2014 5th IEEE RAS & EMBS International Conference on
Biomedical Robotics and Biomechatronics (BioRob)
August 12-15, 2014. São Paulo, Brazil

A Wearable Mechatronic Brace for Arm Rehabilitation*

T. Desplenter, A. Kyrylova, T.K. Stanbury, A. Escoto,
S. Chinchalkar, and A.L. Trejos, *Member, IEEE*

Abstract— **In recent years, the possibility of using smart technologies to enhance rehabilitative therapies has become a reality. Smart technologies can adjust their functionality based on real-time performance to provide the most effective therapy. This paper presents the design, development and testing of a wearable mechatronic brace created to assist in upper limb rehabilitation. The purpose of the smart brace is to provide safe therapy of musculoskeletal disorders, in particular brachial plexus injuries. A control system has been developed that facilitates the retraining of the biceps for individuals who have suffered brachial plexus nerve damage. Electromyography (EMG) data for flexion and extension of the elbow were recorded from three healthy subjects and used to scale velocity profiles. The experiments assessed the performance of the smart brace in its ability to reproduce a motion, to compensate for the effect of muscle disability and to detect fatigue. The results showed that the control system was able to adjust velocities to accommodate for disability or fatigue. This initial implementation provides a control model and logic from which the brace can be improved. Future testing of the brace using subjects with a brachial plexus injury will help solidify the techniques used for brace control.**

I. Introduction

Hundreds of millions of people around the world suffer from musculoskeletal disorders (MSDs) [1] and many struggle to find a therapy that works for them. The lack of appropriate treatment leads to chronic pain, limited range of motion and other unresolved issues. Identifying a proper treatment is difficult because of the complexity of the musculoskeletal system and the variety of possible disorders. This difficulty may be addressed by merging cutting-edge technologies and smart devices that can measure performance and provide customized treatment for each patient.

One particular MSD that requires extensive hand therapy results when the brachial plexus nerves are damaged. The sensory-motor innervation of the upper limb is provided by the brachial plexus nerve, formed by the confluence of the ventral rami of the spinal nerve roots from C5 to T1. The majority of brachial plexus injuries results from traction of the nerves during extra elongation movements (e.g. motor vehicle accident [2] or shoulder dislocation [3]).

Current treatment options include a three stage rehabilitation process [4]. If the patient has not recovered upon completion of the rehabilitation process, the nerves are repaired surgically. Postoperative treatment consists of passive kinesiotherapy to preserve joint mobility [5]. Brachial plexus injury treatment includes daily home exercise programs, as they prevent contractures, joint deformity, and loss of range of motion [6].

The focus of the rehabilitation process is to reeducate the muscles that perform elbow flexion, forearm rotation, and shoulder motion. Patients who cannot lift their forearm at all have difficulty completing standard rehabilitation processes. Mechanical braces are often used to assist in rehabilitation (Fig. 1), but constant adjustment of the elastic bands are needed to ensure that the patient continues to progress.

To address this problem, the objective of this work was to develop a smart brace that can measure musculoskeletal performance using electromyography (EMG) sensors and adjusts brace motion to automatically provide support during therapeutic treatment. The following sections outline some of the background work that must be considered in the development and the assessment of the brace.

II. Prior Art

A. Rehabilitation Mechatronics

Robotic systems [7, 8] and large exoskeletons [9] have already shown to be beneficial to patients during rehabilitation exercises. For example, CADEN-7 [10] and SUEFUL-7 [11] are exoskeletons used for upper body rehabilitation. Although these systems are not designed for constant wear, and due to their size and expense are only available to patients during therapy sessions, they have served to identify design requirements for multiple-link upper-limb rehabilitation exoskeletons.

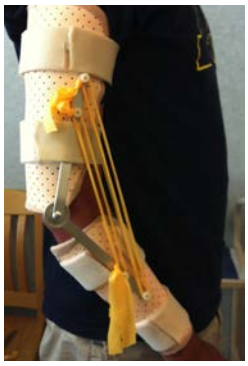

Figure 1. Mechanical rehabilitation brace for brachial plexus injuries.

*This work was supported by the Western Strategic Support for NSERC Success Grant and an Academic Development Fund, Western University. Funding for T.K. Stanbury was provided by an NSERC USRA scholarship.

