Trunk Muscle Transducer

Aidan Friederich, Naji Alibeji, Musa Audu, and Ronald Triolo

Problem Statement/Research Question and Background

An unmet need of the spinal cord injury (SCI) population is reduced control of the trunk muscles which leads to instability when sitting. Trunk stability has been one of the highest priorities in people with SCI [1]. This is often addressed using wheelchair pads and straps to restrict the movement of the trunk. No method exists to both increase support and stability while also allowing for expansion of the effective workspace of the individual, thus increasing independence. Neuroprostheses are able to leverage functional neuromuscular stimulation (FNS) to restore contractions in otherwise paralyzed muscles, [2,3] even specifically targeting the trunk muscles to maintain posture using closed-loop control[4]. Such closed-loop trunk control systems have been reported to be able to withstand external perturbations of up to 45% bodyweight[5].

Previous control applications have been approximated by a linear recruitment curve[5] that can cause destabilizing effects and performance loss in closed-loop control[6]. Nonlinear muscle recruitment curves have shown increased performance in model predictive controllers employing FNS of the leg[6]. These nonlinear recruitment curves have however, never been implemented in control of the trunk. To obtain recruitment curves suitable for trunk muscles, a device must measure the force of the target muscles as they are stimulated *in situ*. It is our conjecture that incorporation of physiologically accurate recruitment curves will potentially improve the performance of controllers for stabilizing the trunk after SCI.

This requires a device that can accurately measure trunk forces which could be further used to determine the loss of trunk strength due to other neurological disorders like stroke that commonly result in hemiparesis. Subjects with hemiparesis use less anterior tilt of the pelvis when moving their trunk[7] and have lateral deviation of the trunk toward the unaffected side[8,9]. This device can be used to estimate the extent of this weakness thus permitting targeted rehabilitation of the weak muscles. Continual use of the device during rehabilitation can provide feedback to the physical therapists possibly resulting in more accurate ways to improve functional outcomes.

Another possible benefit could be seen in lower back pain (LBP) treatment. LBP is experienced at one point by 70% to 80% of the western population[10,11] and has been repeatably associated with a significant reduction in trunk muscle strength[12–15]. Trunk weakness is a risk factor in LBP and correlated with increased pain severity[16]. Characterization of the trunk muscle involvement in LBP can be improved by a complete measurement of trunk strength in multiple directions. Current quantification of trunk strength varies from study to study with many using subjective measures like raising of one leg for a period of time[12] or use of sphygmomanometer placed between a muscle and a surface that outputs a pressure signal[14,17]. Both these measurements are constrained to one directional movement and result in indirect measurements of trunk forces.

To accurately measure trunk muscle force output, a new device was designed to stabilize the pelvis and isolate the actions of the paraspinal muscles. This new apparatus leverages an existing JR3 100M40 six axis load cell to measure forces and moments in multiple dimensions. The rigidity of the attachment to the load cell is imperative for accurate transmission of the muscle forces. For the same reason the device needs to firmly hold the trunk of a wide range of subjects. Thus, the device needs to be highly adjustable. Additionally, the device needs to be easy to use to reduce experiment times and will have to be completely safe for the testing participant. The device called the Trunk Muscle Transducer is detailed below and was designed to fulfill these requirements to characterize the forces of the trunk.

There are currently limited devices on the market to quantify trunk muscle forces and all are limited to one directional movement. For stable joints such as the knee this is acceptable because movement outside of flexion and extension can be neglected with little error. However, the trunk moves in all three dimensions allowing forward, backward, and side to side movement. One commercial device is the BIODEX System 4 Pro designed to measure the torque applied at various human joints such as the knee, ankle, etc. One of the BIODEX attachments, used by Cho *et al.*, is designed to allow for measuring torque at the lumbar joints during trunk anterior/posterior (AP) extension and flexion. This performs well with the assumption that the contribution from both sides of the trunk are nearly equal. However, FNS of different muscles could recruit different amounts of motor units resulting in unequal forces. Also, physiologically accurate recruitment curves of the trunk muscles require measurement of forces in all planes. For these reasons a new design is necessary.

