Technical Note

Prosthetics and Orthotics International

INTERNATIONAL SOCIETY FOR PROSTHETICS AND ORTHOTICS

Prosthetics and Orthotics International
1–5

DOI: 10.1177/0309364615574166
poi.sagepub.com
SAGE

An experimental apparatus to simulate body-powered prosthetic usage: Development and preliminary evaluation

Fan Gao, Johanan Rodriguez and Susan Kapp

Abstract

Background and aim: Harness fitting in the body-powered prosthesis remains more art than science due to a lack of consistent and quantitative evaluation. The aim of this study was to develop a mechanical, human-body-shaped apparatus to simulate body-powered upper limb prosthetic usage and evaluate its capability of quantitative examination of harness configuration.
Technique: The apparatus was built upon a torso of a wooden mannequin and integrated major mechanical joints to simulate terminal device operation. Sensors were used to register cable tension, cable excursion, and grip force simultaneously.
Discussion: The apparatus allowed the scapula to move up to 127 mm laterally and the load cell can measure the cable tension up to 445 N. Our preliminary evaluation highlighted the needs and importance of investigating harness configurations in a systematic and controllable manner.

Clinical relevance:

The apparatus allows objective, systematic, and quantitative evaluation of effects of realistic harness configurations and will provide insightful and working knowledge on harness fitting in upper limb amputees using body-powered prosthesis.

Keywords

Body-powered, grip force, cable tension, cable excursion

Date received: 1 September 2014; accepted: 23 December 2014

Background and aim

The current choices of upper extremity prostheses include cosmetic, body-powered, externally powered, and hybrid,[1] and due to the low cost, ease of maintenance, and reliability, the body-powered prostheses are favorably prescribed.[2,3] The prevalence of body-powered prostheses indicates their importance in clinical settings. To provide quality services to patients, practitioners need to have a good grasp of the functions and mechanisms of upper extremity prostheses. Traditionally, practitioners are educated to test the socket and harness system on the patient with some general fitting and harnessing criteria. However, most of them are subjective. Without consistent harness fitting, it is difficult for practitioners to fully appreciate the effects attributed to changes in the attachment and cable routing. For example, using a voluntary opening (VO) terminal device (TD), each rubber band will provide a pinch force around 6.7 N (1.5 lb), and a fish scale is usually used to check if the tension is sufficient to operate the TD. However, due to the lack of appropriate attachment, it is challenging to get an accurate reading of the cable tension. In addition, the cable is routed around the elbow joint and the placement might significantly affect its curvature which is closely related to efficiency and performance.[4] Due to a lack of consistent and quantitative setting, harness fitting remains more art than science.

UT Southwestern Medical Center at Dallas, Dallas, TX, USA

Corresponding author:
Fan Gao, UT Southwestern Medical Center at Dallas, 6011 Harry Hines Blvd, Dallas, TX 75390-9091, USA.
Email: fan.gao@utsouthwestern.edu

Several attempts have been made to better understand the mechanism of harnessing and efficiency of force transmission.[5–9] Carlson et al.[4] developed a simple mathematical model to predict the Bowden cable and housing efficiency. A motor-driven test apparatus was used to evaluate a mechanical prehensor with variable mechanical advantages.[10] Improvement on cable and harness control was also reported via using a custom apparatus.[11] However, all these studies were conducted solely on the harness system without taking into account the body of the patient. One exception is a recent study by Bertels[12] in which the harness system for a unilateral transhumeral amputee was systematically analyzed using motion analysis with simultaneous registration of cable tension and excursion. However, it remains challenging to evaluate effects of harness configurations consistently and comprehensively. Bench test[9,11] did enhance the consistency of configurations, yet it lacked the interaction between the body and the harness. Ideally, a mechanical simulator composed of basic mechanical structure and properties of the anatomical shoulder and elbow joints integrated with force, linear displacement sensors, will be more appropriate to simulate the fundamental motions such as scapular abduction and shoulder flexion which are essential to operate the TD.