T. Desplenter is with the Department of Electrical and Computer Engineering at Western University (Western), London, ON Canada (e-mail: tdesplen@uwo.ca). A. Kyrylova is in the Biomedical Engineering Program at Western (e-mail: akyrylov@uwo.ca). T.K. Stanbury is in the Mechatronics Engineering Program at Western (e-mail: tstanbur@uwo.ca). A. Escoto is with the Department of Electrical and Computer Engineering at Western and with Canadian Surgical Technologies and Advanced Robotics (CSTAR), Lawson Health Research Institute, London, ON Canada (abeescoto@gmail.com). S. Chinchalkar is with the Division of Hand Therapy, Hand and Upper Limb Centre, St. Joseph's Health Care, London, ON, Canada (e-mail: schinchalkar@hotmail.com). A.L. Trejos is with the Department of Electrical and Computer Engineering at Western and with CSTAR, Lawson Health Research Institute (phone: 519-661-2111 ext. 89281; fax: 519-663-8401; e-mail: atrejos@uwo.ca).

Other devices that have been nonintrusively designed can be found in the literature [12]. For example, a smart brace for elbow rehabilitation can sense motion and torque [13]. Although a large unit attached to the brace makes constant wear impractical, this design shows a solution to achieving portability in a wearable device. Also, a one-axis nonintrusive elbow orthosis allows the user to adjust joint stiffness continuously [14], showing how novel smart materials can provide new actuation solutions.

B. Sensing

Surface EMG sensors have been proposed as natural muscle interfaces for wearable mechatronic devices [11]. As EMG measurements have high sensitivity to muscle activity, they have been successfully used to measure fatigue [15] and to identify the operator's intention [10, 11].

Surface EMG recording provides a safe, easy and noninvasive method that allows objective quantification of muscle energy [16]. The raw EMG signal can be analyzed in one of four different approaches: amplitude, frequency, EMG-force relationship and amplitude probability distributions [17]. The average, root mean square (RMS) and mean spike amplitude of band-passed EMG activity are used to quantify the magnitude of the muscle activity [18].

III. Methods

To achieve the objective of this work and building on the prior art presented above, the need for a wearable mechatronic elbow brace was identified. The brace design needed to consider real anthropomorphic data as those shown in Table I [19]. The specifications of the brace were determined to be as follows:

1) The mechanical design of the brace must support the forearm. Its size must be adjustable to the size of the patient. The mechanism must provide a one degree of freedom motion to move the elbow through the entire range of flexion and extension. The most common daily activities require the movement of the elbow to be between 30° and 130° [20]. The range of motion must be adjustable as needed by the patient. It must also allow the brace to be locked in certain positions and to be released in free motion when required.

2) The actuation mechanism must provide sufficient force to move the entire weight of the forearm and hand. The amount of force applied by the brace and the speed of the motion must be adjustable according to the needs of the patient. To provide this motion, the required torque at the elbow was computed using the bolded values in Table I, considering that selecting the torque for the 95th percentile male would work for the majority of the population. The total required torque was calculated to be equal to the mass of the hand and forearm acting at the centre of mass of the arm (2.3 Kg at a distance of 0.26 m = 5.9 Nm), rounded up to account for mechanism friction gives a final design requirement of 6.0 Nm.

3) The brace must respond to the user's needs. The following sensors must be incorporated into the system: EMG, position, and acceleration.

4) The control system must be developed to ensure that the mechatronic brace responds adequately to the user's needs. Different modes of operation are needed to enhance functionality.

5) Safety and ergonomic considerations include the ability to change direction or stop when the user reaches the limits of their range of motion. The mechanism must move the lower arm through its natural motion and it must be comfortable to wear (no high pressure points, light weight (<1 kg), unobtrusive and quiet).

Considering these specifications, a smart mechatronic brace was designed and built. This brace prototype allowed a preliminary investigation of a suitable control system to be performed. The development of the brace is described in the following section.

