The 2009 IEEE/RSJ International Conference on
Intelligent Robots and Systems
October 11-15, 2009 St. Louis, USA

Mesofluidic Actuation for Articulated Finger and Hand Prosthetics

Lonnie J. Love, *Member, IEEE*, Randall F. Lind and John F. Jansen, *Member, IEEE*

***Abstract*—The loss of fingers and hands severely limits career and lifestyle options for the amputee. Unfortunately, while there have been strides made in advancements of upper arm and leg prosthetics, the state of the art in prosthetic hands is lagging far behind. Options are generally limited to claw like devices that provide limited gripping capacity. The overall objective of this paper is to demonstrate a path towards a low-cost prosthetic hand with multiple articulated fingers and a thumb that rivals the human hand in terms of weight, size, dexterity, range of motion, force carrying capacity and speed. We begin with a description of the functional requirements for a human hand. When comparing requirements with actuation technologies, the fluid power approach has the potential to realize a prosthetic hand that rivals a human hand in size, strength and dexterity. We introduce a new actuation technology, mesofluidics, that focuses on miniaturization of fluid power to the meso-scale (mm to cm). As a novel demonstration of the potential for this technology, we describe a proof-of-principle mesofluidic finger that has intrinsic actuation and control (actuators and control valves within the volume of the finger). This finger weighs 63 grams, is sized to the 50th percentile male finger, has a total of 25 mechanical parts and is capable of providing 10 kg (22 lbs) of pinch force.**

I. Background and Motivation

A. State of the Art

The fingers and hand are the primary link between a human and the physical world. We use our hands to grasp tools, open jars, assemble small components, type, eat, inspect, play music… almost every aspect of our personal and professional lives is impacted by our hands and fingers. Subsequently, the loss of a hand can dramatically impact a person's life. The primary functions of the hand include manipulating, transporting and feeling objects [1]. The flexibility and redundancy of the hand enable a wide variety of configurations for grasping objects of varying shapes and sizes. Napier classified these grips into two basic categories: power and precision grip [2]. Power grip is associated with firmly holding objects within the hand while precision grip focuses on the fine manipulation of objects held between the thumb and index finger. Levanie and Norkin expanded each of these grips into six separate hand configurations shown in Figs. 1a and 1b [3]. These six basic configurations enable a wide range of manipulation capabilities for the hand.

Fig. 1a. Power grip: (a) cylindrical, (b) spherical and (c) hook (from [3]).

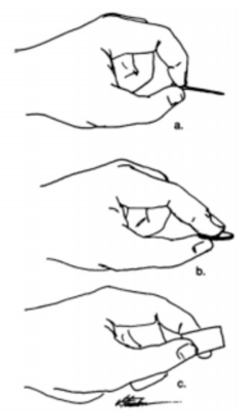

Fig. 1b. Precision grip: (a) tip-to-tip, (b) pad-to-pad, (c) pad-to-side (from[3]).

This wide range of flexibility comes at a cost. The human hand has a total of twenty four degrees of freedom (DOF) packed inside a volume generally less than 500 cubic cm and weighing less than 500 grams. Extrinsic muscles, located in the forearm, transmit forces to the fingers through tendons attached to the hand. The primary role of intrinsic finger muscles, located in the palm of the hand, is to precisely control the direction of fingertip force while the role of extrinsic muscles, located in the forearm, is to provide stability of the joints [4]. Many existing research robotic hand devices, such as the Utah/MIT and Salisbury hands, follow the same basic model of locating actuators in the forearm and using tendons to remotely drive fingers.

This work was supported in part by the Defense Applied Research Projects Agency (DARPA) under the Revolutionizing Prosthetics Program, contract 1868-HH69-X1.

Lonnie J. Love is with the Oak Ridge National Laboratory (ORNL), Oak Ridge, TN 37831-6305 USA (phone: 865-576-4630, fax: 865-574-4624, e-mail: lovelj@ornl.gov). ORNL is managed by UT-Battelle for the U.S. Department of Energy under contract DE-AC05-00OR22725.

