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Design of a Multifunctional Anthropomorphic Prosthetic Hand with Extrinsic Actuation

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***Abstract*—This paper presents an anthropomorphic prototype hand prosthesis that is intended for use with a multiple channel myoelectric interface. The hand contains 16 joints, which are differentially driven by a set of five independent actuators. The hand prototype was designed with the minimum number of independent actuators required to provide a set of eight canonical hand postures. This paper describes the design of the prosthesis prototype, demonstrates the hand in the desired eight canonical postures, and experimentally characterizes the force and speed capability of the device. A video is included in the supplemental material that also illustrates the functionality and performance of the hand.**

I. Introduction

A highly constraining factor in the development of upper extremity prostheses has traditionally been the limited number of communication channels with which the user can control the prosthesis. Specifically, whether the input is a body-powered cable or a myoelectric signal, upper extremity prostheses have typically been limited to a single control input. In the case of a transradial prosthesis, the single control input is typically used to open and close a single degree-of-freedom (DoF) hand. In the case of a transhumeral prosthesis, the single control input is switched between control of the hand and control of the elbow joint. Recent advances in neural interface technology bring to the horizon the potential to significantly increase the number of (electrical) communication channels available to control an upper extremity prosthesis. Specifically, the approach described in [1-2], which utilizes "targeted reinnervation" of peripheral nerves into residual muscle sites, has been shown to provide two channels of myoelectric information on several transhumeral amputees [2], and the approach has the potential in the near future to provide several additional channels of myoelectric information at all typical levels of upper extremity amputation. Given the availability of several additional myoelectric control channels, an upper extremity prosthesis with several independent actuators can be developed to leverage increased functionality with the availability of additional inputs. However, an increased number of control channels generally requires increased complexity, cost, and mass in the prosthesis, increased complexity (and possibly cost) of the surgical intervention, and increased complexity (and most likely cost) in the electrode interface. Therefore, in the development of an appropriate multifunctional prosthesis, one must balance enhanced performance with the consequent increases in cost and complexity that it may imply.

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The prosthetic hand described in this paper and shown in Fig. 1 was designed in an effort to strike this balance. Specifically, the prosthesis was designed to provide the principal functions of a hand through eight canonical grasps (as subsequently defined by the authors) with the fewest number of independent control inputs. Further, it was assumed (and is the case with myoelectric devices) that no conduit exists to bring sensory information to the user. Because of this the device was designed to be force controlled, such that the information contained in the muscle contraction is in essence preserved in the prosthesis. Finally, the device was designed to provide useful levels of force and appropriate speeds of motion relative to the native hand and possesses an anthropomorphic skeletal structure to promote natural function and appearance.

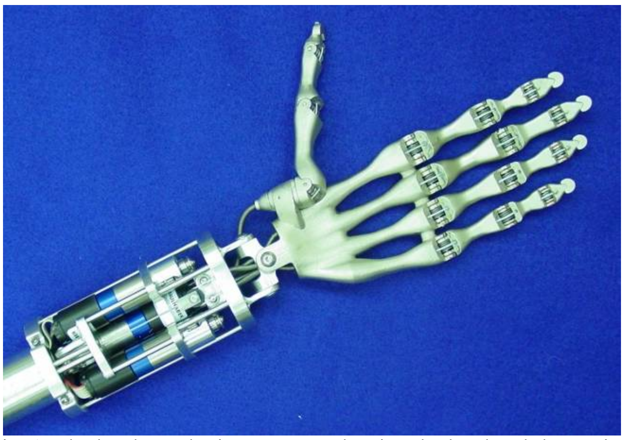

Fig. 1. The hand prosthesis prototype, showing the hand and the extrinsic actuation unit.

Several prosthetic and robotic hands have recently been described in the engineering literature [3-10]. These hands contain between one and six independent actuators and between eight and sixteen joints distributed in various ways in the digits (thumb, index, etc.). In each device, the discrepancy between the number of independent actuators and the number of joints is accommodated either by differential drives, which prescribe a given torque distribution between joints, kinematic linkages, which prescribe a given relative motion between joints, variable compliance couplings, which prescribe given relative compliance between joints, or by a combination of these. Of these devices, the most highly underactuated is the hand described by [3], which is composed of 15 joints and 5 digits, which are differentially driven via a single independent actuator (and as such could be used with current myoelectric interfaces). The Oxford and Manus hands described in [4] contain eight and nine joints, respectively, which are kinematically linked within three (active) digits (two of which are coupled) and driven by two actuators. The HIT/DLR hand described by [5] includes 13 joints, which are kinematically linked within the five digits (three of which are coupled), driven by three actuators either directly or through a combination of differential and compliant transmissions. The Cyber Hand and Smart Hand described by [6] and [7] contain 10 and 16 joints, distributed over 3 and 5 compliantly linked digits, respectively. Each of these devices is driven by four independent actuators either directly (as in [6]) or by a combination of differential and compliant transmissions (as in [7]). The fluidic hand described by [8] utilizes five independent commands to actuate eight joints via a miniature hydraulic actuation system. The prosthetic hand described in [9] incorporates 11 joints within five kinematically linked digits, driven by six independent actuators. Finally, the Southampton hand described in [10] incorporates 16 joints distributed over 5 kinematically linked digits, which are driven by six independent actuators.

