A Review of the Performance of Extrinsically Powered Prosthetic Hands

Revanth Damerla, Yi Qiu, Taylor M. Sun, and Shorya Awtar

***Abstract*— Extrinsically powered prosthetic hands offer the potential to replicate the capabilities of a human hand and thus enable an upper limb amputee to complete activities of daily living. Over the past 20 years however, amputees have consistently indicated that several user needs have not been met. Many of these user needs are related to the hardware of the prosthetic hand, and in particular, its actuators and transmissions. These needs include reduced weight and improved dexterity, hand speed, hand strength, and functionality. To understand why these user needs have not been adequately addressed, we first seek to investigate the state of the art in extrinsically powered prosthetic hands through a comprehensive review of the research, commercial, and open-source literature. This review focuses specifically on actuation of the prosthetic hands because actuation is central to addressing the above user needs. This review, based on actuation strategies, enables a characterization and exploration of the actuation design space. We also compare the performance of the reviewed prosthetic hands with both the human hand and ideal recommendations for prosthetic hands to conclude that existing prosthetic hands do not adequately address user needs. This systematic characterization of the actuation design space helps identify that improvements to transmission pathways are the most promising avenue of further research and innovation to enable future prosthetic hands that adequately address user needs.**

***Index Terms*— Actuators, biomechatronics, mechanical power transmissions, physiology, prosthetics.**

I. Introduction and Motivation

UPPER limb amputations are difficult to overcome because they represent a sudden change in capability and can lead to a significant reduction in independence. There were approximately 41,000 upper limb amputees with an amputation occurring more proximally than the digits in the United States in 2005 [1]. For all these amputees, some form of prosthetic hand would be useful in restoring independence or improving their quality of life. The goal of any prosthesis is to provide the user with the ability to perform activities of daily living (ADLs) in order to regain independence. In cases involving wrist disarticulation (amputation at the wrist) or a more proximal amputation, the prosthetic hand must replace the human hand with an end effector that approximates the hand's functionality to allow the user to complete ADLs. This is a complex objective given the wide variety of tasks that a human hand is capable of. These include highly dexterous tasks such as putting on clothes and using silverware as well as high-strength tasks such as gripping and carrying heavy objects [2]–[7]. Four types of prosthetic hands are currently used to restore varying degrees of this capability: cosmetic or passive, body-powered, extrinsically (i.e. extrinsic to the body) powered, and a hybrid of the latter two. Of these, extrinsically powered prosthetic hands represent the most promising method of replicating human hand functionality because of their potential to provide all of the Degrees of Freedom (DoFs) of a human hand in an intuitive manner without any power input from the user.

Unfortunately, rejection rates for extrinsically powered prosthetic hands have remained consistently high over the past 20 years [8]–[12], with some studies documenting rates higher than 20%. These rejection rates are significant because they suggest that user needs are not adequately met with modern prosthetic hands despite numerous recent technological advancements. In addition, when both non-users (i.e. people who have rejected extrinsically powered prosthetic hands) and users with transradial (at the forearm) amputations were asked how prosthetic hands could be improved, several answers were consistently given over this time period including reduced weight [9]–[15], increased comfort of the interface with the user (e.g. socket or harness) [9]–[13], [16], improved controllability and better ease of use [11]–[14], [16]–[18], increased durability [10]–[14], [16], and better functionality to enable the user to complete desired ADLs [9]–[13], [18]. Better functionality is related to several specific improvements including increased dexterity [12], [13], increased hand strength [16], and increased hand speed [15]. Dexterity requires both the ability to complete small, precise actions (colloquially called "fine-motor skills") and accurately achieve the desired hand posture or grasp.

This list of user needs can be mapped to the three basic domains of the prosthetic hand: the interface with the user, the control (comprised of interpretation of user intent and control of actuators), and the hardware (comprised of structural components, battery and electronics, digits and joints, and actuation). Among the above list of user needs, increased durability, reduced weight, increased hand strength, and increased hand speed are directly related to the hardware while better functionality and increased dexterity are related to both the hardware and control. While addressing the aspects of these user needs related to the interface with the user (see [19]–[22]) and control (see [23]–[27]) are also crucial, we focus on those related to hardware within this paper. With the exception of increased durability, addressing the hardware-related aspects of the other five user needs requires a specific examination of actuation, which consists of the choice and arrangement of

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actuators and transmissions. Actuation accounts for a large percentage of the weight of the prosthetic hand, dictates the maximum hand strength and hand speed, and is directly tied to dexterity and overall functionality. Actuation is therefore central to both the design tradeoffs (tradeoffs across all prosthetic hands) and performance tradeoffs (tradeoffs within the design of a specific prosthetic hand) related to hardware. For example, increasing hand strength typically requires larger actuators and transmissions, limiting the choice of Architecture (i.e. arrangement of actuators – a design tradeoff across multiple designs) and increasing the weight of the prosthetic hand (a performance tradeoff within a given design). To address all five of these user needs, it is therefore critical to understand and improve upon actuation in prosthetic hands.

This paper seeks to present the current state of the art in extrinsically powered prosthetic hand hardware through a comprehensive review of the research, patent, commercial, and open-source literature. It includes and builds upon the findings of several previous papers that have reviewed extrinsically powered prosthetic hand hardware [26]–[32] to provide a more contemporary and complete review of the literature. Several of these papers, in particular [30], [31], have also discussed specific aspects of the actuation and performance of extrinsically powered prosthetic hands. This paper builds upon these findings and offers additional insights by focusing on each of the a) actuators, b) transmissions, and c) Architecture of each prosthesis and emphasizing performance pertaining to all five user needs related to actuation. It also identifies that existing prosthetic hand hardware does not adequately meet these user needs, but also characterizes the actuation design space of prosthetic hands that has been explored. In doing so, we identify further innovation in transmission pathways as the most promising avenue to enable prosthetic hands to adequately address user needs. We also identify several promising alternatives to commonly utilized transmission pathways as well as other avenues within actuation where further research and innovation could eventually enable adequate performance.

The paper is organized into four sections as follows. Section II includes a brief anatomical overview of the human hand and forearm as well as an overview of human hand performance and recommended prosthetic hand performance. This provides the ideal performance to serve as a benchmark for evaluation of prosthetic hand performance. The third section is a review of existing prosthetic hands, organized by their Architecture. Finally, Section IV includes a final discussion and conclusion.

A. Methods

For this review, extrinsically powered prosthetic hands developed between 2000-2020, a time period that adequately covers most modern innovations in the field, were identified using several search engines and the following search terms: (“prosthetic hand”, “prosthetic gripper”, “upper limb prosthesis”, “robotic hand”) by themselves and in conjunction with the terms (“powered”, “extrinsically powered”, “active”); the references cited by each source were also reviewed to ensure this review examined as many prostheses as possible. The attributes and performance of each prosthesis were recorded. In cases where certain values could not be found, the corresponding authors, companies, or creators were contacted to try to obtain the missing values. If no performance attributes could be found, the prosthesis was removed from the review as it could not be adequately compared. Prostheses were also removed from the review if newer versions existed, leaving a total of 96 extrinsically powered prostheses compiled here.

The following attributes were recorded for each prosthesis when available: 1. Transmission pathways from actuator to joint, 2. Total number of actuated DoFs, 3. Number of actuators, 4. Weight of the prosthesis, 5. Maximum hand speeds, 6. Maximum hand forces, 7. Number of grasp patterns, and 8. Type of actuator. These attributes were chosen because they are commonly reported and help convey the hardware’s ability to meet the above five user needs. Weight, hand strength, and hand speed are each addressed by one of the above attributes. However, it is only possible to express dexterity and functionality through a combination of multiple attributes. These include the number of actuators and total number of actuated DoFs as high values in both will lead to more dexterous and anthropomorphic manipulation of digits. The number of achievable grasp patterns, maximum hand speeds, and maximum hand forces are additional indications of the functionality available in the prosthesis.

II. Human Hand Background

A. Architecture of the Human Hand

The human hand consists of five digits – four fingers typically modeled as having four DoFs and one thumb typically modeled as having five DoFs [33]. Each finger is composed of three bones and three joints starting at the end of the palm, which houses the metacarpals (Fig. 1a). As shown in Figs. 1a and 1b, the Proximal Phalanx is connected to the metacarpal via the Metacarpophalangeal (MCP) Joint, the Intermediate Phalanx is connected to the Proximal Phalanx via the Proximal Interphalangeal (PIP) Joint, and the Distal Phalanx is connected to the Intermediate Phalanx via the Distal Interphalangeal (DIP) Joint. Each of these three joints can Flex/Extend (F/E) such as when fingers ball into a fist (Fig. 1c); the MCP joint can also Adduct/Abduct (Ad/Ab), which is the side-to-side motion such as when the fingers spread apart to grasp a wide object (rotation into and out of the page in Fig. 1c). The thumb is composed of two bones – the Proximal Phalanx and Distal Phalanx which are connected via an Interphalangeal (IP) Joint that can F/E. The Proximal Phalanx is connected to the metacarpal of the thumb via an MCP joint that can also F/E and Ad/Ab.

The metacarpal of the thumb, i.e. the First Metacarpal, is significant because it is independently actuated [34] and has a much wider range of motion than any of the other four metacarpals. It connects to the carpal (wrist) bones via the Carpometacarpal (CMC) Joint, which is a saddle joint allowing two DoFs. The first DoF is Opposition (Opp), which is similar to F/E of other joints and enables the thumb to rotate out of the plane of the palm and touch a fingertip. The CMC joint can also Ad/Ab such as when the thumb moves closer or further away from the fingers in the plane of the palm.

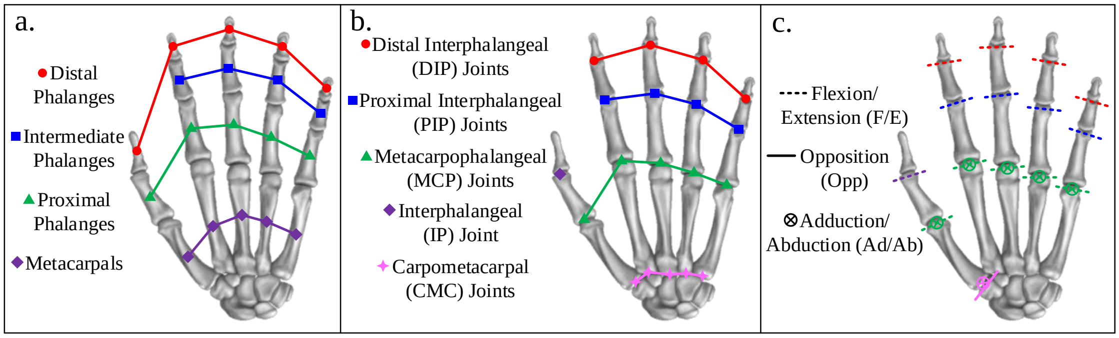

Fig. 1. Hand Skeletal Anatomy and DoFs from a Dorsal View of the Right Hand, Adapted From [35]:
a. Bones of the Digits and Palm, b. Major Joints of the Digits, c. Kinematic Model of the DoFs of the Digits [33]

In total, there are 21 DoFs between the digits that are actuated by a network of muscles located in the hand (intrinsic hand muscles) and the forearm (extrinsic hand muscles) and provide varying degrees of independence. Transmission of the muscle actuation outputs is provided by tendons, which are tough, fibrous tissue that connect each muscle to the bones of the hand. The forearm contains the majority of muscle mass responsible for actuating the hand, with forearm muscle mass concentrated in the proximal half of the forearm [36]–[39]. Each finger is connected to three extrinsic hand muscles that provide F/E, with the index and pinky fingers actuated by an additional muscle each [34]. The thumb is connected to a total of four extrinsic hand muscles that actuate several DoFs [34]. These forearm muscles provide many of the high-torque, -speed, and -power actuations of the digits' DoFs. The forearm is well-shaped for housing muscles for this purpose because it has both a large diameter and a large length. The large diameter means muscles can have large physiological cross-sectional areas (PCSAs), which are associated with large forces [40]–[42]; the forearm's large length enables muscles to contract at higher linear velocities [41]–[43]. The combination of these factors also means that these muscles can produce higher power outputs.