A. Development of the Mechatronic Brace

The mechatronic brace was modeled using the Computer Aided Design software SolidWorks, as shown in Fig. 2. A simulation was completed to ensure that the components were strong enough to withstand the required forces. The brace design allows it to be used on the left or on the right arm by making small adjustments. Metal support sections were built out of aluminum to minimize weight. Customizable and adjustable pieces were built out of acrylonitrile butadiene styrene (ABS) plastic in a Dimension Elite 3D Printer. To provide the one-degree of freedom actuation at a 90 degree angle from the motor shaft (to ensure that the motor is aligned with the arm), a motor with a right-angled gearhead from Maxon Motors was chosen. A right angle gearhead with a 31:1 ratio (RAG #450468) was combined with a planetary one with an 18:1 ratio (GP 32 C). A brushless, 50-Watt motor was used (EC-i 40) with a 2-channel encoder that has a resolution of 512 counts per turn. This motor-gear configuration can produce 6.0 Nm of torque. The final prototype weighed 0.85 kg (Fig. 3).

TABLE I. Sample data used for the design of the smart brace. Bold figures were used to compute the required torque.

	Male			Female		
	5th%	50th%	95th%	5th%	50th%	95th%
Upper Arm Length (m)	0.33	0.36	0.39	0.31	0.33	0.36
Lower Arm and Hand Length (m)	0.45	0.48	**0.52**	0.40	0.43	0.46
Upper Arm Mass (kg)	1.84	2.23	2.67	1.41	1.71	2.07
Lower Arm Mass (kg)	1.14	1.39	**1.66**	0.84	1.02	1.24
Hand Mass (kg)	0.43	0.52	**0.63**	0.34	0.42	0.50

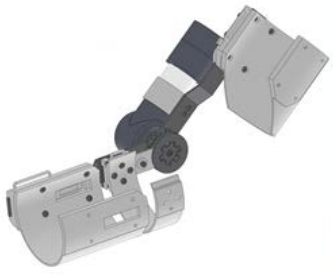

Figure 2. CAD rendering of the wearable brace design.

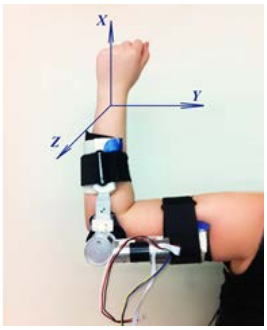

Figure 3. Initial brace prototype worn at 90 degree flexion. Accelerometer coordinates are shown.

Brace actuation is enabled through a motion controller and a desktop computer system. The EPOS 24/5 motion controller (Maxon Motors) was used due to compatibility with the motor and the encoder. Control algorithms were implemented in C++ using the libraries provided with the motion controller. A PID controller was used to control angular position at the low-level. To complete the current design, changes in muscle electrical activity and acceleration were obtained through the Biosignalsplux wearable body-sensing platform (Plux®). Fig. 4 depicts the communication flow between the system components.

B. Experimental Evaluation

An experimental evaluation was performed to test the feasibility of using EMG signals to control brace motion. The focus of the experiments was to control brace velocity based on the EMG input. This section describes how the experiments were completed.

Following the SENIAM's (Surface EMG for a Non–Invasive Assessment of Muscles, 1999) recommendations, the skin was cleaned. Two circular bipolar sEMG electrodes were placed parallel to the muscle fibers (3 cm center to center) over the muscle belly. A reference electrode was placed on the elbow. The resulting voltage waveform corresponds to the difference in potential between the two active electrodes sampled at 1 kHz.

An accelerometer was placed on the inside of the wrist 17 cm from the elbow joint. The motion profile was recorded with help of the OpenSignals (Plux®) software. The accelerometer tracks position changes in the *x*, *y* and *z* directions (see Fig. 3 for coordinate frame assignment).

Prior to the trials, each volunteer was asked to perform a maximal biceps contraction. The EMG signal was recorded for 3 s of rest and 3 s of maximum contraction. The stored data were used to identify the maximum and minimum values of the EMG signal for normalization purposes.