This paper describes an anthropomorphic prosthetic hand designed to fully accommodate a set of grasp and gesture taxonomies with the minimum number of independent actuators. Unlike any of the aforementioned devices, this prosthetic hand possesses five independent actuators and 16 joints, with the discrepancy between the number of joints and number of actuators accommodated by moment isotropy between the joints. Moment isotropy is realized by a combination of tendons spanning multiple joints and a differential pulley mechanism which distributes forces across tendons. This enables fully conformal, highly stable grasping [7] with significantly fewer independent actuators than joints in the hand. Additionally, the hand was designed with the assumption that near-term myoelectric interfaces will not be capable of providing graded, multi-degree-of-freedom force sensing to the user. Because of this the device was designed to preserve open-loop force information (i.e., to preserve the force command from the muscles providing the myoelectric signals) by virtue of force-control and backdrivability. In this manner the force sensing within the user's own muscle can be proportionally mapped to the control of the device. Such an approach provides the possibility to provide to the user some degree of force information, or feedback, with the interface, hardware, and control methods described herein rather than through the use of additional instrumentation.

II. Design Objectives

The essential objective of this work is to develop and characterize an anthropomorphic hand prosthesis for use with a multiple channel myoelectric interface (i.e., with multiple efferent communication channels from the user and no afferent communication) such as that enabled by targeted muscle reinnervation surgical techniques. Because additional channels incur liabilities with respect to cost, complexity, and reliability, the objective is further to obtain the desired functionality with a minimum number of independent actuators through the use of differential drives and underactuated digits. The desired functionality, in terms of varying postures, is to achieve the following set of eight canonical hand postures: a pointing posture (e.g., for punching buttons or keys); lateral and tip grasps between the thumb and forefinger (for grasping small objects); a tripod grasp between the thumb, forefinger, and middle finger (for more stable grasping of small objects); a hook grasp (e.g., for carrying a briefcase); cylinder and spherical whole hand grasps (i.e., for whole-hand grasping of objects); and a platform posture (e.g., for holding a book or plate).

The design objectives have also been influenced by the nature of myoelectric interfaces which do not provide multi-axis force feedback. As reported in [2], targeted motor reinnervation is generally accompanied by some degree of sensory reinnervation (and therefore feedback); but does not directly accommodate force sensing and requires additional device complexity to relay the appropriate sensory information. In the proposed approach, sensory information is not provided to the user, per se, but rather is retained by the actuation method. Specifically, the user's musculature itself (from which the myoelectric command signals are measured) possesses force sensing capabilities. By incorporating force-controlled, backdrivable actuation, force sensing is in essence preserved because the muscle contraction sensed by the user concurrently generates the input for the prosthetic device through electromyogram signals. Backdrivability then allows for a proportional mapping of this input to device output. In this way force feedback is provided to the user. Thus, an additional design objective is to ensure fully backdrivable, force-controlled actuation.

Note that surveys of prosthetic users [8, 11-13] offer the following as important factors in the performance of a prosthetic hand (relative to commercially available prosthetic devices):

1) Increased functionality
2) Natural interaction with the environment
3) Reduced weight
4) Higher grasping speeds and forces
5) Low Noise
6) Better cosmetic appearance

These factors have been incorporated as design criteria in the prosthetic hand in a variety of ways.

