IETI Transactions on Ergonomics and Safety

http://ieti.net/TES/

2017, Volume 1, Issue 1, 12-27

A NEW METHOD AND INSTRUMENTATION FOR ANALYZING SPASTICITY

Lana Popović-Maneski[1, a], Antonina Aleksić[2], Radoje Čobeljić[3], Tadej Bajd[4] and Dejan B. Popović[1,5]

[1]Institute of Technical Sciences of the Serbian Academy of Sciences and Arts, Knez Mihailova 35/IV, Belgrade, Serbia

[2]School of Electrical Engineering, University of Belgrade, Belgrade, Serbia

[3]Clinic for Rehabilitation "Dr Miroslav Zotović", Belgrade, Serbia

[4]Slovenian Academy of Sciences and Arts, Novi trg 3, Ljubljana, Slovenia

[5]Department of Health Science and Technology, Aalborg University, Denmark

[a]lanapm13@gmail.com

Abstract The assessment of spasticity is an important indicator of the course of the recovery in humans with central nervous system lesion. This recovery may be augmented by optimal ergonomic design of wheelchairs, beds, transferring devices and other devices used in activities of daily life. To quantify the influence of different aids and methods in reduction or increase of spasticity, an objective measure is required. The pendulum test was introduced to induce the stretch of the muscle, thereby, trigger spastic response, and the amplitudes of first sinusoidal movements are used as the measure of spasticity. We present a new model of the lower leg during the pendulum test which includes the nonlinear characteristics of the muscles, and a model of the shank and foot that are customized to the patient. The spastic torque component is calculated based on acquired values of angular acceleration, angular velocity, and angle. We developed a new analysis methodology and tool to describe the level and course of reflexive behavior. We describe the instrumentation for data acquisition. The kinematics is captured by two accelerometers, one gyroscope, and a Hall-effect joint angle encoder. Muscle activities of prime knee flexors and extensors are synchronously recorded. The simulation uses user's particular inertial parameters and data from repeated pendulum movements of the lower leg and outputs the reflexive torque. We developed the user-friendly software in MatLab for the analysis of data recorded with the new instrument. We present the application of the method in representative healthy subject and two spastic SCI patients. The main finding is that this method allows distinction between the flexion and extensions types of spasticity and the level of reflexive behavior.

Keywords: spasticity; spinal cord injury; inertial measurement units.

1. INTRODUCTION

As indicated in [1], the spasticity is one of the problems that rehabilitation professionals face in their practice. "Spasticity is a motor disorder characterized by a velocity-dependent increase in tonic stretch reflexes (“muscle tone”) with exaggerated tendon jerk, resulting from hyperexcitability of the stretch reflex, as one of the components of the upper motor neuron syndrome." [2]. Even though it presents great difficulties for the patient and care workers, spasticity should not be considered as a condition that must be eliminated at all cost; namely, spasticity at the level not interfering with the patient’s residual functional capacities is preferable compared with a flaccid paralysis [3]. Therefore, the aim of spasticity evaluation is to differentiate between the spastic components among the other various neurologic disorders and to determine the extent to which spasticity is causing discomfort. The result of an objective spasticity evaluation could be used in rehabilitation ergonomics for optimal design of various aids for activities of daily life (ADL), such as wheelchairs [3], beds, transferring devices etc. A standard clinical measure of the level of spasticity is the Ashworth modified scale. The grading procedure involves a human manipulation of the bodily segment; thereby, it is subjective. In parallel, the range of only six discrete values is limiting the subtle gradation of the changes of spasticity which result of the progression of the impairment or the recovery.

As defined, spasticity is manifested as an uncontrolled muscle response to the stretch due to the absence of efferent inputs to the neural circuitry bellow the lesion in patients with central nervous systems (CNS) lesion. A method to generate the stretch is brisk perturbation of muscle. For studying the uncontrolled muscle responses, a joint movement can be induced which triggers muscle spindles and Golgi's tendon organs [5]. A simple method to trigger stretch is to provoke free swinging of a large enough segment of the body in the gravity field (e.g., the lower leg swinging about the knee joint). The pendulum test is a two-sequence event: the examiner lifts the relaxed lower limb to the full knee extension (horizontal position of the lower leg), and lets than the lower leg to swing freely [6]. This swinging of the leg ultimately stretches two groups of muscles (quadriceps m. and hamstrings m.) that are extending and flexing the knee. Burke *et al*. [7, 8] studied the stretch responses of the quadriceps m. and hamstrings m. when subjected to a sinusoidal force acting at the lower leg and demonstrated that the sensitivity rises with the speed of stretch and initial elongation of the muscles.

1.1. The biomechanical model of the swinging lower leg

The pendulum test has been used to assess knee muscle spasticity in patients with neurological disorders [9, 10]. The pendulum test has been used in other central nervous systems lesions [3]. Bohanon *et al*. [11] used the Polhemus tracking system (http://polhemus.com/motion-tracking/) in patients with chronic stroke and demonstrated substantial differences between the nonparetic and paretic leg, and Fowler at al. [12] analyzed the variability of the spasticity in persons with spastic cerebral palsy.

The model of the pendulum swinging of the lower leg was introduced by Bajd and Bowman [13], and minimally modified in their subsequent research [14]. The sketch of the test and the model of the swinging lower leg are shown in Figure 1.

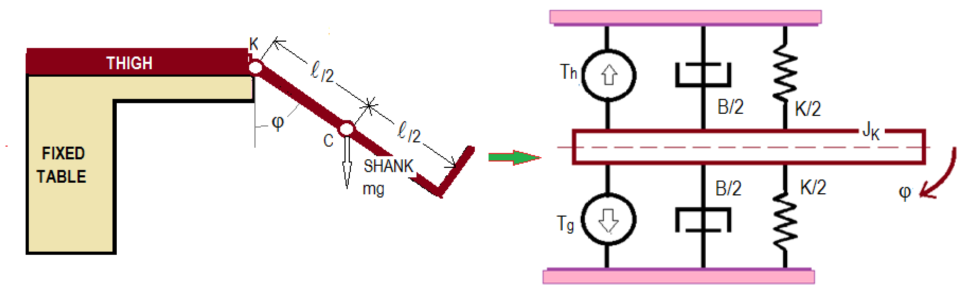

Figure 1. The biomechanical model of the lower leg in the gravity with the stretch reflex opposing the free swinging.