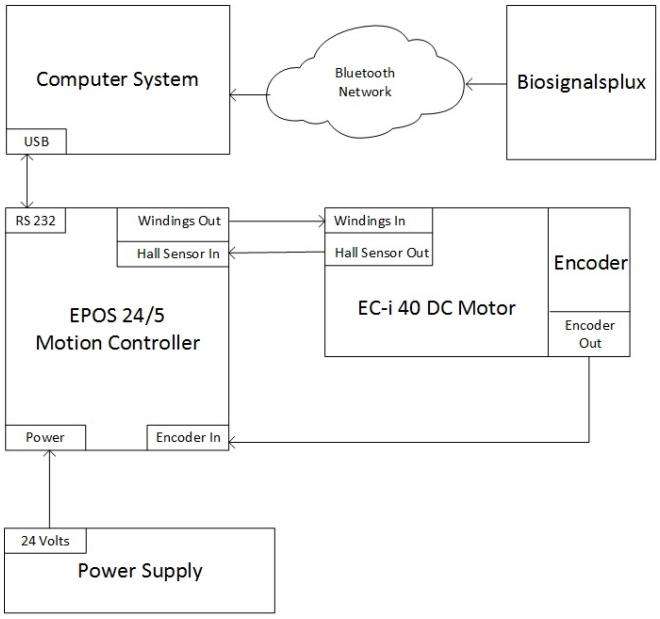

Figure 4. Component level communication diagram of the control system developed for the preliminary experiments.

1) Setup Description

To complete the trials, the smart brace was mounted to a fixed support instead of to the user's arm. Since the brace is still in the developmental stage, its safety has not been validated. Mounting the brace to a fixed support allowed the control system to be assessed without limitations. A rigid L-shaped support was clamped to the top of a desk. The upper-arm support of the brace was clamped to the L-shaped support. A stop was used to keep the upper-arm of the brace in a position perpendicular to the ground.

The brace was loaded with 1.5-kg weights located at the average center of mass of the forearm (14 cm distal from elbow joint) for each subject. This load was selected based on the arm mass of the 50% percentile male and female as listed in Table I. An accelerometer was attached to the brace 17 cm from the center of the joint axis. In the experiments described below, the velocity profile, derived from the subject's EMG signal, was used as input to the motion controller. The velocity commands were sent at a frequency of 4 Hz. The Biosignalsplux accelerometer was used to track brace motion as it completed the desired trajectory.

2) Experiment 1 – Motion Reproduction

Experiment 1 aimed to show that EMG data collected for a biceps flexion motion could be processed and used as a velocity control mechanism for the brace. The brace completes a full flexion motion at maximum speed with a 1.5 kg load in approximately 2 s. The subjects were asked to complete the flexion motion lasting longer than or equal to 2.5 s. One trial, in which each subject was closest to completing the motion in 3 s, was used for EMG to velocity scaling. The velocity profile of each subject's arm was recorded but remained hidden until the experiments were completed.

3) Experiment 2 – Disability Response

The goal of Experiment 2 was to show that the velocity profile of the brace could be adjusted based on the magnitude of the EMG signal. A linear relationship between the EMG signal magnitude and the velocity was assumed. The velocity profile for two subjects from Experiment 1 was used as baseline. To represent a brachial plexus disability, a disability factor was included in the algorithm from Experiment 1. The baseline velocity was multiplied by the disability factor, which ranges from 0 to 1, where 0 is no movement and 1 is the movement of a healthy individual. The result of the multiplication is a velocity profile representing an individual with sub-optimal elbow motion. The brace receives the resultant velocity profile as input. The velocity profile of the brace is recorded for comparison.

4) Experiment 3 – Fatigue Detection

The effectiveness of the therapy is maximized when patients rest immediately after exerting themselves to the point of fatigue. Therefore, the focus of Experiment 3 was to show that the brace could respond to fatigue detection. The goal of the system was to stop brace motion when fatigue was detected. Each subject was asked to complete an elbow flexion and extension motion in repetition beyond the onset of fatigue. The subject held a 5 lb dumbbell during the motion. When the subject felt fatigued, he or she was

instructed to announce the repetition number they had completed up to the point of feeling fatigued.

As it was defined in [21], biceps fatigue is best detected from the RMS and maximum peak of the EMG signal, whereas averaging EMG values is inadequate for detection. Thus, the magnitude of the peak EMG signal of any repetition before fatigue occurs can be used as the threshold value. Fatigue detection occurs when the magnitude of the EMG signal exceeds the threshold. The threshold is unique to each user and varies among trials.