While the hand contains significantly less muscle mass, these intrinsic hand muscles still actuate several important DoFs. Finger Ad/Ab, some auxiliary finger F/E capabilities, and several DoFs of the thumb are actuated by muscles located in the palm [34]. Since these muscles are significantly smaller than those located in the forearm, they are more useful for actuating DoFs that do not require significant power outputs or for providing additional dexterity and capabilities in DoFs that do. Thus, actuation in the human hand is distributed in an intelligent manner between the hand and forearm based on the required torque, speed, and power outputs of the various joints.

While this muscle architecture leads to remarkably diverse hand capabilities, it presents an important challenge in designing a prosthesis. As noted above, the majority of the actuation (i.e. muscles) for the degrees of freedom of the hand is located in the forearm, and more specifically in the proximal half of the forearm. From an evolutionary standpoint, this is a great advantage because this reduces the joint torques required at both the elbow and shoulder to move the arm due to a smaller inertia and proximal center of mass. However, this poses a great challenge in the design of prosthesis. The residual limb for most amputees with wrist disarticulation or transradial amputations will retain most of this portion of the forearm, which no longer serves its biological purpose of providing hand actuation. The volume and mass available for a prosthesis targeting these amputees is therefore limited to what is primarily used for low-power actuation and structural components in the human hand (e.g. hand bones, joints, and intrinsic muscles). Thus, a key challenge in designing a prosthesis lies in the fact it must contain all the necessary actuation for each DoF in a much smaller volume and mass compared to the biological hand.

B. Human Hand Dimensions and Weights

The mean weight and volume of the hand and forearm for men and women are shown in Table I. Percent of body weight for the hand and forearm were found across multiple studies [36]–[39] and were averaged to obtain the values shown in Table I. The mass of the hand and forearm were then estimated using data from the median weights of men and women living in the United States between 2011-2014 [44]. Two studies also measured the volume of the hand and forearm for men [38], [39] but similar values could not be found for women.

Various dimensions of the hand and forearm are listed in Table II, which were obtained from several studies [45]–[49] that measured these dimensions for people serving in the U.S Armed Forces. This may lead to mean values that reflect people who are younger and more muscular than the typical prosthesis user. In studies that listed the mean height and weight of study participants [46], [47], the median weight of participants in the studies was in some cases over 10kg less than the median U.S. adult while the median heights of participants were within 1 cm of median U.S. adults. This difference in height is significantly smaller than the difference in weight. Thus, circumference and width dimensions in Table II, which are based on data from US Armed Forces members, may vary significantly from the

TABLE I
HAND AND FOREARM WEIGHT AND VOLUME FOR MEN AND WOMEN

	Dimensions	Men	Women
Hand	% Body Weight	0.63 [36]–[39]	0.53 [36], [37]
	Mass of Hand for Median US Adult by Weight (g)	540	380
	Volume (cm^3)	363 [38], [39]	
Forearm	% Body Weight	1.66 [36]–[39]	1.48 [36], [37]
	Mass of Forearm for Median US Adult by Weight (g)	1420	1060
	Volume (cm^3)	909 [38], [39]	

TABLE II
MEAN FINGER, HAND, AND FOREARM DIMENSIONS FOR MEN AND WOMEN

	Dimensions	Men	Women
Hand	Index Finger Length (cm) [47]–[49]	7.53	6.93
	Index Finger PIP Joint Width (cm) [47]–[49]	2.23	1.91
	Palm Length (cm) [45]–[47]	10.89	9.98
	Hand Thickness at the Knuckle of the Middle Finger (at MCP Joint) (cm) [45]	3.02	2.62
	Hand Length (cm) [45]–[47]	19.17	17.71
	Hand Width (cm) [45]–[47]	8.82	7.71
Forearm	Wrist Width (cm) [47]–[49]	6.68	5.77
	Flexed Forearm Circumference (cm) [46], [47]	30.70	25.91
	Radial-Stylion Length (cm) [46], [47]	26.86	24.25
	Length of Forearm and Hand (cm) [46], [47]	48.21	44.17

general U.S. population while length dimensions are more likely to represent the general population well.

In spite of such variations, the values listed in these two tables can be used to provide basic guidelines for the weight and dimensions of a prosthesis based on the targeted user demographic (e.g. target gender and amputation types). However, additional factors should be considered. For example, amputees may not be satisfied by a prosthesis that weighs the same as the portion of the limb that they lost. Factors such as quality of socket fit and irritation of the residual limb [10]–[13], [16] may limit the maximum weight of a prosthesis that an amputee is able to wear. In addition, the distribution of weight in the prosthesis should also be considered. A prosthesis with a more distal center of mass than another of the same weight could be perceived as heavier because of the longer moment arm at the amputee's socket and the larger inertia.

C. Human Hand Performance

Due to inconsistencies in reporting in prosthesis hardware, three different metrics representing maximum hand forces are commonly measured: maximum cylindrical grasp force, maximum pinch force (either index tip pinch, chuck pinch, or index pulp pinch), and maximum fingertip force. These are described in further detail below. Similarly, three different values related to maximum hand speeds are often reported (also described in further detail below): maximum joint angular speed, maximum linear finger velocity, and time for hand to move from fully open to fully closed. Hand performance in terms of these values is reported in the literature [50]–[60], making it a useful benchmark for evaluating prosthesis performance. However, prostheses often do not need to achieve the same level of performance to enable an amputee to complete most of their ADLs. In several cases, lower capabilities are recommended by clinicians and prosthesis designers as acceptable performance for prostheses [28]–[30], [61]. For example, prostheses that can move as quickly as the biological hand can be difficult to control. Likewise, grasp strength that is as high as the biological hand may be unnecessary for completing ADLs and challenging to control without sensory feedback. Values for anthropomorphic capabilities of a biological hand and recommendations for prostheses are listed in Table III (hand forces) and Table IV (hand speeds).

Grasp strength is commonly evaluated with a grasp dynamometer, which measures the force applied by the entire hand to an object during a cylindrical grasp. The average values for healthy men and women are significantly higher than the grasp strength for a prosthesis recommended by prosthesis designers and clinicians. While these recommended capabilities may be acceptable for completing many common tasks, it could prevent amputees from completing tasks requiring high grasp strength that a human hand could complete.

Three different types of pinches are often measured clinically: index tip pinch, chuck pinch, and key pinch, which are typically measured with a pinch dynamometer. Tip pinch is a pinch performed between the index finger and thumb. Chuck pinch is a three-digit pinch involving the index finger, middle finger, and thumb. Key pinch is the grasp when the thumb pad is placed on the lateral portion of the index finger.

Index fingertip force is the magnitude of force measured at the fingertip when the index finger, in a fully extended configuration, attempts to flex. This force is not measured as commonly in clinical settings, and fewer sources were found reporting these values. However, fingertip forces are commonly reported for prostheses as they are simple and inexpensive to measure and can still be a good indicator of prosthesis strength.

While the hand strength data listed above is useful in understanding the overall performance of a human hand and for comparing performance to a prosthesis, it can be more practical for a prosthesis designer to understand the strength at individual joints. This can be done by estimating the maximum torque exerted at each joint. A search of the literature did not reveal any studies that systematically measured the maximum joint torque of any digit by joint. However, one study measured the forces at and orientations of each finger phalanx during grasps of cylinders of different diameters for four men [62]. These forces and orientations are used to roughly estimate the finger joint torques with mean phalanx lengths [63]. The maximum calculated joint torques for each finger joint across the cylindrical grasps of various diameters are reported in Fig. 3, assuming only F/E joint rotations (see Fig. 2a [64]). Thumb joint torques were estimated using the force measured in key pinch, the pinch with highest average force. Mean phalanx and metacarpal lengths were used along with this force to estimate maximum thumb joint torques for men, assuming the bones are in a straight line (see Fig. 2b). Separate joint torque values for Opp and CMC Ad/Ab could not be obtained using this estimation technique. The values plotted in Fig. 3 should only be treated as estimates of the average finger joint torques for men; error bars in the figure indicate the range in joint torques based on variations of a single standard deviation in force, joint

TABLE III
MEAN GRASP, PINCH, AND FINGERTIP FORCES FOR MEN AND WOMEN

Hand Forces	Men (N)	Women (N)	Recommended for a Prosthesis (N)
Grasp Force	496 [50]–[55]	301 [50]–[55]	45-68 [28], [61]
Index Tip Pinch	73 [51]–[55]	50 [51]–[55]	
Key (Lateral) Pinch	104 [52]–[56]	72 [52]–[56]	
Chuck (Palmar) Pinch	96 [53]–[56]	68 [53]–[56]	
Index Fingertip Force	48.7 [55], [57]	42.4 [55]	

TABLE IV
ESTIMATES OF MAXIMUM HAND SPEEDS

	Peak	Continuous	Recommended for a Prosthesis
Maximum Index Finger F/E Joint Speeds	MCP: 8.9-12 rad/s (509-690 °/s) PIP: 10.6-20 rad/s (609-1100 °/s) [58], [59]	MCP: 5.2 rad/s (300 °/s) PIP: 5.2 rad/s (300 °/s) [60]	3.0-3.5 rad/s (172-200 °/s) [28], [29]
Estimated Maximum Index Finger Speed for Men from MCP Flexion (cm/s)	76-100	45	26-30
Estimated Maximum Index Finger Speed for Men from MCP and PIP Flexion (cm/s)	120-190	67	39-45
Time for Hand to Close (s)			0.8-1.5 [30]

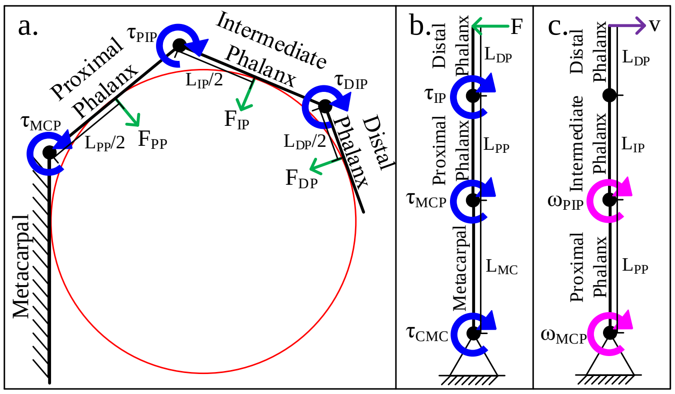

Fig. 2. Methods for Calculating Maximum Joint Torques and Fingertip Speeds: a. Finger Joint Torques, b. Thumb Joint Torques, c. Index Finger Speeds.