2017, Volume 1, Issue 1, 12-27

The model of the system presented in Figure 1 is:

$$T_h - T_g = J_K\ddot{\varphi} + B\dot{\varphi} + K\varphi. \tag{1}$$

where T_h is the muscle generated knee joint torque (reflex), J_K the moment of inertia of the lower leg-foot complex with respect the knee joint axes, and B and K are the damping and stiffness parameters (passive) at the knee joint, and φ the angle between the direction of the lower leg and gravity line. T_g is the gravity caused torque (Eq. 2)

$$T_g = \frac{mg\ell}{2}\sin\varphi. \tag{2}$$

where m is the mass of the lower leg and foot, $\ell/2$ the distance between the center of mass of the lower leg-foot complex and the knee (axes of rotation), and g the gravity acceleration. Bajd and Vodovnik [9] used the electro-goniometer aligned with the knee axes for the recording of the movements (Figure 1). They also measured surface EMG from the knee extensors to determine the onset and time course of the contractions. The patient was sitting on a tilt table in a supine position with both upper legs resting on the table, and the lower leg was free to swing. The examiner brings the shank to a horizontal position and then releases it to allow free swinging (Figure 2). The values $\dot{\varphi}$ and $\ddot{\varphi}$ are calculated numerically. The values R_1, R_2, and R_{2n} are the measures showing the differences between the healthy and a pathologic movement.

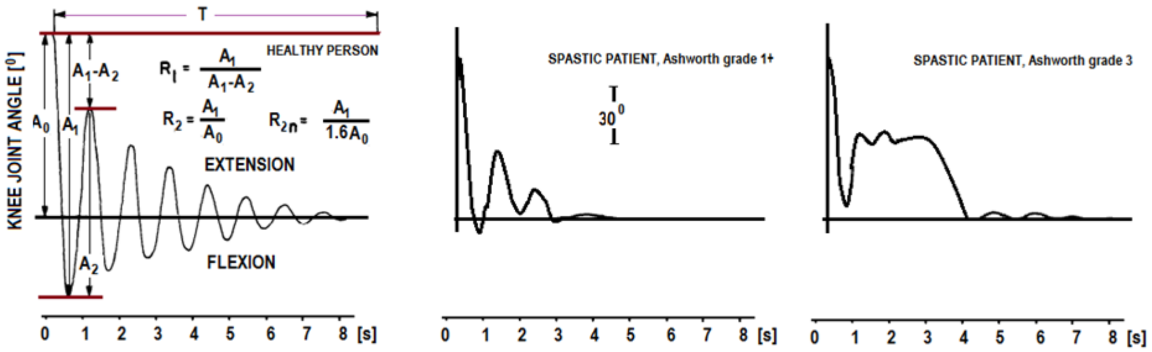

Figure 2. The results from the pendulum test in healthy and paraplegic patients with different spasticity. Modified from Bajd and Vodovnik [8].

The index used as the measure of spasticity is the ratio between the size of the first backward swing and the difference between the starting angle and resting angles [9] as shown in Figure 2. This index has been verified as a reliable indicator of spasticity [10, 15, 16]. The pendulum test should be repeated in the same individual to assess the initial response and the plateau that will be reached after few repetitions. The resting intervals of 15 seconds between the repeated tests are desirable for improved reliability [17]. Shorter intervals result with a decreased muscle tone [14].

2017, Volume 1, Issue 1, 12-27

The model of a pendulum used by Bajd and Bowman [14] was expanded by Le Cavorzin et al. [18]. The novelty in the work of Le Cavorzin et al. [19] was the addition of the reflexive component (Eq. 3) in the torque as shown in Figure 3.

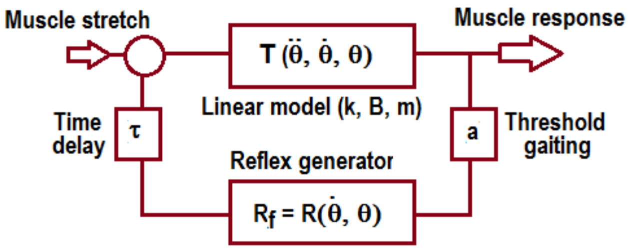

Figure 3. The model of the spastic response of the muscles which follows the spinal cord injury. Modified from Le Cavorzin et al. [20]

The reflex generator considers the biological behavior of the threshold and the delay that lump together the agonist and antagonistic muscles (Figure 3), pointing mostly on the contribution of the quadriceps activation.

$$T_h + R_f - T_g = J_K \ddot{\varphi} + B\dot{\varphi} + K\varphi.$$

$$R_f = C\exp(D|t - t_0|^n). \tag{3}$$

The additional member R_f follows the experimental work (Figure 4) and the heuristics.

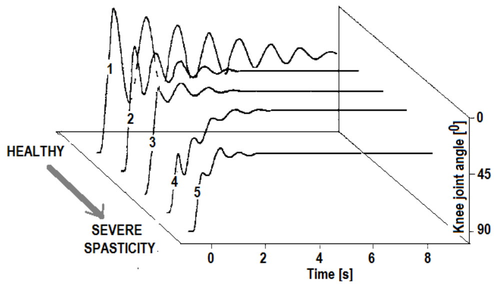

Figure 4. Set of the recordings in patients with spinal cord injury. Modified from Le Cavorzin *et al*. [19].

The measures considered by Le Cavorzin *et al*. were the estimated mechanical ratios K/J_K and B/J_K, as well as the neural parameters (R_f/J_K). The Student's test and the linear regression analysis proved that the model used is valuable for grading the level of spasticity.

Le Cavorzin et al. [20] introduced in their subsequent research a different form of the reflexive behavior:

$$R_f = f * \left(\delta\varphi - \frac{\alpha}{n}\dot{\varphi}\right). \tag{4}$$

2017, Volume 1, Issue 1, 12-27

where δφ is the angular variation during the stretch, n is the heuristically determined scaling factor, α is the so-called clasp-knife phenomenon intensity (positional sensitivity to the stretch) which can be positive or negative, and φ is the angle between the shank and the gravity direction. The difference between his original model in Eq. 3 and the one defined by Eq. 4 is mostly due to the difficulties in matching the recordings of the model and experimental data in the early phase of pendulum swinging and follows the concept of spastic hypertonia.

In the current study of evaluating spasticity [21] by using the pendulum test we found the spastic behaviors which cannot be described well enough with models given in Eq. 1, Eq. 3, and Eq. 4. Namely, the said models consider mainly the spasticity triggered in the knee extensors. We show original recordings (Figure 5) from patients in whom the dominant component of reflex response was coming from the knee flexors, and these reflexes were triggers when the knee was extending after the initial drop from the full extension.