5) EMG to Velocity Scaling

The process of using EMG signals as velocity commands in the control system was split into two parts (Fig. 5). The first section of the process involves creating a velocity profile from the collected EMG data. The OpenSignals software was used to visually determine which section of the EMG data was related to the desired flexion or extension motion. The elbow motions were parsed further into sets of 250 samples. The peak EMG signal for each sample set was recorded. The data set containing the peak EMG signals was normalized and scaled to the desired velocity range. The result of this process produces a velocity profile corresponding to a flexion or extension motion. Although the process can be used for both elbow flexion and extension, Experiment 1 and Experiment 2 focused solely on elbow flexion.

The second section of the process involves using the derived velocity profile as input to command the motion of the brace. A loop is used to command the velocities in the order in which they are listed from the velocity profile. When the last velocity has been sent to the motion controller, the program logic waits until the maximum or minimum position is reached. A velocity of zero is finally sent to the brace to halt the motion.

IV. Results

A. Subjects

Data were collected from three subjects. All subjects are known to have no injury to the brachial plexus nerves. All three subjects participated in Experiment 1 and Experiment 3. The data gathered from subject 1 was not used in Experiment 2 due to time constraints. The results of these experiments are presented below.

B. Experiment 1 Results

To assess the ability of the brace to reproduce a motion, three or more trials were completed by each individual. The trials that shared a common duration (3.1s) across the subjects were used. Only one trial for each subject was used in order to verify that the system could respond appropriately for any given EMG flexion signal. A comparison between the position and velocity profiles of the subject and the brace is presented in Figs. 6 and 7 respectively. The brace was able to follow a similar but not exact trajectory to that of each subject. The velocity profiles of the human and brace must correlate well to ensure safety. In order to address this issue, further research will be conducted.

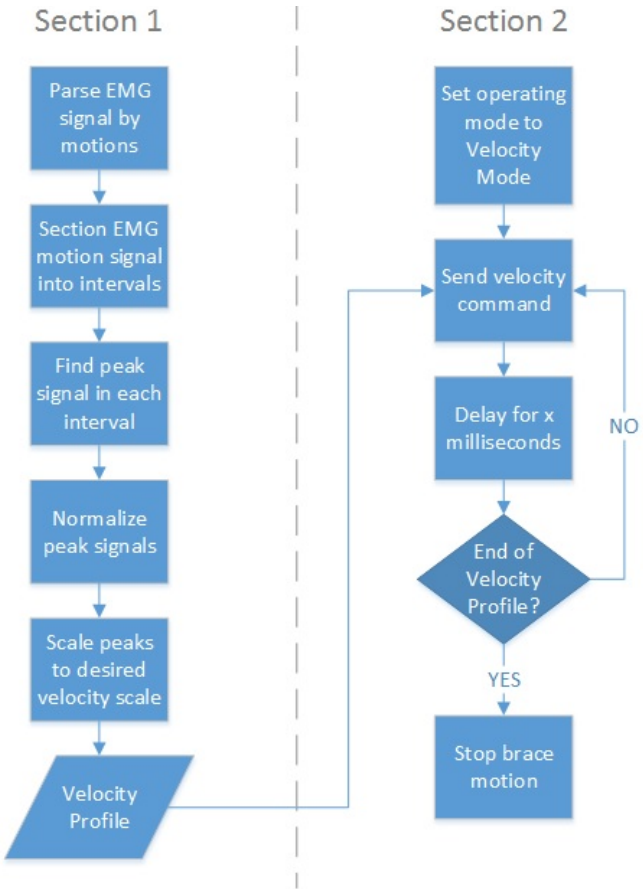

Figure 5: EMG to velocity conversion process broken into sections.

The varying displacement in both directions occurred because the velocity profile of the brace was not similar to that of the subject. These results were expected as the velocity profile was directly scaled from the EMG signal and was not influenced by knowledge of the velocity profile of the subject's arm.

C. Experiment 2 Results

To assess the response to a disability, one trial from Experiment 1 was selected at random to represent the baseline velocity profile for two of the subjects, Six trials for each subject's velocity profile were conducted using the brace: three trials at 50% and three trials at 25% of the baseline velocity. The velocities at each disability factor were averaged and compared to the baseline. Fig. 8 provides sample velocity data of the brace when each disability factor was applied. The brace was able to complete the correct motion when subjected to a reduced velocity signal. For both subjects, decreasing the magnitude of the velocity by a certain factor leads to an increase in duration by approximately the same factor. The desired velocity profile can be tuned based on the disability level of the user.

