DEVELOPMENT AND PERFORMANCE EVALUATION OF A PROTOTYPE LINEAR ACTUATOR DRIVEN ARTIFICIAL HEART

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الملخص

تعرض هذه الورقة استحداث نموذج عملي لجهاز تدعيم للبطين الأيسر باستخدام نوع جديد من المحركات الخطية الكهرومغناطيسية. يتميز المحرك المقترح بقلة أجزاءه وبساطة تراكيبه ويستطيع أداء الحركة الأساسية لقلب صناعي نبضي دون الحاجة لمحول حركة. معظم أجزاء النموذج إما أنشئت باستخدام وسائل تصنيع اعتيادية أو أخذت من الموارد المتاحة.

تم استخدام خلية أحمال لقياس القوة الدافعة الساكنة للمحرك، كما تم إنشاء دورة دموية زائفة مبسطة لغرض تقييم الأداء الحركي للنموذج. تم تشغيل النموذج لأكثر من 72 ساعة، تشمل أكثر من ثلاث ساعات من التشغيل المستمر. تمكن النموذج من ضخ حوالي 1.14 لتر في الدقيقة ضد ضغط يعادل 100 ملم زئبقي عند إمداده بطاقة كهربائية تقارب 20 وات وذلك عند معدل نبضات يعادل 60 نبضة في الدقيقة. تم إثبات وظائفية المحرك المقترح عمليا و لكنه لم يتمكن من تحقيق المتطلبات كقلب صناعي. بعض المشاكل التي تسببت في انخفاض أداء النموذج تم مناقشتها في نهاية هذه الورقة.

ABSTRACT

This paper presents the development of a new prototype of left ventricular assist device (LVAD) with linear electromagnetic actuator. The proposed actuator consists of very few components and has a very simple structure. The proposed actuator can provide the back and forth motion required for the basic operation of a pulsatile artificial heart without any movement converter. The prototype pump components are fabricated either using ordinary machining techniques or taken from available resources.

Static force produced by the actuator is measured using a load-cell and a simple mock circulatory loop is built to evaluate the steady state dynamic performance of the prototype pump. Accumulative operation hours were more than 72.0 hrs with more than 3.0 hrs of continuous operation. When a steady power of ~ 20 W was fed to the pump, at a pumping rate of 60 Pulse per minute, the pump output was ~1.14 litre /min against ~ 100 mm Hg. The functionality of the actuator is experimentally validated; however, the prototype pump could not clear the requirement of an artificial heart. Problems encountered in developing the prototype resulted in its low performance and are discussed at the end of this paper.

KEYWORDS: Pulsatile artificial heart; Ventricular assist device; Performance evaluation; Linear reluctance motor

INTRODUCTION

Mechanical circulatory support devices are blood pumps that support or replace the native heart. The most common application is the left ventricular assist device, which supports the native left ventricle. When applied as a total artificial heart, the native heart -in most designs- is removed and two pumping chambers are used.

The use of temporary mechanical circulatory assist devices in adults and children is now routine, and the development of permanent ventricular assist devices (VADs) and total artificial hearts (TAHs) is the subject of ongoing research in many countries. It is possible to maintain a paraphysiological circulation for effective periods using continuous and pulsatile flow devices.

The majority of circulatory support devices in use today are pulsatile, volume displacement pumps, which show superiority of survival rates over continuous flow devices [1]. Energy sources for the currently used devices are pneumatic and electric. Some investigators [2,3] state that the potential for size reduction to permit implantation of a pneumatic system is not as great as the potential for an implantable electrically activated device. While implantation of pneumatic pumps continues, development work is heavily devoted to electrical motors to enable implantation for long periods with improved mobility and quality of life.

Many different direct drives (including linear actuators) were tried and used for driving a VAD or a TAH. Such drives aim to generate pulsatile flows with simple construction, and high reliability. Some of them have just been tried experimentally and others have entered clinical applications and gained potential success.

Kitamura et al. [4] employed a commutator type motor with brushes for driving a pneumatic system. A pulsatile airflow is produced by a piston moving back and forth, which, at the same time, causes the pump diaphragm to move back and forth. Thus, pulsatile blood flows are generated throughout its unidirectional valves.

Portner et al. [5] used a pair of laminated core magnets and coils that face one another. Energizing the coils produces a magnetic field, attracting the magnets toward one another. As the magnets move one another, they bush on pressure plates, which squeeze the sac. Blood then squirts through the outflow port until the sac empties. Pivoted beam springs aids in refilling the blood sac again. The described solenoid actuator is used in Novacor VAD.

J. Cathey et al. [6] has built a tubular self-synchronous motor for artificial heart pump drive as an experimental model. In his model, the motor directly drives a piston that acts on a silicon fluid. The fluid in turns acts on diaphragms, which actually pump the blood. Yamada et al. [7] built a linear pulse motor to drive a total artificial heart. The TAH pumps the blood by expanding and compressing the sacs according to the reciprocation motion of the mover.