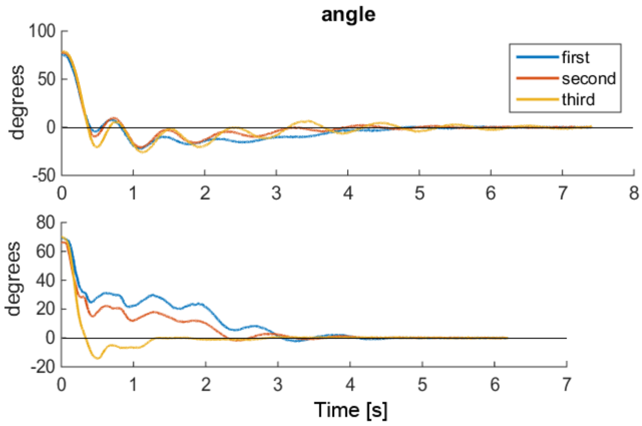

Figure 5. Three repetitions of the pendulum test recorded in series with the pause of 15 seconds between the trials for two patients participating in the clinical study. The top panel shows that the damping of the swing is taking place from the flexed knee (negative joint angles) after the initial swing from the fully extended position. The bottom panel shows that the damped movements during the returning to the neutral line are in the domain of positive angles.

To overcome the limitation of models described we assumed a more complex model of the pendulum movements where the rotation is affected by the reflexive response of extensors (T_E) and flexors (T_F) as shown in Eq. 5:

$$J_K \ddot{\varphi} = T_E(\varphi, \dot{\varphi}) - T_F(\varphi, \dot{\varphi}) - B(\dot{\varphi}) - K(\varphi) - m\,g\,d \sin\varphi\,. \tag{5}$$

The extension and flexion torques include nonlinear characteristics of muscles that have been used in many studies, based on the model of Shue et al. [22] as shown in Figure 6. J_K is the moment of inertia of the shank and foot complex treated as the single rigid body for the axes through the knee joint perpendicular to the plane of rotation, m is the mass of the foot and shank complex, d is the distance of the center of mass from the knee joint, B and K are the parameters determining the passive opposition (elasticity, friction, and air resistance). The passive component is modeled as the first order system, like the previous models of the pendulum.

2017, Volume 1, Issue 1, 12-27

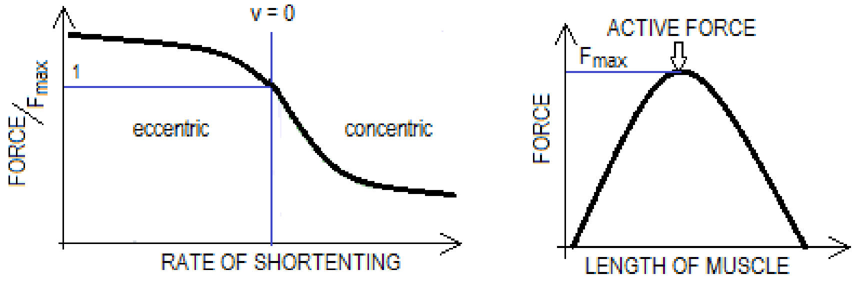

Figure 6. Sketch of the muscle force with respect the rate of shortening (left panel) and the muscle length (right panel).

The muscle force is usually modeled as shown in Eq. 6 [22].

$$F = b_1(L - L_0)^2 + b_2(L - L_0) + b_3$$

$$\frac{F}{F_{max}} = \begin{cases} k_1\,, \dot{L} < 0 \\ k_1(1 - k_2\dot{L})\,, 0 \le \dot{L} < k_2 \\ 0\,, k_2 \le \dot{L} \end{cases} \tag{6}$$

where L_0 is the length of muscle in a resting position, L is the length of the muscle when stretched, and parameters b_i and k_i (i = 1,2,3) are specific for each muscle. These parameters can be estimated based on methods presented in Stein et al. [23] and Chizeck et al. [24].

The model is highly nonlinear unlike the previously used models of the pendulum. The particular parameters b_i and k_i (i = 1, 2, 3) cannot be determined for the single muscle in a noninvasive procedure. The modeling that we suggest is the effort to overcome the difficulty with the fitting procedures in previous linear models and to avoid potential over-fitting with multiple members such as R_f in Equations 3 and 4.

The second novelty is the inclusion of the customized model of the lower leg for a particular patient.

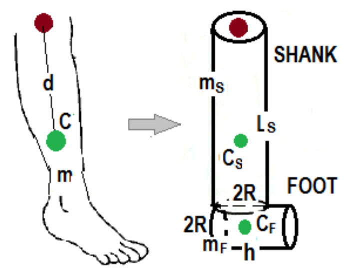

Figure 7. Model of the shank and foot for the simulation of the pendulum swinging.

2017, Volume 1, Issue 1, 12-27

The lengths L_s and h are measured and the radius R calculated from the perimeter of the shank above the ankle joint. The masses m_S and m_F are determined based on the volume of the shank and foot (Figure 7) and the density of tissues from the literature (≈1100 kg/m^3). The distance of the center of mass of the shank and foot complex from the knee (d), and the moment of inertia (J_K) were calculated as shown in Eq. 7:

$$d = \frac{m_s L_S/2 + m_F(L_S + R)}{m_S + m_F}$$
$$J_K = \frac{m_S L_S^2}{3} + \frac{(m_S + m_F)R^2}{4} + \frac{m_F h^2}{12} + m_F(h - R)^2 + m_F(L_S + R)^2 \quad (7)$$

We developed a program in Matlab environment that fits the equation of motion into the data recorded by assuming that there are no active or reflexive torques ($T_B = 0$).

2. METHODS AND MATERIAL

2.1. Subjects

Five healthy subjects participated in the evaluation of the system, and the recorded data were used for the estimation of parameters in Eq. 5 (K and B). Two chronic spinal cord injured patients (traumatic spinal cord injury) have been selected among about 20 patients participating in an evaluation of the effects of vestibular galvanic stimulation on spasticity since they are both spastic (Ashworth modified score >3), yet exhibit very different behavior during the pendulum test. Both subjects signed the informed consent approved by the local ethics board of the Clinic for Rehabilitation "Dr. Miroslav Zotović," Belgrade. Patient 1 has a T5/T6 complete paraplegia (ASIA A), while the patient 2 has a C4 lesion resulting with incomplete tetraplegia (ASIA C).