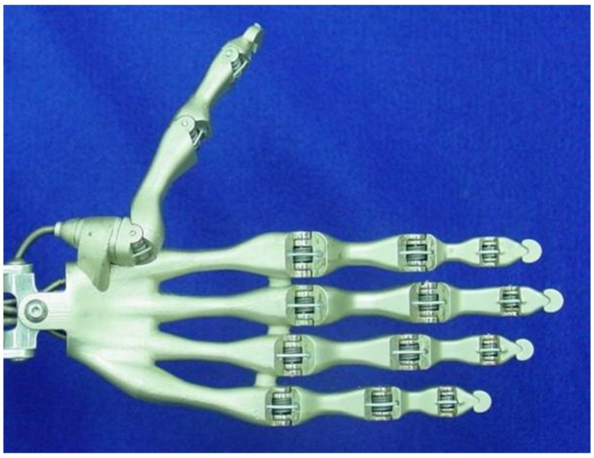

Fig. 2. The 16-joint anthropomorphic hand.

The first two criteria are accommodated in the anthropomorphic hand by the previously discussed design objectives. Specifically, increased functionality is attained through the achievement of eight canonical grasps and natural interaction is facilitated via implicit force sensing and anthropomorphic design. The eight grasps described here represent a grasp taxonomy which considerably expands upon the (1 DoF) state of the art. Backdrivable, force controlled actuation enables implicit force sensing (i.e., by preserving native force sensing in the muscles) and allows both the generation and absorption of power. This in turn allows smooth and natural interaction with high impedance kinematic constraints (i.e. stiff objects in the environment) [14]. Finally, by utilizing a scalable, anthropomorphic design, it is believed that the prosthesis will be better suited to interact naturally with the environments associated with the activities of daily living that are designed for the human hand (e.g., doorknobs, steering wheels, dinnerware, etc.).

In addition to utilizing underactuation to reduce weight, the third criterion is addressed by utilizing hollow structural elements in the hand, visible in Figs. 4 and 5, and a space-frame to house the actuation units in the forearm, visible in Figs. 9 and 10. The third and fourth criteria, low weight and higher grasping speeds and forces, are generally competing objectives although both, in addition to the fifth criterion (low noise), are served by ensuring an efficient transmission (i.e., minimizing friction in the system). Finally, the sixth criterion is achievable with conscientious design. In the case of the hand presented here, an anthropomorphic design based on the skeletal characteristics of the human hand has been utilized. In doing this it is believed that the appearance of the device (in addition to its interaction with objects) will be as natural as possible, particularly when covered with a life-like outer layer as discussed below.

The remainder of this paper describes the hand design, demonstrates the ability to provide the aforementioned canonical postures with the prosthesis, and experimentally demonstrates the force and speed characteristics of the hand in operation.

III. Hand Design

The anthropomorphic hand, shown in Fig. 2, has 16 joints which are driven by five independent actuators. The tendon-based actuation units for the hand reside in the forearm, similar to the native human anatomy. The actuation units, which are described in the following section, use DC motors coupled with low-ratio gearheads and small diameter pulleys to pull hand tendons. Each joint in the hand includes embedded torsional springs as discussed subsequently. The five actuators are allotted to the 16 joints in the hand as described in Table I.

TABLE I
DISTRIBUTION OF ACTUATION AND MECHANISM OF COUPLING IN THE HAND.

Actuator	DoF's	Coupling	Mechanism of Coupling
Forefinger Flexion	3	Same Moment Across All Joints	Tendon Spans Multiple Joints
Index finger Flexion	3	Same Moment Across All Joints	Tendon Spans Multiple Joints
Ring & Little Finger Flexion	6	Same Moment Across All Joints	Two Tendons Coupled by Pulley, each Tendon Spans Multiple Joints
Thumb Flexion	3	Same Moment Across All Joints	Tendon Spans Multiple Joints
Thumb Opposition	1	Direct Drive	Direct Drive

In all cases, the underactuation is governed by moment isotropy (i.e., differential coupling), rather than by kinematic constraints. In other words, the hand will reach a configurational equilibrium when all joint moments are (essentially) equal, excepting tendon friction and nonlinearities in the relationship between tendon force and joint moment as a function of joint angle. Note that this is achieved by a combination of having the tendon span multiple joints and using a pulley differential to split the force of the actuator output equally into two tendons, as briefly described in Table I.

Each joint in the hand incorporates embedded torsional springs, as shown in Fig. 3. The use of torsional springs in the joints serves several important purposes. First, and perhaps most obviously, the compliance provides a return force, which simplifies tendon actuation of the joints, since active extension of the joints via an extensive tendon is not necessary. Second, the compliant joints map joint motion to tendon force in free space, thus 1) eliminating the need for position sensing in the hand and 2) eliminating the need to switch between motion control and force control. Specifically, the hand frequently engages in both motion control (e.g., when gesturing or reaching) and force control (e.g., when grasping or squeezing). In the hand, the tendons are always under (open-loop) force control. When the fingers are not in contact with an object, the springs in the fingers map the tendon force to finger position, such that the fingers are in effect under position control. When the fingers come in contact with a rigid object, the force-controlled tendons map directly to force control, and thus the switching between position and force control is natural and seamless.