The aim of this study was to fill this gap and develop a mechanical, human-body-shaped apparatus to simulate body-powered upper limb prosthetic usage and evaluate its capability of quantitative examination of harness configuration.

Technique

Development

A torso of a wooden mannequin was used as the base of the simulator (Figure 1(a)). A base metal plate was attached via a roller guide block and rail (i.e. a linear guide) which allowed it to translate freely only in the mediolateral direction (Figure 1(b)). On the base plate, an aluminum plate (in lieu of the anatomical scapula) was attached via two heavy-duty hinge joints with locking mechanism. This configuration allowed three degrees of freedom motion of the shoulder joint (i.e. flexion/extension, abduction/adduction, and internal/external rotation). A figure-9 harness configuration was adopted for its simplicity and a Therapeutic Recreation Systems (TRS) GRIP 3 voluntary closing (VC) hook was selected due to its good force transmission efficiency.[9] A subminiature load cell was placed in-line with the Bowden cable to register cable tension, and an electrical linear displacement sensor was connected in parallel to the Bowden cable to measure the cable excursion. A FlexiForce sensor was attached to the hook to measure the grip force. The detailed sensor specifications are summarized in Table 1. The joints can be adjusted and locked in incremental positions. A loading jack was used to manually move the mechanical scapula laterally to mimic the scapular abduction. A custom LabVIEW™ graphical interface program was used for data acquisition. All signals including cable tension, cable excursion, and grip force were fed into a National Instruments (NI) PCI-6221 card via BNC 2120 terminal box and sampled at 100 Hz (National Instruments Corporation, Austin, TX USA).

Evaluation

Four configurations (1 and 2 were commonly used in checkout test, 3 was to mimic the TD operation via shoulder flexion, and 4 was to mimic TD midline operation[13]) were selected and evaluated (Figure 2). In addition, for each configuration, the ring was placed in such a way that the upper edge was either against the upper edge of the torso in the midline (on C7) or an inch below it (below C7). It should be noted that there was no true anatomical C7 in the apparatus, and it was estimated as closely as possible based on the measurement of a real person with the same height (180 cm, 75th percentile male) of the wooden mannequin.

With elbow and shoulder locked in the prescribed positions, the mechanical scapula was translated laterally up to 120 mm to simulate the scapular abduction maneuver. Loading was conducted stepwise with a step size of 10 mm before the gripper was closed (i.e. the thumb was touching the finger) and a step size of 5 mm was used thereafter. At each step, the loading was held for 5 s. Compared to continuous loading, the stepwise loading eliminated the effect of loading rate. When the grip force approached 35 N[14] (35 N was selected by both referring to previous study in which 30 and 40 N were used and taking into account the loading capacity of the current setup), the above procedure was reversed. Five complete loading and unloading cycles were conducted for each condition. All analog signals were low-pass filtered using a fourth-order Butterworth filter with zero lag and a cut-off frequency of 5 Hz. Signals were then averaged across the last 3-S of each 5-S window for each cycle. The work of closing and opening was obtained and the area enclosed was quantified as the hysteresis (i.e. energy loss) for both grip force and cable tension as a function of cable excursion.[9,15] Specifically, the work done by grip force in relation to cable excursion represented the work done by finger pinching. In addition, cable excursions at selected grip forces (10, 15, 20, and 30 N) were obtained. Outcome measures were averaged across trials. Data analyses were conducted in MATLAB (Mathworks Inc., MA, USA). The results of preliminary evaluation were summarized in Table 2.