2.2. Data acquisition

The scissors-like system made of two thin aluminum bars was fixed at the thigh and shank cuffs by using the Velcro (Figure 6). The bars were connected by a low friction hinge joint. A Hall-effect joint angle encoder was mounted at the hinge joint to measure the rotation angle. One gyroscope and two analog accelerometers at the distance of 14.5 cm with their x-axes aligned with the long axis of the shank are bonded on one scissor hand (y-axes output of the accelerometer shows the tangential acceleration). Two pairs of GS26 pre-gelled disposable surface EMG electrodes (Bio-medical Instruments, MI, USA) were positioned over the quadriceps m. and hamstrings m. and connected to the inputs of two EMG amplifiers (Biovision, Germany). The same type of electrode (GS26) positioned over the bony part of the knee was used as the ground. The NI 6009 USB A/D card (National Instruments, TX, USA), 16-bit resolution connected via cable to the laptop digitized data from all four kinematic sensors and EMG amplifiers. Sampling frequency was 1 kHz. The EMG recordings were used to measure the reflexive reaction during the stretch. The reason for using two accelerometers positioned over a single segment was to eliminate the gravity component from the signals [25, 26].

2017, Volume 1, Issue 1, 12-27

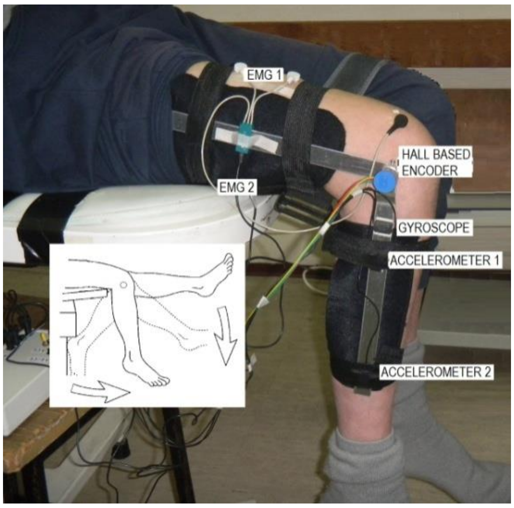

Figure 8. The new instrument with the accelerometers, gyroscope, Hall-effect encoder and two channels of EMG during the test in a paraplegic patient.

2.3. Procedure

The subject was comfortably sitting on a custom made thigh support with a square hole under the hamstrings m. to prevent from sitting on the electrodes and reduce the motion artifacts. Subjects were leaning on the backrest with their hip flexed at 45^0 with respect the gravity line. The examiner held the foot and slowly fully extended the knee joint, and then released it. The shank-foot complex started swinging and eventually stopped in the neutral position. The examiner repeated the said maneuver for at least three times within a single recording session with a pause of 15 seconds between the trials. The session was captured on the video for later inspection, and the data were acquired throughout the session. The data was processed, and the results saved for the analysis.

2.4. Data processing

The EMG signals were band-pass filtered between 30 and 300 Hz and notch filtered at 50 Hz with 3^{rd} order Butterworth filter. The signals from the Hall encoder, gyroscope and accelerometers were filtered with the moving average filter in 20 samples. Angular acceleration was calculated as the ratio of the difference between the tangential components of acceleration measured by the distal and the proximal accelerometers and the distance between the sensors. The size of the dataset for a single trial was composed of signals from the time when the angular velocity first became lower than zero to one second after the last swing. The last swing was defined as the oscillation with the amplitude smaller than 3% of the difference between the first maximum and minimum of angular velocity. The mean value of the knee angle during the last second of the swing was considered as neutral (zero) angle position and it was subtracted from the values of that dataset. This zeroing was included to eliminate the problem occurring because the resting position of the shank in relation to thigh is larger than 90^0 (see Figure 8). For calculation of parameters K and B in patients, we used high-pass filtered signal from the encoder, with the cutoff frequency of 1 Hz.

2017, Volume 1, Issue 1, 12-27

2.5. Matlab application

We developed the simulation program in Matlab environment that fits the equation of motion into the data recorded in healthy and spinal cord injured subjects for determining the parameters K and B (Eq. 5) with the particular body parameters of each subject by assuming that there is no active or reflexive torques, $T_h = 0$. The interactive window in the MatLab environment allowed the selection of up to three trials from each recording session. The examiner selected only the trials which do not include voluntary contractions, as suggested in Figure 9.

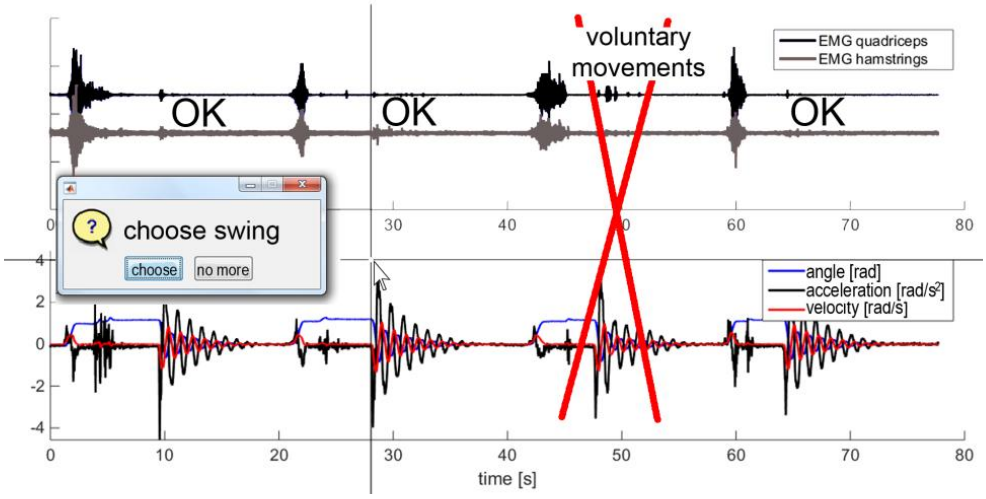

Figure 9. The first screen in Matlab program that shows angular acceleration, angular velocity, knee-joint angle and EMG activities during consecutive repetitions of pendulum test in a healthy subject. The third trial with excessive EMG activities during the swing was not selected for the later processing.

As spastic patients usually exhibit very different movement compared with the healthy-like pendulum motion (Figure 10), it was not possible to estimate parameters K and B from Eq. 5; therefore, we used only the high-pass filtered angle signal to calculate parameters K and B. The program outputs on the screen the torque T_h superimposed on the joint angle and EMG recordings from the quadriceps and hamstrings muscles.

2.6. Outcome values

The final outputs of the program are sets of numbers that characterize the exponential fit for the absolute values of T_h:

$$T_{h\_fit} = a e^{-bt} \tag{8}$$

and numbers that represent positive and negative areas between the T_h curve and the threshold lines parallel to the time axes. PP was calculated as the area between the positive values of T_h and the threshold line at 1 Nm, while the NP is the area between the negative values of T_h and the threshold line at -1 Nm. These two sets give a comprehensive representation of the type and level of spasticity.