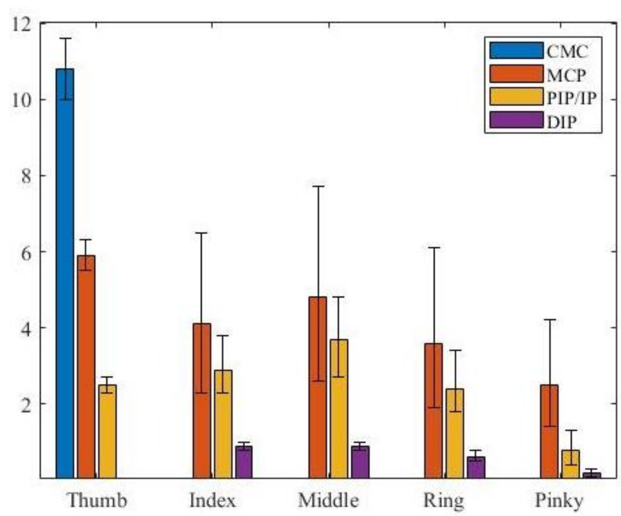

Fig. 3. Estimated Maximum Joint Torques for an Average Man (Nm).

angle (when applicable), and phalanx length. Studies that measure these values would aid prosthesis designers in creating prostheses with more anthropomorphic joint performance.

Despite the approximate nature of these estimates, several important observations can still be made. Firstly, the maximum torque capabilities decrease for joints that are more distal. This is logical given that finger joints that are more proximal may need to counteract forces applied with larger moment arms (e.g. when pinching an object). It also means that more proximal phalanges can apply larger forces, which can play a significant role during a cylindrical grasp. Joints that are more distal can also play an important role in grasping. For example, DIP joints can contribute greatly to the stability of a grasp and enable the hand to complete many grasps. Similarly, the joints of the ring and little fingers provide smaller torque output than those in other fingers but can also contribute greatly to grasping. These observations are supported by analyzing usage of grasp patterns, which can help to better understand the importance of certain fingers and joints [2], [6], [65]. While the thumb and index finger are the most commonly used in performing tasks [2], [6], [65], the majority of grasp patterns involve all five fingers [2], [6], [65], [66], though not for the same functions. In many cases, the thumb and first two fingers are used when high forces and precision manipulation are needed while the ring and little fingers often provide additional grasping force or conform to objects to stabilize them [66], [67].

Maximum joint speeds of the index finger are recorded in three forms in Table IV. Peak maximum joint speed refers to the maximum speed during one cycle of movement from fully extended to fully flexed. In contrast, continuous joint speed refers to a maximum speed during typical use of the hand while completing ADLs. MCP and PIP joint speeds of the index finger were found in the literature, but similar values could not be found for the DIP. Due to significant variation between sources in the reported speeds, these values are reported as a range. Since the index finger is generally one of the quickest fingers, the values given could serve as an upper bound for the recommended corresponding joint speeds for all the digits. We converted the finger joint speeds into linear speeds for the index finger in two motion cases: purely MCP flexion and with both MCP and PIP flexion (see Fig. 2c). Both values can be useful depending on the arrangement of joints and actuation (or underactuation – when a single actuator drives multiple DoFs) that are chosen. While anthropomorphic capabilities may be useful in certain cases, it may be more useful to understand finger and joint speeds while completing ADLs. Evaluations of the maximum joint speeds in fingers while the human hand performs various grasps have been used to recommend maximum joint speeds for prostheses [28], [29] and can be used to calculate recommended maximum index finger speeds. The time for the hand to close (i.e. time for the hand's digits to move from fully extended to fully flexed) is also commonly reported for existing prostheses (but not the human hand). Therefore, several resources have recommended a range of values for a prosthesis [30].

For the hand prosthesis designer, this section can help inform the selection of actuators and transmissions in the design of a prosthesis. While the prosthesis may not have to meet human hand performance in forces and speeds, the targeted capabilities will impact the overall design and functionality of the prosthesis. As noted above, a prosthesis that is not able to meet the force capabilities of a human hand may not be able to perform all the tasks an amputee would like it to. Joint speed, however, can often be more related to convenience. A prosthesis that is unable to meet joint speed requirements may simply require the user to wait longer for the hand to reach a certain orientation before a task can be completed. This will likely not prevent the user from completing most tasks but would impact the time needed to complete it. This difference in practical consequences between insufficient force and speed capabilities is important when deciding on the transmission pathway that translates actuator output to the force and speed capabilities of the prosthesis.

III. Review of Prostheses

The 96 reviewed extrinsically powered prostheses fall into four Architectures based on where actuators driving DoFs of the hand are placed: 1. Actuators housed in the digits, 2. Actuators housed in the palm, 3. Actuators housed in both the digits and palm, and 4. Actuators housed in the forearm. Each of these Architectures presents different sets of advantages and performance tradeoffs. In certain cases, these advantages and tradeoffs enable prostheses employing certain Architectures to better address certain user needs or be more suitable for certain amputees. Within a certain Architecture, prostheses can vary in several ways related to actuation, including differences in number of actuators, number of DoFs, and allocation of actuation for driving these DoFs. These differences lead to several Design Strategies within each Architecture that in turn present additional advantages and performance tradeoffs.

Within this section, the reviewed prostheses are organized based on the Architecture and Design Strategy they employ. This approach to organization enables a review and evaluation of each prosthesis in comparison with similar prostheses. Furthermore, it enables a systematic, qualitative analysis of the advantages and tradeoffs associated with each Architecture and Design Strategy that can inform the design of future extrinsically powered prostheses. Tables V-XI organize the prostheses based on Architecture and Design Strategy and describe their capabilities and performance. These tables also describe the various methods of transmission from actuator to rotation about a joint (called the transmission pathways) utilized in each prosthesis using the following nomenclature: Actuator (A), Joint (J), Unspecified transmission elements (which may not include any additional transmissions elements in certain cases) (_), Unspecified type of gears, gearbox, or drive (G), Planetary gearbox (P), Cycloidal drive (C), Harmonic drive (H), Spur gears (S), Worm-wormwheel pair (W), Bevel gears (B), Helical gears (Hg), Rack and pinion (R), Linkage (L), Screw and nut (e.g. lead screw, ball screw, etc.) (Sc), Cam (Ca), Whippletree-type mechanism (Wh), Tendon or other form of open cable system (T), Belt or other form of closed cable system (Bt), and Other (O); parentheses indicate coupling of various DoFs. For example, the transmission pathway driving the thumb's DoF in the prosthesis shown in Fig. 4a would be APHgJ. In Fig. 4b, a single motor and planetary gearbox drive MCP and PIP F/E of three fingers via three separate tendons (one for each finger); this transmission pathway is labelled APTJ(T). 'APTJ' indicates that an actuator and planetary gearbox drive a tendon that in turn drives actuation of at least one DoF while '(T)' indicates that at least one additional DoF is also driven by a tendon. Finally, the numbers of grasp patterns listed in the tables are not standardized using a common grasp taxonomy; instead, the number of grasps reported for each prosthesis is listed.

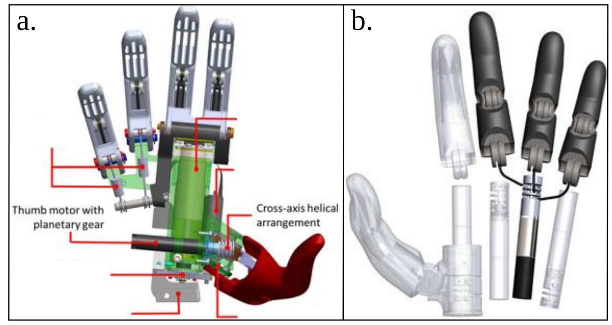

Fig. 4. Examples of Transmission Pathways in Prostheses: a. Uncoupled Thumb DoF [68], b. Coupled MCP and PIP F/E of Three Fingers [69].

A. *Architecture 1 – Actuators Housed in the Digits*

Prostheses using Architecture 1 house their actuators in the prosthesis' digits but may include an additional actuator located in the palm, typically for Opp. The Tactile Sensor Hand [70] is one such example, as shown in Fig. 5a. Each finger consists of two phalanges, with the proximal phalanx of each finger containing a Brushed DC motor (BDC) and planetary gearbox that run along the length of the phalanx. The axis of rotation of the actuator is reoriented via a worm gear combination to be aligned with the desired MCP joint rotation. In addition to this reorientation, the worm gear prevents the phalanx from being backdriven, which is useful for holding heavy objects or providing large forces without requiring significant power input to the actuators; the motion of the distal joint is coupled to that of the MCP via a tendon (APWJ(T)). While utilizing fingers with only two phalanges may limit the capabilities of the Tactile Sensor Hand, particularly in the grasps it can perform, additional phalanges can also increase the complexity of the transmission pathway and the prosthesis' weight. The thumb is actuated by two separate motors to enable independent F/E (via the same transmission pathway as the fingers) and Ad/Ab (using the same actuator, planetary gearbox, and worm gear combination but located within the palm – APWJ).

The Tactile Sensor Hand meets the anthropomorphic size and weight parameters well, with anthropomorphic dimensions and a weight lower than a median male's. However, it is unable to achieve anthropomorphic performance. The hand's fingertip force is significantly lower than anthropomorphic values for both men and women and its reported maximum MCP F/E joint speed is also significantly lower than recommended joint speeds. The prosthesis' limited performance capabilities are likely due to a couple of factors, including a conscious decision to prioritize limiting the weight of the prosthesis. This decision places strong constraints on the performance capabilities of the actuators and transmissions, including limiting the maximum power output. Limited motor power output creates a difficult optimization problem where designers must decide how to divide the output into speed and torque capabilities. An attempt

([a] – MAXIMUM FINGER JOINT SPEED (°/S), [b] – MAXIMUM FINGER LINEAR SPEED (MM/S), [c] – TIME FOR HAND TO CLOSE (S),
[d] – MAXIMUM GRASP FORCE, [eI] – MAXIMUM INDEX TIP PINCH FORCE, [eII] – MAXIMUM KEY PINCH FORCE, [eIII] – MAXIMUM CHUCK PINCH FORCE, [f] – MAXIMUM FINGERTIP FORCE)

TABLE V
ARCHITECTURE 1 PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
Tactile Sensor Hand [70]	2018	APWJ, APWJ(T)	11	6	450	80[a]	12[f]	7	BDC
I-Limb Quantum (comm.) [71]		A_WJ, A_WJ(T)	11	6	432-518	0.8[c]		24	DC motor
VINCENTevolution3 (comm.) [72], [73]		A_W_J(_)	10	6	386			14	
Unnamed [74]	2011	A_WJ(L), A_J	15	6	272-336 (w/ wrist)	2.5[c]	20[d]	4	BLDC, AC
FiMec Hand [75]	2020	AGJ	15	15	328.45				DC motor
F3Hand II [76]	2020	AJ	5	6	245 (w/ socket)	1.0[c]	7.9[d], 1.8[eI], 3.0[eII]	6	Pneumatic

