35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Seated Balance During Pitch Motion with and without Visual Input

Mohsen Shafeie, *Member, IEEE*, Nika Zolfaghari, *Member, IEEE*, Kristiina M. Valter McConville, *Senior Member, IEEE*

Abstract— **The study of seated balance and postural control, specifically in relation to wheelchair propulsion, has been an area of interest for quite some time. In biomedical and rehabilitation research this has led to the potential of treatment and prevention of spinal cord and musculoskeletal injuries. To date, little study has been done which analyzes the activity of lower trunk muscles for seated balance, as opposed to upper limb and shoulder muscles. For the purpose of this study, motorized rotational movement in the forward and backward directions was simulated and the corresponding lower back and abdominal muscle activity was recorded by surface electromyography (EMG). A comparison of how muscle activity was affected by visual input was also conducted. This pilot study was performed on two healthy individuals, recording two of their abdominal muscles, and two lower back muscles. Electrodes were placed on the right and left rectus abdominis, external oblique, thoracic erector spinae, and lumbar erector spinae. Each trial consisted of twelve randomized tests that were performed twice on each subject. The results showed that the speed of rotational motion was the dominant factor in abdominal muscle activity. The results also suggested that motion of the subject with respect to the visual display had an inhibitory effect on the motion perception. Furthermore, challenges to wheelchair patients on a slightly rough terrain were highlighted. Finally, the results also suggested that visual effects during rotational motion had a small effect on the subject, which was possibly caused by placing focus on something else rather than on balance issues.**

I. Introduction

Understanding seated balance and the factors which affect it have become common areas of interest in rehabilitation research. One of the areas in which the analysis of seated balance and posture come into play is when dealing with individuals in wheelchairs. To a certain extent, wheelchairs have revolutionalized the way disabled individuals navigate through the world by regaining their mobility, and most importantly, their independence. However, these users commonly experience discomfort and injuries with prolonged wheelchair use.[1] Moreover, the force which is repetitively applied to these users when going over uneven surfaces has been associated with various injuries such as back pain, carpal tunnel syndrome and tendinitis to name a few.[2,3,4,5,6] Wheelchair users commonly utilize ramps for mobility in order to overcome the difficulties associated with changes in ground levels, and experience vibrations when travelling over uneven surfaces. In order to prevent backwards tipping when ascending a ramp, wheelchair users compensate by leaning forwards to shift their centre of mass.[7] Most times, these individuals have weakened muscles due to atrophy or spinal cord injury,[8] and it has been suggested that the amount of directional leaning may be related to abdominal muscle strength.[2]

Mohsen Shafeie is with the Institute of Biomaterials and Biomedical Engineering (IBBME) at University of Toronto.
Nika Zolfaghari is with the Department of Electrical and Computer Engineering at Ryerson University, Toronto, Canada,
Kristiina M. ValterMcConville, is with the Departmentof Electrical and Computer Engineering at Ryerson University, the Institute of Biomaterials and Biomedical Engineering at the University of Toronto and the Toronto Rehabilitation Institute, Canada. kmcconvi@ee.ryerson.ca

To date, numerous studies have been done which examine the upper limb and shoulder muscles in relation to wheelchair propulsion,[9,10] but very little research has been done which investigate the trunk and abdominal muscles. In fact, it has been suggested that the lower trunk and abdominal muscles surrounding the spinal cord are essential during wheelchair propulsion when aiming to maintain a sturdy base and upright posture.[11,12] In a study conducted by Yang et al.[11] lower back and abdominal muscles were assessed during manual wheelchair propulsion at varying speeds and accelerations on a level surface. An interesting finding of this study was that muscle activity was highest during the initial stages of motion as recorded by electromyography (EMG). This made us want to further investigate the activity of lower back and abdominal muscles during forward and backward movement. Most of the studies in this field have been done in relation to manual wheelchair movement. However, as technology is advancing, more and more disabled individuals are using motorized wheelchairs and scooters. A novel aspect of this study is the combination of motorized motion accompanied by virtual reality.

