JRRD
Journal of Rehabilitation Research & Development

Journal of Rehabilitation Research and Development Vol. 35 No. 4, October 1998
Pages 388-395

The design and development of a gloveless endoskeletal prosthetic hand

Rajiv Doshi, BS; Clement Yeh, BS; Maurice LeBlanc, CP, MSME

Rehabilitation Research and Development Center, Department of Veterans Affairs Palo Alto Healthcare System, Palo Alto, CA 94304; Stanford University School of Medicine, Stanford, CA 94305

This project is based upon work supported by the Rehabilitation R&D Center at the Department of Veterans Affairs Healthcare System in Palo Alto, the Temesvary Fund of the Rehabilitation Engineering Center at Packard Children's Hospital at Stanford University, and the Stanford University Medical School Medical Scholars Program.
Address all correspondence and requests for reprints to: Maurice LeBlanc, Chief of Design and Development, Rehabilitation Research and Development Center (153), VA Palo Alto Healthcare System, Palo Alto, CA 94304; email: leblanc@roses.stanford.edu.

Abstract--Current prosthetic hands, although functional, have the potential of being improved significantly. We report here the design and development of a novel prosthetic hand that is lighter in weight, less expensive, and more functional than current hands. The new prosthesis features an endoskeleton embedded in self-skinning foam that provides a realistic look and feel and obviates the need for a separate cosmetic glove. The voluntary-closing mechanism offers variable grip strength. Placement of joints at three locations (metacarpophalangeal and proximal and distal interphalangeal) within each of four fingers affords realistic finger movement. High-strength synthetic cable attached to the distal phalanx of each finger is used to effect flexion. A multiposition passive thumb provides both precision and power grips. The new prosthesis can securely grasp objects with various shapes and sizes. Compared to current hands, weight has been reduced by approximately 50%, and cable excursion required for full finger flexion by more than

50%. The new endoskeletal prosthesis requires approximately 12-24% less force input to grasp a variety of everyday objects, largely due to its adaptive grip. Production cost estimates reveal the new prosthesis to be significantly less expensive than current prosthetic hands.

Key words: endoskeleton, prosthetic hand, self-skinning foam, upper-limb amputation.

INTRODUCTION

Background

Interest in prosthesis design that coincided with the return of wounded American soldiers from World War II has decreased in recent decades. The design of body-powered upper-limb prostheses in particular has experienced few, if any, major breakthroughs since the early 1960s (1). Persons with amputation frequently express dissatisfaction with the current state of upper-limb prosthesis technology (2), noting numerous deficiencies with their prostheses. Yet continued advances in materials science make more functional and realistic-appearing prostheses increasingly possible.

Upper-limb prostheses are either hook or hand-shaped, and are actuated by body or external power. In the United States, approximately 70 percent of users wear hooks (3). Outside the United States, especially in developing countries, there is a greater preference for hand-shaped prostheses. Compared to hooks, prosthetic hands generally offer less function and durability at greater weight and cost (4). Nonetheless, many individuals still choose hands over hooks, primarily for cosmetic reasons. Our goal was to design a new prosthetic hand that addresses these deficiencies by increasing function and appearance while decreasing cost and weight.

Areas for Improvement

Identifying deficiencies in current prosthetic hands was a priority. Surveying adults and the parents of children with amputation regarding current limitations of prostheses and suggested improvements was particularly helpful.

Finger Movement

Most current prosthetic hands only bend at the metacarpophalangeal joint in each of the first two fingers. The remaining two fingers are passive. Finger flexion, therefore, does not accurately mimic the movement of the human hand. Past designs using multiple phalanges and joints within each finger to improve finger movement have proven disappointing (5).

External Glove

The external glove (usually made of PVC) serves an important cosmetic function and provides protection from abrasion and corrosion. However, the glove requires frequent replacement (6). The expected service life of a glove is 3 to 6 mo, although this varies depending on the conditions of use. More gloves may be discarded because they are stained than because they are worn out (7). The glove also restricts finger motion, accounting for an estimated 50 percent of the inefficiency in the prosthesis (8). Moreover, current gloves may not fit properly, have acceptable appearance (2), or feel soft, like human tissue (9).

Function

Current body-powered prosthetic hands are functional, but there is room for improvement (2). A recent survey of persons using such prostheses revealed a preference for a new prosthesis that is better able to hold both small and large objects (6). Interestingly, the top two preferences of those using electric hand prostheses included bending fingers and a thumb that can be positioned (6).

Weight

Users frequently express a preference for lighter prosthetic hands (2). The interface between the prosthetic socket and residual limb is soft tissue. Even with a solid fit, the attachment may feel insecure because there is a false joint (10). Minimizing prosthesis weight can reduce this common problem.

Energy Expenditure

Current prosthetic hands can require high energy expenditure (11). Since bending is typically restricted to two joints that cannot move independently, the grip is not adaptive. That is, the fingers do not wrap around the object as fingers in the human hand do. Consequently, there is little contact area between hand and object, requiring high pinch forces at the fingertips to grasp the object. Therefore, the user must expend more energy than with an adaptive grip that increases contact area.