D. Experiment 3 Results

To assess fatigue, each subject performed one trial. Two repetitions of data were considered from each trial: the repetition before fatigue identification and the fatigued repetition. The magnitude of the maximum EMG signal of the first repetition was considered as the threshold value.

When the EMG signal of the fatigued repetition exceeded the threshold, the brace was commanded to stop. A sample velocity response of the brace was plotted against the EMG signal in Fig. 9. The fatigue identification method used performed as expected for all subjects: upon fatigue detection, the velocity of the brace decreased towards zero.

V. Discussion

A first prototype of a wearable mechatronic brace for biceps rehabilitation following a brachial plexus injury was designed and built. The set specifications of the brace were successfully met. Further improvements and additional requirements have been recognized for future work.

The results of the experiments show a promising future for the use of smart braces in rehabilitation. The EMG to velocity scaling process was able to provide velocity profiles that can achieve flexion motion in similar durations to those of the subjects. In the current trials, the extension motion was conducted using the biceps EMG signal, as opposed to using the triceps signal. A more accurate extension could be integrated into the device using the methods described above, additional EMG sensors on the triceps and changes to the control software. Tuning the scaling process will help to provide velocity profiles that more closely resemble those of the subjects. The brace was able to compensate for a decreased EMG signal by increasing the duration of the motion. Further experiments using subjects who have a brachial plexus injury will provide a more accurate assessment of the ability of the brace to adjust the control logic to accommodate disability. The threshold method for fatigue detection was easy to implement and the brace responded as expected. Adjustments to the threshold method will be discovered upon further testing using disabled subjects. The control system was able to accommodate all experiments with only minor adjustments to the velocity scaling on a per trial basis. Generalization of the control software will provide a platform for which the brace can be enhanced for future use in rehabilitation.

A. Sources of Error

The expectations of the experiments have been verified by the results. However, the accuracy of the results has been affected by several sources of error. For example, each subject performing the trials was asked to wear a standard brace on their arm while performing the motion, in order to limit the range of the motion. The brace restricted the motion to a 110° range but may have altered the trajectory and velocity profile of the arm. Increased friction and weight, due to the brace, will cause the EMG activity to be greater than it would have been for a non-restricted movement. In future experiments, subjects will be wearing the mechatronic brace thereby eliminating the need to limit the range of motion. In addition, the subject will be able to immediately see the response of the brace to the contraction of the muscle and adjust accordingly.

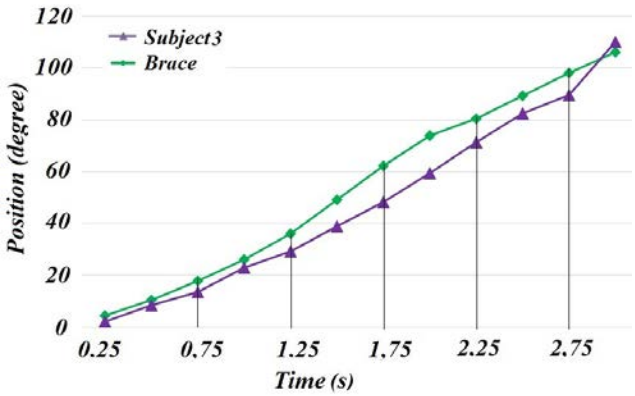

Figure 6: Sample angular position comparison between the brace and Subject 3 for a trial from Experiment 1.

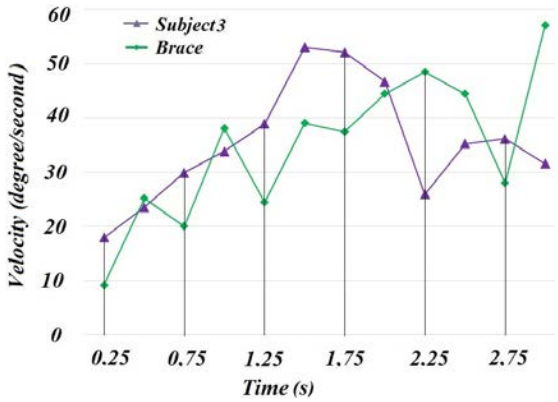

Figure 7: Sample angular velocity comparison between the brace and Subject 3 for a trial from Experiment 1.