Virtual reality has become a vital component in assessing real world concerns in various domains, as it not only allows for traditional biomechanical assessments, but it also allows for testing which is accompanied by real world visuals. A lot of the time, the individual's perception of the world is enhanced when accompanied by visuals, as it provides an additional dimension and depth perception to interpret the surrounding environment. A further application of this study can be applied to motion theatre rides. Many wheelchair users enjoy motion theatre rides, as it does not exert intolerable forces on the user. It may be worthwhile for designers of these rides to know the thresholds and ranges of motion which wheelchair users can withstand.

For the purpose of our pilot study, we used seated motorized rotary motion at various speeds to assess lower back and abdominal muscle activity recorded by surface EMG. The specific muscles we looked at were the Rectus Abdominis (RA), External Oblique (EO), Thoracic Erector Spinae (TE), and Lumbar Eector Spinae (LE). In addition, we also combined this motion with visuals scenery. This will allow us to perform a comparison of the results to determine if the virtual environment affected muscle activity.

II. Methods

A. Subjects

In this pilot study, the tests were performed on the first two authors of this paper (one male 31 years old, and one female 22 years old). At the time of study neither of them were wheelchair users, nor did they have a history of lower back injury or abdominal pain.

B. Experimental Setup

Subjects were seated on a motorized rotating device, as shown in Figure 1. This pneumatic system was made up of an air compressor, a solenoid valve and a linear actuator.

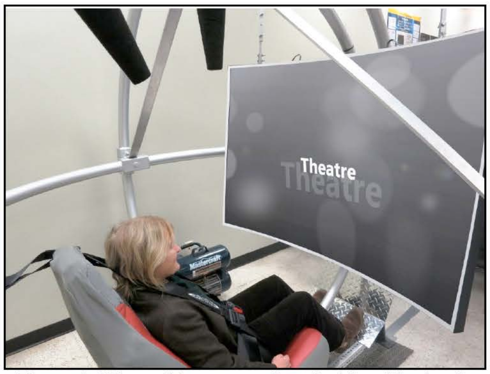

Figure 1: View of the system from inside with visuals

The compressor was capable of storing 18.9 litters (5 gallons) of air under a rated pressure of 861.8 kPa (125 psi) and could deliver 1.18 liters per second (2.5 cubic foot per minute) at 620.5 kPa (90 psi). The compressed air is sent directly to the solenoid valve, and it operates using an electrical motor. The cylinder has a stoke length of 50 cm (19.7 inches) and a bore size of 4 cm (1.6 inches), and is capable of operating under 896 kPa (130 psi). With this cylinder a pitch motion was achieved from -17 to 14 degrees from the horizontal, which yields a total movement of 31 degrees. Furthermore, the solenoid valve used was a 5-way 3-position valve; meaning that there were five different ways the air would flow inside the solenoid. The solenoid could set in three positions depending on the movement of the mechanical core inside. When air is transferred into such a valve the mechanical core moves directing the air to the appropriate ports of the solenoid. The two outputs of the solenoid were sent directly to the pneumatic actuator cylinder, where extension and retraction movements could be performed. The solenoid worked off of a 12V supply, and a Darlington pair transistor circuit as shown in Figure 2 at the rated current (250 mA).

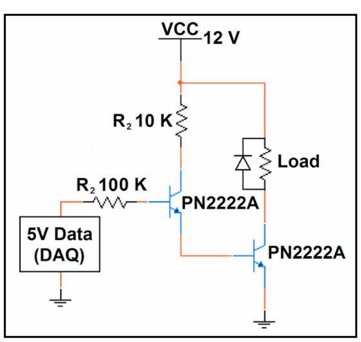

Figure 2: Darlington circuit used for controlling the solenoid

The circuit was hooked up to the NI 6008 DAQ, which allowed for customized control by PWM from LabVIEW. The program was specifically designed for this experiment in order to allow for control of the speed of the cylinder.