TABLE VI
ARCHITECTURE 2 MULTIGRASP-TYPE PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
Multigrasp Hand [69]	2015	APTJ, APTJ(T)	9	4	546		29[eI], 30[f]	6	BLDC
SmartHand Transradial Prosthesis [77]	2011	APSOTJ(T), APSScTJ(T), APJ	16	4	530	1.47[c]		3	BDC
LUKE Arm (comm.) [78], [79]	2013		12	4	1400 (w/ wrist)			6	
Prensilia MIA (comm.) [80]	2018		6	3	480	0.5[c]	70[d]	3	
Hero Arm (comm.) [81], [82]	2018	AGSc_TJ, AGSc_TJ(T)	10	3 or 4	280-346	1[c]		4 or 6	BDC
HIT/DLR Prosthetic Hand [83]	2006	APBJ(L), ABtHJ(BJL),	14	3	500		10[f]	4	BDC
Unnamed [84]	2008	AG_J, AGTJ(L)	11	4	400.72			3	DC motor
AR Hand III [85]	2009	APJ(L), APBJ(L)	15	3	500		10[eI]	4	Stepper motor
TU Biomimetic Hand [86]	2014	AGJ, AGTJ(T)	15	5	520 (w/ wrist)			6	DC motor
ISR-Softhand [87]	2014	AGTJ, AGTJ(T)	9	3	530			10	DC motor
X-hand [88]	2016	APSScLTJ(T), APHgJ, APTJ(T)	16	4		1.2[c]	12.1[d]	30	BDC
Unnamed [89]	2017	APBSJ(T), APSTJ(T), APSJ	15	4				6	BDC
MORA HAP-2 [90]	2017	AGWJ, AGWJ(L)	15	4	250			5	DC motor
SSSA-MyHand [91]	2017	APWJ, APWSJ(L), APWLJ(L, O)	10	3	478	170[a], 0.37[c]	14.6[f]		BLDC
Brunei Hand 2.0 (open-source) [92]	2018	A_TJ, A_TJ(T)	9	4	332				BDC
Unnamed [93]	2019	AGJ, AGTJ(T)	11	3	132.5			13	Servo motor

TABLE VII
ARCHITECTURE 2 RIC-TYPE PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
RIC [68], [94]	2016	APSOScLJ(L), APHgJ	9	2	383	180[a], 0.4[c]	84[e]	3	BLDC
Ottobock Michelangelo (comm.) [30], [95]	2012	AG_LJ(L), A_J	6	2	420	86.9[a], 325[b]	60[eII]	5	BLDC, Unknown
KIT Prosthetic Hand [96]	2018	APTWhTJ(T), APTJ(T)	10	2		120.92[a], 1.32[c]	24.19[d]		BDC
Unnamed [97]	2013	APWTJ(T, WhT, O)	11	1	350	2.4[c]	5.1[d], 4.7[eII]	4	BDC
Yale MyoAdapt Hand [98]	2018	APWTJ(T, WhT)	11	1	290	145[a], 1.113[c]	15.2[d], 18.2[eII], 3.6[eIII]	3	BDC
UT Hand I [99]	2014	APTWhTJ(TL) APTJ(T), APWJ	15	3		3-4[c]	12[eII], 5[eIII]	3	BLDC
Leverhulme/Oxford Southampton Hand (3 digits) [100]	2001	ASBt_J, ASBtSJ(_)	6	2	964	1.2[c]	45[e]		DC motor
MANUS-HAND (3 digits) [101]	2004	A_SJ(Bt), A_SBJ(_, O)	9	2	800 (w/ wrist)		60[d]	4	BLDC
SPRING Hand (3 digits) [102]	2004	AGBtScTJ(T)	8	1	400		9[d]		BDC

Unnamed [103]	2005	AGTJ(T)	15	1			30[d]	2	DC motor
KNU Hand [104]	2008	APSJ(L), AWLJ(L, SO)	16	2	800			6	DC motor
sim-PH [105]	2015	AGJ, AGJ(L)	4	2	150			3	Servo motor
Rehand [106]	2015	A_LJ(L)	11	1	467	1.5[c]	13.2[e]	6	
Softhand 2 [107]	2018	AGTJ(T)	19	2				>12	BDC
Unnamed [108]	2018	AHLJ(L)	10	1	980	1.7[c]	34.5[eIII]	2	BLDC
MERO Hand [109]	2019	ASBtJ(T), AGTJ(T)	16	2	336		3.6[d]	3	BDC, DC motor
MGM Hand [110]	2020	AGBLO(LJ(L), OLJ(L))	10	1	470	0.58[c]	25.7[d], 6.0[eIII]	10	BDC

TABLE VIII
ARCHITECTURE 2 GREIFER-TYPE PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
Ottobock System Electric Greifer (comm.) (2 digits) [111], [112]		A_SRLJ(RL)	4	1	520	200[b]	160[d]		
Ottobock AxonHook Hand (comm.) (2 digits) [113]			1	1	400	173[b]	110[d]		
Motion Control ETD2 (comm.) (2 digits) [114]			1	1	408-454		107[e]		BLDC
Ottobock MyoHand Variplus Speed (comm.) (3 digits) [115]			2	1	460	300[b]	100[d]		
Motion Control ProPlus Hand (comm.) (3 digits) [116]			2	1	431-479		100[e]		BLDC
Steeper MyoSelect Hand (comm.) (3 digits) [117]			2	1	470-520	0.8[c]	80[d]		DC motor
Unnamed (3 digits) [118]	2013		2	1	230		54[d]		DC motor

TABLE IX
ARCHITECTURE 2 ONE ACTUATOR PER DIGIT PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
Unnamed [119]	2015	APBJ(L)	15	5	440	118[a]	10[eIII], 10[f]	5	DC motor
Psyonic Ability Hand (comm.) [120]–[122]		A_WJ, A_WJ(L)	10	6	460	0.2[c]		5	BLDC
Prensilia IH2 Azzurra (comm.) [123]		A_J, A_TJ(T)	11	5	640	1[c]	35[d], 7[eII]	10	BDC
Ottobock Bebionic3 Hand (comm.) [124]		A_Sc_J(L)	10	5	369-557	1[c]	140.1[d], 26.5[eII], 36.6[eIII]	14	DC motor
TASKA Hand (comm.) [125]	2017		10	6	616	98[a]		7	
Covvi Nexus Hand (comm.) [126]	2020		11	6	570	0.7[c]	80[d], 22[eII], 45[eIII]	13	
Touch Hand II [127]	2016	APBtJ, APBtJ(Bt)	14	6	451	0.826[c]	60.6 [d], 8.0[eII]	10	BDC
Unnamed [128]	2017	APSJ, APWJ, APWJ(L)	10	6		180.76[a]	1.64[f]		BDC
Tact Hand (open-source) [129]	2015	AGTJ(L), AGJ	11	6	350	249.8[a]	4.21[f]	5	BDC
X-Limb [130]	2020	ASTJ, ASTJ(T)	13	5	253	1.3[c]	21.5 [d], 10.2[eI]	3	BDC
Modular Prosthetic Limb [131]	2020	ASCLJ(L), ASPJ, ASPLJ(L)	19	10	1300 (w/ wrist)	360[a], 0.3[c]	310[d], 67[eI], 110[eII], 110[eIII]		BLDC
Southhampton-Remedi Hand [132]	2001	AGWLJ(L), AGW_J	14	6	400	96[a]	9.2[f]	5	DC motor
Harada Hand [133]	2001	AG_TJ(T), AG_ScTJ(T)	14	5	369	45[a], 2.39[c]		7	BDC
Unnamed [134]	2011	ATJ, ATJ(L)	11	6	272		4.52 [d]	3	SMA
Unnamed [135]	2012	A_BtJ(Bt)	14	5			20[f]	2	Servo motor
Unnamed [136]	2012	ASWJ, ASWJ(L)	10	6	410	1.4[c]		4	BDC
Dextrus Hand [129], [137]	2013	APTJ(T), AGJ	15	6	428	175.4[a]	1.71[f]		BDC, DC motor
Unnamed [138]	2015	ASWJ, ASWTJ(T)	15	6	1600 (w/ wrist, socket)	180[b], 2[c]		≥ 5	BDC
Ada 1.1 Hand (open-source) [139]	2016	A_TJ(T)	10	5	380			4	BDC
Unnamed (open-source) [140]	2016	APBJ, APBSJ, APBSJ(Bt)	10	6	584	128[a]	4.12[f]		BDC

Galileo Hand [141]	2017	ASTJ(T), ASBHJ	15	6	<350		50[d]	4	BDC
Touch Hand 3 [142]	2017	A_J, A_J(L)	9	5	593		18.6[d]	6	DC motor
Unnamed [143]	2017	A_J, A_TJ(T)	11	6	450				DC motor
Unnamed [144]	2018	AGTJ(L), AGSJ, AGSJ(L), AGTJ(T)	18	7					Servo motor
UHVAT Hand (3 digits) [145]	2018	A_WLJ(L)	6	3	600			2	DC motor
DUFAB [146]	2019	ASLJ(L)	14	5				4	BDC
AstoHand v 3.0 [147]	2019	A_TJ(T)	10	5	375			5	BDC
Grasp Bionic Hand (open-source) [148]	2020	A_LJ, A_LJ(L), AGJ	10	6	335	71.96[a]	6.82[f]	6	BDC, BLDC
Smart Bionic Hand [149]	2020	A_LJ(L)	11	5		0.2[c]	53.27[d], 6.88[eII]	7	DC motor
Unnamed [150]	2020	A_J(_)	10	5	531		20[d]		DC motor

TABLE X
ARCHITECTURE 3 PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
Unnamed [151]	2010	ASBJ, ASBJ(Bt), ATJ(T)	16	15			8.1[eII]	≥ 3	BDC, SMA
Fluidhand III [152]	2009	N/A	8	9	400	1[c]	45[f]	5	Hydraulic
Biomechatronic Hand (3 digits) [153]	2002	APScLJ, APScLJ(L)	8	6			1[f]	2	BLDC
Unnamed [154]	2018	N/A	15	6			>0.9[f]	≥ 21	Pneumatic

TABLE XI
ARCHITECTURE 4 PROSTHESES

Prosthesis Name	Year	Transmission Pathway	# of Actuated DoFs	# of Actuators	Mass of Prosthesis (g)	Hand Speed	Hand Force (N)	# of Grasp Patterns	Type of Actuator
CyberHand [155]	2006	APSScTJ(T), APSWJ	16	6	360 (hand), 1800 (hand w/ forearm)	45[a]	70[d], 5[eI]	4	BDC
Unnamed (3 digits) [156]	2007	AP_ScTJ(T), AGWJ	10	4	320 (hand), 920 (hand w/ forearm)	6[c]	35[d], 15[eI]		BDC
Unnamed [157]	2015	ASTJ(_)	10	5	240 (hand w/ forearm)				BDC
Unnamed [158]	2009	APTJ, APTJ(T)	16	5	580 (hand w/ forearm)	0.4[c]	80[d], 11.7[f]	6	BDC
TBM Hand [159]	2001	ASSc(LJ(L), TLJ(L))	14	1	280 (hand w/ forearm)	4-5[c]	12[eI], 8.5[eII], 14[eIII]	1	BDC
Unnamed [160]	2008	ATWhTJ(T)	15	1	730 (hand w/ forearm)	0.8[c]	37[d]		Ultrasonic Motor
Unnamed [161]	2006	AGJ, AGTJ, AGTJ(T)	15	10	204 (hand), 1207 (hand w/ socket, wrist)	200[a]		≥ 6	Servo motor
Unnamed [162]	2016	AGTJ(T)	15	5				4	Servo motor
Unnamed [163]	2018	AGTJ(T)	16	5	286 (hand w/ forearm)				Servo motor
TN Hand [164]	2020	A_TJ, A_TJ(T)	18	9	160 (hand), 670 (hand w/ forearm)			15	Servo motor
Unnamed [165]	2020	AGOBSJ(SLJ(SL))	15	5	475	487.30[a]			DC motor
Unnamed [166]	2014	ATJ, ATJ(T)	11	6	310 (hand w/ forearm)	2.1[c]	11.5[f]	4	SMA
Unnamed (3 digits) [167]	2014	ATJ(T)	8	6			1.4[f]	≥ 1	SMA
Unnamed [168]	2016	A_TJ(T), ATJ	24	21				5	BLDC, SMA
Unnamed [169]	2008	ATJ(T)	17	5	2000 (hand w/ forearm, wrist, elbow)			≥ 2	Pneumatic
Unnamed [170]	2018	AJ	10	10	950 (hand w/ forearm)	0.1[c]		6	Pneumatic

to achieve a middle ground can lead to insufficient performance in both, as in this case. Insufficient power output can be further exacerbated by the efficiency of transmission elements. While a worm gear enables the non-backdrivable feature of the fingers as discussed above, it can do so at the cost of relatively low efficiency. This in turn places further limits on the force and speed capabilities of each finger.