Cable Systems

Biscapular abduction and/or glenohumeral flexion effect finger flexion by way of a cable. Control cable technology has not changed significantly in the past 50 years (12). Traditionally, body-powered prostheses have used stainless steel cable, with end fittings attached by special crimping tools. Users of body-powered hands report replacement of the cable as their most frequent maintenance problem (6), and one that cannot be solved at home, because specialized tools are required. Additionally, cable stiffness makes following sharp radius bends difficult, reducing efficiency (12).

METHODS

With the above deficiencies and suggested improvements in mind, the design objectives for the new body-powered prosthetic hand were established.

Design Objectives

1. Simple body-powered voluntary-closing mechanism between the fingers and thumb. Variable grip strength (graded prehension) should be offered.
2. Flexion of four fingers against a passive thumb. Finger movement in the prosthesis should realistically mimic the movement of human fingers.
3. Replacement of the external glove with a material that looks and feels more realistic. Material should be inexpensive, nontoxic, and inert.

4. Provision for precision grips (i.e., holding a key) and cylindrical power grips (i.e., holding a soda can) using a passive thumb that can be positioned as needed.
5. Lighter weight than current prosthetic hands.
6. Improved efficiency and decreased input forces required to grasp everyday objects, compared to those of current prosthetic hands.
7. A harness cable that extends into the center of the prosthesis through a standard threaded attachment. The unit should be interchangeable with other terminal devices.
8. Greater ease of manufacturing and lower costs compared to current prostheses, allowing for widespread production and use of the new prosthesis.

General Design Features

Our overall goal was to design a prosthesis that mimics the appearance and movement characteristics of the human hand, and we decided that an endoskeleton surrounded by self-skinning foam would be a viable solution. The endoskeleton would be composed of four active fingers, each capable of bending at the metacarpophalangeal (MCP), proximal interphalangeal (PIP), and distal interphalangeal (DIP) joints. The endoskeleton would also contain a passive thumb that could be positioned by the user to provide different grips, power and precision.

The prosthesis is voluntary closing. The fingers remain extended until the harness cable is displaced. Harness cable actuation causes flexion of all four fingers against the stationary thumb. Upon release of the cable, the fingers extend, opening the grip. This voluntary-closing mechanism offers the benefit of graded prehension (13), allowing the user to vary grip force depending on harness cable force. This feature assists in grasping fragile objects. A maximum harness cable displacement of approximately 5 cm with 270 N is assumed available through biscapular abduction and/or glenohumeral flexion (14).

Endoskeletal Fingers and Thumb

Each finger consists of plastic phalanges with pinned-hinge joints at the MCP, PIP, and DIP locations (**Figure 1**). These joints offer low-friction bending while resisting lateral deflection. With three joints in each of four fingers, this design represents a significant change from current prosthetic hands that bend at only two MCP joints and at no PIP or DIP joints. The 10 additional joints offer the hand more authentic finger movement than current prostheses, while also providing an adaptive grip.

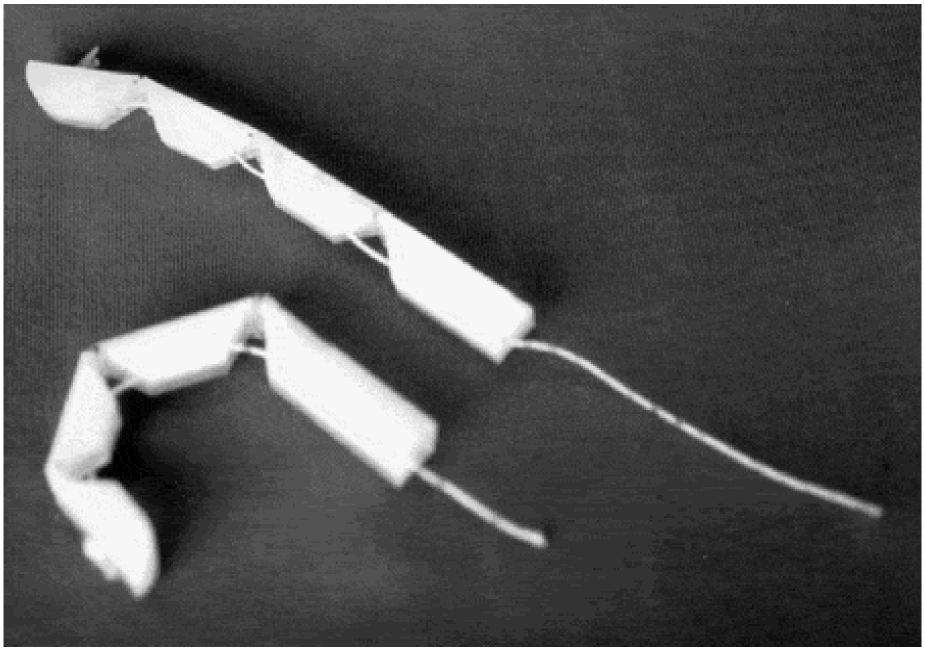

Figure 1. Endoskeletal plastic fingers offer bending at the MCP, PIP, and DIP joints, providing realistic finger movement.