J. F. Jansen is with the Oak Ridge National Laboratory, Oak Ridge, TN 37831-6305 USA (e-mail: jansenjf@ornl.gov).

R. F. Lind is with the Oak Ridge National Laboratory, Oak Ridge, TN 37831-6305 (e-mail: lindrf@ornl.gov).

Compounding the complexity of the human hand, there is an additional constraint in the design of a prosthetic hand and arm. A survey of existing commercial prosthetic hands shows similarity in weight between the prosthetic device and the human hand: Otto Bock SensorHand (460 g), Touch Bionics i_Limb (200 g), TRS Lite Touch (284 g), RSLStepper (500 g). The entire prosthetic device (structure, actuation and power source) must weigh less than the actual hand or arm that it is replacing. The primary motivation for the reduction in weight has to do with discomfort associated with load forces and torques transmitted through the prosthetic socket. There are three possible paths for transferring loads: through the skin, an external structure, or through the bone. To date, most prosthetic devices transfer the load through a socket to the skin on the stump. However, skin pressure exceeding a few pounds per square inch results in discomfort and skin abrasions and even damage to healthy tendons. Active prosthetic devices will further increase the loads on the soft-tissue interface. An external structure, such as an orthotic device, provides a mechanical link for distribution of the load and could provide an immediate solution for extending the load-bearing capacity of prosthetic devices. Clearly, the ideal interface would transfer loads directly to the skeletal structure. The most promising procedure to date, osseointegration, consists of implanting a titanium shaft in the bone which will serve as the mechanical interface between the prosthetic and skeletal structure. Properties of pure titanium promote integration of the bone tissue and the implant. This procedure is still in the research phase but is rapidly approaching a reality in the U.S. For now, if no internal (e.g. osseointegration) or external structure (e.g. orthodic device) is utilized, then any new prosthetic design must continue to weigh significantly less (~ 2/3 less) than a human arm which means that any actuation system that will provide equivalent performance as a healthy arm must have a power density exceeding that of human muscle to compensate for the significant reduction in overall weight.

Weight constraints have driven most recent advancements in prosthetics in the materials area. However, almost all commercial prosthetic hands and arms are limited to just one to two DOF. The impact of structural weight reduction has done little to impact actuation and DOF. Other practical problems exist such as cost and robustness. Many advances have been made in the area of prosthetics but the costs are prohibitively expensive for the majority of the population. Furthermore, the systems must be capable to stand up to everyday use and abuse. Clearly there is a strong need for additional DOF while constraining the mass and cost of the system. The following section reviews the basic requirements for a prosthetic hand in terms of specific metrics and targets. These targets, coupled with existing constraints (volume and weight), provide clear motivation for advancements in the area of actuation.

II. FUNCTIONAL REQUIREMENTS

As with any manipulation system, there are a number of specifications that dictate capabilities and performance. These specifications generally include degrees of freedom, range of motion, peak force, maximum speed, bandwidth and stiffness. A survey of hand performance concluded that working in cold environments, using vibrating machines, handwriting, picking up small objects, opening lids, lifting and carrying activities are considered the most difficult to accomplish [5].

We surveyed the literature and tabulated target requirements that include range of motion, speed, tip force, torque and stiffness. Range of motion data for the fingers and thumb are well reported in the literature [6], [7], [8]. There is some variability in the literature but the values in Table 1 capture the general trend for most healthy fingers.

TABLE 1
FINGER JOINT RANGE OF MOTION (UNITS IN DEGREES)

	Thumb	Index	Middle	Ring	Little
MCP abb./add.	0 to 90	-30 to 30	-20 to 20	-30 to 30	-30 to 30
MCP flex	0 to 100	-30 to 90	-30 to 90	-30 to 90	-30 to 90
PIP		0 to 110	0 to 110	0 to 110	0 to 110
DIP	0 to 90	0 to 70	0 to 70	0 to 70	0 to 70