Fig. 3. Exploded view of a finger.

To achieve a high strength to weight ratio, the skeletal/structural components of the hand were designed in CAD as hollow, one-piece (monocoque) components, as shown in Figs. 4 and 5. The CAD files were then used to directly form parts out of a high-strength thermoplastic coated with copper and nickel using a laser direct manufacturing method. The resulting structure weighs less than 80 grams, with springs and connecting pins adding an additional 20g. This high strength, low weight, and low cost approach to constructing each hand, in conjunction with a parametric code-based approach to CAD modeling, facilitates tailoring the dimensions of each hand to the individual user.

Fig. 4. Cross-section of a proximal phalanx, showing the hollow structure, the structural interface for the torsional spring, and the tunnel that guides the tendon.

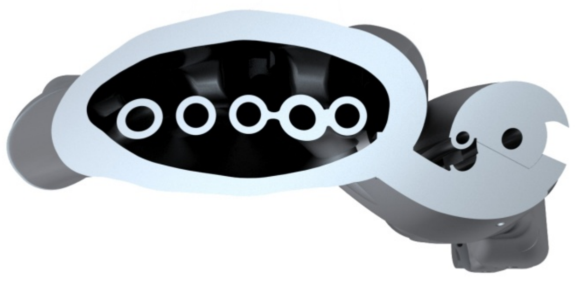

Fig. 5. Cross-section through (base of the) palm of the hand, showing the hollow structure and the suspended tunnels that guide the tendons through the palm. The base of the thumb is shown on the right.

The structure of the prosthetic hand is designed to fit well within the user's expected anatomical envelope by mimicking the dimensions of the skeletal structure of a healthy hand with high anthropometric fidelity. Although not shown here, it is important to note that the prosthesis is designed to be used with a life-like outer layer, which will greatly facilitate grasping in addition to anthropomorphism. Specifically, the artificial skeleton leaves room for mimicking the soft tissue (skin and muscle) components of the natural hand with a skin of silicone rubber wrapped around other soft materials such as a synthetic viscoelastic urethane polymer (to mimic the feel of muscle) and viscoelastic polyurethane foam (to mimic softer tissues). It should also be noted that the approximate stiffness of such a cover was considered in the specification of the actuator requirements (section IV).

As previously discussed, the hand was designed to achieve a set of eight grasp postures, which are the tip, lateral pinch, tripod, spherical, cylindrical, and hook grasps, in addition to the pointing and platform postures. Figure 6 demonstrates the ability of the hand to achieve these grasps and postures. The addition of a soft "tissue" outer layer would presumably further facilitate such grasps. Figure 7 shows some additional grasps and postures representative of the activities of daily living. Note also that a video is included with the supplemental material that illustrates several of the grasps represented in Figs. 6 and 7.

Fig. 6. Eight canonical hand postures, which constitute one of the primary design objectives of the hand prototype.

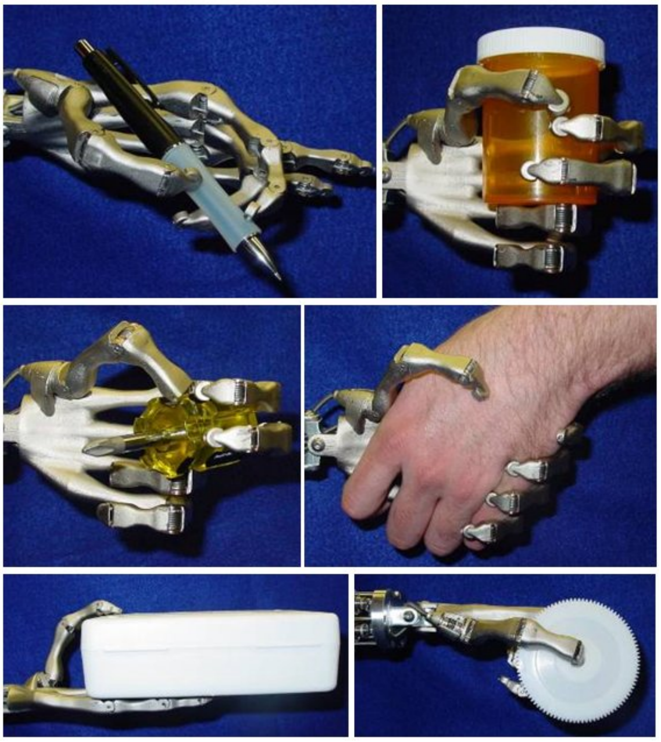

Fig. 7. Additional grasps and postures representative of typical activities of daily living.