IETI Transactions on Ergonomics and Safety
http://ieti.net/TES/

2017, Volume 1, Issue 1, 12-27

3. RESULTS AND DISCUSSION

Figure 10 shows the recordings from sensors for a healthy and two patients.

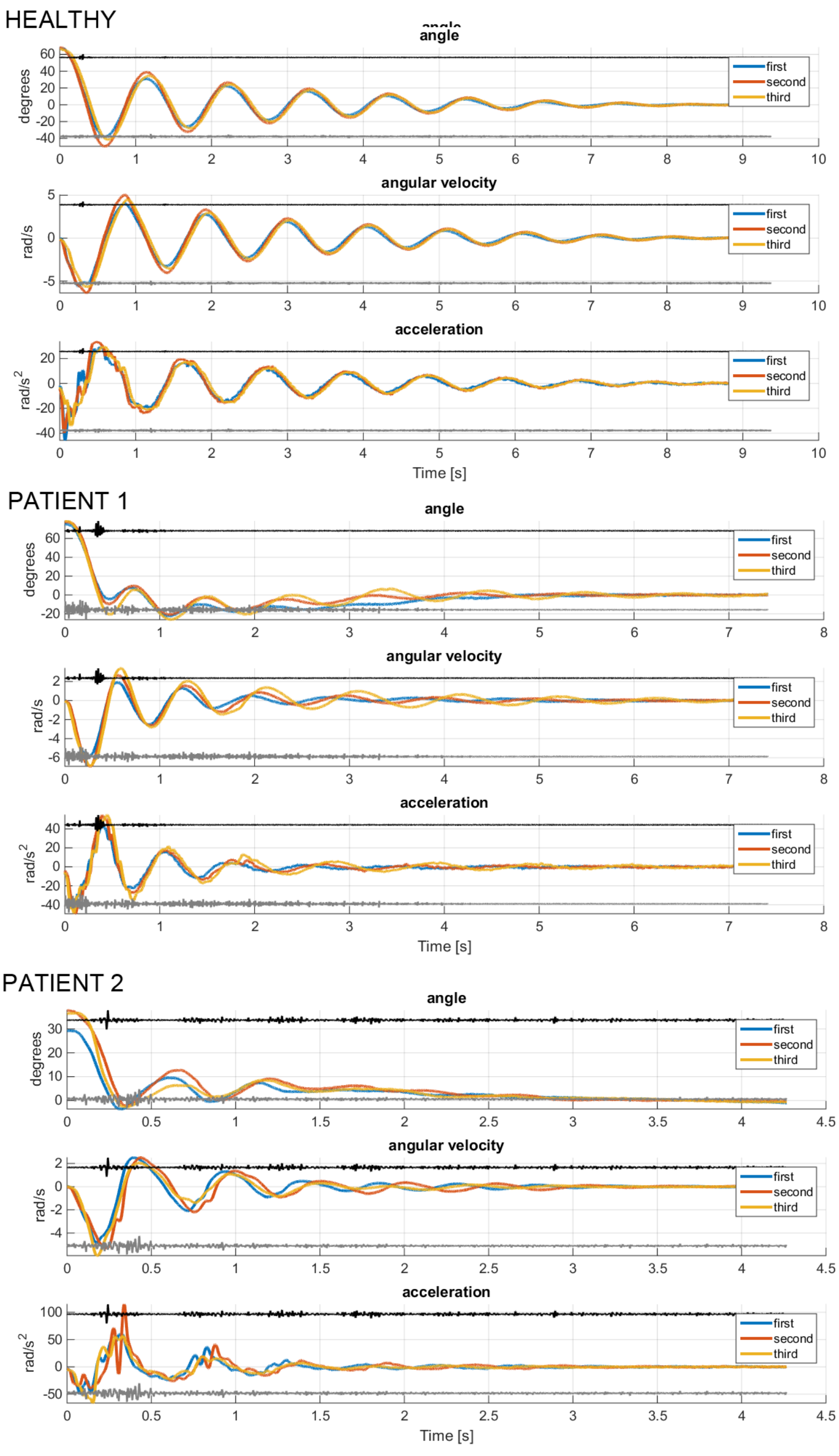

Figure 10. Acquired signals from Hall encoder, gyroscope, and two accelerometers during three consecutive pendulum tests.

2017, Volume 1, Issue 1, 12-27

Figure 11 shows the results of simulation for the same subjects. Parameters K and B are determined by Nelder-Mead simplex algorithm [27] with Matlab function *fminsearch*, when assumed that T_h is equal to zero.

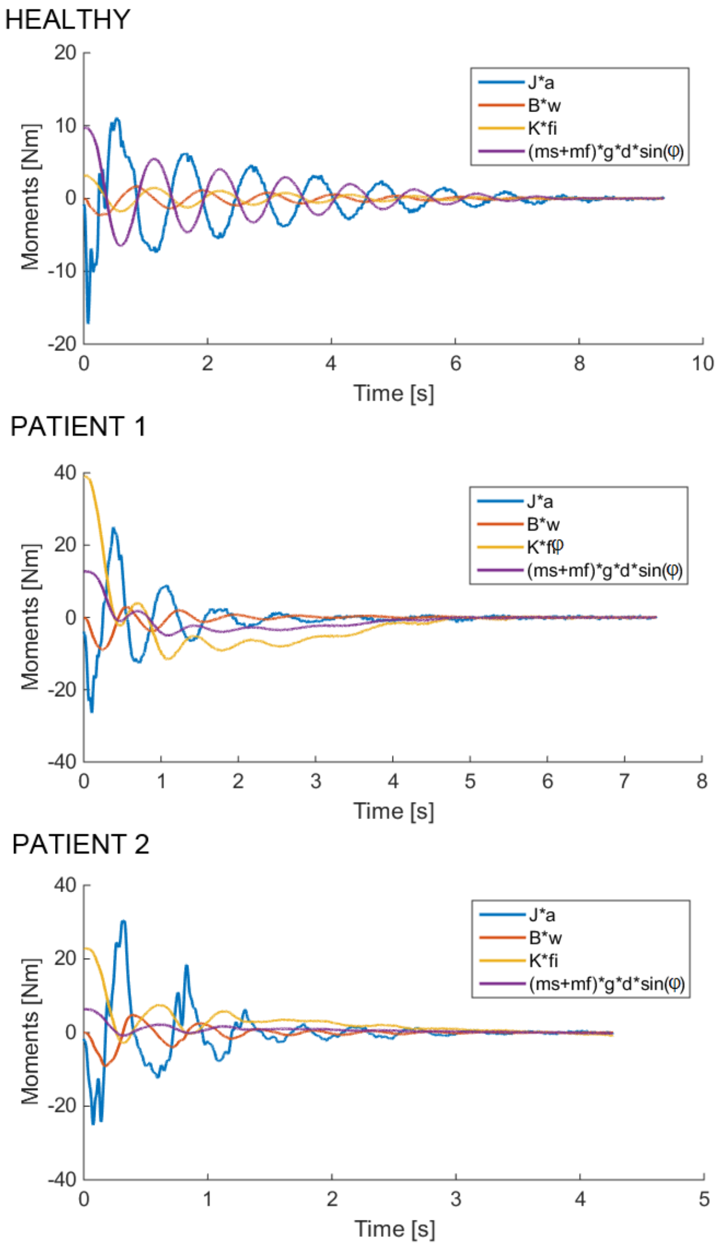

Figure 11. Components of Eq.5 resulting from leg inertia, elasticity, friction and air resistance.