Discussion

To the best knowledge of the authors, this was the first mechanical, human-body-shaped apparatus to mimic body-powered upper limb prosthetic usage and to allow systematic and comprehensive evaluation of effects of harness configurations. The linear guide of the apparatus

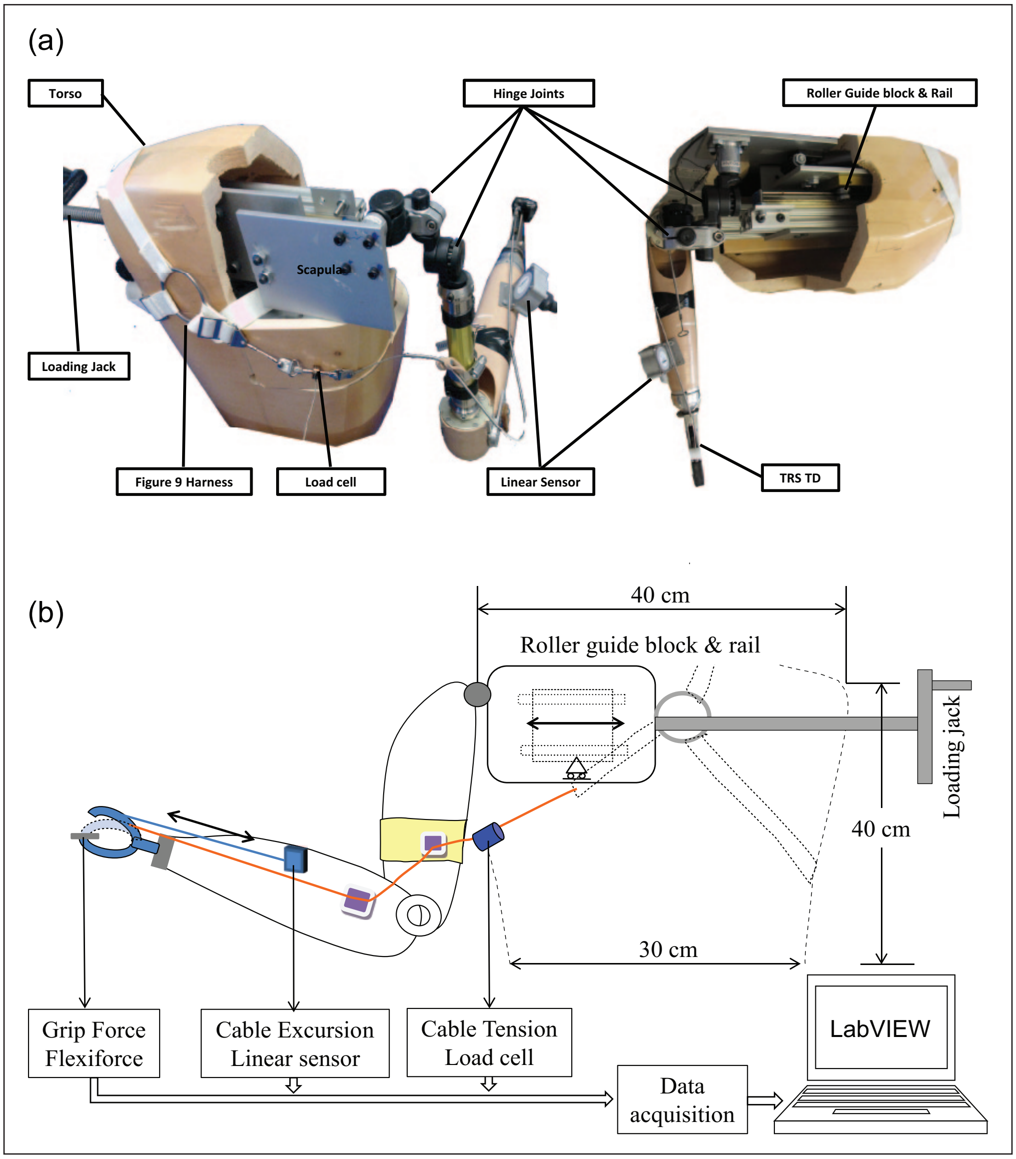

Figure 1. (a) Experimental apparatus with figure-9 harness and (b) schematic drawing of experimental setup.

Table 1. List of sensors.