Each finger is actuated by a cable running along its palmar surface and attaching to the distal phalanx. This configuration is similar to that of the flexor tendons in the human hand. The four finger cables converge at a single node within the palm. The node attaches to a traditional fitting that is connected proximally to the harness cable. When the harness cable is displaced, tension in the four finger cables causes the fingers to flex (**Figure 2**).

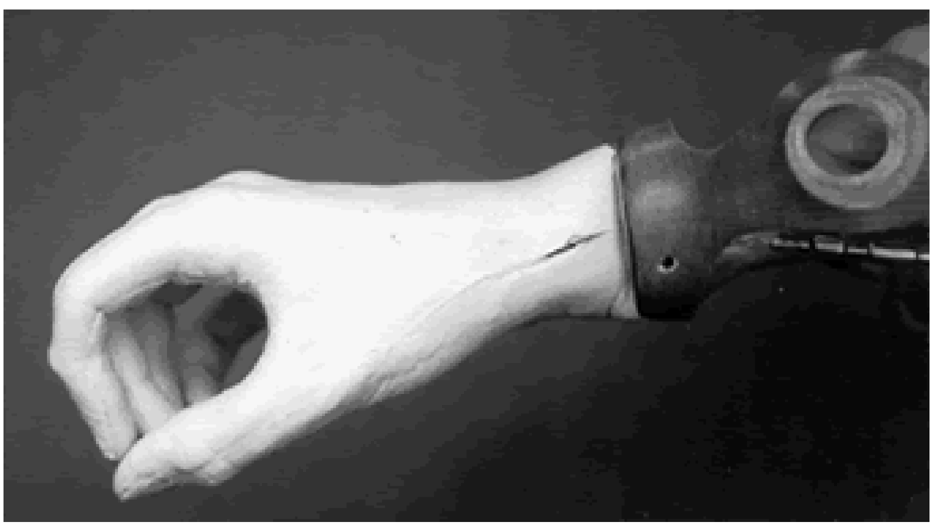

Figure 2. Displacement of the harness cable causes flexion of the four fingers.

By manipulating the distance between cable and hinge at each joint within the fingers, the magnitude of the lever arm at each joint is controlled. By progressively decreasing the lever arms proceeding distally, finger flexion begins proximally and continues distally. This design allows the fingers to sweep the largest arc possible, maximizing grasping abilities. Equally important, this design offers realistic finger flexion. The fingers return to the extended position, due to the elasticity of the surrounding foam (**Figure 3**).

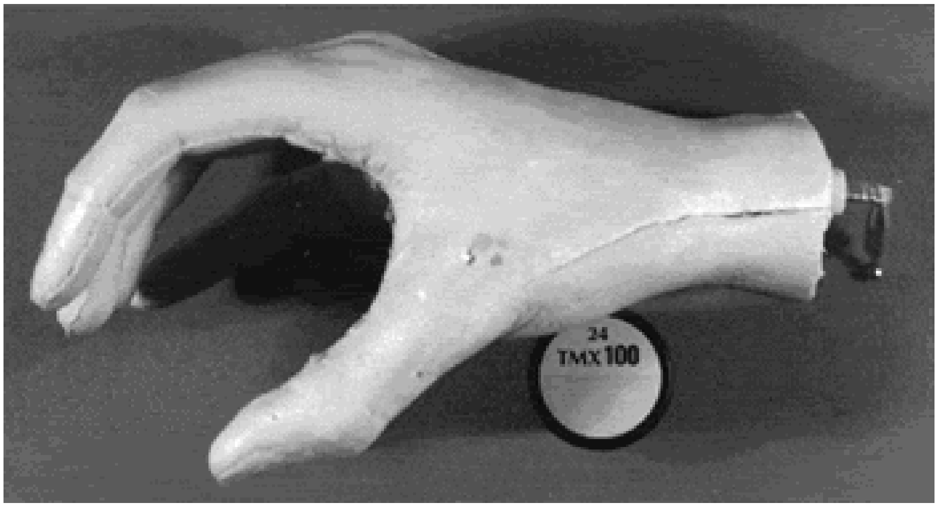

Figure 3. When the harness cable is released, the fingers return to an extended position due to the elasticity of the surrounding foam.

The passive thumb consists of a thick copper armature surrounded by three rigid plastic tubes that simulate the metacarpal, proximal, and distal phalanges. These rigid tubes restrict bending to the joints, ensuring the thumb maintains an appropriate shape. The thumb can be positioned depending on whether a cylindrical power, precise, or other grip is required. The gauge of the armature has been chosen to provide adequate resistance during grasping, while not making positioning the thumb too difficult.