Figure 12 shows the value of reflexive torque T_h (Eq. 1) calculated with the estimated values of parameters K and B.

2017, Volume 1, Issue 1, 12-27

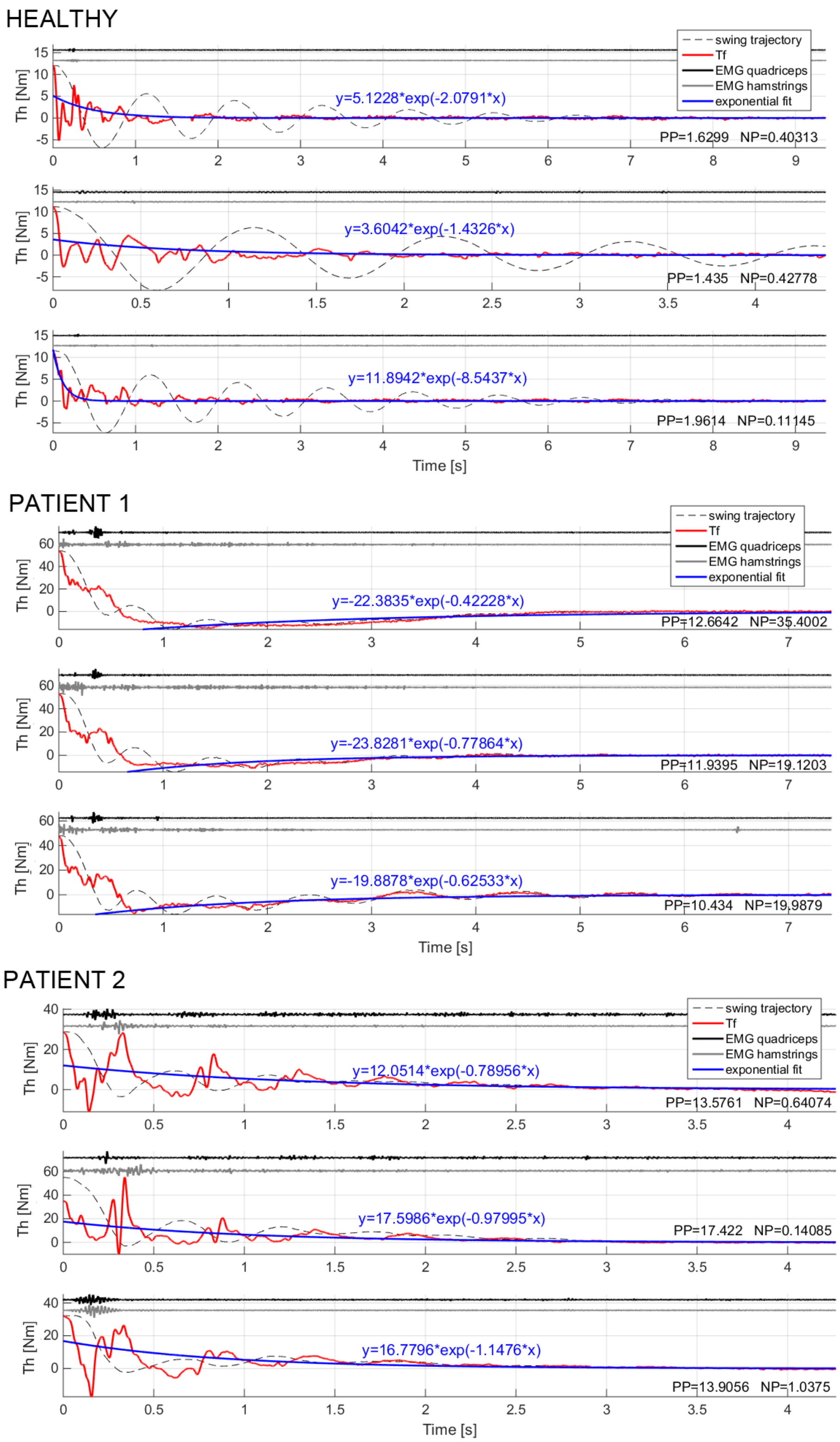

Figure 12. Reflexive torque T_h (red), swing trajectory (dashed lines, arbitrary values), EMG activities from the knee extensors (black) and flexors (gray), the exponential fit of absolute values of T_h (blue) and surfaces under positive (PP) and negative (NP) side of the T_h curve.

2017, Volume 1, Issue 1, 12-27

One would expect small values of T_h at any time in healthy; however, there is a strong intensity of T_h just after the beginning of the first swing. This can be explained, by considering that in the initial phase, when a very fast rate of the stretch occurs, nonlinear components from Eq. 6 cannot be neglected. This behavior results in the low values of the ratio between the constants *a* and *b* (Eq. 8). For the healthy subjects, the ratio is typically $|a/b|<2.5$.

Figure13 shows the result of simulation of the simplified form of the Eq. 5, where T_F and T_E are equal to zero and $\sin\varphi = \varphi$, with the calculated values of parameters K and B. The dashed line in plots for patients shows high-pass filtered values of the angle which were used in Nelder-Mead simplex algorithm for obtaining parameters K and B.