Sensor	Vendor	Range	Accuracy of full scale
SP1-12 linear displacement sensor	Celesco Transducer Products Inc., USA	317.5 mm (12.5 in)	0.25%
Model 11 BR subminiature load cell	Honeywell Inc., USA	444.8 N (100 lb)	0.5%
Flexiforce A201-100	Tekscan Inc., USA	444.8 N (100 lb)	±3%

allowed lateral translation up to 127 mm (i.e. 5 inch) and the load cell can measure the cable tension up to 445 N (i.e. 100 lb). As reported, to operate TD, the cable tension was typically in the range of 10–90 N and the cable excursion was between 30 and 70 mm.[12] Our design capacity met the needs and even allowed us to simulate the operation of TD over a wide range of body postures and/or conditions not commonly seen in the clinic (e.g. to stimulate young and strong amputee).

Our preliminary evaluation, though not comprehensive and conclusive, did highlight the importance of investigating harness configurations in a systematic and controllable manner. Our results indicated that configurations with the ring closer to C7 required slightly less cable excursion to reach the desired grip force. In addition, the hysteresis of cable tension–excursion was lower when the ring was placed on C7 in comparison to lower ring placement across configurations with an exception

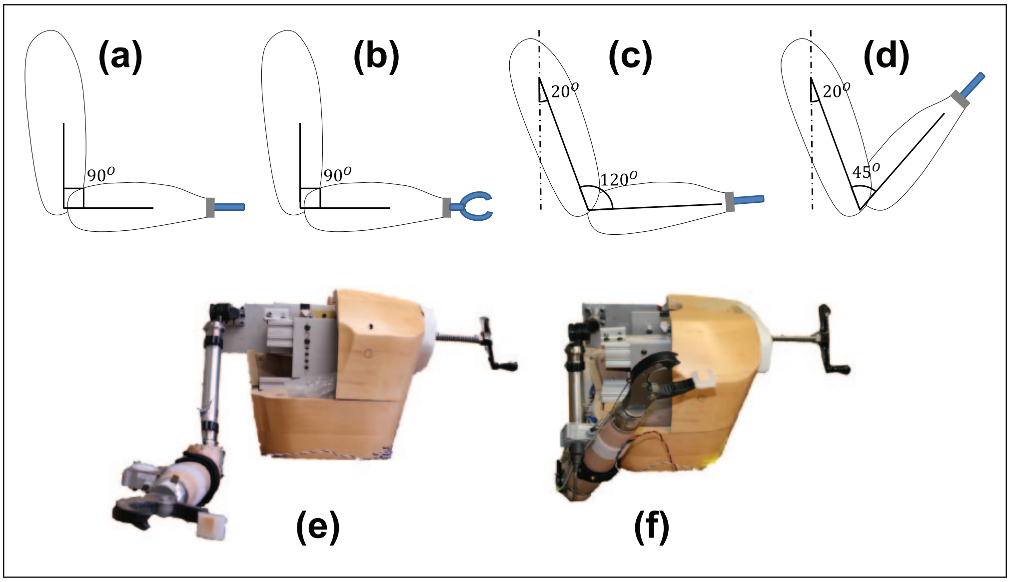

Figure 2. (a) Configuration 1—neutral shoulder position, elbow joint flexed at 90° with TRS hook parallel to the transverse plane; (b) configuration 2—neutral shoulder position, elbow joint flexed at 90° with TRS hook parallel to the sagittal plane; (c) configuration 3—shoulder flexed at 20°, elbow joint set at 120° with TRS hook parallel to the transverse plane; (d) configuration 4—shoulder flexed at 20°, internally rotated 30° with elbow joint set at 45°; (e) photo of configuration 3; and (f) photo of configuration 4.

Table 2. Summary of preliminary evaluation (mean ± *SD*, N = 5).