Foam

Soft polyurethane foam surrounds the endoskeletal fingers and thumb. The foam opposes finger motion less than traditional gloves, potentially improving efficiency. Through the use of plasticizers, desired softness can be achieved and opposition to finger motion minimized. Extra foam has been placed within the thenar and hypothenar eminences and within the finger pads to simulate the feel of the human hand. The foam also conforms to the shape of the grasped object, increasing the area of contact.

The foam is self-skinning, forming a skin on the surface during polymerization within the mold. The skin offers good detail, including palm wrinkles and fingerprints (**Figure 4**). The foam is inert, nontoxic, resistant to abrasion, and treatable with flame retardant. Pigments can be added to achieve various skin shades. Since the foam is lightweight, prosthesis weight is reduced.

Figure 4. The skin offers good detail, including palm lines and fingerprints.

Cable and Adapter

Spectra^{TM} (Allied-Signal, Inc., Petersburg, VA), a polyethylene fiber used in stunt kite flying, is used to actuate the fingers. Carlson et al. (12) had previously detailed its potential applications in upper-limb prostheses. Spectra has adequate tensile strength and, when used in a teflon liner, offers longer fatigue life than 1/16 in steel cable. Thinner and more flexible than steel, Spectra can be used within the endoskeletal fingers that assume tight angles of bending during flexion. The low coefficient of friction of Spectra, particularly in conjunction with its teflon liner, minimizes frictional losses in the internal mechanisms.

The new prosthesis uses a standard male 1/2-in, 20-pitch attachment to the wrist unit. The harness cable runs through the centerline of this fixture. This routing allows rotation of the hand about the fixture axis without displacing the harness cable and moving the fingers.

RESULTS

Most of the design objectives outlined earlier have been met. The new endoskeletal prosthesis features flexion of four fingers against a passive thumb. Graded prehension is offered, allowing the user to vary grip force depending on the object being grasped. The traditional external glove has been replaced by self-skinning foam, potentially reducing maintenance requirements and increasing hand softness.

Finger Movement

Primarily because of the placement of three joints (MCP, PIP, and DIP) within each of four fingers, finger movement appears more realistic than in current prosthetic hands. A survey of 17 nonimpaired adult controls compared finger movement in the endoskeletal hand to that in voluntary-closing hands made by Otto Bock (Stock No. 8K15. Otto Bock Orthopedic Industry Inc., Minneapolis, MN) and the Army Prosthetics Research Laboratory (APRL: Stock No. 52540, Hosmer Dorrance Corp., Campbell, CA). Each of the two latter prostheses was covered with a traditional PVC glove. Those surveyed were asked to rate finger movement on a scale of 1 to 10, with 1 being "not lifelike" and 10 being "extremely lifelike." Ratings for each prosthesis were then averaged. Finger movement in the endoskeletal hand was rated markedly higher than in the Otto Bock and APRL hands (6.7 versus 4.0 and 5.4, respectively).

Foam Properties

By varying polyurethane density and quantities of plasticizer and pigment, acceptable skin thickness, foam softness, and color have been achieved. The foam is sufficiently firm to resist abrasion, yet soft enough to feel realistic and not excessively oppose finger movement. Because the endoskeletal hand prototype has not been tested by persons with amputation for extended periods, long-term durability and maintenance requirements are unknown. However, it is the opinion of the authors that the foam is sufficiently robust to withstand daily wear and tear without requiring frequent maintenance.

Grip Variability

To assess grasping ability, a specially designed demonstration prosthesis has been used to allow a control to operate the endoskeletal hand (**Figure 5**).

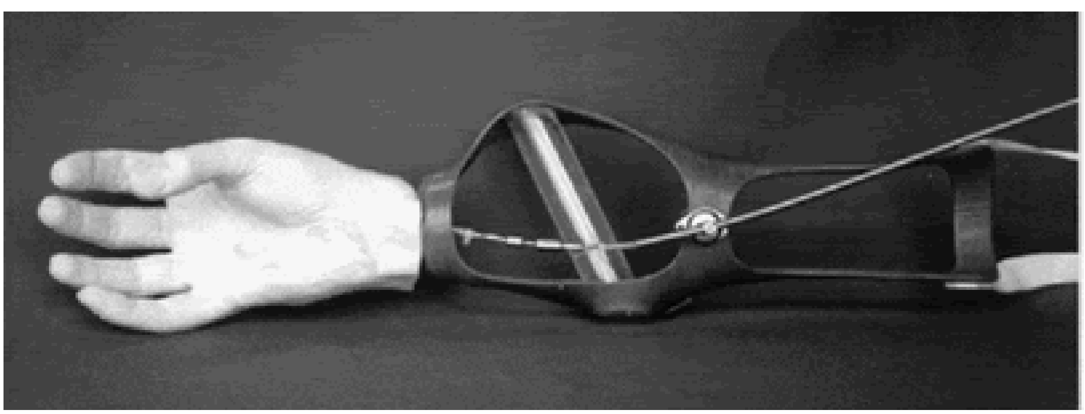

Figure 5. A specially designed demonstration prosthesis allows a non-amputee to operate a prosthetic hand. Biscapular abduction and/or glenohumeral flexion provide the requisite cable displacement.