There have been many studies that have examined finger forces during various activities. Schenk and Mai conducted a survey in which they identified the average grip force of a healthy adult during handwriting activities and showed approximately 10 N of peak grip force with a peak speed of 0.1 m/s [9]. Another example of a daily activity involving the fingers is opening containers. Crawford et al. showed torque requirements ranging from of 1.3 N-m (opening a coffee container) to 6.2 N-m (cap on a jar of marmalade) [10]. Nikonovas et al. used a novel instrumented glove to measure finger forces while hitting a golf ball [11]. The highest force (ring finger at prior to impact) was 15.8 N with the overall (summed) grip force peak of 21.2 N. The distal and proximal phalanx of the thumb experienced peak forces of 16.5 and 9.5 N respectively. As an example, torques as high as 0.6 N-m and 1 N-m have been recorded for the proximal interphalangeal (PIP) and metacarpolphalangeal (MCP) joints respectively during jar cap opening activities [12]. However, precision requires not only control of forces, but regulation of stiffness. Finger stiffness is controlled by varying the joint angles and/or the combination of muscle forces to achieve both objectives [13]. Milner and Franklin provided an excellent study of finger stiffness and force capacity as a function of finger posture [14]. Their studies focused on the right index finger in both the extended and flexed posture, varying the direction of the force in the finger plane. The results showed a maximum finger force ranging from 10.5 to 49.6 N as a function of finger posture and force direction. They used these tests to estimate both peak joint torque and joint stiffness, summarized in Table 2.

TABLE 2
PEAK INDEX FINGER TORQUE AND STIFFNESS

	MCP		PIP		DIP	
	Torque (N-m)	Stiffness (N-m/rad)	Torque (N-m)	Stiffness (N-m/rad)	Torque (N-m)	Stiffness (N-m/rad)
Extended Finger	0.8	7.25	0.75	1.9	0.14	0.35
Flexed Finger	0.52	5.08	0.35	2.48	0.16	0.28

Many activities require not only control of forces, but control of speed. The most common examples include playing music and typing. Almost all careers require some interaction with computers through keyboards. Subsequently there have been numerous studies focusing on finger articulation and keyboard layout [15],[15]. Proficient typists can easily exceed 500 keystrokes per minute. If all fingers are utilized equally and a keystroke required total angular transition of 180 degrees, this results in close to one keystroke per finger per second or 2.6 rad/sec. Darling performed a more rigorous analysis of finger joint speed in which he found the maximum joint speed for the MCP joint was approximately 18 rad/sec while the maximum joint speed for the PIP joint was 12 rad/sec. Finally, we can combine the peak joint speed with the peak joint torque to establish an upper bound on joint power requirements assuming zero speed at peak torque and zero torque at peak speed. The peak power for the MCP and PIP joints are 7.2 W and 4.5 W respectively. While these peak powers are typically of short duration, they can impose a serious demand on the power source.

A. Actuation Comparison

The previous section highlighted the functional requirements for finger and hand prosthetic actuation. The fact that the complete prosthetic system must not only weigh less than the healthy limb, but also include the energy source, suggests that the form of actuation must have a power density and stress capacity that significantly exceeds the human muscle by almost an order of magnitude. Furthermore, some procedures (such as the transradial disarticulation) require that the actuators be collocated with the joint (compared to the tendon drive on the human finger). The literature is rich with comparisons of actuation technologies [17]-[21]. Typical examples include conventional actuators (electromagnetic, pneumatic, hydraulic) and smart materials (piezoelectric, magnetostrictive, shape memory alloys and electroactive polymers). Metrics used to compare technologies include stress (normalized force), strain (normalized displacement), specific power (normalized mechanical power), bandwidth (responsiveness) and stiffness (load holding ability). Table 3 summarizes these metrics for the above actuators including the human muscle. Many of the smart materials (piezoelectric, magnetostrictive and shape memory) provide superior stress but suffer with extremely low strain. While there are examples of interesting mechanical amplification techniques to expand strain, the size and weight of these devices tend to overwhelm any advantage over other actuators. Pneumatic actuators are attractive in terms of stress and strain but the compliance of the air impacts accuracy and load holding capacity. One of the more interesting advancements in recent years is Electroactive Polymers (EAP). Like pneumatics, they tend to be very compliant but also require very high voltage (~150 MV/m).

