) | [18] | National Instrument & Vernier | Max sampling rate 48,000 samples/second           |
|                            |      |                         | I/O Connectors 3 Analog, 1 Digital                  |
|                            |      |                         | Resolution 32 bits (16 bits for output)             |
|                            |      |                         | Operating system Windows 2000/XP                    |
| Electrode amplifier (Vernier EA-) | [19] | Vernier                 | Power 7 [mA] @ 5V DC                                |
|                            |      |                         | Input range -450 to +1100 [mV]                      |
|                            |      |                         | Impedance 100 [MΩ]                                 |
All the equipment listed has different functions in the data collection system. The main data to be collected and analysed for upper limb spasticity are the angle of elbow and the corresponding torque. The function of goniometer is to track and record all the instances of different angle of the elbow to be paralleled with the other parameters. Secondly, the muscle strength meter is used to sense, record, and transmit the data of the force exerted by the physician upon the patient. This is only valid at the fact that the force exerted by the physician is equal to the resistance force of the patients’ spastic upper limb. The existence of the DAQ interface in the system is for the connection between Vernier goniometer and a computer while the electrode amplifier is mainly acting as a converter from Bayonet Neill–Concelman (BNC) to British Telecom Connector (BTA) which is able to amplify the voltage into a monitorable range for the interface.

Apart from the main equipment and system, a video recording system is used to record the process of collecting the data for further reviewing purpose in any case when the video is required as a proof. The clinical measurement process is divided into three different phases: the pre-assessment phase, assessment phase, and post-assessment phase. The pre-assessment phase is the preparation and setting up of equipment including all the sensors, testers, and the recording equipment. The assessment phase is when the therapist engages with the patient for the slow stretch and fast stretch followed by the evaluation of severity level according to the Modified Ashworth Scale (MAS). The post-assessment phase is about analysing the data to extract the valuable information for training. The overall process flow is summarised in Figure 2.

The pre-assessment phase is the preparation of all the equipment for the assessment including the whole measurement system of goniometer and muscle strength meter, together with the whole video recording setup. The assessment process is as such: the therapist will conduct 3 slow stretches with the patient to determine the angle and torque through the whole elbow extension, followed by 3 fast stretches to detect the angle and torque at the catch, and then the therapist will determine the severity level of the particular patient based on MAS level. The post-assessment includes the visualisation of the torque and elbow angle side by side for analysis purpose. Figure 3 shows the instance of conducting the assessment on the patient by a trained clinician.
3.2 Simulation program

The mathematical model produced by the HESS team [9] is reconstructed in MATLAB accordingly for simulation purpose. The mathematical model is mainly divided into three main phases according to their definition, which is the pre-catch phase, catch phase, and post-catch phase. The three main equations for the torque are connected as a piecewise function. The three main equations are as followed:

\[\begin{align*}
\tau_{\text{pre}} &= m\ddot{\theta} + b\dot{\theta} + k\theta \\
\tau_{\text{catch}} &= H\dot{\theta}_{\text{catch init}}\delta(t) + \tau_{\text{pre end}} \\
\tau_{\text{post}} &= m\ddot{\theta} + b\dot{\theta} + k_{\text{post}}(\theta - \theta_{\text{post init}})
\end{align*}\]

(1)  
(2)  
(3)

The symbols above are as followed: \(\tau_{\text{pre}}\) is pre-catch torque, \(\tau_{\text{catch}}\) is catch torque, \(\tau_{\text{post}}\) is post-catch torque, \(\theta\) is the angle of elbow, \(m\) is mass of the forearm, \(b\) is damper constant, \(k\) is spring stiffness, \(k_{\text{post}}\) is spring stiffness in post-catch phase, \(H\) is the catch torque constant and \(\delta(t)\) is the coefficient constant.

Some of the variables in the equations above are not clarified in this paper as all the declaration has been made by the authors in [9]. The three equations listed are only for reference purpose in this paper. All the constants in the equations for the simulation are according to the figures given by the original paper as well.
A MATLAB script is used to simulate the control system of the upper limb spasticity simulator. The main input of the script (angle) is set based on the clinical data obtained from previous study. Microsoft Excel is used for storing the input data and the program will read this data by extracting the specific Excel files.