Methods/Approach/Solutions Considered

Several methods were considered to accomplish this task. First option was to create a custom cast of the subject's torso and attach it to the loadcell. If made of fiberglass, the same material used to cast broken bones, the corset (Figure 1A) would be rigid and specifically designed to prevent unwanted movement of the upper body and shoulders. The casting process would take several hours with practice and would be a truly custom fit that could be reused. Holes would be drilled in the back of the cast to allow mounting to the measurement load cell. However, this option would require a lengthy customization time for every participant and there would be no guarantee of it fitting from session to session. The approach would also fail to stabilize the pelvis and isolate trunk muscle forces.

Another option was use of rope or strap anchored to the load cell and then wrapped around the participant's trunk. A similar method was used by McNeill *et al*. This would require aligning the subject with the load cell, pulling the rope completely tight and then stimulating the muscle. While inexpensive and simple, the downside would be large variability from test to test and a loss of information with any slack in the rope. This would also be limited to one direction as all other information would be difficult to extract with a statics analysis.

The last design was inspired by exercise machines that target the external obliques. Exercise machines have been modified to detect muscle strength in one direction before[18]. These devices hold the user at the armpit with pads and use a pulley system to transfer the rotation of the trunk into lifting of weights. Modification of this idea led to a design that would use a T-bar structure to attach to the load cell this would then extend outward with pads to hold the subject's chest under the arms (Figure 1B). This would also prevent forward and backward movement to allow data to be collected in all directions. Making the construct out of 80/20 aluminum beams would increase the rigidness of the system, allow for easy machining, and maintain a manageable weight. This design would be more expensive than the others and require more engineering effort.

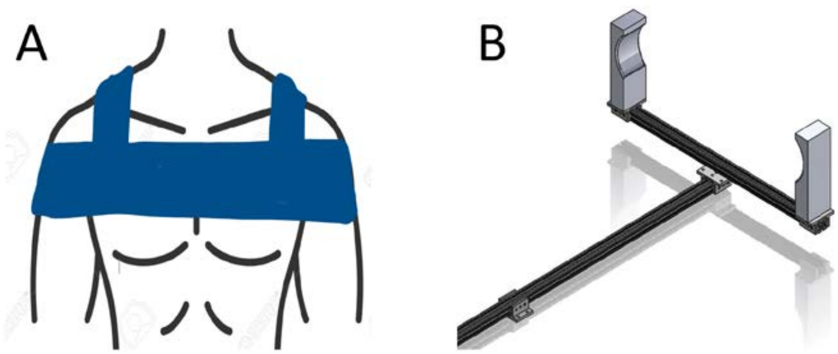

Figure 1: Initial concept designs of possible solutions. A) Initial idea of a fiberglass cast (blue) fitted around the chest. B) First CAD drawing of the trunk holding device.

To determine the best design choice of the three discussed above, a weighted decision matrix (Table 1) was used to qualitatively assess each device in terms of safety, adjustability to various body habitus, comfort, ease of use, rigidity, and cost. The chest holder achieved the highest score due to high scores in adjustability and rigidity; allowing it to comfortably hold the subject and relay all movement back to the load cell. The detailed design was carried out in Solidworks with an emphasis being placed on parts that can be easily bought off the shelf to reduce cost and ease in-house manufacturing/assembly.

Table 1: Weighted decision matrix of the possible solutions.