Watada et al. [8] proposed a linear oscillatory actuator for total artificial heart drive system. The actuator has single-winding excitation coil on the stator and two permanent magnets. Forward and reverse motions are available by alternating the exciting current. Fukunaga et al. [9] developed another type of linear oscillatory actuator -moving magnet- in driving an LVAD. The mover moves back and forth when forward and backward electric current is supplied to the excitation coil.

The magnetic drive used in Novacor II LVAS [10] is yet another example of direct drive-driven artificial heart. The bias field (permanent magnet) is modulated by a coil driven field with high-energy conversion efficiency. Each of the two chambers alternately pumps by the shared pusher plate.

METHODOLOGY

In the present work, a simple construction linear electromagnetic actuator is proposed to drive a pulsatile displacement pump. Preliminary design calculations were carried out, under simplified assumptions, to predict its feasibility to perform the specified job before hardware is produced. A prototype was fabricated for experimental work purposes and a simple mock circulatory loop was built to check its dynamic performance characteristics. Description of the prototype pump is presented thereafter:

a) Actuator

Figure (1) is a schematic drawing of the proposed actuator to show its basic configuration. The basic elements are the frame (stator or yoke), the magnetic plunger, the refluxing spring, and the exciting coil.

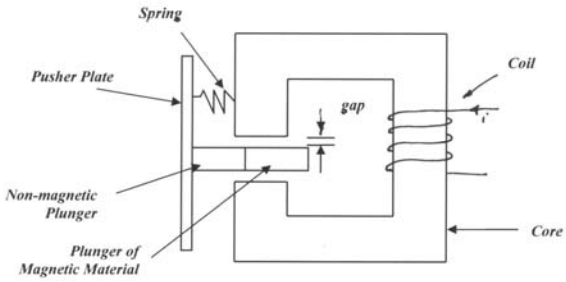

Figure 1: Schematic drawing of the proposed actuator

When excited, the magnetic plunger tends to align itself with the pole pair. As the plunger travels, it bushes on pusher plate, which squeezes the pumping chamber diaphragm (blood sac in actual systems). When the current is turned off, a reflexive spring forces the plunger to its unaligned position. Instead of using a separate spring, the pumping chamber diaphragm has afforded the stiffness property (see section b below). Mechanical and electrical specifications of the actuator are listed in Table (1).

Table 1: Mechanical and electrical specifications of the actuator

Item	Nominal value (unit)
Pole depth	5 (cm)
Pole width (stroke direction)	2 (cm)
Plunger slot height	2 (cm)
Air-gap length	~ 0.38 (mm)
Number of turns	1150 (turn)
Wire gauge used	AWG 18, AWG 19
Winding resistance (room temp.)	~ 7.2 (Ohm)
Core material	Silicone steel
Magnetic plunger material	Mild steel

b) Pumping chamber

The pumping chamber consists of two unidirectional valves in a cylindrical cavity. The cavity inner diameter is made to accommodate for a maximum pusher plat of 8 cm in diameter. A diaphragm is used as the moving boundary and is hermetically attached to the pumping chamber. The diaphragm stiffness produces the repelling force necessary to fill the pumping chamber again. A number of available rubber rags were tested as the diaphragm. Two jellyfish valves of domestic water pump were used as the unidirectional valves. Figure (2) shows the main parts of the prototype pump (actuator and pumping chamber).

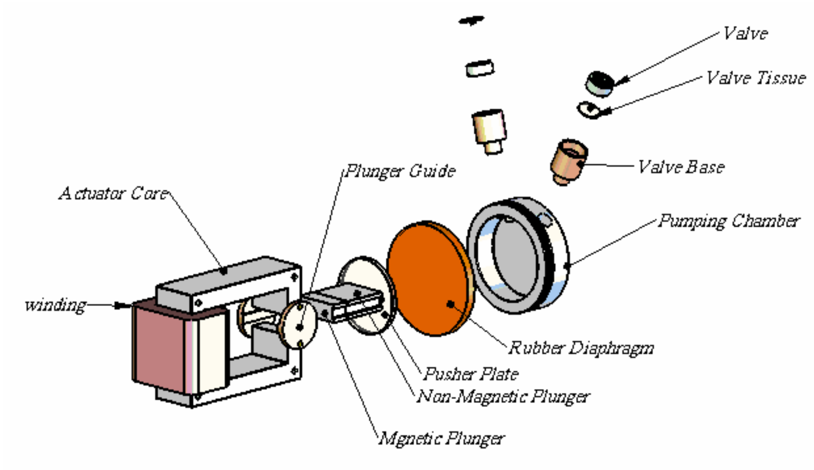

Figure 2: Prototype pump components

c) Power electronics

As a DC power supply, a full-wave rectifier with a smoothening capacitor was used. A function generator was used to allow for controllable firing to a switching relay. The power consumed was estimated by recording the current pulse via a sensing resistor and oscilloscope. A freewheeling diode is used to provide the stored magnetic field with a path during commutation of the phase winding thus protecting the power semiconductors (switches). Figure (3) shows a schematic diagram of the power supply circuit.