Mounted onto the setup was a custom build curved screen, which displayed a projected video during some of the tests. The visuals were an effective virtual reality video from the smartphone game “Boost 2” as it gives the subject the feeling that they are moving forward in a tunnel. The speed of the visuals was edited to match the motion speed, with the lowest speed being 1.5 times slower then a normal video clip playing, and the high speed being twice as fast as a normal video clip being played. Although motion sickness is induced by peripheral vision, the video is structured in such a way that the focal point is directly in the centre of the screen, reducing any possible peripheral distractions. Figure 3 illustrates a screen shot of the video.

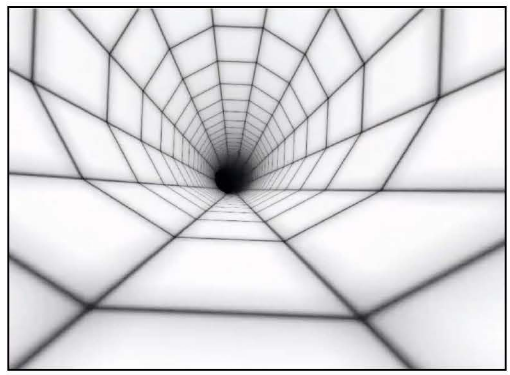

Figure 3: Screen shot of the visual display

C. Surface Electromyography

Surface EMG activity of the lower back and abdominal muscles were recorded by placing two Ag-AgCl electrodes (3M™ Red Dot™ Monitoring Electrodes) on each muscle, with an approximate inter-electrode distance of 3cm.[13] Before the electrodes were placed, the skin was wiped down with alcohol wipes, and similar to a study performed by Howarth et al.[14], two electrodes were placed on each of the following left and right muscles: rectus abdominis (3cm from the umbilicus), external oblique (at the level of the umbilicus), thoracic erector spinae (5cm from the T9 spinal disk), and lumbar erector spinae (3cm from the L4 spinal disk). The electrodes were then connected to snap-leads, which were hooked up to the CleveMed Bioradio 150 which in turn was connected to the computer by a wireless receiver from the CleveMed data acquisition system.

D. Trials

Each trial consisted of twelve tests, with each test consisting of a combination of speed, direction, and visual factors. Specifically, there were three speeds; 4.5 cm/s, 9 cm/s and 50 cm/s of the cylinder arm stroke, which respectively represented 2.82 deg/s, 6.89 deg/s and 31.0 deg/sec. The test was done in two directions (forward and backwards pitch), and with two visual options (with visual video, and without visual video). The order in which each test was run was randomized and the subject was not told which test was about to be performed, so that it was a blind study. Furthermore, each trial was performed twice. The acceleration of the circular rotational path was also measured by the CleveMed Bioradio's Accelerometer.

E. Signal Processing

The sampling frequency of data was 960 Hz at 12 bit/sec. A 4th order Low-pass Butterworth filter with a cut-off frequency of 2.5 Hz was used in order to filter the raw data.

$$|H(\Omega)|^2 = \frac{1}{1+(\Omega/\Omega_C)^{2N}} = \frac{1}{1+\varepsilon^2\,(\Omega/\Omega_P)^{2N}} \quad (1)$$

Where N is the order of filter, Ω_c is the corner frequency, Ω_p is the pass-band edge frequency, and $1/(1+\varepsilon^2)$ is the band edge value of $|H(\Omega)|^2$.

Once the data was filtered, the envelope of the rectified EMG signal was obtained to quantify the muscle activity. After that, the RMS value was calculated which represents 0.707 of one half of the peak-to-peak value.

$$x_{rms} = \sqrt{\frac{1}{n}(x_1^2 + x_2^2 + x_3^2 + \cdots + x_n^2)} \quad (2)$$

Next, the amount of EMG activity in the muscles (iEMG) was obtained by integration of the enveloped signal. This integrated average represents 0.637 of one-half of the peak-to-peak value.

$$iEMG = \int_0^t EMG_{envelope}\, dt \quad (3)$$

Finally, the data was analyzed by Discrete Wavelet Transform (DWT) to express the signal as a linear combination of a set of functions, where the signal is translated at time u and scale s.