Criteria Description	Safety	Adjustability	Comfort	Ease Of Use	Rigidity	Low Cost	
	Criteria 1	Criteria 2	Criteria 3	Criteria 4	Criteria 5	Criteria 6	Weighted Score
Weight	5	4	3	3	4	2	21
	24%	19%	14%	14%	19%	10%	100%
Options	Scores	Scores	Scores	Scores	Scores	Scores	
Fiberglass	7	6	4	2	8	6	57.6%
Rope	4	7	9	8	3	10	62.4%
Chest Holder	8	9	6	8	10	3	**78.1%**

Description of Final Approach and Design

The Trunk Muscle Transducer (TMT), shown in Figure 3, was constructed using 80/20 aluminum bars with a customizable design that allows each of the parts to be adjusted along the beam to fit a wide variety of body habitus. Once adjusted each of the joints can be locked into place to make the device as rigid as possible. The interface between the subject and the device uses modified commercial wheelchair pads to provide a comfortable and moldable surface. This allows maximum transfer of force and moments from the point of contact with the subject to the load cell. It was observed during experimentation that when leaning to one side the opposite hip can lift off the chair causing incorrect measurements. Additionally, adjustment of sitting position during experiments increased inter-test variability. A chair was developed address these issues using orthotic thigh holders to prevent movement of the legs and to stabilize the pelvis. The seat of the chair also contains a force plate for optional center of pressure measurements. This is covered in foam padding to distribute interface pressures and help mitigate the potential for developing pressure sores.

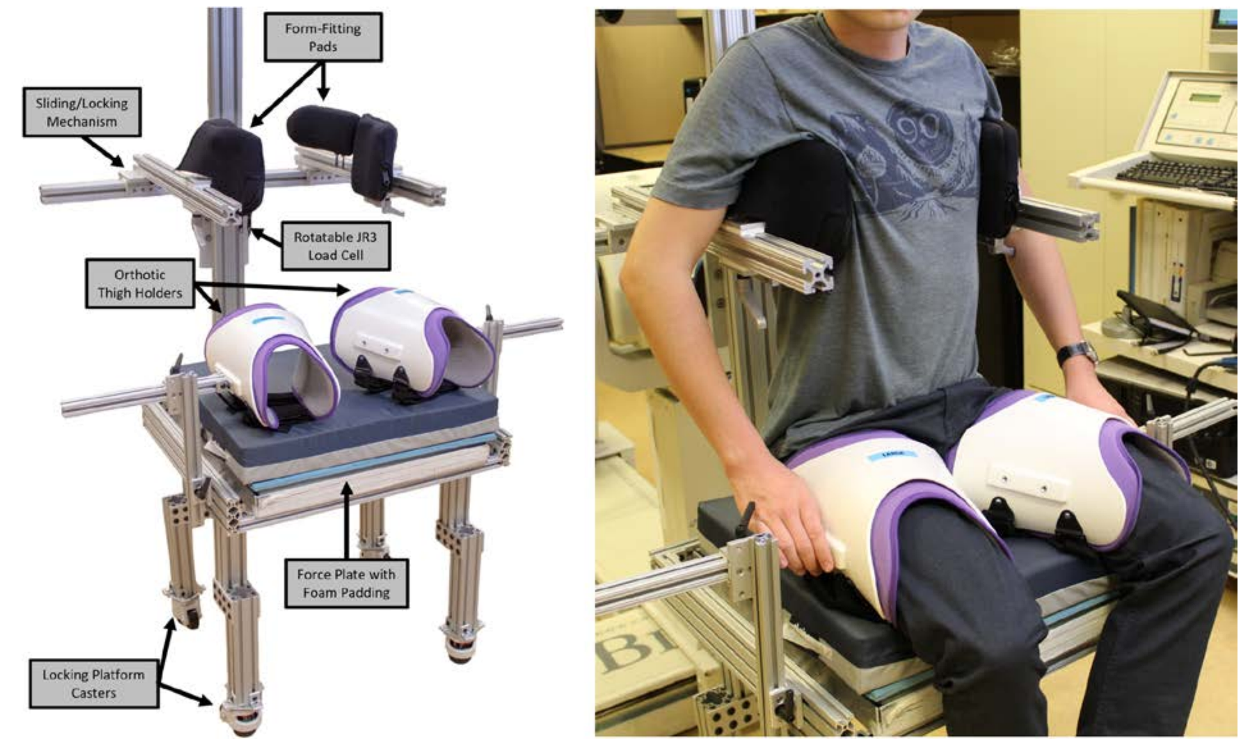

Figure 3: The Trunk Muscle Transducer with and without a subject.