Outcome measures		**Configuration 1**		**Configuration 2**		**Configuration 3**		**Configuration 4**	
		On C7	**Below C7**	**On C7**	**Below C7**	**On C7**	**Below C7**	**On C7**	**Below C7**
Grip force vs. cable excursion	Work closing (N mm)	11 ± 1	8 ± 1	23 ± 2	19 ± 2	13 ± 1	15 ± 3	9 ± 1	14 ± 1
	Work opening (N mm)	8	6 ± 1	17 ± 2	13 ± 1	9 ± 1	12 ± 2	8 ± 1	13
	Hysteresis (N mm)	2 ± 1	3 ± 1	6 ± 2	6 ± 2	4 ± 1	4 ± 2	1 ± 1	1 ± 2
Cable tension vs. cable excursion	Work closing (N mm)	168 ± 4	189 ± 15	190 ± 4	227 ± 5	136 ± 5	169 ± 7	171 ± 2	184 ± 5
	Work opening (N mm)	108 ± 8	126 ± 5	135 ± 6	165 ± 4	107 ± 7	122 ± 3	110 ± 3	109 ± 5
	Hysteresis (N mm)	59 ± 4	63 ± 10	54 ± 3	61 ± 7	30 ± 5	47 ± 9	61 ± 4	75 ± 4
Cable excursion at selected grip force (mm)	10 N	18	18	17	18	18	18	19	19
	15 N	18	19	17	18	18	18	19	19
	20 N	18	19	17	18	19	19	19	19
	30 N	19	19	18	18	19	19	19	19
Cable tension at selected grip force (N)	10 N	108 ± 3	147 ± 3	43 ± 7	74 ± 6	77 ± 4	82 ± 10	95 ± 6	101 ± 10
	15 N	129 ± 4	177 ± 6	64 ± 6	105 ± 3	97 ± 7	111 ± 8	132 ± 6	138 ± 10
	20 N	141 ± 11	200 ± 10	83 ± 7	136 ± 7	118 ± 11	137 ± 13	159 ± 5	172 ± 14
	30 N	167 ± 9	234 ± 8	126 ± 8	195 ± 12	164 ± 12	190 ± 13	211 ± 0	256 ± 12

SD: standard deviation.
SD is not shown if it is 0.

of configuration 2 in which the gripper was set in the sagittal plane. The required cable tension to achieve preselected grip forces was in general lower with ring placement at C7 compared to ring placement below C7 with only one exception in configuration 4 under 30 N of grip force. Previous study using bench test showed that for TRS gripper, 33 N of cable tension was needed to achieve a pinch force of 15 N.[9] By taking into account a wide range of configurations, our results, however, showed that the required cable tension was much higher and ranged from 64 to 177 N. Additionally, though the hysteresis of cable tension versus cable excursion was similar to that reported in the literature,[9] our results showed much lower work of opening and closing and cable excursion. This discrepancy might be attributed to the difference in configuration (i.e. in-line layout vs. more realistic harnessing used in the current setup). Further systematic investigations of factors such as types of the ring, placement of the ring, cable routing, and joint positions (i.e. wrist, elbow, and shoulder) are underway.

Limitations of this study are acknowledged. The joint locking mechanism allows discrete incremental displacement at a step size of 10°–15°, which is rough and significantly reduces the number of possible postures to study. In addition, the torso lacks the coverage of soft tissue which likely either under- or over-estimates the measures. A jack is used as the loading mechanism, and it has to be operated manually. The manual operation is arduous and significantly limits the possible number of configurations to explore. The current apparatus primarily uses ipsilateral scapular abduction to operate TD to simplify the setup. Although it is also possible to operate TD via shoulder flexion (e.g. pre-set shoulder at various postures and then load scapula), the procedure is tedious and discrete. Further improvement might accommodate motors to make the operation more easily. The selected configurations, though representative, are far from comprehensive. More configurations will be investigated in the future study to establish a more realistic model of the harness which in turn can be used in computer simulation to further explore the musculoskeletal involvement.

In summary, we have developed a mechanical, human-body-shaped apparatus capable of simulating body-powered upper limb prosthetic usage via simple yet effective mechanical mechanism such as linear guide and locking joints. With integrated sensors, it allows systematic and comprehensive evaluation of effects of realistic harness configurations. A thorough, objective, and quantitative evaluation will provide insightful and working knowledge on harness fitting.