$$Wf(u,s) = \langle f, \psi_{u,s} \rangle = \frac{1}{\sqrt{s}}\int_{-\infty}^{+\infty} f(t)\,\psi\left(\frac{t-u}{s}\right) dt \quad (4)$$

The wavelets used involve a scaling function also called the "father wavelets" that must be orthogonal in translation on itself and in dilation.

$$\psi_{u,s}(t) = \frac{1}{\sqrt{a_0^m}}\,\psi\left(\frac{t-nb_0a_0^m}{a_0^m}\right) \quad (5)$$

Since DWT captures both frequency and location in time, it would tell us when and how often a muscle was contracted during the performed tests.

III. RESULTS

Based on the analysis of the processed data, the results indicate that the abdominal muscle activity was least at the lowest speed 2.82 deg/s. At the medium speed of 6.89 deg/s, the muscle activity increased and at the highest speed it was the highest. This can be seen from Figures 4a to 4f where the muscle activity of one of the subjects is graphed from all eight channels during the downward pitch motion. The percentage muscle activity was calculated based on the intensity of the muscle activity compared to maximum voluntary contraction being 100%.

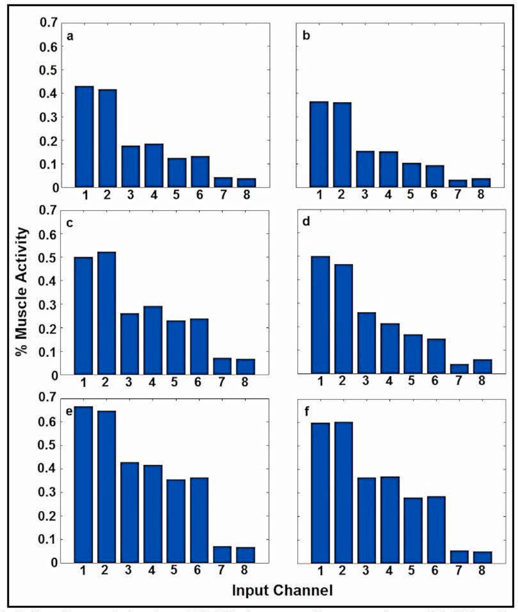

Figure 4: Muscle activity by iEMG (area under envelope EMG) of subject 1 during downwards pitch motion, a) 2.82 deg/s without visuals, b) 2.82 deg/s with visuals, c) 6.89 deg/s without visuals, d) 6.89 deg/s with visuals, e) 31.0 deg/s without visuals, d) 31.0 deg/s with visuals, Channels 1 = RA right, 2 = EO right, 3 = RA left, 4 = EO left, 5 = LE right, 6 = LE left, 7 = TE right, 8 = TE left.

It is clear from the plots on the left side compared to those on the right in Figure 4 that there is a small difference in the muscle activity with the visual display present compared to muscle activity without visuals. This small decrease of activity suggests that the motion of the subject with respect to the visual display may have had an inhibitory effect on the motion perception. The fact that the highest activity was observed at the highest speed was to be expected as there is a sudden start and stop in the motion, and thus a corresponding reaction in terms of acceleration and deceleration respectively. In general, all of the results were similar in this context regardless of whether the motion was forwards or backwards.

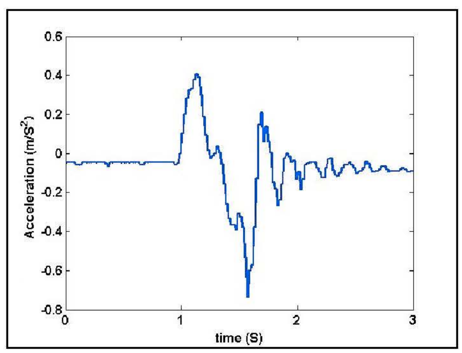

Figure 5: Acceleration profile of upwards pitch motion (1 deg/s).