Outcome

The device was bench tested by applying known forces in various directions and rotations by placing a 5kg weight on the pads at various angles. This resulted in four conditions to test the ML forces. Two of the conditions represented ML left and right trunk leaning. The other two represented ML left and right trunk leaning if the subject was also pushing down on the device by applying a force off center 45°, this is a suboptimal testing condition. Only one condition replicated direct application of AP extension force due to the mounting constraints of the load cell. The measured force was compared to the applied force and resulted in a maximum error of 7.5% (Table 2) when the device was suboptimal more complicated loading scenario. Testing conditions that were more indicative of a straight upright trunk resulted in smaller errors.

Table 2: Bench testing data of the trunk muscle transducer (n=5).

Condition	Applied Force (N)	Measured Medial/ Lateral Force (N)	Percent Error
Direct Force (Left ML Lean)	49.05	50.0	1.9%
Direct Force (Right ML Lean)	-49.05	-51.7	5.3%
Off Center Force (Left ML Lean)	34.7	37.3	7.5%
Off Center Force (Right ML Lean)	-34.7	-34.0	-2.1%
Condition	Applied Force (N)	Measured Anterior/Posterior Force (N)	Percent Error
Direct Force (Backward AP Lean)	-49.05	-48.0	-2.1%

Further testing was performed with able bodied subjects to observe trunk forces in the AP and ML directions. The subjects were placed in the device and instructed to apply force in various directions using only their trunk muscles. The directions tested were left and right ML flexion (lateral bending), AP flexion and extension, and combinations of these directions. The last test involved increasing amounts of subject effort to simulate increases in stimulation to the neuroprostheses (Figure 5).

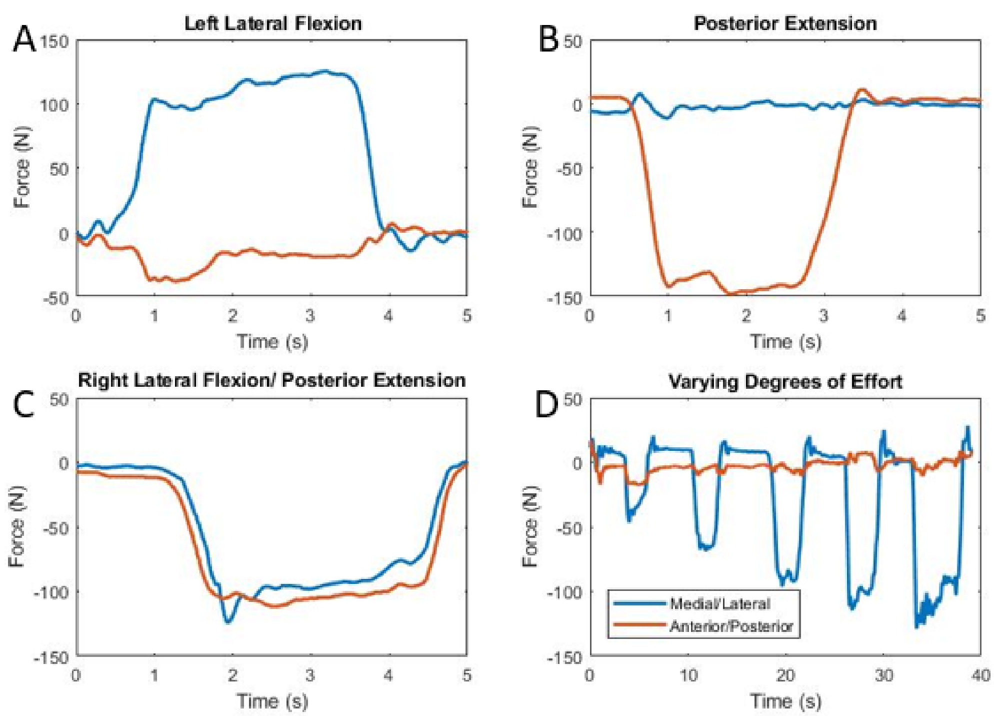

Figure 5: Force measurements with the device in able body tests. A) Lateral flexion to the left side of the subject. B) Posterior extension of the subject. C) Right lateral flexion combined with backward extension. D) Varying subject effort in lateral flexion to the right side of the subject.