These grasps were achieved in real-time using the fingertip forces of a healthy hand for input as measured by force sensitive resistors.

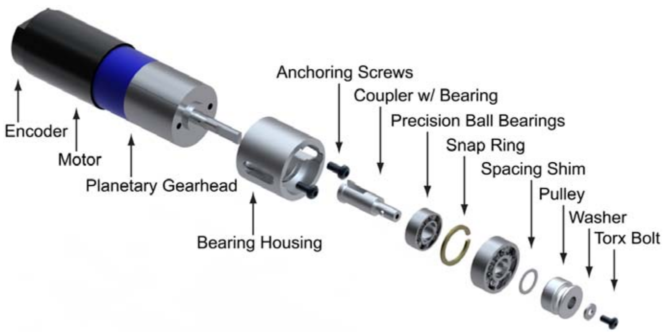

Fig. 8. Exploded view of actuation unit.

IV. Forearm/Actuation Design

The actuation units were designed to provide the fingers with sufficient force and speed to perform activities of daily living and to be as light weight and efficient as possible. Exact specifications of "sufficient" force and speed are difficult to obtain. Based on typical capability of the native hand the authors utilized a continuous finger tip force of 20 N and a finger motion bandwidth of 2 Hz as force and speed design objectives. Given the nominal geometry of each finger and the respective tendon paths, a 20 N finger tip force requires a tendon force of approximately 120 N and a peak tendon excursion speed of approximately 15 cm/sec. In order to achieve these specifications, the actuation units shown in Fig. 8 were developed. Each unit is composed of a brushed DC servomotor (Faulhaber 1724 SR12) with integrated magnetic encoder (Faulhaber 1E2 516) and planetary gearhead (Faulhaber 16/7 43:1), coupled to a bearing housing across the structural plates of the extrinsic actuation housing. Within the bearing housing are two miniature, high precision (ABEC 7) radial bearings (New Hampshire Ball Bearing SSR3 & SSRI614). A coupling shaft cantilevered within the bores of these bearings transmits torque from the motor to the pulleys about which the tendon material is wrapped. Note that each unit is fully backdrivable (the return force being provided by the torsion springs embedded in the joints of the digits, Fig. 3) as per the design objectives previously stated.

The actuation units actuate the digits via braided spectra cable (the "tendon") which runs from the motor pulleys, around idler pulleys for redirection, and through the distal-most plate of the forearm (see Fig. 10). From here, the tendons pass through Bowden cables between the forearm and hand (allowing for rotational positioning of the wrist) and then through tunnels in the hand (shown in Figs. 4 and 5). These tendon paths are lined with Teflon tubing to reduce friction. Braided spectra cable of 0.75 mm (0.03 in) diameter and 668 N (150 lbf) rated strength was chosen for tendon material due to its high strength, high tensile fatigue resistance, low creep and low stretch characteristics.

Figures 9 and 10 show the five actuation units situated within the forearm. An open space-frame structure consisting of four plates rigidly connected by stainless steel tubing houses the actuators and is visible in Figs. 9 and 10. The space-frame is scalable, lightweight and provides easy access to the actuation units; allowing for quick maintenance and modification.

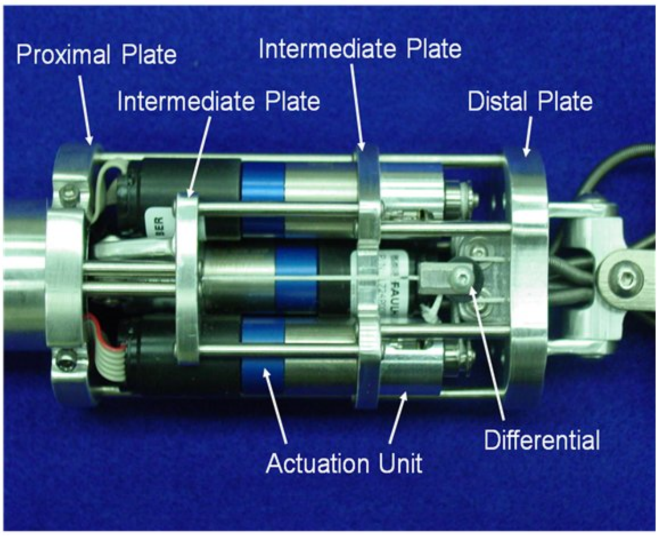

Fig. 9. Anterior view of the actuation unit housing showing structural plates, actuation units, and differential mechanisms.