When we look at the envelopes and the RMS values of the data, the results become even clearer to us. As we can see in Figures 5 and 6, the envelope and RMS of the same muscle (LE right) in both tests are very similar in amplitude, when visuals were present and absent. However, one can also easily notice the rippling of the envelope. This was most probably caused by the small steps of movement of the cylinder as it made the system vibrate just enough so that it would be felt by the subject. This was due to the low frequency of 3 Hz that was used to control the solenoid. This phenomenon is particularly interesting because it highlights the challenges existing for wheelchair patients while they go over slightly rough terrain, such as a cobblestone or cracked sidewalk. The significance of visual scenery was somewhat noticed here, however the difference in the results was not large enough to conclude it as being a major influence.

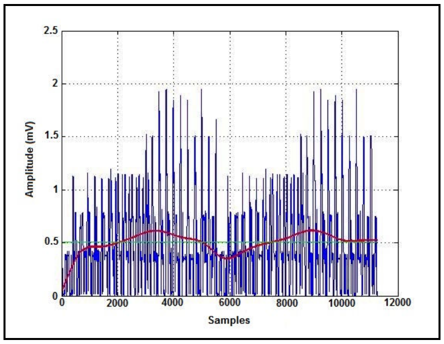

Figure 6: Plot of EMG data of subject 2 during upwards pitch motion at 6.89 deg/s (LE right) without visuals, Blue = rectified EMG, Red = Envelope EMG, Green = RMS value

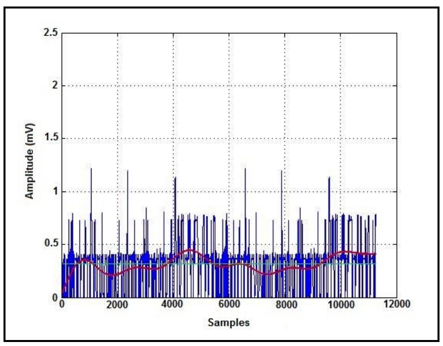

Figure 7: Plot of EMG data of subject 2 during upwards pitch motion at 6.89 deg/s (LE right) with visuals, Blue = rectified EMG, Red = Enveloped EMG, Green = RMS value

When we analyzed the wavelet transforms it was very easy to distinguish the more dominant muscle contractions and their time locations. As we can see from Figure 7, the previous results are yet again confirmed and the transform highlighted the effect of speed; where speed contributed to more muscle activity at the beginning and end of the motion. Also, here the visuals only had a small effect when compared to tests without visuals.

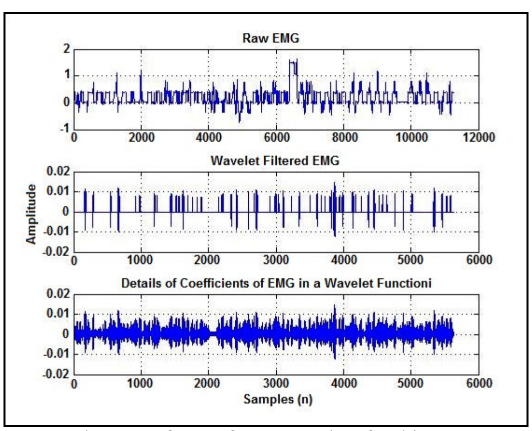

Figure 8: Wavelet Transform of EO muscle of subject 1 at 6.89 deg/s without visuals (forward pitch motion).

IV. Conclusion

The results showed that the seated balance was mostly affected by speed rather than visual scenery. Visuals did have a small affect on the muscle activity possibly by placing focus on something else rather than on balance issues. The fact that small vibrations affect the abdominal muscle activity was also highlighted. For future work, different combinations of pitch dynamics and speed of the visual scene, as well as the effect of sound cues will be investigated.

V. Acknowledgments

We thank the Natural Sciences and Engineering Research Council of Canada and Technical Standards and Safety Authority for funding the study. We also thank David Halliday and Dynamic Structures, Port Coquitlam, Canada. The motion system was provided by Laurence Harris and Michael Jenkin, York University. We also thank Elmira Navidbakhsh and Luka Milosevic for their assistance.