Cost Estimate

The total cost of the TMT was $656, most of the cost accrued from attempts to design for adjustability, as the components for that are the most expensive. The accompanying chair cost $1094, most of this cost was acquired in ensuring the chair is stoutly stable while still adjustable. The overall cost of the device and chair was $1750. This price included any revisions to the design and could be much reduced in mass-production. Parts were predominantly bought off the shelf to reduce custom manufacturing cost. The aforementioned BIODEX attachment cost $9000 and is designed only to measure lumbar torques in AP flexion and extension. The TMT is both drastically less expensive and can measure forces in multiple directions.

Significance

The TMT was bench tested to accurately reflect an applied weight with less than 7.5% error, with lower errors in orientations replicating an upright trunk. This accuracy lends confidence to measurements obtained experimentally. Showing that the rigidity of the device was sufficient to transmit forces with minimal loses due the elastic properties of the material. This was achieved in tandem with adjustability using sliding joints that can easily lock position and with pad components that are molded to the general shape of the body. These pads increased comfort while providing an interface from the body to the trunk for maximum force transfer. The device is easy to use, requires minimal straps and can be quickly adjusted to the participant on site while setting up the device and chair. Safety of the device was carefully checked by a physical therapist.

When tested on an able-bodied subject the trunk force transducer was able to measure and distinguish forces in the AP and ML directions separately. When the subject was told to both lean backward and to the right the device measured a force in both the AP and ML directions, appropriately measuring the real-life condition. The resultant of these forces is larger than either the AP or ML muscles groups could produce alone. It is possible these muscles are combining to produce a larger force together however this would have to be confirmed with electromyography to examine muscle activity. This can be further analyzed to determine what muscles are required to obtain different trunk angles and positions. The device was also able to clearly determine differing amounts of muscle effort (Figure 5D), a critical function when controlling the stimulation level of intramuscular electrodes.

Future use of this device will be in characterizing the performance of implanted intramuscular electrodes to match a recruitment curve to the level of applied stimulation. Implementation of this recruitment curve can possibly improve the performance of a trunk specific feedback controller to improve stability in individuals with SCI. Further, this information can be applied to increase the workspace of the individual and allow them to lean reliably to reach objects at greater distances or assist in grabbing of objects off the floor. This device and subsequent characterization will provide robust control to improve the performance of trunk controllers.

The TMT can additionally be used to determine the loss of trunk strength due to other conditions such as hemiparesis and LBP. In Trunk muscle weakness may reduce the ability to protect the spinal cord and to relieve disk pressure[19]. This device can be used to estimate the extent of this weakness thus permitting targeted rehabilitation of the weak muscles. Continual use of the device during rehabilitation can provide feedback to the physical therapists possibly resulting in more accurate ways to improve functional outcomes.

Acknowledgements

This research was funded with help from the National Institutes of Health and the Department of Defense. Resources were utilized from Case Western Reserve University and the Louis Stokes Cleveland VA Medical Center. All testing was performed in the Advanced Platform Technology Center.

References

1. Anderson, K. D. Targeting Recovery: Priorities of the Spinal Cord-Injured Population. *J. Neurotrauma* **21**, 1371–1383 (2004).

2. Triolo, R. J., Bieri, C., Uhlir, J. & Kobetic, R. Implanted FNS systems for assisted standing and transfers for individuals with cervical spinal cord injuries. *Arch Phys Med Rehabil* **77**, 1119–1128 (1996).

3. Marsolais, E. B., Kobetic, R., Ph, D. & Kobetic, R. Implantation techniques and experience with percutaneous intramuscular electrodes in the lower extremities. *J. Rehabil. Res. Dev.* **23**, 1–8 (1986).