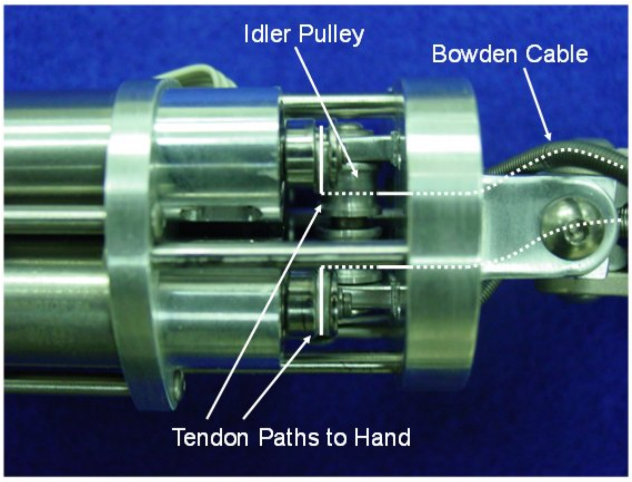

Fig. 10. Close-up lateral view of the actuation unit housing showing idler pulleys, Bowden cables, and tendon pathways.

V. PERFORMANCE

Figures 6 and 7, in addition to the video provided in the supplemental material, illustrate the capability of the hand prosthesis to achieve the desired grasps and postures but do not provide quantitative information regarding the hand's force and speed capabilities. The dynamic performance of the hand prosthesis was characterized quantitatively in terms of closed-loop position tracking, open-loop force tracking, and maximum fingertip normal force as a function of position (tendon excursion) for the index finger.

A. Closed-loop position tracking

The position tracking capability of the prosthesis, shown in Fig.11, was investigated by commanding the position of the index finger around the mid-flexion point (corresponding to 50% of total tendon excursion). Specifically, the index finger tracked a sinusoidal position signal which varied around the mid-flexion point by ±25% of total tendon excursion for various frequencies. The bandwidth was determined using the integrated encoders to send position information to MATLAB software by way of a Humusoft 624 DAQ card. The results indicate a bandwidth of 4.5 Hz in position tracking. This bandwidth is representative of similar amplitude motion in the native hand and surpasses the design objective of 2 Hz.

B. Open-loop force tracking

The force tracking capability of the prosthesis was characterized by commanding a positive 0.5A peak-to-peak amplitude sinusoidal current (current being linearly related to tendon force by the motor torque constant and pulley diameter) through the motor at various frequencies using open-loop force (i.e., current) control. The experimental setup consisted of a load cell (Measurement Specialties model ELFM-T2E-025L) connected to MATLAB software via an analog signal conditioning circuit and Humusoft 624 DAQ card. Force normal to the index fingertip at the fully open (0% excursion) position was then measured using the calibrated load cell. The results, shown in Fig. 12, indicate a

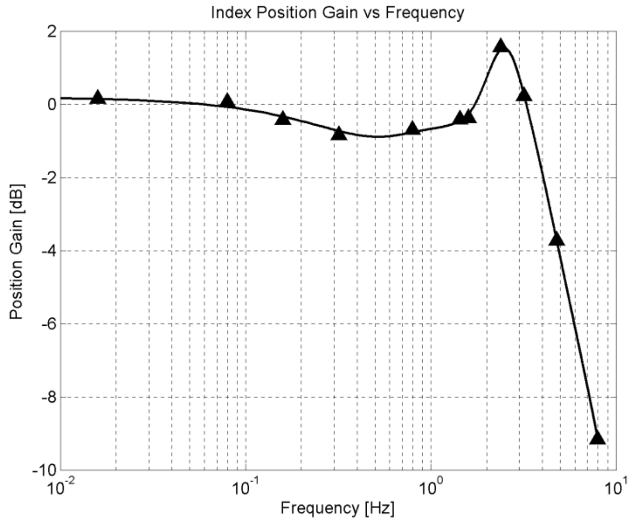

Fig. 11. Index position gain vs. frequency for ±25% tendon excursion about mid-flexion point indicating a bandwidth of 4.5 Hz.