TABLE 3
ACTUATOR COMPARISON

Actuator	Strain (%)	Stress (MPa)	Specific Power (W/kg)	Bandwidth (Hz)	Stiffness (MPa)
Muscle	20	0.35	50	30	20
Electromagnetic	50	0.035	200	30	0.1
Pneumatic	50	0.69	200	50	0.1
Piezoelectric	0.2	110	0.1	kHz	400
Magnetostrictive	6	9	5	kHz	29
EAP	380	3	35	10	1
Shape Memory	8	200	6	1	$83e^3$
Hydraulic	70	20.8	2000	50	1380

However, scaling clearly plays a role in actuation and some of the conclusions drawn for traditional sized actuators may not hold for meso-scaled system. Almost all prosthetic devices today are electromagnetic. However, no clear path for miniaturization exists for electric drives with cable transmission systems. Figures 2 and 3 show the rotor stress and power density for commercial electric motors as a function of scale. It is clear that the rotor stress (torque) decreases with scale. The only way to increase power is by operating the motor at higher speeds. This introduces a subtle problem for prosthetic fingers: transmission. Electric motors are low torque/high speed actuators. Therefore the motors must also include a transmission with a high transmission ratio to transform the low torque/high speed of the motor to the high torque/low speed of the joint. Steltz shows that the power density of electric motors not only significantly drops with scale, but the addition of the transmission further reduces the power density by an average factor of 2 and efficiency suffers [22]. The high transmission ratio introduces significant amounts of friction and compliance in the joint.

It should be clear that new actuators are needed to provide higher power density and direct drive capability to enable future intrinsically actuated fingers. In comparing existing actuation technologies, no present method of actuation can provide the required force, displacement and bandwidth (high stiffness with low mass) required for finger joints within the volumetric constraints of the hand. The metrics clearly indicate that fluidic systems has the near-term potential to achieve the required stress, strain, stiffness and bandwidth while also providing low friction (direct drive), low effective mass, and compact packaging required for future prosthetic devices. The major technical challenges with fluidic actuation at this scale relate to scaling down the technology to loads under 5 kg (~10 lbs) compared to hundreds and thousands of pounds for commercial applications and to sub-fractional horsepower levels. In essence, what are the limits of performance of hydraulics at these scales and how does one precisely control the minute amount of fluid flow? The following section provides an overview of miniaturized fluid based actuation for prosthetics. Emphasis is placed on the motivation, advantages and challenges associated with mesoscale fluid power, leaving details on control for subsequent publications.

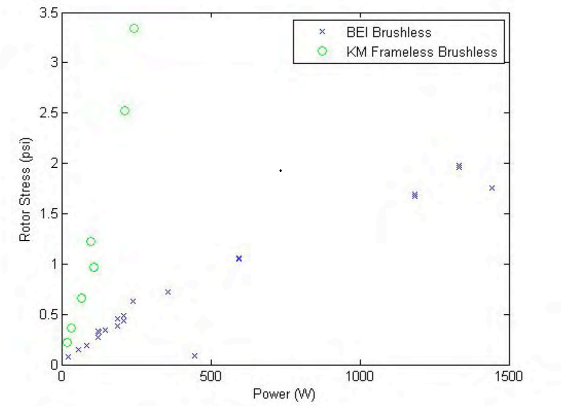

Fig. 2. Rotor stress vs. motor power.

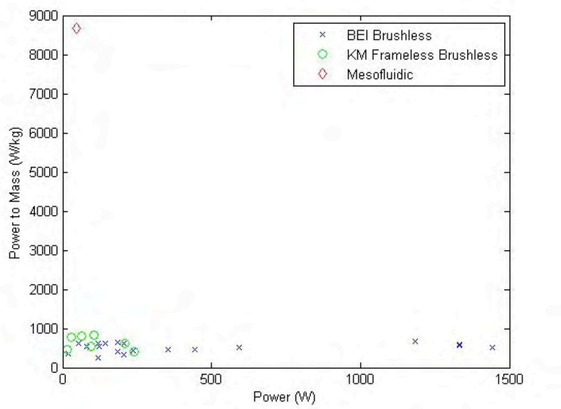

Fig. 3. Power density vs. motor power.

A. *Mesofluidics*

Mesofluidic actuators are fluid based actuators that range from a few millimeters to centimeters in size and use pressurized fluid for the motive force (see Fig. 4). Mesofluidics can provide high force density (>1000 psi), low friction, direct drive, high mechanical bandwidth and use a variety of working fluids ranging from oil to water. As an example, the smaller actuator in Fig. 4 is 2.3 mm (0.09") in diameter, provides 1.09 kg (2.4 lbs) of force with 7.6 mm (0.3") of displacement while the larger actuator, 9.6 mm (0.38") in diameter, provides 8.9 kg (19.8 lbs) of force with 25.4 mm (1.0") of displacement. Both have a dynamic response exceeding equivalent human muscle actuation. In fluid-powered systems, pressure controls force while flow rate controls speed. Clearly, the flow control element is the single most important device in fluidics. Figure 5 shows that there are ample examples of high pressure/high flow valves and low pressure low/flow valves. The enabling technology for mesofluidics is the development of high pressure/low flow valves (see Fig. 6). In terms of prosthetic fingers and thumbs, mesofluidic actuation is one of the most promising new enabling technologies in terms of providing high performance actuation within the volumetric constraints of the human fingers and hand. Furthermore, technologies developed for fingers and hands have the potential to be scaled up to impact larger joints such as ankles, wrists, elbows, shoulders and knees.