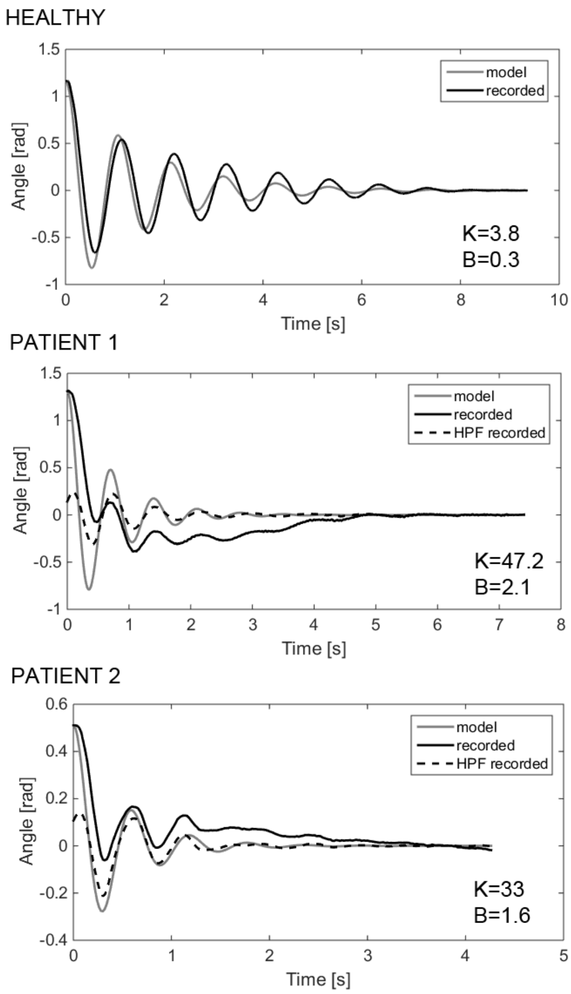

Figure 13. Simulation of the knee joint angle during the pendulum test with calculated values of parameters K and B for the healthy, Patient 1 and Patient 2.

There is a substantial difference of time courses of torques T_h between the two patients presented in this paper. The torque course in Patient 1 (middle panel, Figure 12) is generally in negative domain (dominant flexion spasticity), compared with Patient 2 (bottom panel, Figure 12) in whom positive trend dominates (extension type spasticity).

IETI Transactions on Ergonomics and Safety
http://ieti.net/TES/

2017, Volume 1, Issue 1, 12-27

A comparison between the reflexive EMG activities in two patients is shown in Figure 14.

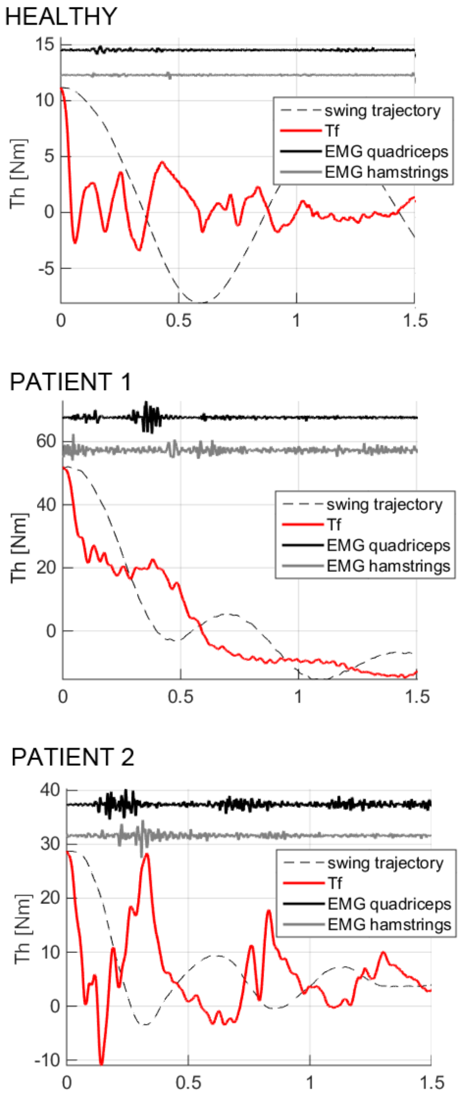

Figure 14. First 1.5 seconds of the simulation results shown in Figure 12 in parallel with the recorded EMG activities. The black lines on the top of panels are the recordings from the quadriceps m. and the gray lines below are the recordings from the hamstrings m. The EMG activities are largest at the first swing since the stretch is the fastest and largest. The quadriceps m. in Patient 2 shows cyclic activation during later swings.

The recordings from first the first 1.5 seconds in Patient 1 (the stretch was sufficient to trigger reflexive response) shows that flexor EMG was dominant, and in Patient 2 the extension activity was dominant, although declining as expected. The frequencies of swinging are similar between the two patients and much higher than in healthy.

We also show the exponential fit into data for the joint torques T_h (blue lines in Figure 12). The mean exponential fit (three trials) for two patients is characterized with the Eq. 9:

$$T_{h\_fit\_P1} = -21e^{-0.6t} \text{ and } T_{h\_fit\_P2} = 15.5e^{-0.97t} \tag{9}$$

2017, Volume 1, Issue 1, 12-27

suggesting the extension spasticity (a=15.5) in the Patient 2, compared to flexion spasticity (a=21) of the Patient 1. The typical values for healthy subjects are about 5, with large damping constant about 2.5, meaning that the value of T_h rapidly decreases (during the initial drop). The damping is more expressed in the Patient 2 (b= 0.97) compared with the damping in Patient 1 (b= 0.6). The $|a/b|$ ratios are 1.7, 35 and 15.98, for the healthy, Patient 1 and Patient 2, respectively.

The presented areas between the fitted torque and the threshold lines are: $PP_H = 1.96$, $NP_H = 0.11$ $PP_{P1} = 11.67$, $NP_{P1} = 24.8$ and $PP_{P2} = 14.96$, $NP_{P2} = 0.6$ for the patients 1 and 2 respectively. This directly gives that the total area in Patient 1 is negative $TP_{P1}+PP_{P1}= -13.13$ (spasticity in flexor muscles), while in Patient 2 it is $TP_{P2}= 14.36$ (spasticity in extensor muscles). In healthy subjects, the total area is typically below $TP_H < 2$.

4. CONCLUSION

The new method for acquiring data during the pendulum test (lower leg swinging in the gravity field) allows a detailed analysis of the spasticity. The message we send is that the user customized model, apparatus and data acquisition/processing that we developed allows the distinction of the type of spasticity, relaxation rate, and level of spasticity. This tool can help care workers, physicians and therapists in rehabilitation ergonomics and other fields to optimize designs of devices and modify therapy progress. The clinical validity of the presented method is in progress.

Acknowledgements

The work on this project was partly supported by the grants III 44008 and TR35003 from the Ministry of Education, Science and Technological Development of Serbia.