Numerical differentiation is used to find a numerical value of a derivative of given function at a given point. In the ideal situation, the derivative of a certain mathematical function \( f(x) \) can be estimated directly using the common differentiation method. However, there are always cases that a specific formula for \( f(x) \) is not known in the real engineering applications. There are only a set of data points acquired to represent the dependent relationship between \( x \) and \( f(x) \). Hence, the finite difference approximation can be used to figure out the estimate, \( f'(x) \) when estimating the rate of change of \( f(x) \) with respect to \( x \). There are basically three ways to calculate the approximating derivatives, i.e. forward, backward and central difference method. For this study, 3-points backward difference method is used. The first derivative and second derivative of this method is written as follow.

\[
\frac{f(x) - 2f(x-h) + f(x-2h)}{2h} \quad (4)
\]
\[
\frac{f(x) - 2f(x-h) + f(x-2h)}{h^2} \quad (5)
\]

where \( h \) = interval gap

Due to the fact that the collected clinical data is still insufficient for a thorough and complete simulation for all MAS level, only MAS level 1+ will be focused in this study. A total of 3 patients with ULS of MAS level 1+ have been engaged for data collection session, and the data are used for the simulation purpose in this study.

4. Result
The simulated results are compared with the real clinical data to assess and verify the applicability of the mathematical model in the context of Malaysian patients. 2 clinical data samples of MAS level 1+ are used for the comparison purpose. The main parameter that indicates spasticity is the catch angle. However, the definition of catch according to the professional clinician is not something that is obvious in the data set due to its qualitative nature and it is more sensational rather than by certain parameters. Hence, the peak torque and the peak torque angle (angle at which the peak torque occurs) are identified for verification purpose instead of the catch angle.

Three main datasets are collected from the clinical data measurement: torque, elbow angle, and their corresponding time. For every patient, 3 slow stretches and 3 fast stretches are applied on each clinical data collection session, thus there are 3 sets of data collected from each patient for the fast extension. The average values of peak torque and angle of peak torque are recorded in Table 2.

| Data | Average Peak Torque [Nm] | Average Angle of Peak Torque [°] |
|------|-------------------------|---------------------------------|
|      | Real | Sim | Difference | Real | Sim | Difference |
| 1    | 13.287 | 11.642 | -1.645 | 134.20 | 107.37 | -26.83 |
| 2    | 12.052 | 15.734 | +3.682 | 169.04 | 120.18 | -48.86 |

From Table 2, it is seen that the mathematical model is able to imitate the peak torque and angle of peak torque with certain tolerance. For Patient 1, the simulated peak torque is 1.645 Nm short of the real torque value and it occurs 26.83° earlier. As for Patient 2, the simulated peak torque is 3.682 Nm
higher, and it occurs 46.86° earlier than the real clinical data. The elbow angle corresponding to the torque value could be referred to in the subplot immediately below the torque figure.

![Graph of Torque and Elbow Angle against Time for Patient 1](image1)

![Graph of Torque and Elbow Angle against Time for Patient 2](image2)

**Figure 4.** Graph of Torque and Elbow Angle against Time for Patient 1  

**Figure 5.** Graph of Torque and Elbow Angle against Time for Patient 2

From the graphs plotted, it is observed that the mathematical model is able to reproduce the trend of the slow stretches and fast stretches with the bell-shape curve for the slow stretches and the sudden surge for the fast stretches. It is observed that the simulated slow stretches (the first 3 torque hikes) and simulated fast stretches (the subsequent torque hikes) are able to follow the increment and decrement trend albeit the difference in magnitude. The shape of slow and fast stretches is able to be perceived clearly from the graphs.

Overall, the simulation results based on the clinical data collected from Malaysian patients are capable of imitating the real clinical data with some tolerance. The simulated results also resemble the definition according to [15] which stated that for MAS level 1+, the increase in muscle tone and catch would happen at the second half of the range of motion (ROM), and will be followed by minimal resistance throughout the remaining ROM.

5. Conclusion
In order for patients with ULS to return to normal activities of daily living, it is very important to have more trained professional therapists and clinicians to cater to the demands in physical and rehabilitation medicine, which in turn requires better medical education as the foundational support. Thus, ULS part-task trainer would be a very vital tool in producing healthcare providers of higher quality by allowing a safe zone for numerous practices for the trainees. In producing a ULS part-task trainer with higher fidelity, the overall program needs to be improved so that the trainees could undergo a top-tiered training before being launched into clinical training. In the current program employed by BITA (ULS part-task trainer previously developed), the value of torque is directly mapped onto the elbow angle as a discrete function and value, which is primitive in nature.

In this study, the mathematical model produced by other researchers is being verified by using the real clinical data collected from the Malaysian patients. The clinical data used are of MAS level 1+ as the clinical data of other MAS levels are insufficient. The mathematical model is being programmed into MATLAB script for simulation purpose with the angle of elbow as the input and the torque value produced as the output. From the simulated results, it can be concluded that the trend for slow and fast
stretches is able to be reproduced successfully albeit the tolerance in the difference of value of peak torque and angle of elbow at which the peak torque occurs. In the future study, more clinical data will be collected so that simulation for other MAS levels could be carried out for a more thorough simulation and verification purpose.

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