The actuator and transmission are also constrained by the choice of Architecture, with both needing to fit within the limited volume and shape of the first phalanx of the finger. For example, the width of the finger places a fundamental limitation on the maximum torque output of the motor and may therefore require the transmission pathway to provide a larger reduction ratio to achieve a high enough torque output. Similarly, the natural axis of rotation for the BDC is along the length of the phalanx as this allows for a larger motor and gearbox. As a result, the transmission pathway

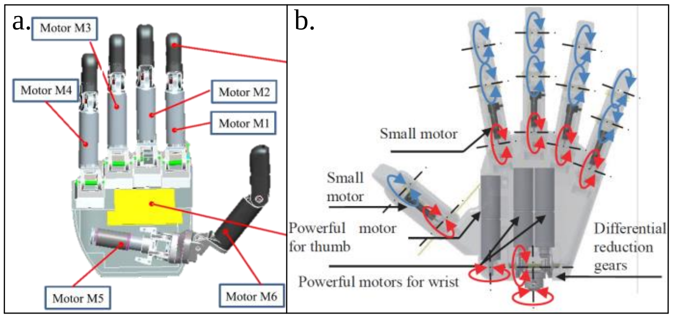

Fig. 5. Examples of Architecture 1 Prostheses: a. Tactile Sensor Hand [70], b. Unnamed [74]

must include an element that reorients the axis of rotation correctly for F/E. A previous version of the Modular Prosthetic Limb (see Section B4) was able to avoid this challenge by using a small and power-dense Brushless DC motor (BLDC) whose axis of rotation was parallel to finger joint F/E [171], [172]. Another limitation of the Architecture is that some of the power output of the actuator and transmission (among the heaviest components of the prosthesis) must be consumed in moving themselves. The inertia that comes from this can reduce maximum hand speeds and impact the responsiveness of the hand (related to joint acceleration). Finally, placement of the actuators distally within the prosthesis may lead to a larger perceived weight for the amputee. Thus, aiming for a prosthesis mass that is less than a human hand's (to account for the greater inertia) is a common strategy for this Architecture.

Two commercially available prostheses, the I-Limb Quantum [71] and VINCENTevolution3 [72], [73] utilize Architecture 1. Both prostheses have similar transmission pathways to the Tactile Sensor Hand and use digits with two phalanges (except for the thumb in the VINCENTevolution3 which only has one phalanx). They each utilize DC motors with a transmission housed in the proximal phalanx of each digit. The output of this combination is connected to a worm gear to enable MCP F/E for each digit, with F/E of the distal joints coupled to the MCP joints of their respective fingers. Both prostheses also contain an actuator with a similar transmission pathway within the palm to provide a DoF like Opp.

As with the Tactile Sensor Hand, each of these prostheses weighs less than the median male hand and is offered in multiple sizes that may also have different weights. This demonstrates a clear emphasis on addressing the user need of reduced weight, potentially at the cost of performance (e.g. better functionality). These prostheses also provide several functions not found in most research and open-source prostheses, including a large number of achievable grasps, the ability to program additional grasps, waterproof options, and the ability to connect a prosthetic wrist [71], [72]. Unfortunately, both prostheses have few listed performance attributes, making it difficult to compare their force and speed capabilities to the human hand. However, data collected on the performance of previous versions of these prostheses [30] can provide some insight and comparison. The iLimb Pulse and Vincent Hand, both released in 2010, had measured maximum finger MCP F/E speeds of 110.6 and 103.3 °/s and maximum fingertip forces of 11.18 and 8.44 N, respectively. These speed and force values are near those of the Tactile Sensor Hand and are well below recommended and anthropomorphic capabilities, suggesting that these prostheses are likely limited by similar considerations, tradeoffs, and challenges as the Tactile Sensor Hand. However, these capabilities are likely to be lower than for the i-Limb Quantum and VINCENTevolution3, which both claim performance improvements over their respective previous models.

Three other prostheses utilize Architecture 1. The first [74] (Fig. 5b) has a similar transmission pathway to the Tactile Sensor Hand, with a small BLDC motor (housed in each digit's proximal phalanx) that eventually connects to a worm gear to provide MCP F/E of each digit. A compliant linkage provides coupling to the two distal joints in each finger and one distal joint in the thumb (A_WJ(L)). This prosthesis is unique for its use of three larger and more powerful AC motors that are housed in the palm. One enables thumb Opp while the other two provide wrist Pronation/Supination and Flexion/Extension. While the use of small BLDC motors for digit F/E DoFs leads to both slow speeds and low grip strength relative to other prostheses, it enables the use of heavier actuators for DoFs that require higher forces and speeds to complete ADLs. By utilizing Architecture 1, there is significant space in the palm to house these additional actuators with few constraints. This space could also be used for housing batteries or electronics.

The second is the FiMec Hand [75], which uses small DC servo motors at each joint to provide F/E; it also uses another identical motor for Opp, leading to a total of 15 motors. The high number of identical motors places limits on the mass, size, and overall performance of each motor (and therefore the overall performance of the hand). However, it could enable a larger range of achievable postures and grasps that are not possible with joint coupling. The final prosthesis, the F3Hand II [76], uses novel pneumatic artificial muscles to actuate the fingers, but requires an external CO_2 cylinder to operate.

One of the key features of Architecture 1 is that since the actuators are within the digits themselves, it is straightforward to ensure that each digit will be independently actuated. A greater number of actuators also increases the number and complexity of grasps that the hand can be capable of. However, these actuators will require more electrical components (e.g. drivers and microcontrollers) and can be a greater burden to the amputee. User burden due to multiple independently driven joints primarily arises when triggering the actuation of these joints. For example, a myoelectric prosthesis that only features one actuator for flexion/extension may only require one or two EMG sensors to trigger this motion. However, in a hand with more actuators that can achieve many more grasps and gestures, a remote (e.g. smartphone app) or other technique may be needed to enable distinct triggers for each of them. One method that has been suggested to reduce complexity in control is to provide coupled actuation for the ring and little fingers [70].

B. *Architecture 2 – Actuators Housed in the Palm*

Prostheses in Architecture 2 place actuators within the palm of the hand. This Architecture offers significantly more flexibility in number and type of actuators and their respective functions than Architecture 1, enabling a wider array of Design Strategies. The differences across Strategies are related to several factors, including differing prioritizations of user needs. These distinct approaches to utilizing the available space and weight therefore lead to varied performance and functionalities.

1) MultiGrasp-Type Prostheses

One such Design Strategy is demonstrated by the Multigrasp Hand [69] (Fig. 6a). This hand has five digits and utilizes four identical BLDC motors and planetary gearboxes housed in the palm that actuate 9 DoFs via tendons connected to pulleys. Two of the motors actuate Opp and thumb MCP F/E while a third actuates index finger MCP F/E. Both the thumb and index finger do not have joints more distal to their respective MCP joints (APTJ). The three remaining fingers, which are capable of MCP and PIP F/E, are actuated by the final motor via three separate tendons (one per finger) with integrated compliance to enable adaptive grasping capabilities (APTJ(T)).

This allocation of articulation separates the functions of the hand's digits based on how humans grasp objects. As discussed in Section 2, the thumb and first two fingers are used in most grasps and play an especially important role in precision grasps such as pinches. The ring and pinky finger meanwhile are more generally used to provide additional grasping force and primarily stability. This prosthesis functions similarly by placing the focus of articulation on the thumb and index finger and ensuring each DoF therein is independently actuated (i.e. no coupled joints). This enables the two digits to have a higher degree of dexterity and more strength than the other three. The use of underactuation and compliance in coupling the joints of the final three fingers provides the important additional grasping force and stability these fingers provide in the human hand. This choice of actuation enables the prosthesis to perform grasps that are either precise (e.g. index tip pinch) or conformal (e.g. cylindrical grasp) while ensuring that each digit has the dexterity needed to grasp objects similarly to a human hand.

This choice of Design Strategy greatly reduces the number of actuators needed to achieve the functionality targeted by the hand's designers while maintaining a weight slightly above that of a median male. However, the hand's force performance does not match a human hand's. The pinch and fingertip force capabilities are lower than their respective anthropomorphic values for both men and women. In addition, while the maximum fingertip force of the index finger is 30 N, the combined maximum fingertip force for the last three fingers is 23 N. This could place a significant limit on the maximum grasp force of the prosthesis and illustrates a clear tradeoff to this Design Strategy. The emphasis on dexterity and capability of the thumb and index finger comes at the cost of grasp force performance. However, these fingers could still contribute to grasping and stabilizing an object, especially if the object is lightweight or if little force is needed. The prosthesis may therefore have acceptable force capabilities for some amputees.

Several other prostheses within Architecture 2 employ similar Design Strategies, but with variations in the number of joints in the fingers and thumb. For example, the SmartHand Transradial Prosthesis [77] contains four BDCs arranged

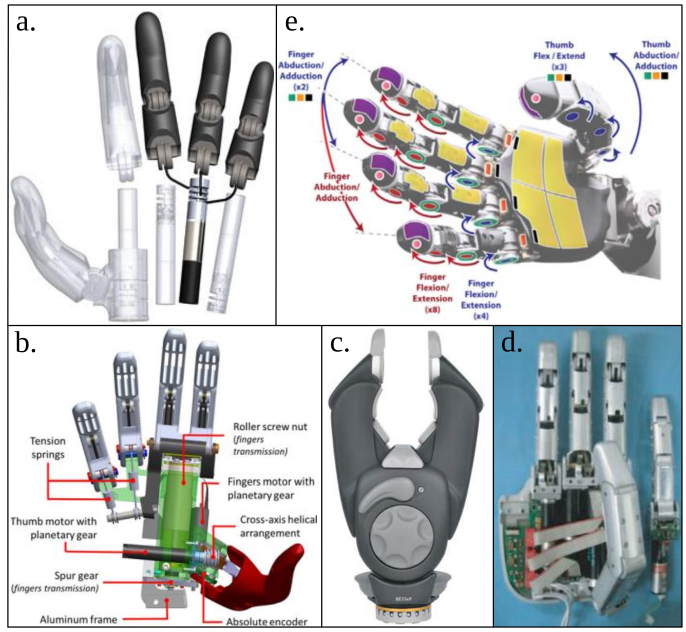

Fig. 6 Examples of Architecture 2 Prostheses: a. Multigrasp Hand [69], b. RIC hand [68], c. Ottobock System Electric Greifer [111], d. Unnamed [119], e. Modular Prosthetic Limb (MPL) [131]

similarly to the Multigrasp Hand. However, each finger contains three joints capable of F/E and the thumb can complete CMC Ad/Ab and CMC, MCP, and IP F/E. The SmartHand therefore places similar emphasis on the digits and DoFs that should be actuated but makes use of underactuation to enable slightly different functionality. This can include better capability in conformal grasps compared to the Multigrasp Hand but potentially at the cost of lower dexterity.