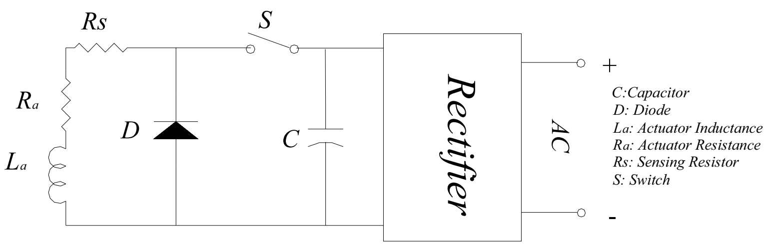

Figure 3: Schematic diagram for the actuator's power supply circuit

RESULTS

Static Performance

A load-cell was used to measure the static force produced by the actuator at various positions and excitation currents. Figure (4) shows the static force characteristics with respect to plunger displacement at different excitations. The position at the start of overlap between the magnetic plunger and actuator poles is denoted as zero. The target static force by design calculations was set to 150 N (as an LVAD) along the plunger travel distance. It can be noticed that at peak current provided by the supply (5 A) the actuator produced a force slightly higher than design requirement; however, this region is limited to few millimeters.

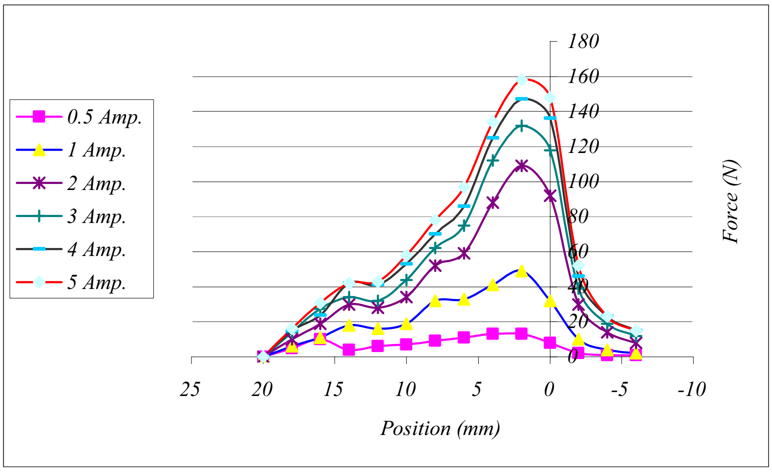

Figure 4: Measured static force characteristics of the actuator

In contraction to the preliminary design equation [11] the force is position dependent. The differences are owing, of course, mainly to saturation effects in the magnetic circuit, fringing flux and more importantly the non-equal air gaps which causes unbalance in the normal forces, which consequently increases the fiction force; similar conclusions are reported by [12,13]. The produced force per step current is decreased when the excitation current is higher than 2.0 Amperes, which also confirms the presence of bulk saturation of the core.

Dynamic Performance

a) Mock circulatory loop structure

A simple mock circulatory loop was constructed which consists of two reservoirs connected to the pumping chamber via tubes Figure (5). The constructed loop can hydraulically simulate the systemic or the pulmonary conditions of the normal human. by applying a head of water column at the inlet and out let of the pumping chamber. The pressure applied at inlet or outlet conditions is determined by the corresponding height of water columns and it can be simply adjusted by changing the elevation of the reservoirs.

The pumping chamber -artificial heart- to be tested is connected to the plastic reservoirs by flexible transparent plastic piping. Water circulates in a close loop from the inlet reservoir through the inlet valve to the pumping chamber. It is then being pumped through the outlet valve to the outlet reservoir. The water level in both reservoirs is held constant by draining excess amount of water introduced into the outlet reservoir back to the inlet reservoir. The drain tube is also used to measure the flow rate by collecting the amount of excess fluid at certain time (during this process makeup water has to be added to keep the inlet head constant).