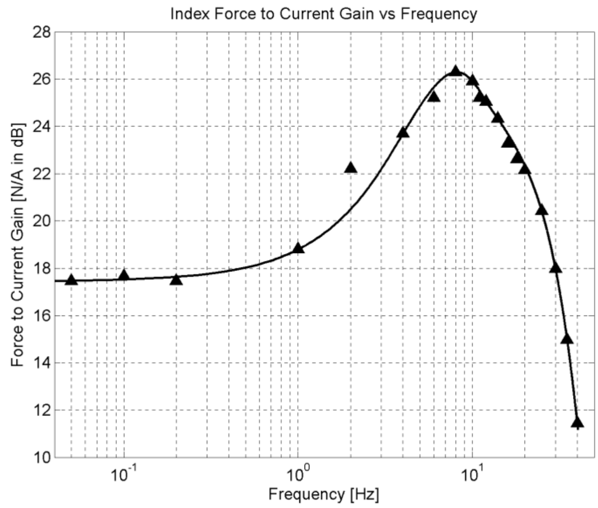

Fig 12. Index Force-to-Current Gain vs. Frequency indicating a bandwidth of 36 Hz and a DC gain of 17.5 N/A in dB (7.5 N/A or 1.7 lbf/A).

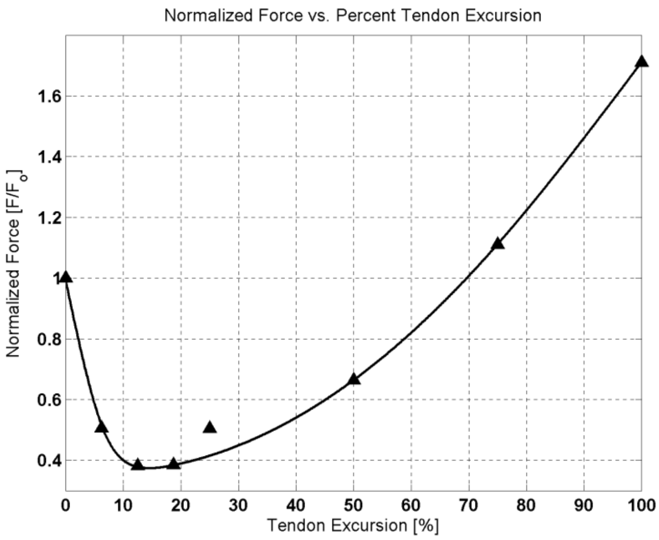

Fig. 13. Normalized Fingertip Normal Force vs. Percent Tendon Excursion

force tracking bandwidth of approximately 36 Hz. This data also shows the DC force to current gain at the open (0% excursion) position as being 17.5 N/A in dB (7.5 N/A or 1.7 lbf/A). Note that these forces are averaged over several cycles and do not represent peak values.

C. Maximum force as a function of position

The maximum fingertip normal force was measured as a function of percent-tendon-excursion for the index finger using an Extech Instruments 475044 tension and compression force gauge rigidly mounted in a variety of positions relative to the palm. The index finger was flexed and held in place for each position to be measured. The force gauge was then mounted with its sensing tip placed adjacent and perpendicular to the fingertip face. With the finger in position, a current of 1A (the maximum) was passed through the motor so that the maximum force could be recorded with a minimum amount of initial travel. This was repeated three times and an average of the forces was taken. These values were then normalized by the initial maximum force at the fully open position, F_o=11.7 N, the results of which are shown in Fig. 13.

The curvature in the data points of Fig. 13 may be explained as follows. At small tendon excursions there is no considerable counteracting force from the springs in the joints of the finger. As tendon displacement increases, the springs provide a (linearly) increasing opposing force, causing the fingertip force to initially drop. Simultaneously, as the finger traverses its range of motion, the tendons act through increasingly greater moment arms (across the finger joints) and by 20% of tendon excursion the maximum fingertip force begins to increase again. The drop in maximum force due to spring resistance is therefore eventually overcome by greater mechanical advantage as tendon excursion, and finger position, increases.

The maximum force is found by multiplying the initial force by the normalized force at full excursion. This results in a maximum force of 19.9N, indicating that the hand can attain the fingertip normal force of 20N (at full flexion) specified in the actuation unit design.

VI. Conclusion and Future Work

This paper describes a prototype prosthesis designed for use with a multiple channel EMG interface. By utilizing underactuation governed by moment isotropy the hand was shown to achieve a complete grasp taxonomy consisting of eight desired hand postures utilizing five independent actuators. The hand is operated in an open-loop force control configuration and the actuation units are backdrivable. This permits force sensation from the user's native muscle to be preserved in the device and facilitates natural motion control while providing some degree of force feedback. This feedback methodology does not require additional instrumentation. The hand is capable of providing a 20 N maximum fingertip force at the forefinger which is also representative of the thumb and middle fingers (the other two digits in general are capable of one half of that force, since the tendon force is split between the two with a pulley differential). The continuous force is somewhat less than the capability of a native hand but likely adequate for most activities of daily living. It was further demonstrated that the hand is capable of tracking forefinger position at 4.5 Hz, which is representative of the native human hand. Finally, the prosthesis design embodies anthropomorphic features and a hollow structure to facilitate user acceptance.