Three commercially available prostheses [78]–[81] also employ this Design Strategy. While other important considerations impact the design and success of commercially available prostheses including cost of development, product price, reception by clinicians (e.g. prosthetists), and product regulations, this helps demonstrate the Strategy's appeal and that capabilities of the thumb and index finger can be an important consideration for amputees. In addition, while none of the prostheses employing this Design Strategy achieve anthropomorphic force capabilities, most weigh about the same or less than the median male hand. Improvements to these capabilities may therefore require considering actuator and transmission options that can provide larger force and torque outputs for the same mass and dimensions (i.e. better force and torque densities).

One alternative approach that could enable anthropomorphic force capabilities is to utilize fewer actuators. This would allow each motor to be larger and heavier. Larger motors can naturally achieve higher force and torque densities because they can have a larger diameter and a smaller percentage of the total weight of the actuator would be used for structural components (e.g. frames and bearings). While this comes at the cost of limiting other functionality such as dexterity and precision grasping capabilities, these may be reasonable tradeoffs for amputees who place a greater emphasis on high force and speed capabilities. This approach is used in two separate Design

Strategies within Architecture 2.

2) RIC-Type Prostheses

The first uses a single actuator housed in the palm to actuate the four fingers together in coupled F/E. Generally, either an additional actuator or the same actuator will actuate the thumb's DoFs. One example is the RIC hand [68], [94] (Fig. 6b). This hand contains two different BLDCs connected to planetary gearboxes. The first motor is significantly larger, with the output of the planetary gearbox connected to a non-backdrivable clutch via a pair of spur gears. The clutch drives a roller screw that in turn actuates MCP and PIP F/E of the four fingers via a linkage with integrated compliance (APSOScLJ(L)). The thumb has one DoF that is a combination of Opp and Ad/Ab, optimized over several trials to find the most desirable axis of rotation. This joint is connected to a helical gear pair driven via the smaller BLDC and planetary gearbox that is naturally non-backdrivable (APHgJ).

This choice of actuation leads to a prosthesis that is approximately the weight of a median woman's hand. The hand is also capable of a pinch force that is between the capabilities of an average man and woman. However, this may not translate to anthropomorphic grasping capabilities because the motor provides the pinch force output of 1-2 fingers instead of all four. Finally, it can also achieve a maximum joint speed and time for hand to close that are both within the recommended range.

By reducing the number of actuators within the prosthesis and shifting the focus away from maximizing dexterity (as in Multigrasp-type hands), the hand was able to achieve more favorable force and speed capabilities at a low weight; the low weight also made it possible to include a prosthetic wrist into the overall prosthesis [68], [94]. Furthermore, fewer actuators can also translate to easier controllability and a lower product price for the user. While motor and transmission capabilities are still limited by the dimensions of the palm in this prosthesis, they are able to achieve higher performance. It also increases the set of transmission options that can be used, including a non-backdrivable clutch.

This Design Strategy can also be found in a commercial prosthesis, the Ottobock Michelangelo [95]. The hand contains a large BLDC located in the palm to actuate F/E of all five digits and uses a smaller actuator to enable thumb Ad/Ab [30]. While the hand does not achieve the same speed and force capabilities as the RIC hand, it demonstrates the commercial viability of an approach that emphasizes speed and force capability, a potentially lower price, easier controllability, and other relevant commercial considerations (see Section B1), at the potential cost of dexterity and precision grasping capabilities. Several other hands use a similar Design Strategy with varied degrees of underactuation and transmission pathways, with some using whippletree-type structures to enable underactuation [96]–[99] and most employing planetary, spur, or other gears as the first part of the transmission pathway.

3) Greifer-Type Prostheses

A second Design Strategy utilizes a single actuator housed in the palm. In this case, the hand contains 2-3 digits, with one digit representative of a thumb that opposes 1-2 fingers. One example is the Ottobock System Electric Greifer [111], [112] (Fig. 6c), a commercially available prosthesis that consists of two opposing digits each composed of two phalanges. A single motor housed in the palm, with axis of rotation parallel to the fingers, drives both digits. The two phalanges of each digit are coupled via a linkage, ensuring they move in a motion that is convenient for pinching and grasping. The transmission pathway utilizes curved rack gears attached to the first phalanx of each finger. These gear racks are driven by a pinion gear that is actuated by the motor via a spur gear reduction and a mechanical automatic transmission (possibly [173]). This automatic transmission toggles between two reduction ratios based on the load applied to the fingers, enabling the prosthesis to switch between a high-speed mode and a high-force mode. The ability to switch between reductions automatically means a lighter, less powerful motor can be used to achieve the desired capabilities. As a result, the prosthesis achieves a grasp force significantly above what is suggested as adequate and hand speeds close to the recommended value at a weight less than a median man's hand. While this performance does not meet anthropomorphic capabilities, it is likely to be adequate for some amputees. In addition, the limited number of digits, actuators, and DoFs may contribute to making this prosthesis easier to use and therefore more dexterous in practice. However, the hand's appearance is far from anthropomorphic. Thus, while its functionality and weight may be desirable, some amputees may reject this prosthesis based on its appearance.

Several prostheses within this Design Strategy address this issue and maintain an anthropomorphic appearance [115]–[118] by making it possible to fit a covering resembling a hand over them. Unfortunately, the transmission pathways of these prostheses are not well-documented in the literature. However, most of these options can also achieve adequate grasp or pinch forces and similar hand speeds to the Ottobock System Electric Greifer while weighing less than the median male hand.

As with the RIC-type hands, prostheses that use this Design Strategy can take advantage of the benefits that using fewer actuators offer. The limited number of DoFs and digits compared to the RIC-type hands also translates to less mechanical complexity and additional mass available for actuation. These prostheses also do not require the same degree of underactuation, which can contribute to better dexterity and capabilities in certain precision grasps, easier controllability, and lower price. However, use of a single actuator places unique limitations on the set of achievable grasps and the set of tasks the prostheses can help achieve. For example, the digit representing the thumb is only able to F/E, preventing the prostheses from achieving grasps such as lateral pinch.

All except one of the hands within this Design Strategy are commercially available. The number of commercially available options suggests this Strategy offers a strong commercial case and that high hand strength and speed capabilities, lower price, and easier controllability, among other factors, can be more attractive than total number of DoFs to amputees. These prostheses can also offer certain additional functionalities that are beneficial to users. For example, the Ottobock AxonHook features a quick-disconnect wrist that allows it to be easily switched out with the Ottobock Michelangelo Hand. This allows users to take advantage of the complementary sets of functionalities the two prostheses offer and avoid some of the tradeoffs inherent to a single Design Strategy. The prosthesis options that are non-anthropomorphic in appearance also offer hooks at the end of each digit [111]–[114] which make it possible to passively carry objects such as bags. This Design Strategy therefore presents several options to address the user need of better functionality.

4) One Actuator per Digit Prostheses

The final Design Strategy in Architecture 2 seeks to provide better functionality via independent actuation of each digit. This strategy naturally requires at least five actuators and shares strong similarities with Architecture 1 hands. One such example [119] (Fig. 6d) contains five DC motors within the palm connected to planetary gearboxes arranged such that one motor actuates each digit. The thumb, index finger, and middle finger are actuated by identical motors and planetary gearboxes while the ring and pinky fingers are actuated by smaller ones. The output of each planetary gearbox is connected to a bevel gear that reorients the axis of rotation to drive the most proximal joint of each digit. Each of the four fingers is capable of MCP, PIP, and DIP F/E, whose rotations are coupled via planar linkages (APBJ(L)) while the thumb is capable of a combined CMC Opp and Ad/Ab rotation coupled with MCP and IP F/E via a combination of spatial and planar linkages (APBJ(L)).

Actuating each digit independently may help enable additional dexterity over the RIC-type hands, especially for the last three fingers. Employing smaller motors to actuate the ring and pinky fingers is an intelligent approach to reducing prosthesis weight since these two fingers generally do not need to match the force capabilities of the first three digits. This decision also enables each of the first three digits to be actuated with a larger motor, leading to higher pinch forces without compromising on dexterity. However, the spatial constraints from housing five motors within the palm limit maximum pinch forces to well below anthropomorphic capabilities and the prosthesis' maximum joint speeds are also below recommended values. Given the prosthesis' weight (between the weights of the median man and woman's hands), actuators that are more torque and power-dense may be needed to improve force and speed capabilities to recommended levels.

Many prostheses employing this design strategy also utilize a sixth actuator to provide additional independent actuation to the thumb. In many cases [125]–[129], either Opp, Ad/Ab, or a combination of the two were actuated independently of MCP and IP F/E. The additional actuators were often housed within the phalanges of the thumb to enable a serial connection (and simpler mechanical construction) and to avoid the spatial constraints that accompany housing an additional actuator within the palm [128]–[131]. Several commercially available hands also employ this Design Strategy [120]–[126], demonstrating that independent actuation of each finger is an appealing feature to amputees, despite the additional control complexity and higher price that can be associated with these prostheses. These prostheses also generally achieve better force capabilities than the other hands using this Design Strategy while still maintaining weights below the median male hand. In particular, the Ottobock Bebionic 3 Hand can achieve a maximum grasp force well above recommended grasping capabilities. These commercially available prostheses likely utilize actuators and transmissions with better force and torque densities to achieve these capabilities.

The Modular Prosthetic Limb (MPL) [131] (Fig. 6e) is a highly articulated prosthesis containing 10 actuators. Each finger is actuated by a BLDC located just under the MCP joint, providing coupled F/E of the three finger joints. The BLDCs have axes of rotation parallel to the MCP joint and are connected to spur gears that in turn drive a cycloidal reduction. The output of the cycloidal reduction is connected to a novel linkage mechanism located in the finger that enables adaptive, coupled rotation of the MCP, PIP, and DIP joints based on external loads applied to the finger [174] (ASCLJ(L)). The thumb has four DoFs that enable Opp (through a combination of CMC F/E and Ad/Ab), CMC Ad/Ab, and MCP and IP F/E. Each DoF is independently actuated by a BLDC connected to spur gears that drive a multi-stage planetary gearbox connected to each joint (ASPJ). These actuators and transmissions are connected serially, with the first located in the palm and the final three placed within the thumb. Finally, two additional BLDC actuators located in the palm with the same transmission pathway as used in the thumb provide Ad/Ab of the finger MCP joints via additional linkage mechanisms (ASPLJ(L)).