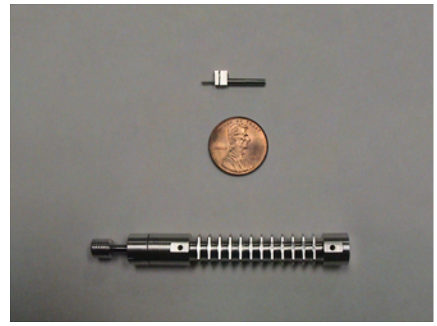

Fig. 4. ORNL mesofluidic actuators.

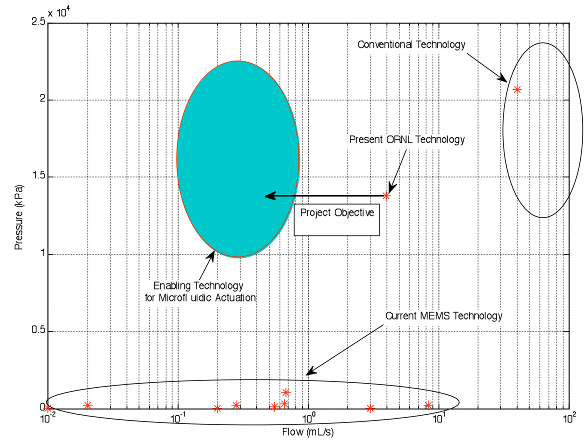

Fig. 5. Flow control technologies.

Fig. 6. ORNL mesofluidic valve.

III. Mesofluidic Finger

The human hands, as well as almost all prosthetic hands, are based on extrinsic actuation where the muscles, or actuators, are located in the limb prior to the joint. Muscles for most of the fingers are located within the forearm. This presents a fundamental problem in prosthetics. A transradial disarticulation (loss below the elbow) limits the volume of the prosthetic device to the wrist and palm of the hand. There is simply no volume to locate the actuators as well as power source. There is strong motivation to create an intrinsic finger (locating the actuators within the volume of the fingers). The advantages of such a design go well beyond volume and weight. Performance and low complexity are benefits as well. Since the actuators are collocated with the joint, there are no tendons or cables which add complexity and compliance. The mesofluidic finger, shown in Fig. 7, is designed to provide 9.06 kg (20 lbs) of pinch force with an operating pressure of 13.8 MPa (2000 psi). Each joint has antagonistic actuation which eliminates backlash. The flexing actuator (above the joint) is controlled by the valves (one valve regulates high pressure flow into the actuator, the second valve regulates flow out of the finger). The extending actuators (below the joint) always see system pressure. This approach to controling the joint reduces the complexity of the controls and simplifies assembly. There are no rod seals since both pistons only see pressure on one side of their pistons (both push against the joint). Table 3 lists the design characteristic for each of the three joints on the finger. It is clear, based on the sizing of the actuator (bore and stroke), that fludics is an ideal fit for prosthetic fingers that rival human fingers in size, dexterity and strength.

TABLE 3
FINGER JOINT PARAMETERS

Parameter Joint	Torque (N-m)	ROM (deg)	Speed (deg/s)	Moment arm (mm)	Actuator Force (N)	Actuator Stroke (mm)	Actuator Bore (mm)
MCP	6.8	120	120	8.7	980	18.5	9.5
PIP	3.4	90	120	7.4	430	11.7	6.3
DIP	1.7	70	120	4.0	430	5.0	6.3

To demonstrate the strength of the finger, we set up a test stand, see Fig. 7, where the finger presses down on a load cell. We varied the pressure between 6.9 MPa (1000 psi) and 10.3 MPa (2000 psi). The finger is commanded to move to the load cell, provide pressure, then return. We repeat this multiple times at multiple speeds. The results are displayed in Fig. 8. The peak force at 6.9 MPa (1000 psi) is 5.9 kg (13 lbs), at 10.3 MPa (1500 psi) the peak force is 8.6 kg (18.9 lbs) and finally at 13.8 MPa (2000 psi) the peak force is 10.2 kg (22.5 lbs). Figure 9 displays the finger on a scale, showing it's weight at 62.7 grams (0.14 lbs). The resulting payload to weight ratio of the finger is 163:1.