4. Vanoncini, M., Holderbaum, W. & Andrews, B. J. Electrical Stimulation for trunk control in paraplegia: A feasibility study. *Control Eng. Pract.* **20**, 1247–1258 (2012).

5. Audu, M. L. *et al.* A neuroprosthesis for control of seated balance after spinal cord injury. 1–12 (2015).

6. Alibeji, N., Kirsch, N., Farrokhi, S. & Sharma, N. Further Results on Predictor-Based Control of Neuromuscular Electrical Stimulation. *IEEE Trans. Neural Syst. Rehabil. Eng.* **23**, 1095–1105 (2015).

7. Messier, S., Bourbonnais, D., Desrosiers, J. & Roy, Y. Dynamic analysis of trunk flexion after stroke11No commercial party having a direct financial interest in the results of the research supporting this article has or will confer a benefit on the author(s) or on any organization with which the author(s) is/are associated. *Arch. Phys. Med. Rehabil.* **85**, 1619–1624 (2004).

8. Duclos, C., Nadeau, S. & Lecours, J. Lateral trunk displacement and stability during sit-to-stand transfer in relation to foot placement in patients with hemiparesis. *Neurorehabil. Neural Repair* **22**, 715–722 (2008).

9. Lecours, J., Nadeau, S., Gravel, D. & Teixera-Salmela, L. Interactions between foot placement, trunk frontal position, weight-bearing and knee moment asymmetry at seat-off during rising from a chair in healthy controls and persons with hemiparesis. *J. Rehabil. Med.* **40**, 200–207 (2008).

10. Svensson, H. O., Andersson, G. B., Johansson, S., Wilhelmsson, C. & Vedin, A. A retrospective study of low-back pain in 38- to 64-year-old women. Frequency of occurrence and impact on medical services. *Spine (Phila. Pa. 1976).* (1988).

11. Waddell, G., Somerville, D., Henderson, L. & Newton, M. Objective clinical evaluation of physical impairment in chronic low back pain. *Spine (Phila. Pa. 1976).* (1992).

doi:10.1097/00007632-199206000-00001

12. Ashmen, K. J., Swanik, C. B. & Lephart, S. M. Strength and flexibility characteristics of athletes with chronic low-back pain. / Caracteristiques de force et de souplesse d'athletes atteints de douleurs lombaires chroniques. *J. Sport Rehabil.* (1996).
13. Hemborg, B. & Moritz, U. Intra-abdominal pressure and trunk muscle activity during lifting. II. Chronic low-back patients. *Scand J Rehabil Med* (1985).
14. Nourbakhsh, M. R. & Arab, A. M. Relationship Between Mechanical Factors and Incidence of Low Back Pain. *J. Orthop. Sport. Phys. Ther.* **32**, 447–460 (2013).
15. McNeill, T., Warwick, D., Andersson, G. & Schultz, A. B. Trunk strengths in attempted flexion, extension, and lateral bending in healthy subjects and patients with low-back disorders. *Spine (Phila. Pa. 1976).* (1980). doi:10.1097/00007632-198011000-00008
16. Cho, K. H. *et al.* Trunk muscles strength as a risk factor for nonspecific low back pain: A pilot study. *Ann. Rehabil. Med.* (2014). doi:10.5535/arm.2014.38.2.234
17. Helewa, A., Goldsmith, C. H. & Smythe, H. A. Patient, observer and instrument variation in the measurement of strength of shoulder abductor muscles in patients with rheumatoid arthritis using a modified sphygmomanometer. *J. Rheumatol.* (1986).
18. Kienbacher, T. *et al.* Age-related test-retest reliability of isometric trunk torque measurements in patients with chronic low back pain. *J. Rehabil. Med.* **48**, 893–902 (2016).
19. Ebenbichler, G. R., Oddsson, L. I. E., Kollmitzer, J. & Erim, Z. Sensory-motor control of the lower back: Implications for rehabilitation. *Med. Sci. Sports Exerc.* (2001). doi:10.1097/00005768-200111000-00014