The MPL's performance in both force and speed capabilities is much closer to meeting anthropomorphic performance than most prostheses. In particular, its force capabilities are significantly higher than those reported for any other prosthesis reviewed in this paper. The hand's maximum grasp force surpasses the mean grasp strength for women. In addition, pinch capabilities either surpass or almost surpass those of the average man for all three types of pinches. The MPL's high strength does not come at the cost of low speeds as its maximum finger joint speeds exceeds recommended capabilities and is within the range of maximum continuous joint speed capabilities for the human index finger. These performance capabilities come partially at the cost of greater mass. The reported combined mass of the hand and wrist is 1300 g, which is significantly greater than mass of the average male hand (540 g) but under the mass of the average male forearm (1420 g). Unfortunately, the mass of the hand is not reported independent of the wrist, so it is difficult to speculate on the suitability of the prosthesis by itself for amputees with wrist disarticulation. However, the prosthesis (including the wrist) may attain the desired reductions in weight for some transradial amputees while also providing significant improvements in functionality, dexterity, hand strength, and grasping speeds.

The speed and force capabilities of this hand are the product of innovative actuator and transmission solutions. The chosen actuator is a compact, frameless BLDC that is small enough to be oriented such that its axis of rotation is parallel with most joints of the human finger. This is different than in most prostheses in Architectures 1 and 2, which typically have actuators oriented with their axes of rotation orthogonal to those of the joints they are actuating. By doing so, these other hands must reorient the axis of rotation through the use of additional transmission elements such as bevel gears, worm gears, linkages, or cable-based methods that can occupy significant volume and weight and impact efficiency. The chosen actuator in the MPL avoids this predicament, helping to enable higher performance. The choice of custom, compact cycloidal drives and planetary gearboxes are also important in enabling the prosthesis' performance capabilities. In particular, they provide large reductions (≥ 60:1) that enable the large output torques needed for the uniquely high-force capabilities of the prosthesis. Cycloidal drives can inherently provide large torques for a small size and weight because of larger regions of contact between the elements of the drive. Several design choices made in the planetary gearbox, including using more planets in the final stage, similarly enable the hand's high force capabilities within a small, lightweight package. Both of these transmissions appear to be more compact and potentially lighter than many commonly used, commercially available options with similar torque and speed output capabilities [175], [176].

Another important feature of the MPL is the high number of actuators housed within the hand. The prosthesis has more actuators than any other prosthesis within Architecture 2 and more than nearly every prosthesis across this review. This will likely enable the prosthesis to complete many grasp patterns that would not be possible for other prostheses; the decision to actuate each DoF of the thumb independently can be especially beneficial in this regard. Although there may be some hand gestures that the MPL hand will not be able to achieve, it is still able to adapt its grasps to irregularly shaped objects and demonstrate impressive joint speeds and joint forces [174].

The MPL demonstrates that Architecture 2 hands can attain a high level of performance, but at the cost of greater mass. One factor that may contribute to this greater mass is that the MPL team chose to use the same BLDC motor for all actuated joints within the hand. Instead of using the same actuator and transmission for every finger, it may be more desirable to optimize both actuator and transmission choice to the practical function of the thumb joints and fingers. Using the same actuator for all joints does however have the benefit of the same electrical interface and fabrication procedure throughout the hand, which can help reduce costs.

Overall, Architecture 2 presents significant flexibility in Design Strategy when compared to Architecture 1, owing to the palm's rectangular and relatively large volumetric space (compared to the digits). The diversity in Design Strategies demonstrates that an ideal solution that addresses all user needs has not been reached and is still needed. Instead, most designers have attempted to best address user needs with solutions that achieve a subset of the human hand's capabilities and functionality. It's likely that the varied performances, especially for commercially available prostheses, are most attractive for certain groups of amputees based on their user needs and other practical considerations such as lifestyle and cost. The vast majority of prostheses reviewed in this study (70 of the 96 reviewed prostheses) use this Architecture, most likely because of the flexibility it provides and its ability to be used by any transradial amputee regardless of the location of amputation (i.e. applicable for those with amputations near the wrist and near the elbow). These factors and the promising performance of several prostheses demonstrate that the Architecture should be investigated further and that new designs may eventually adequately resolve all user needs.

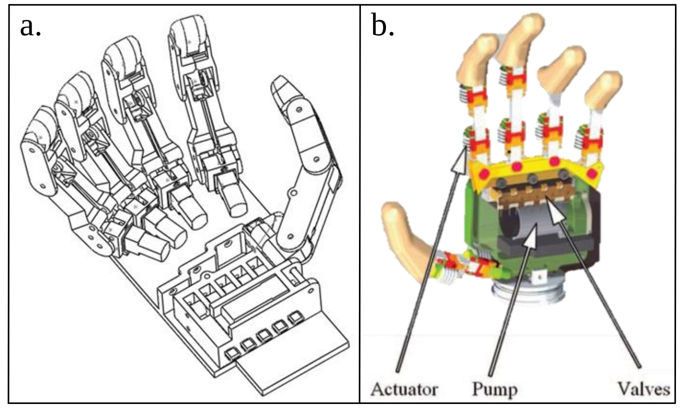

Fig. 7 Examples of Architecture 3 Prostheses: a. Unnamed [151], b. Fluidhand III [152]

C. Architecture 3 – Actuators Housed in Both the Digits and Palm

Architecture 3 is a hybrid of the first two Architectures that involves placing actuators in both the digits and the palm. In one example [151] (Fig. 7a), the prosthesis contains ten identical BDCs connected to spur gearboxes, with five in the palm and the other five located in the proximal phalanx of each digit; an additional five SMA actuators are housed within the palm. Each finger is capable of MCP, PIP, and DIP F/E while the thumb is capable of Opp, CMC Ad/Ab and MCP and IP F/E. One motor in the palm actuates the MCP of each finger via a set of bevel gears (ASBJ) while the motor in the proximal phalanx similarly actuates the PIP, which is coupled to the DIP via a closed cable system (ASBJ(Bt)). One SMA actuator provides additional F/E articulation to all three DoFs of each finger via a tendon connected to the intermediate phalanx (ATJ(T)). The thumb is actuated similarly, with a motor in the palm actuating CMC Ad/Ab (ASBJ) while the motor in the proximal phalanx actuates MCP and IP F/E (ASBJ(Bt)). An SMA actuator located in the palm provides coupled actuation of Opp, MCP, and IP F/E (ATJ(T)).

This arrangement of actuators provides a high degree of independent actuation to each DoF while maintaining a relatively simple mechanical design. This can lead to a greater number of achievable grasps and a higher degree of dexterity. The arrangement of actuators is also a great option when spatial constraints (limitation of both Architectures 1 and 2) impact the number and size of actuators that can be used. However, a prosthesis employing this Architecture and Design Strategy requires compact actuators and transmissions that are torque and power-dense in order to achieve desirable force and speed capabilities while maintaining a reasonable weight. This prosthesis demonstrates this challenge as it achieves a maximum key pinch force well below anthropomorphic capabilities. Unsurprisingly, this Architecture is subject to many of the same advantages and tradeoffs of both Architectures 1 and 2. However, its ability to also help

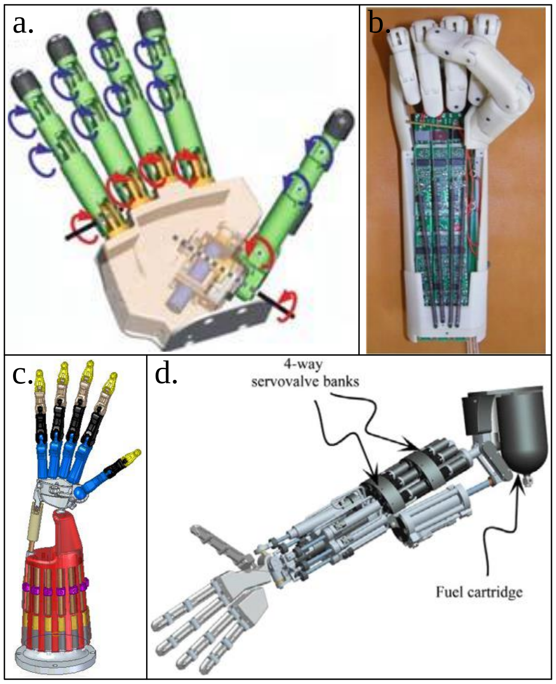

Fig. 8 Examples of Architecture 4 Prostheses: a. CyberHand [155] (red arrows indicate the most proximal joint driven by a single actuator), b. Unnamed [166], c. Unnamed [168], d. Unnamed [169]

overcome certain limitations of either of the two previous Architectures presents many interesting possible Design Strategies that have yet to be explored fully.

The Fluidhand III [152] (Fig. 7b) demonstrates an innovative Design Strategy that attempts to overcome these limitations through a novel approach to actuation involving hydraulic actuators. This prosthesis contains a single hydraulic pump within the palm of the hand that can modulate pressure and flow rate of water in a closed-loop system. Hydraulic valves housed in the palm control fluid flow into flexible fluidic actuators located at the joints they actuate. These actuators provide Opp, MCP F/E for all five digits, and PIP F/E for the index and middle fingers. The valves enable semi-independent motion of these DoFs that is mainly limited by the pump. Since the pump can only enable flow in a single direction at a time, the digits can only move together in a single direction. For example, it is not possible to flex one digit while simultaneously extending another. However, the use of hydraulics does enable a fingertip force that matches anthropomorphic capabilities and a hand closing time within the recommended range. The hand is also only slightly heavier than a median woman's. This prosthesis therefore demonstrates that hydraulics can offer a lightweight form of transmission that enables high forces. However, its force capabilities may be limited when multiple fingers are required in a grasp. Improvements such as using a pump that can provide higher output pressures may be needed in cases where the force output in various grasps is not satisfactory.

D. Architecture 4 – Actuators Housed in the Forearm

Architecture 4 consists of prostheses that house their actuators within the forearm, with transmission elements that transmit the actuator outputs from the forearm to the digits of the hand. Compared to the first three architectures, this design more closely resembles a natural human hand in terms of the location and distribution of actuators and transmissions. This helps naturally reduce inertia, with bulky and heavy actuators and transmissions in more proximal locations. However, they can only be used by those users with amputations proximal enough to accommodate these components.

One example is the CyberHand [155] (Fig. 8a), an anthropomorphic hand with five BDCs located in the forearm that each drive one underactuated digit and achieve adaptive grasping. Each motor is connected to a planetary gearbox that in turn drives a non-backdrivable leadscrew via a pair of spur gears. A slider driven by the leadscrew is connected to a tendon that terminates in the distal phalanx of each digit (APSScTJ(T)). Each motor drives a different digit through this transmission pathway, enabling coupled MCP, PIP, and DIP F/E in the four fingers and coupled CMC Ad/Ab and MCP and IP F/E in the thumb. An additional, smaller BDC and planetary gearbox are placed in the palm and drive Opp through a pair of spur gears connected to a worm gear (APSWJ). This approach leads to a very favorable distribution of weight in the prosthesis. The hand weighs less than a median woman's while the total weight of the prosthesis, including the actuators in the forearm, is lighter than the combined mass of a median man's hand and forearm. This prosthesis may therefore be suitable for some transradial amputees, especially those whose amputations are more proximal. The distribution of weight may also improve comfort and enable an amputee to wear the prosthesis longer. The maximum grasp force of the prosthesis is greater than the recommended minimum, but its joint speeds are well below recommended capabilities. Thus, actuators that are more power-dense may be needed to enable functionality that would make this prosthesis more attractive to amputees.

The Design Strategy of the CyberHand resembles that of the majority of prostheses in Architecture 4, which use various numbers of BDCs [155]–[159], BLDCs [168], or unspecified types of motors [160]–[165] housed within the forearm and connected to gearboxes; these actuate underactuated digits through tendon transmissions. While this approach may lead to difficulties in precision grasps and limited overall dexterity, many of these prostheses have similarly favorable weight distributions that may make them attractive options for transradial amputees. In certain cases, the prostheses are also able to achieve recommended joint speeds and grasp forces.