Future work includes the incorporation of an outer layer designed to mimic the soft tissue properties of the natural hand, location of high power density actuation units in the hand (i.e., conversion of the current prototype to an intrinsic hand with comparable speed/force performance), the addition of a two degree-of-freedom wrist, the implementation of embedded power and electronics, and the use of myoelectric signals to control the prosthesis.

VII. References

[1] T. A. Kuiken, G.A. Dumanian, R. D. Lipschutz, L.A. Miller, and K. Stubblefield, "The use of targeted muscle reinnervation for improved myoelectric prosthesis control in a bilateral shoulder disarticulation amputee," *Prosthetics Orthotics Int.*, vol. 28, pp. 245-253, 2004.

[2] L.A. Miller, K.A. Stubblefield, R.D. Lipschutz, B.A. Lock, and T.A. Kuiken. "Improved Myoelectric Prosthesis Control Using Targeted Reinnervation Surgery: A Case Series." *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 16, no. 1, pp. 46-50, 2008.

[3] Y. Kamikawa, and T. Maeno, "Underactuated Five-Finger Prosthetic Hand Inspired by Grasping Force Distribution of Humans," 2008 *IEEE/RSJ Int. Conf. on Intelligent Robots and Systems*, pp. 717-722, 2008.

[4] P.J. Kyberd, and J.L. Pons, "A Comparison of the Oxford and Manus Intelligent Hand Prostheses," *Proc. 2003 IEEE Int. Conf. on Robotics & Automation, pp. 3231-3236, 2003.*

[5] H. Huang, L. Jiang, Y. Liu, L. Hou, and H. Cai, "The Mechanical Design and Experiments of HIT/DLR Prosthetic Hand," *Proc. 2006 IEEE Int. Conf. on Robotics and Biomimetics*, pp. 896-901, 2006.

[6] L. Zollo, S. Roccella, E. Guglielmelli, M. C. Carroza, P. Dario, "Biomechatronic Design and Control of an Anthropomorphic Artificial Hand for Prosthetic and Robotic Application," *IEEE/ASME Trans. on Mechatronics*, vol. 12, no. 4, pp. 418-428, 2007.

[7] M. Controzzi, C. Cipriani, M. C. Carrozza, "*2008 IEEE/RSJ Int. Conf. on Intelligent Robots and Systems*, pp. 576-581, 2008.

[8] A. Kargov, C. Pylatiuk, R. Oberle, H. Klosek, T. Werner, W. Roessler, and S. Schulz, "Development of a Multifunctional Cosmetic Prosthetic Hand," *Proc. 2007 IEEE 10th Int. Conf. on Rehabilitation Robotics, pp. 550-553, 2007.*

[9] S. Jung, and I. Moon, "Grip Force Modeling of a Tendon-driven Prosthetic Hand," *Int. Conf. on Control, Automation, and Systems*, pp. 2006-2009, 2008.

[10] C.M. Light, and P.H. Chappel, "Development of a lightweight and adapatable multiple-axis hand prosthesis," *Medical Engineering and Physics*, vol. 22, pp. 679-684, 2000.

[11] D.H. Silcox, M.D. Rooks, R.R. Vogel, and L.L. Fleming, "Myoelectric prostheses. A long-term follow-up and a study of the use of alternate prostheses," *J. Bone Joint Surg. Am.*, vol. 75, pp. 1781-1781, 1993.

[12] D. J. Atkins, D. C. J. Heard, W. H. Donovan, "Epidemiologic Overview of Individuals with Upper-Limb Loss and Their Reported Research Priorities," *J. Prosthetics and Orthotics*, vol. 8, no. 1, pp. 85-92, 1996.

[13] P.J. Kyberd, J. J. Davey, J. D. Morrison, "A Survey of Upper-Limb Prosthesis Users in Oxfordshire", *J. Prosthetics and Orthotics*, vol. 10, no. 4, pp. 85-92, 1998.

[14] J.E. Colgate, and N. Hogan, "Robust control of dynamically interacting systems," *International Journal of Control*, vol. 48, no. 1, pp. 65-88, 1988.