The endoskeleton hand has been shown to grasp large and small objects in a variety of shapes. Positioning of the thumb by the user provides both precision and power grips (**Figures 6-8**).

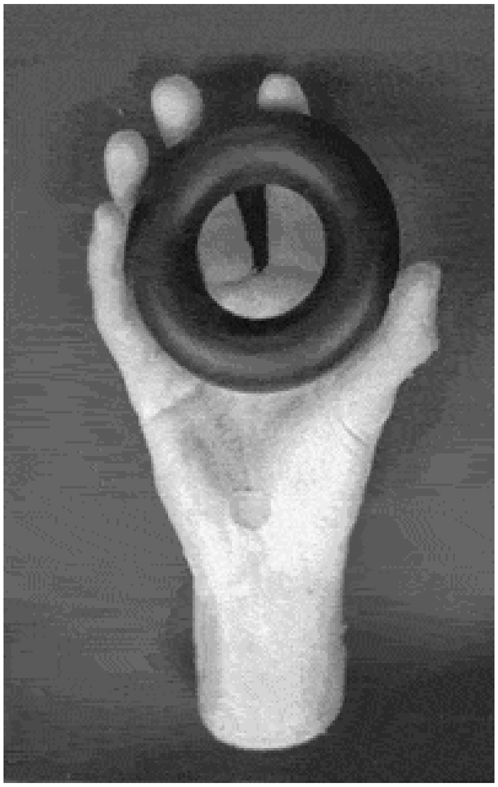

Figure 6. The thumb can be positioned as needed to grasp objects with different sizes and shapes.

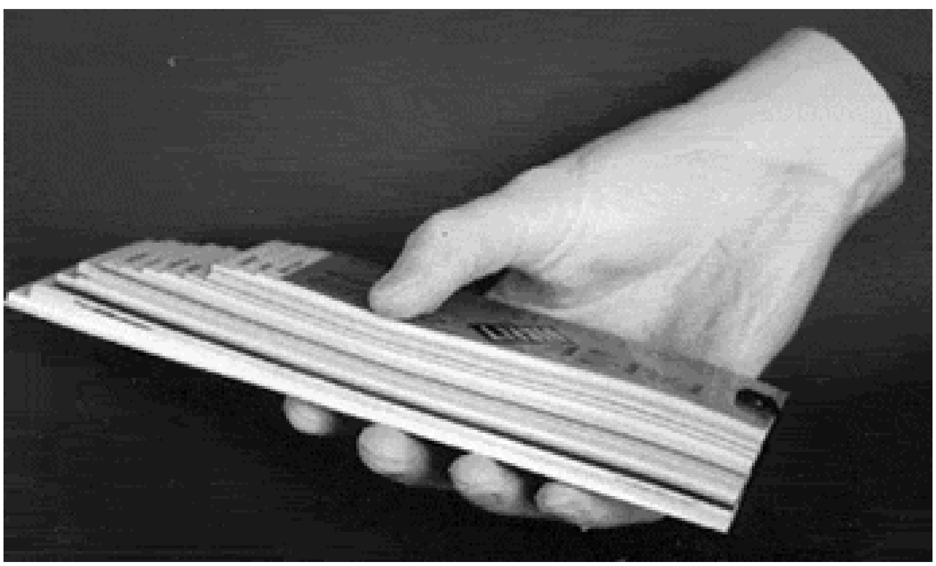

Figure 7. The thumb can be positioned inward to grasp smaller objects. This is an example of a precise grip.

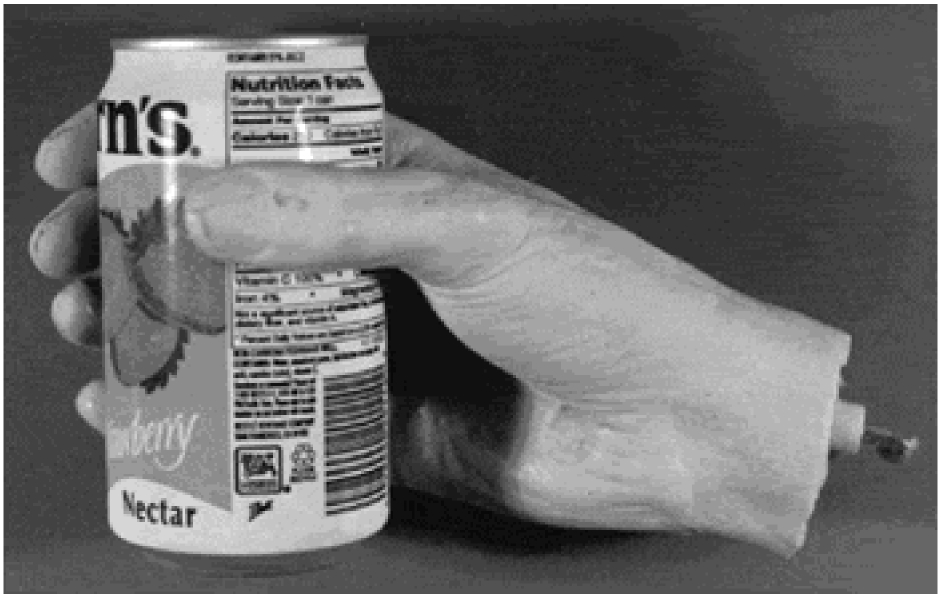

Figure 8. The thumb can be positioned outward to grasp larger objects. This is an example of a power grip.