References

[1] Lin S. J., 2009, Energy cost considerations in common disabilities: scientific basis and clinical perspectives, In: Kumar S. (Ed.), *Ergonomics for rehabilitation professionals*, pp. 41-70, CRC Press, Boca Raton.
[2] Lance J. W., 1980, Pathophysiology of spasticity and clinical experience with baclofen. In: Lance J. W., Feldman R. G., Young R.R. and Koella W. P. (Eds.), S*pasticity: disordered motor control*, pp. 185-203, Year Book Medical Publishers, Chicago.
[3] Sindou M., Georgoulis G., and Mertens P., 2014, Evaluation of Spasticity in Adults. In: Sindou M, Georgoulis G, Mertens P. (Eds.), *Neurosurgery for Spasticity: A Practical Guide for Treating Children and Adults*, pp. 31-58, Springer, Vienna.
[4] Dicianno B. E., Arva J., Lieberman J. M., Schmeler M. R., Souza A., Phillips K., Lange M., Cooper R., Davis K. and Betz K. L., 2009, RESNA position on the application of tilt, recline, and elevating legrests for wheelchairs, *Assistive Technology*, 21 (1), pp. 13-22.
[5] Sinkjaer T., Toft E., Andreassen S. and Hornemann B. C., 1988, Muscle stiffness in human ankle dorsiflexors: Intrinsic and reflex components. *J Neurophysiol*, 60, pp. 1110-1121.
[6] Wartenberg R., 1951, Pendulousness of the legs as a diagnostic test, *Neurology*, 1,18-24.
[7] Burke D. A., Gillies J. D. and Lance J. W., 1970, The quadriceps stretch reflex in human spasticity, *Journal of Neurology, Neurosurgery & Psychiatry*, 33 (2), pp. 216-223.
[8] Burke D., Gillies J. D. and Lance J. W., 1971, Hamstrings stretch reflex in human spasticity, *Journal of Neurology, Neurosurgery & Psychiatry*, 34 (3), pp. 231-235.

2017, Volume 1, Issue 1, 12-27

[9] Bajd T., and Vodovnik L., 1984, Pendulum test of spasticity, J *Biomed Eng*, 6, pp. 9-12.
[10] Robinson C. J., Kett N. A. and Bolam J. M., 1998, Spasticity in spinal cord injured patients: 2. Initial measures and long-term effects of surface electrical stimulation, *Archives of physical medicine and rehabilitation*, 69 (10), pp. 862-868.
[11] Bohannon R. W., Harrison S. and Kinsella-Shaw J., 2009, Reliability and validity of pendulum test measures of spasticity obtained with the Polhemus tracking system from patients with chronic stroke, *Journal of neuroengineering and rehabilitation*, 30:6(1), pp. 1-12.
[12] Fowler E. G., Nwigwe A. I. and Ho T. W., 2000, Sensitivity of the pendulum test for assessing spasticity in persons with cerebral palsy, *Developmental Medicine & Child Neurology*, 42(03), pp. 182-9.
[13] Bajd T., and Bowman B., 1990, Measurement of skeletal muscles spasticity, In: Popović D. B. (Ed.), *Advances in External Control of Human Extremities X,* Yugoslav Committee for Electronics and Automation, Nauka, Belgrade.
[14] Bajd T., and Bowman B., 1982, Testing and modeling of spasticity, *J. Biomed. Eng.*, 4, pp. 90-96.
[15] Brown R. A., Lawson D. A., Leslie G. C., MacArthur A. L., MacLennan W. J., McMurdo M. E., Mutch W. J. and Part N. J., 1988, Does the Wartenberg pendulum test differentiate quantitatively between spasticity and rigidity? A study in elderly stroke and Parkinsonian patients, *Journal of Neurology, Neurosurgery & Psychiatry*, 51 (9), pp. 1178-1186.
[16] Jamshidi M., and Smith A. W., 1996, Clinical measurement of spasticity using the pendulum test: comparison of electrogoniometric and videotape analyses, *Archives of physical medicine and rehabilitation*, 77 (11), pp. 1129-1132.
[17] Bohannon R. W., and Smith M., 1987, Interrater reliability of a modified Ashworth scale of muscle spasticity, *Physical Therap*, 67 (2), pp. 206.
[18] Le Cavorzin P., Hernot X., Bartier O., Allain H., Carrault G., Rochcongar P. and Chagneau F., 2001, A computed model of the pendulum test of the leg for routine assessment of spasticity in man, *ITBM-RBM*, 22 (3), pp. 170-177.
[19] Le Cavorzin P., Poudens S.A., Chagneau F., Carrault G., Allain H. and Rochcongar P., 2001, A comprehensive model of spastic hypertonia derived from the pendulum test of the leg, *Muscle & Nerve*, 24 (12), pp. 1612-1621.
[20] Le Cavorzin P., Carrault G., Chagneau F., Rochcongar P. and Allain H., 2003, A computer model of rigidity and related motor dysfunction in Parkinson's disease, *Movement disorders*, 18 (11), pp. 1257-1265.
[21] Petrović M. M., Čobeljić R., Popović D.B., 2016, The Use of Inertial Measurements Units for the Evaluation of Spasticity: Pendulum Test, *III IcETRAN*, pp. ME.1.9-12, Zlatibor, Serbia.
[22] Shue G., Crago P. E. and Chizeck H. J., 1995, Muscle-joint models incorporating activation dynamics, moment-angle and moment-velocity properties, *IEEE Transactions on Biomedical Engineering*, 42, pp. 212–223.
[23] Stein R. B., Zehr E. P., Lebiedowska M. K., Popovic D., Scheiner A. and Chizeck H. J., 1996, Estimating mechanical parameters of leg segments in neurologically intact and humans with disabilities, *IEEE Transactions Rehabilitation Engineering*, 4, pp. 201-211.
[24] Chizeck H. J., Chang S., Stein R. B., Scheiner A. and Ferencz D. C., 1999, Identification of electrically stimulated quadriceps muscles in paraplegic subjects, *IEEE Transactions on Biomedical Engineering*, 46 (1), pp.51-61.
[25] Nielsen, S. I., Popovic, M. B. and Popovic, D. B., 2008, Estimation of joint angles with capacitive accelerometers, *2008 Annual IEEE Conference 2008*, pp. 1-4, IEEE Press.
[26] Djurić-Jovičić M. D., Jovičić N. S., Popović D. B. and Djordjević A. R., 2012, Nonlinear optimization for drift removal in estimation of gait kinematics based on accelerometers, *J Biomechanics*, 45 (16), pp. 2849-54.
[27] Lagarias J. C., Reeds J. A, Wright M. H., and Wright P. E., 1998, Convergence Properties of the Nelder-Mead Simplex Method in Low Dimensions, *SIAM Journal of Optimization*, 9 (1), pp. 112–147.