References

[1] Samuelsson K, Larsson H, Thyberg M, Gerdle B, "Wheelchair seating intervention," *Disabil Rehabil*, vol. 23, pp. 677-682, 2001.

[2] Sanderson DJ, Sommer HJ, "Kinematic features of wheelchair propulsion," *J Biomech*, vol. 18, pp. 423-429, 1985.

[3] Curtis, K.A., Drysdale, G.A., Lanza, R.D., Kolber, M., Vitolo, R.S., West, R., "Shoulder pain in wheelchair users with tetraplegia and paraplegia," *Arch. Phys. Med. Rehabil.,* vol. 80, pp. 453–457, 1999.

[4] Rintala, D.H., Loubser, P.G., Castro, J., Hart, K.A., Fuhrer, M.J., "Chronic pain in a community-based sample of men with spinal cord injury: prevalence, severity, and relationship with impairment, disability, handicap, and subjective well-being," *Arch. Phys. Med. Rehabil.,* vol. 79, pp. 604–614, 1996.

[5] Samuelsson, K.A., Tropp, H., Nylander, E., Gerdle, B., "The effect of rear-wheel position on seating ergonomics and mobility efficiency in wheelchair users with spinal cord injuries: a pilot study," *J. Rehabil. Res. Dev.,* vol. 41, pp. 65–74, 2004.

[6] Sinnott, K.A., Milburn, P., McNaughton, H., "Factors associated with thoracic spinal cord injury, lesion level and rotator cuff disorders," *Spinal Cord*, vol. 38, pp. 748–753, 2000.

[7] Chow JW, Millikan TA, Carlton LG, Chae WS, Lim YT, Morse MI., "Kinematic and electromyographic analysis of wheelchair propulsion on ramps of different slopes for young men with paraplegia," *Arch Phys Med Rehabil.,* vol. 90, pp. 271-278, 2009.

[8] Bjerkefors A, Carpenter MG, Cresswell AG, Thorstensson A., "Trunk muscle activation in a person with clinically complete thoracic spinal cord injury," *J Rehabil Med*, vol. 41, pp. 390-392, 2009.

[9] Mulroy SJ, Gronley JK, Newsam CJ, Perry J., "Electromyographic activity of shoulder muscles during wheelchair propulsion by paraplegic persons," *Arch Phys Med Rehabil*, vol. 77, pp. 187-193, 1996.

[10] de Groot S, Veeger HE, Hollander AP, van der Woude LH., "Short-term adaptations in co-ordination during the initial phase of learning manual wheelchair propulsion," *J Electromyogr Kinesiol,* vol. 13, pp. 217-228, 2003.

[11] Yang YS, Koontz AM, Triolo RJ, Mercer JL, Boninger ML., "Surface electromyography activity of trunk muscles during wheelchair propulsion," *Clin Biomech*, vol. 21, pp. 1032-1041, 2006.

[12] Dallmeijer, A.J., van der Woude, L.H., Veeger, H.E., Hollander, A.P., "Effectiveness of force application in manual wheelchair propulsion in persons with spinal cord injuries," *Am. J. Phys. Med. Rehabil.,* vol. 77, pp. 213–221, 1998.

[13] Yu-Sheng Yang, Alicia M. Koontz, Ronald J. Triolo, Rory A. Cooper, Michael L. Boninger, "Biomechanical Analysis of Functional Electrical Stimulation on Trunk Musculature During Wheelchair Propulsion," *Neurorehabilitation and Neural Repair*, vol. 23, pp. 717-725, 2009.

[14] Samuel J. Howarth, Jan M. Polgar, Clark R. Dickerson, Jack P. Callaghan, "Trunk Muscle Activity During Wheelchair Ramp Ascent and the Influence of a Geared Wheel on the Demands of Postural Control," *Arch Phys Med Rehabil,* vol. 91, pp. 436-442, 2010.