Weight

At 203 g, the endoskeletal prosthesis weighs roughly half as much as the Otto Bock (390 g) and APRL (421 g) hands (**Figure 9**). Current hooks made by Hosmer Dorrance including the aluminum model 5XA and stainless steel model 5X weigh 113 and 213 g, respectively.

Efficiency

A major objective involved improving prosthesis efficiency. Most current body-powered prosthetic hands have work efficiencies of 20-30 percent (11,15). Increased efficiency potentially reduces required force generation via the harness. The technique for measuring work efficiency used by LeBlanc et al. (15) proved unsuccessful when applied to the endoskeletal prosthesis. The technique uses force versus excursion curves to calculate work efficiency. However, the independent movements of each of the four fingers, coupled with the inherent nonreproducibility of the arcs traced by them (due to multiple joints in each finger) prevent use of this testing method.

Force Transmission

Force transmission ratios were calculated for the endoskeletal, Otto Bock, and APRL hands. Testing involved measuring pinch forces at the fingertips given a constant force applied to the harness cable. The endoskeletal hand offered a force transmission ratio of approximately 1:6, approximately 30-40 percent less than the other hands. A higher force transmission ratio is preferred, since a lower harness cable force is required to achieve a desired grip force.

Grasping Test Objects

Input forces required to grasp everyday objects were measured for each of the hands (**Figure 9**). Objects included a pen, small book, coin, newspaper, telephone handset, notebook, film canister, and tennis ball. On average, the endoskeletal hand required 33 N to pick up each object. The Otto Bock and APRL prostheses required an average of 37 N and 41 N respectively. Thus, to grasp these objects, the endoskeletal hand required approximately 12 percent less input force than the

Otto Bock, and 24 percent less than the APRL hand. These results are slightly surprising, given the relatively low force transmission ratio of the endoskeletal hand.