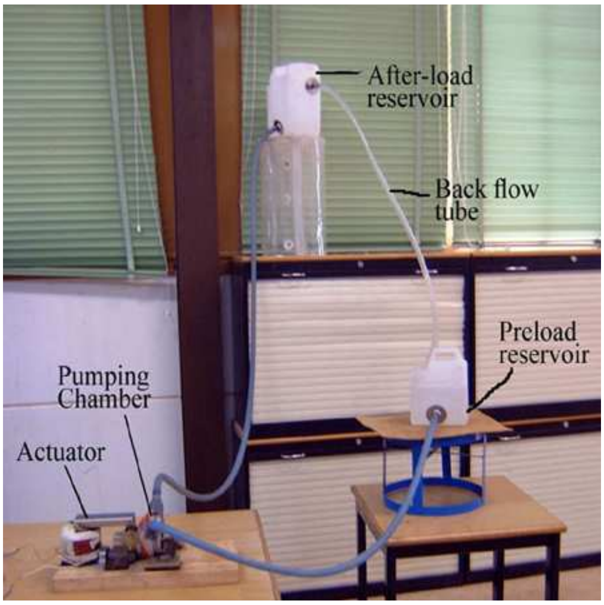

Figure 5: Simple mock circulatory loop used in dynamic testing

b) Dynamic test results

Extensive testing was conducted on the mock circulatory loop to demonstrate the actuator's pumping capabilities. Many parameters and operation conditions were changed manually. The impact of varying some parameters on the prototype performance is also investigated. Among these parameters: the pumping chamber position, used diaphragm, beat-rate, pusher plate size, preload, after-load, and input power.

To check the pumping capability of the actuator as an LVAD the preload was maintained at 19 cm of water (14 mm Hg). Although slightly higher than mean a trial pressure this value is considered by some research group to simulate the preload of both the right and left ventricles [7]. The after load was set to 135 cm of water (100 mm Hg) to simulate the mean aortic pressure. The supply voltage was limited to 30 V because of problems encountered with the switching relay and excessive temperature rise.

The maximum flow rate obtained was ~ 1.8 litre /min, against ~ 98 cm of water (72 mm Hg). Such results were not encouraging to check the performance at higher after-load pressure. It was judged that pressure drop across the valve bases of the pumping chamber reduced the apparent performance. Therefore, a second pumping chamber was fabricated with reduced pressure drop across it [11]. The maximum flow rate obtained at approximately the same conditions was ~ 2.4 litre/min.

Using the second pumping chamber, a flow rate of ~1.4 litre/min against an after-load of 100 mm Hg was recorded at starting up of the system at a beat rate of 72 BPM. However, the resulting flow rate decreases as the winding temperature increases. The pump output is related to the peak current pulse in the coil winding which seams to stabilize at 3.1 A with a maximum corresponding input power of ~ 20 W. To check the steady pump performance the peak current pulse of 3.1 A was maintained throughout the test and the existing flow rate was measured for over 1.0 hr and was $1.14^{\pm 0.09}$ litre/min at 60 BPM the corresponding overall efficiency* is ~ 1.28%.

Accumulative operation hours were more than 72.0 hrs with more than 3.0 hrs of continuous operation. During the mock circulation period, tear of diaphragm, failure of relay**, and movement of intake valve from place occurred at approximately 20.0, 49.0 and 54.0 operation hours respectively. However, no critical problem related to the actuator found sufficient to stop pumping occurred. This may indicate some sort of the actuator reliability. The actuator operated at a beat rate ranging from 60-72 BPM with reasonable input power fed to the actuator during dynamic operation (Maximum input power ~ 22 W). However, the prototype pump could not clear the requirement of an LVAD at 100 mm Hg.

DISCUSSIONS AND CONCLUSIONS

Some problems encountered in developing the prototype pump resulted in its low performance; the following factors are most likely the main causes:

- Lack of advanced manufacturing techniques resulted mainly in relatively large air gap and improper eccentricity of the magnetic plunger in the air-gap region (non-uniform unequal air gaps) resulted in significant friction force.
- Lack of resources led to the use of available materials which were chosen because it was difficult to find other suitable ones (diaphragm, valves ...etc.).
- Unused window area of the core produces excessive core losses due to the unnecessary magnetic path length of the core [14].
- Improper placement of winding results in bulk saturation of the core yoke rather than at the pole face, thus force is only developed at the start of the overlap [15]***.
- The mechanical power output is actually slightly higher than estimated because losses of pressure drop throughout the tubes, the tube joints, and valve bases are not considered.

The functionality of the actuator has been experimentally validated; a logical next step is to modify the system to achieve better results. There are many areas, as mentioned, which could be improved upon to enhance the motor performance. The level of improvement of the actuator performance is dependent on method of fabrication of actuator's components and on the available resources. The unique simple structure of the actuator may make it interesting for further development and evaluation for use in ventricular assist applications.

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*Overall efficiency is defined as the mechanical power to the fluid (after load pressure times flow rate) by the input power

** Relay failure occurred when the supply voltage increased to more than 30 V

*** Saturation is assumed in a zone in a magnetic circuit of ideal iron and zero air-gap

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