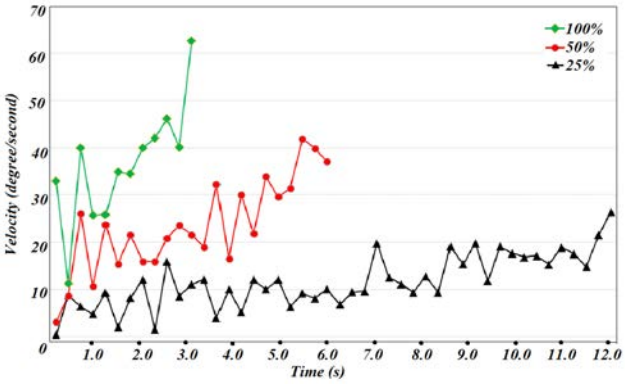

Figure 8: Sample velocity profile adjustment based on varying disability factors for Subject 3.

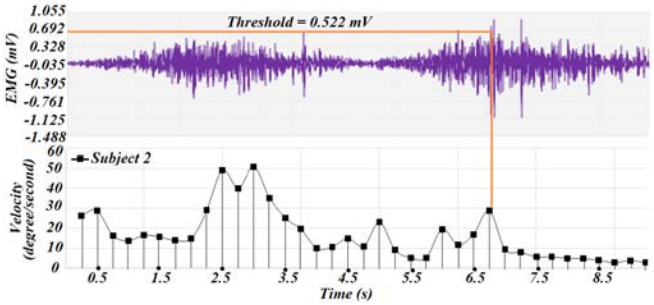

Figure 9: Sample velocity response of the brace during the fatigue detection trial of Subject 2.

The EMG analysis was simplified in order to adjust the Biosignalsplux system to the designed setup. A more accurate algorithm for processing the EMG signal should be considered. The recorded EMG data were used with no filtering or rectifying. Before filters can be tuned, further experiments are needed to differentiate between natural variability of the EMG signal and noise.

B. Model Inaccuracy

The model of the lower arm used in the torque calculation is inaccurate. Evidence of the inaccuracy was discovered when the brace could not complete flexion motions at high velocities. The result forced the time constraints of the motion for Experiment 1. Not considering user dimensions, it is estimated that the brace can be used for subjects with a body weight of 68 kg or less using the lower arm torque model. Improvements to the model will allow for more accurate torque requirements for which the controller parameters can be adjusted.

C. Control Improvements

The EMG to velocity scaling resulted in trial dependent scaling. Customizing the brace to work for each user would require a very long calibration process. The maximum EMG signal per interval method was unable to provide a velocity profile that resembled the velocity profile of the subject. The result was expected but undesired. Analysis of other conversion methods should be explored to better understand if EMG to velocity scaling can be used to provide safe therapy.

The response and stability of the brace is greatly affected by the controller gains. The instability caused by velocity fluctuations can be attributed in part to the controller configuration. Further testing will be conducted to tune controller gains and try other controller configurations in an attempt to maximize performance.

VI. CONCLUSION

A first prototype of a wearable mechatronic brace for biceps rehabilitation was designed and built to match the specifications. The ability of a wearable mechatronic brace to control elbow flexion and extension motion via EMG signals was proven. This solution provides a novel alternative to standard rehabilitation programs for patients with brachial plexus injuries. Although the brace response to disability factors and fatigue was tested separately and without being directly attached to the user's arm, further development will allow for the brace to directly interact with individuals suffering from upper limb disabilities. Adjustments to the mechanical design must be made in order to constrain the motion to the desired trajectories and to improve overall comfort. Distinguishing between common and unique EMG signals and arm motion characteristics will help to understand user customization protocols and improve future designs. EMG processing techniques from the literature will be used during the next development phase. This preliminary investigation proved to be very useful but future work must be conducted in order to reduce error, increase model accuracy and ensure the safety of the individuals using the brace.

ACKNOWLEDGMENT

The authors would like to acknowledge the support from Dan Sweiger at University Machine Services at Western.

REFERENCES

[1] A.D. Woolf, B. Pflege, "Burden of major musculoskeletal conditions," in *Bulletin of the World Health Organization*, vol. 81(9), 2003, pp. 646–656.