Several hands within Architecture 4 explore alternative actuation methods to conventional DC motors, which is partially enabled by the forearm's large volume and convenient shape. SMA actuators located in the forearm are utilized in [166]–[168] because of their low weight and size and high force capabilities. In [166] (Fig. 8b), six SMA actuators connect to tendons to actuate coupled MCP and PIP F/E of each of the four fingers (ATJ(T)), coupled MCP and IP F/E of the thumb (ATJ(T)), and independent CMC Ad/Ab of the thumb (ATJ). Each SMA actuator was made from NiTi wire and could only produce a force in a single direction. A bias mechanism was therefore employed to enable the actuator and digit to move back to their original positions. While this prosthesis weighed well below the combined weight of a median woman's hand and forearm, its force and speed capabilities were below recommended and anthropomorphic capabilities. A drawback of the actuators is the time needed for them to cool down (up to ~3s). This limits the speed with which the prosthesis can switch between grasps or postures (i.e. bandwidth). Another drawback of SMA actuators is low efficiency (possibly below 5% [166]), which leads to high power consumption during operation.

One prosthesis [168] (Fig. 8c) sought to overcome these disadvantages by pairing BLDCs with SMA actuators, utilizing five sets of opposing SMA actuators placed in the palm to drive Ad/Ab of the five digits and 11 BLDCs placed in the forearm that drive finger F/E and several thumb DoFs. This approach utilizes SMA actuators for DoFs that require high force outputs, but with low speeds, small displacements, and low bandwidth. By also using opposing sets of actuators, the prosthesis avoids some of the bandwidth limitations associated with cooling time in the above prostheses. Unfortunately, the performance of the SMA actuators and prosthesis Ad/Ab capabilities were not reported. However, the approach to providing Ad/Ab capabilities with lightweight, small-stroke, and high-force actuators should be explored further and may enable additional functionality that is useful for an amputee.

Another alternative actuator to DC motors is pneumatic actuators, which are utilized in [169] (Fig. 8d). This prosthetic arm is designed for transhumeral amputees and uses pneumatic actuators powered by monopropellant hydrogen peroxide. All the actuators are housed in the forearm, with five responsible for actuating the 17 joints in the hand through a tendon-pulley system (ATJ(T)). This actuation approach is attractive because it can produce higher force outputs than many electromagnetic actuators. However, the actuators require storage of fuel or a pressurized gas to power the actuators, which can be unsafe and can add significant weight to the prosthesis. Unfortunately, the force and speed capabilities of the prosthesis are not reported, making it difficult to compare these actuators to DC motors. However, the prosthesis, which also incorporates fuel storage and actuation for the wrist and elbow weighs 2000 g. This approach may therefore be lightweight enough to be feasible for some amputees if safety concerns can be mitigated.

Architecture 4 presents several advantages over the previous three Architectures. Among these, the most important may be the additional volume and weight available for actuators and transmissions, which could enable higher performance. The forearm also presents a cylindrical cross-section that may be more conducive for housing motors and other types of actuators. The favorable weight distribution seen in many of the prostheses in this Architecture also demonstrate that hybrid Architectures that place actuators in the forearm, palm, and fingers (similar to [168]) may also be feasible while still addressing the user need of reduced weight. This flexibility and the above advantages may make this Architecture the best option for adequately addressing user needs. While these prostheses can only be used by those with more proximal amputations, amputees may eventually choose to have suitable amputations if a prosthesis using this Architecture is able to address these improvements while other Architectures cannot.

IV. Discussion and Conclusion

The prostheses reviewed in this paper demonstrate that designers have been mindful of user needs. The intentions behind Design Strategies and an obvious emphasis on maintaining an anthropomorphic weight and size make this mindfulness clear. Despite these intentions, the vast majority of the prostheses are unable to achieve both recommended force and speed capabilities and none can achieve both anthropomorphic force and speed capabilities while maintaining an anthropomorphic weight. Thus, the required improvements in actuation to achieve this while maintaining a desirable weight and size remain an important design challenge for future prostheses. In many cases however, prostheses achieved acceptable speed capabilities but fell well short of desired force capabilities. A special focus may therefore be needed to provide improvements to actuation that enable sufficient force capabilities.

Further evaluation using a consistent grasp taxonomy and control architecture are also needed to understand whether the prostheses achieve acceptable levels of dexterity and functionality beyond force and speed capabilities. While many prostheses do have a similar number and arrangement of actuated DoFs as the human hand, most do not include enough actuators to enable the same level of independence in these DoFs. Further development in actuation may therefore be needed to achieve the levels of dexterity and functionality that users desire. Thus, adequately addressing user needs in future prostheses requires continued investigation and innovation to improve the performance of actuators and transmissions in prostheses. To do so, an examination of similarities in actuation across prostheses is a logical first step.

DC motors are used as the primary actuator for more than 72% of the reviewed prostheses. This is not surprising since DC motors offer specific powers (i.e. power output per unit mass) and power densities (i.e. power output per unit volume) that exceed the capabilities of human muscle [41], [42], [175], [176]. While other actuators can offer sufficiently high power densities and/or specific powers, few can also achieve the power outputs and displacements needed to actuate finger joints. Furthermore, DC motors are conveniently shaped, allowing them to be used in any of the Architectures discussed above. There are also many commercially available options, which can be cost-effective, convenient to use, and customizable (including customizable transmissions). These reasons, among others, make DC motors very attractive for use in a prosthesis. However, certain inherent disadvantages have helped prevent prostheses using DC motors from adequately addressing user needs.

A survey of two motor manufacturers' (Faulhaber and Maxon) catalogs of BDCs and BLDCs [175], [176], which were commonly used in the reviewed prostheses, revealed that for all motors under 60 mm in diameter (approximately the width of the wrist) and under 550 g in weight (approximately the weight of the human hand), maximum continuous torque outputs ranged from $1\text{x}10^{-5}$-0.2 Nm and maximum speed outputs ranged from 200-7800 rad/s. Thus, compared to recommended prosthesis capabilities, DC motors typically achieve much higher speeds and lower torques. This means that a significant reduction ratio, potentially greater than 100:1, is needed to convert the power output of each motor into the torque capabilities of any of the human hand joints [77], [109], [110], [165]. For DC motors with sufficiently high specific powers and power densities, the accompanying transmission pathway's weight, volume, efficiency, reduction ratio, and output force and speed performance will dictate whether the prosthesis can adequately address user needs. Unfortunately, many of the most commonly used transmission solutions do not provide adequate performance in these metrics to enable the prosthesis to address user needs. Thus, further examination of and innovation in transmission pathways offers the most promising avenue to enable prosthetic hands to adequately address user needs.

Geared transmissions, and in particular planetary gearboxes, are commonly used in the reviewed prostheses' transmission pathways. Planetary gearboxes are favored for their compact size, high efficiency, ability to achieve high reduction ratios, and commercial availability, among other reasons. However, an analysis of Maxon and Faulhaber catalogs reveal that if each joint were to be actuated by a separate DC motor-planetary gearbox combination that can achieve anthropomorphic torque (Fig. 3) and recommended speed (Table 4) capabilities, the total required mass for actuators and gearboxes would be over 2kg. This is greater than the combined weight of a median man's hand and forearm and thus would not be an attractive method of actuation for a prosthesis. However, only 30-40% of the mass required for actuation would be occupied by the motors, meaning transmissions would account for most of the mass. These results are not surprising as planetary gearboxes utilize small gears, which limit the maximum torque output. In order to increase torque output, the size of the gears must increase, leading to gearboxes that are larger and heavier. This makes planetary gearboxes better suited for DoFs requiring lower torques, or only as the first steps in a transmission pathway. However, gears are used as the last step in a transmission pathway in at least 33% of the reviewed prostheses; gears were also frequently combined with underactuated tendon systems (which often offer low mechanical advantages) as the last two steps in a transmission pathway. This placement of gears at the end of transmission pathways helps to explain why so many prostheses fall short of desired force capabilities.

The above tradeoff of planetary gearboxes, between torque output and both weight and volume, suggests that further investigation of transmission options that can achieve higher force/torque outputs at a smaller weight may offer better alternatives that in turn can lead to prostheses that adequately address user needs. Options such as linkages, lead- and ball-screws, and parallel kinematic mechanisms (PKMs) may be able to offer these capabilities. Linkages are found in many prostheses but are usually used to ensure phalanges rotate in certain sequences or achieve specific postures [177]. However, linkages can also provide significant reduction ratios while transmitting large forces/torques in a lightweight, compact package [178], [179]. Lead- and ball-screws can also achieve relatively large force outputs for small input torques. The small travel range required from these screws in a prosthesis means they can provide these capabilities within a small, lightweight package. PKMs can offer the ability to articulate multiple DoFs with ground-mounted (and therefore more proximal) actuation in a compact, lightweight package [180]–[182]. These mechanisms can also offer significant performance improvements over Serial Kinematic Mechanisms for joints with multiple independently actuated DoFs (e.g. CMC, MCP joints) [182], [183]. Variable or automatic transmissions, which have been used previously for more than one actuator in robotic hands [184], [185], offer another alternative and can reduce actuator mass. This is because the primary actuator no longer needs to provide the desired torque or speed outputs with a single reduction ratio. Instead, two or more different reduction ratios can be employed to achieve different desired outputs. However, this option is only attractive if the automatic functionality can be provided for sufficiently small mass and electrical power. Most of the above-suggested transmission options can enable a lower mass, but potentially at the cost of a larger required volume. This could make Architecture 4 and other hybrid Architectures utilizing the forearm, which naturally offer a larger volume, attractive options.

These Architectures may be necessary simply because of the weight of actuators required to actuate each joint – at least 650g in the analysis of DC motors and planetary gearboxes described above. Further innovation in DC motor design may enable a significant reduction in weight that makes accommodating a large actuation weight unnecessary. However, both BDCs and BLDCs have been studied extensively, making the required improvement unlikely. Innovation in alternative forms of actuation could therefore offer a more viable path forward. In particular, actuators that offer performance more similar to a human muscle (i.e. higher output force/torque and lower output speed) would be desirable as they would concurrently reduce the required size, weight, and reduction ratio of the required transmission pathway. Most current actuator options with these traits (e.g. piezo actuators, hydraulic/pneumatic actuators, and artificial muscles) do not yet offer high enough power densities, specific powers, and power outputs with sufficient force, speed, strain, displacements, bandwidth, precision, and efficiency when combined with required power supplies, drivers, and additional equipment [41], [186].

Finally, innovations in Architecture and Design Strategy can play a critical role in satisfying user needs. For example, Design Strategies optimized for specific sets of tasks (similar to the Michelangelo Hand and Axon Hook) may offer the simplest solution to achieving desired force and speed capabilities while also addressing other user needs. Design Strategies for new hybrid Architectures (including those utilizing the forearm) can also offer many new solutions and improved performance. Despite the large number of prostheses that have been created in just the past 20 years, the field is far from saturated. Many of the reviewed prostheses share significant similarities across Architecture, Design Strategy, actuation, and transmission pathway. This leaves the door open, especially for new Design Strategies (including those using different actuators) and transmission pathways, that can achieve significantly more than incremental improvement. This may require novel approaches to the design problem or a detailed examination of the fundamental capabilities of different types of actuators and transmissions.

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