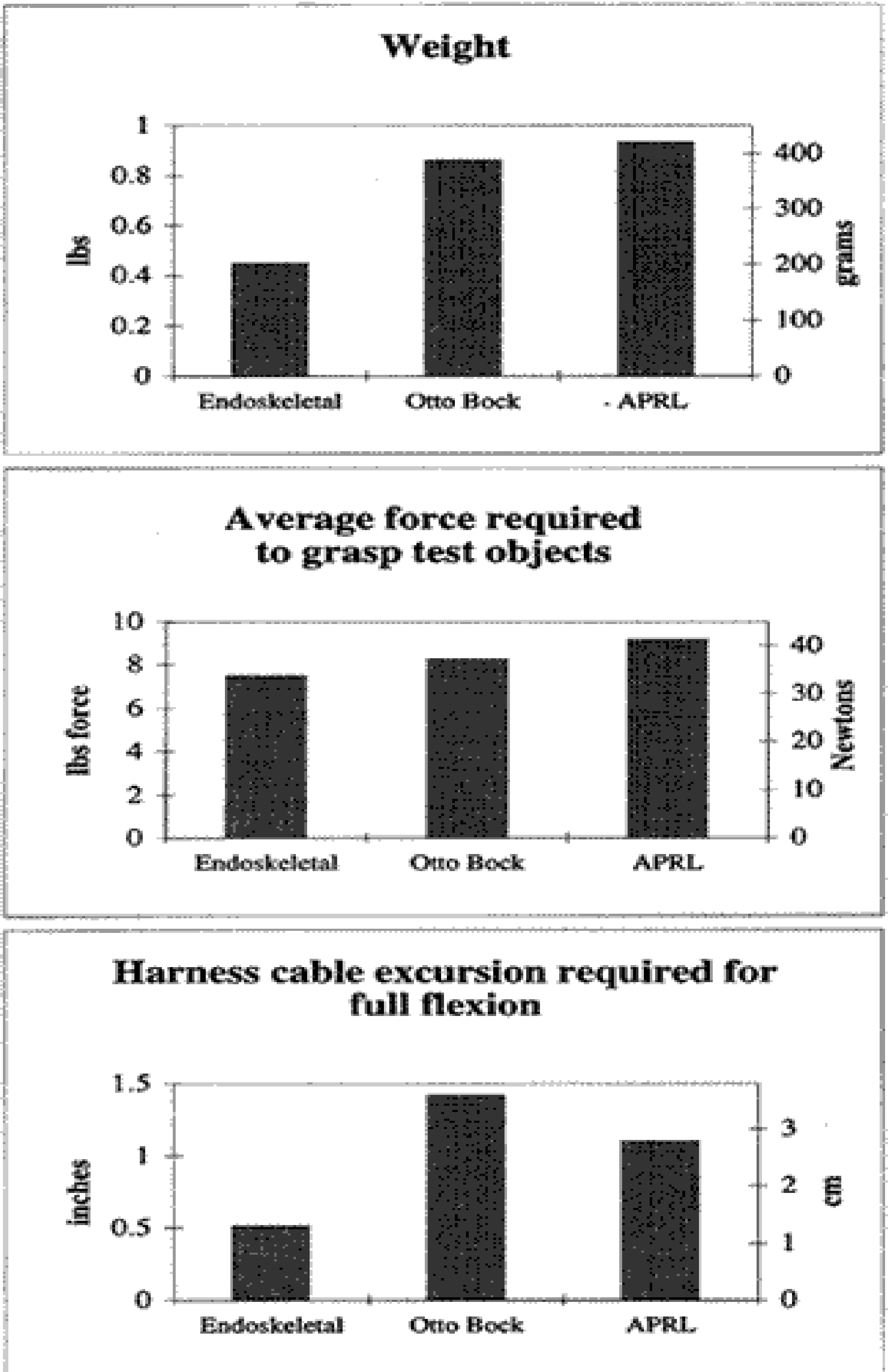

Figure 9. The endoskeletal prosthetic hand was compared with Otto Bock and APRL hands in the areas of weight, average cable force required to grasp a variety of everyday objects, and required

harness cable excursion to effect full finger flexion.

Cable Excursion

The cable excursion required to effect full finger flexion was calculated by measuring the difference in harness cable length at full flexion and full extension (**Figure 9**). The endoskeletal hand has the lowest required excursion (1.3 cm), indicating that the user has to displace less cable to fully flex the fingers than with the Otto-Bock (3.6 cm) and APRL hands (2.8 cm).

Estimated Cost

A preliminary cost analysis was conducted for the endoskeletal hand. For a production run of 1,000 units, the estimated cost of production (material and labor) is $50 per hand. Compared to current hands that cost several hundred dollars each to produce, the endoskeletal hand is much less expensive, mainly because less skilled labor is required.

DISCUSSION

Our results to date have been encouraging. The new endoskeletal prosthesis has satisfactory appearance and feel. Finger movement appears realistic and input forces required to grasp common objects are decreased compared to current prostheses. The new prosthesis is approximately 50 percent lighter than current hands, which may significantly improve comfort. Additionally, the prosthesis is inexpensive to produce.

Perhaps the most important feature of the new prosthesis is its realistic finger movement. By bending at the MCP, PIP, and DIP joints in each of four fingers, the endoskeletal prosthesis offers an adaptive grip that conforms to the shape of the grasped object (**Figure 10**). Compared to current hands, there is greater area of contact with the object, reducing required pinch forces. This increased area of contact explains why the endoskeletal prosthesis, despite its lower force transmission ratio, requires less force input to grasp objects than the Otto Bock and APRL prosthetic hands.

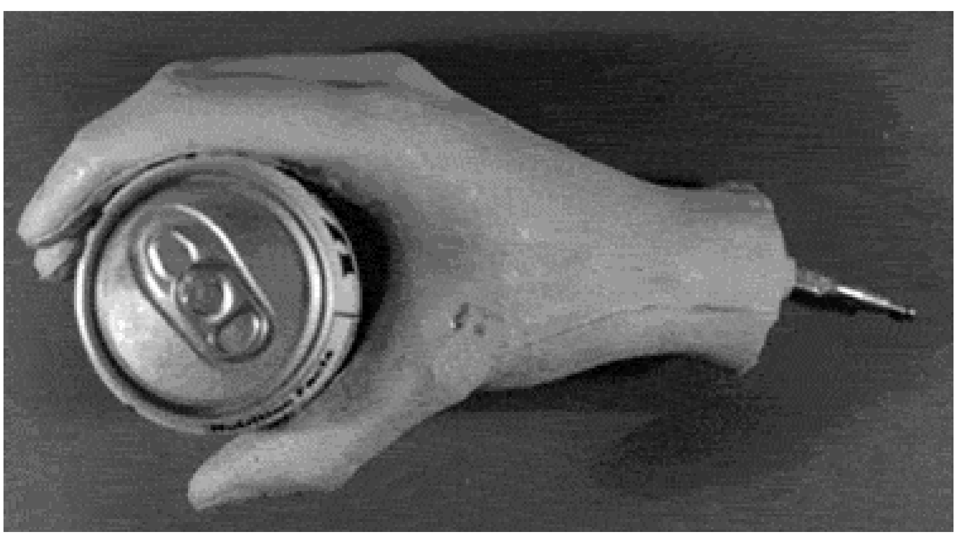

Figure 10. Bending at three joints in each finger provides an adaptive grip with large contact area. This increased contact area decreases the required harness forces provided by the amputee. Energy expenditure is therefore potentially reduced.