Key points

- Harness fitting in the body-powered prosthesis remained more art than science due to a lack of consistent and quantitative evaluation.
- We developed a mechanical, human-body-shaped apparatus to simulate body-powered upper limb prosthetic usage with embedded sensors and evaluated its capability of quantitative examination of harness configuration.
- Our preliminary evaluation highlighted the needs and importance of investigating harness configurations in a systematic and controllable manner.
- The apparatus allowed objective, systematic, and quantitative evaluation of effects of realistic harness configurations and can provide insightful and working knowledge on harness fitting in upper limb amputees using body-powered prosthesis.

Author contribution

F.G. was responsible for drafting the manuscript, experimental setup and test, and data processing and revision; J.R. was involved in the experimental setup and data processing; and S.K. was also involved in the experimental setup.

Declaration of conflicting interests

The authors declare that there is no conflict of interest.

Funding

This work was supported by UT Academy of Health Science Education Small Grants; Green fellowship (J.R.); American Heart Association (09SDG2080460).

References

1. Smith D, Michael J and Bowker J. *Atlas of amputations and limb deficiencies: surgical, prosthetic, and rehabilitation principles*. 3rd ed. Rosemont, IL: American Academy of Orthopaedic Surgeons, 2004.
2. Biddiss EA and Chau TT. Upper limb prosthesis use and abandonment: a survey of the last 25 years. *Prosthet Orthot Int* 2007; 31: 236–257.
3. Shaperman J, Landsberger S and Setoguchi Y. Early upper limb prosthesis fitting: when and what do we fit. *J Prosthet Orthot* 2003; 15: 11–17.
4. Carlson LB, Veatch D and Frey DD. Technical forum—efficiency of prosthetic cable and housing. *J Prosthet Orthot* 1995; 7: 96–99.
5. Carlson L and Long M. Quantitative evaluation of body-powered prostheses. In: *American Society of Mechanical Engineers, dynamic systems and control division*, Chicago, IL, 27 November–2 December 1988, pp. 1–16. New York: ASME.
6. Corin J, Holley T, Hasler R, et al. Mechanical comparison of terminal devices. *Clin Prosthet Orthot* 1987; 11: 235–244.
7. LeBlanc M, Setoguchi Y, Shaperman J, et al. Mechanical work efficiencies of body-powered prehensors for young children. *J Assoc Child Prosthet Orthot Clin* 1992; 27(3): 7–75.
8. Smit G, Bongers RM, Van der Sluis CK, et al. Efficiency of voluntary opening hand and hook prosthetic devices: 24 years of development? *J Rehabil Res Dev* 2012; 49: 523–534.
9. Smit G and Plettenburg DH. Efficiency of voluntary closing hand and hook prostheses. *Prosthet Orthot Int* 2010; 34: 411–427.
10. Frey DD and Carlson LE. A body powered prehensor with variable mechanical advantage. *Prosthet Orthot Int* 1994; 18: 118–123.
11. Kitayama I, Matsuda M, Nakajima S, et al. Improvement of control cable system of trans-humeral body-powered prostheses. *Prosthet Orthot Int* 1999; 23: 123–129.
12. Bertels T. Functions of the body harness for upper extremity prostheses. *Med Orthop Tech* 2001; 121: 13–17.
13. Atkins JD and Meier RH. *Comprehensive management of the upper-limb amputee*. New York: Springer-Verlag, 1989, p. 260.
14. Plettenburg DH, Hichert M and Smit G. Feedback in voluntary closing arm prostheses. In: *MyoElectric controls/powered prosthetics symposium*, Fredericton, New Brunswick, Canada, 14–19 August 2011. New Brunswick, Canada: The University of New Brunswick.
15. Gao F, Carlton W and Kapp S. Effects of joint alignment and type on mechanical properties of thermoplastic articulated AFO. *Prosthet Orthot Int* 2011; 35: 181–189.