[2] F. Altaf, K. Mannan, M. Sinisi, R. Birch, "Severe brachial plexus injuries in rugby: mechanisms and management," *Inj Extra,* vol. 39(5), 2008, pp. 206–206.

[3] A. Kosiyatrakul, S. Jitprapaikulsarn, S. Durand, C. Oberlin, "Recovery of brachial plexus injury after shoulder dislocation," *Injury*, vol. 40, no. 12, 2009, pp. 1327–1329.

[4] R.A. Donatelli, *Physical therapy of the shoulder*. Ed. 5th ed. St. Louis, MO: Elsevier Churchill Livingstone, c2012.

[5] G.E. Omer, Jr., M. Spinner, A.L. Van Beek, *Management of peripheral nerve problems*. Ed. 2nd ed. Philadelphia: Saunders, c1998.

[6] S. Riddick-Grisham, L.M. Deming, *Pediatric life care planning and case management*. Ed. 2nd ed. Boca Raton, FL: CRC Press, c2011.

[7] B. Tsai, *et al*., "An articulated rehabilitation robot for upper limb physiotherapy and training," *IEEE/RSJ International Conf. on Intelligent Robots and Systems*, 2010, pp. 1470–1475, Taipei, Taiwan.

[8] H. S. Lo, S. Q. Xie, "Exoskeleton robots for upper-limb rehabilitation: state of the art and future prospects," *Medical engineering & physics*, vol. 34(3), 2012, pp. 261–268.

[9] K. Anam, A.A. Al-Jumaily, "Active Exoskeleton Control Systems: State of the Art," *Procedia Engineering*, 41(Iris), 2012, pp. 988–994.

[10] J.C. Perry, J. Rosen, S. Burns, "Upper-Limb Powered Exoskeleton Design," *IEEE/ASME Transactions on Mechatronics*, vol. 12(4), 2007, pp. 408–417.

[11] R.A.R.C. Gopura, K. Kiguchi, Y. Li, "SUEFUL-7: A 7DOF upper-limb exoskeleton robot with muscle-model-oriented EMG-based control," *IEEE/RSJ International Conference on Intelligent Robots and Systems,* 2009, pp. 1126–1131, St. Louis, MO, USA.

[12] Chiri, A., *et al*. (Aug. 28–Sept. 1, 2012). On the design of ergonomic wearable robotic devices for motion assistance and rehabilitation. *34th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, pp. 6124–6127, San Diego, CA, USA.

[13] Mavroidis, C., Nikitczuk, J., Weinberg, B., Danaher, G., Jensen, K., Pelletier, and P., Yasevac, D. (2005). Smart portable rehabilitation devices. *Journal of Neuroengineering and Rehabilitation*, 2(1): 18.

[14] Ohba, T., Kadone, H., and Suzuki, K. (Oct. 7–12, 2012). An elastic link mechanism integrated with a magnetorheological fluid for elbow orthotics. *IEEE/RSJ International Conference on Intelligent Robots and Systems,* pp. 2789–2794, Vilamoura, Algarve, Portugal.

[15] T. Oberg, "Muscle Fatigue and Calibration of EMG Measurements," *J. Electromyogr. Kinesiol.*, vol. 5(4), 1995, pp. 239–243.

[16] J. R. Cram, G.S. Kasman, J. Holtz, *Introduction to surface electromyography*. Gaithersburg, Md.: Aspen Publishers, 1998.

[17] A. Freivalds, *Biomechanics of the upper limbs: mechanics, modeling, and musculoskeletal injuries*. 2nd ed. Boca Raton, Fla.: CRC Press, c2011.

[18] G. Kamen, D.A. Gabriel, *Essentials of electromyography*. Champaign, IL: Human Kinetics, c2010.

[19] R. Huston, *Principles of Biomechanics*. CRC Press, 2008, p. 442.

[20] A. Celli, L. Celli, B.F. Morrey, *Treatment of elbow lesions: new aspects in diagnosis and surgical techniques*. Ed. Milano; New York: Springer, c2008.

[21] N. Ahamed, *et al*., "Non-invasive electromyography-based fatigue detection and performance analysis on m. biceps brachii muscle," *ICCSCE*, 2012, pp. 302–306, Penang, Malaysia.