Future Improvements

Despite the initially promising results, there is still room for improvement. First, force transmission could be improved. If the DIP joints were stationary, force transmission would increase, resulting in greater pinch forces available at the fingertips. This change would add function without dramatically detracting from finger movement realism.

Cable actuation could also be improved. Since harness cable excursion required to effect full flexion is only 1.3 cm, it is possible to use mechanical advantage to double the force available at the fingertips while also doubling the required input cable excursion. This increased cable excursion would offer the user more fine control over finger flexion.

Cosmetic improvements in the foam are also required. Several seams are apparent along the sides of the fingers where the two halves of the mold meet. Improving the mold in which the foam is cast would correct these flaws.

Inexpensive Passive Hand

A passive version of the endoskeletal hand has also been constructed. Surrounded by self-skinning foam, the endoskeleton consists of a copper armature with plastic phalanges placed at appropriate locations. The plastic phalanges restrict finger movement to the joint areas, ensuring the fingers maintain appropriate shapes. As in the active hand, significant detail can be achieved, including fine wrinkles and fingerprints. Weight is approximately equal to that of the active hand. This hand is expected to be an extremely inexpensive option for those interested in passive hand prostheses with good cosmetic characteristics. With continued health care reforms aimed at cost containment, there will likely be a sizable market for both the passive and active prosthetic hands in coming years.

CONCLUSIONS

There are multiple potential applications for a low-cost endoskeletal hand prosthesis. This hand could be manufactured and distributed in developing countries, where persons with amputation who could not otherwise afford a prosthesis may find it a welcome option. Additionally, the prosthesis could be used by growing children who require a larger size every 6 months or year: it could be cheaply replaced. Adults could also use this prosthesis as a disposable: if torn or stained, it can be replaced inexpensively.

ACKNOWLEDGMENTS

We would like to thank Stanford University mechanical engineering students Ajit Chaudhari, Michael Clarke, and Justin Davis for their contributions during the preliminary stages of this project. We would also like to thank Dr. Robert Chase, Stanford University Medical School Emeritus Professor of Surgery for serving as a secondary advisor for the project. Additional thanks to Michael Gianandrea, owner of MVG Molded Products, Petaluma, CA, for his expertise in mold making and foam casting.

REFERENCES

1. Wilson B. History of amputation surgery and prosthetics. In: Bowker JH, Michael JW, editors. Atlas of limb prosthetics: surgical, prosthetic, and rehabilitation principles. St. Louis: Mosby Year Book; 1992. p. 3-15.
2. Melendez D, LeBlanc M. Survey of arm amputees not wearing prostheses: implications for research and service. J Assoc Child Prosthet Orthot Clin 1988:23:62-9.
3. LeBlanc M. Use of prosthetic prehensors. Prosthet Orthot Int 1988;12:152-4.
4. LeBlanc M, Bechtol CO. Endoskeletal upper extremity prosthesis. University of California at Los Angeles Prosthetics-Orthotics Program Feasibility Study; 1968.
5. Fletcher MJ, Leonard F. The principles of artificial-hand design. Artif Limb 1955;2:78-94.
6. Donovan WH. The next generation myoelectric prosthesis. Houston, TX: The Institute for Rehabilitation and Research; 1994.
7. Gwynne G. Mechanical components. In: Santschi WR, editor. Manual of upper extremity prosthetics. Los Angeles: University of California at Los Angeles; 1958. p. 33-68.
8. LeBlanc M, Yeh C, Doshi R. Development of an endoskeletal hand with skin and no glove. Proceedings of the 23rd Annual AAOP Meeting & Scientific Symposium; 1997 Mar 12-15, San Francisco; 1997.
9. Bilotto S. Upper extremity cosmetic gloves. Clin Prosthet Orthot 1986;10:87-9.
10. Schurr DG, Cook TM. Prosthetics & orthotics. Norwalk, CN: Appleton & Lange; 1990.
11. Corin JD, Holley TM, Hasler RA, Ashman RB. Mechanical comparison of terminal devices. Clin Prosthet Orthot 1987;11:235-44.
12. Carlson LE, Radocy B, Marschall PD. Spectron 12 cable for upper-limb prostheses. J Prosthet Orthot 1991;3:130-41.
13. Fryer CM, Michael J. Upper limb prosthetics: body-powered components. In: Bowker JH, Michael JW, editors. Atlas of limb prosthetics: surgical, prosthetic, and rehabilitation principles. St. Louis: Mosby Year Book; 1992. p. 107-31.
14. Taylor C. The biomechanics of the normal and of the amputated upper extremity. In: Klopsteg P, Wilson P, editors. Human limbs and their substitutes. New York: McGraw-Hill; 1954. p. 169-221.
15. LeBlanc M, Setoguchi Y, Shaperman J, Carlson L. Mechanical work efficiencies of body-powered prehensors for young children. J Assoc Child Prosthet Orthot Clin 1992;27:70-5.

Submitted for publication July 2, 1997. Accepted in revised form August 26, 1997.

Last revised Thu 01/21/1999