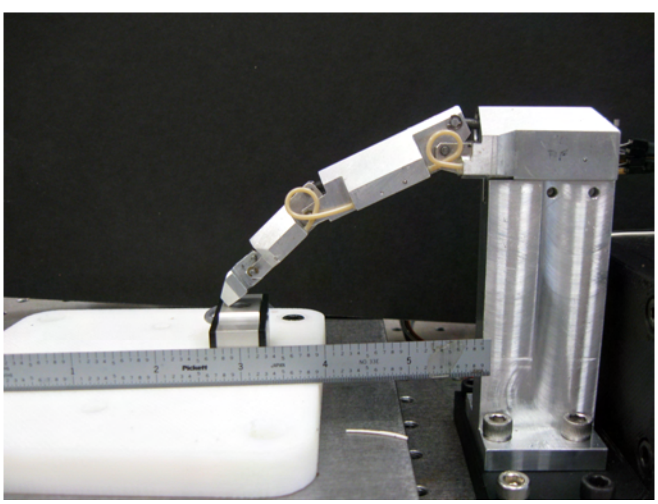

Fig. 7. Mesofluidic finger.

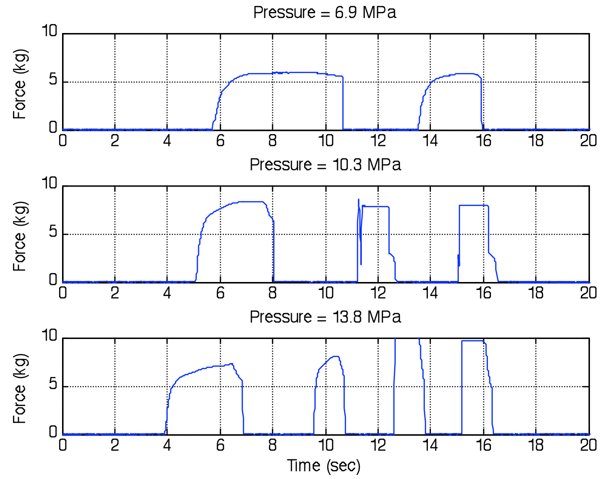

Fig. 8: Finger Force as a function of pressure

Fig. 9: Finger on scale

IV. CONCLUSION

The motivation for this paper is to provide the reader with the vision of the development of a prosthetic hand that rivals the human hand in terms of size, dexterity and force capacity. We provide a description of the functional requirements that, when compared to actuation technology, strongly suggests that the fluid power approach has the potential to realize these requirements. This is especially significant when considering transradial disarticulations where the actuator must be collocated with the finger joints. We describe a new actuation technology, mesofluidics, that focuses on enabling the miniaturization of conventional fluid power. We conclude by describing a new prosthetic finger, developed under DARPA's Revolutionizing Prosthetics Program, that rivals the human finger in dexterity and force capacity while achieving the goal of intrinsic actuation [23]. Present research efforts are focusing on integration of the finger into a mesofluidic hand (see Fig. 10) with applications in the areas of prosthetics, robotics and explosive ordnance disposal.

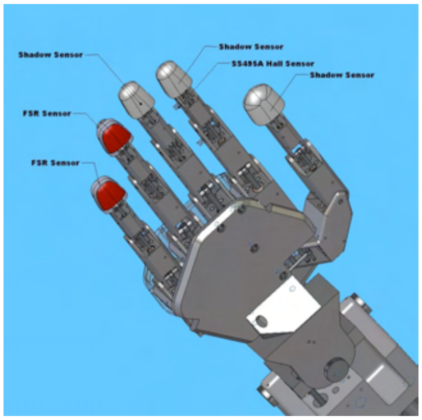

Fig. 10. Mesofluidic hand

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