Validation of the electromagnetic design of a total artificial heart under physical load conditions

Thomas Finocchiaro
Chair of Applied Medical Engineering, RWTH Aachen University, Aachen, Germany
André Pohlmann
Institute of Electrical Machines, RWTH Aachen University, Aachen, Germany
Elena Cuenca Navalon
Chair of Applied Medical Engineering, RWTH Aachen University, Aachen, Germany
Marc Leßman
Institute of Electrical Machines, RWTH Aachen University, Aachen, Germany
Ulrich Steinseifer
Chair of Applied Medical Engineering, RWTH Aachen University, Aachen, Germany, and
Kay Hameyer
Institute of Electrical Machines, RWTH Aachen University, Aachen, Germany

Abstract

Purpose – The purpose of this paper is to introduce the RWTH's total artificial heart, ReinHeart, focusing on the design of the unique drive system.

Design/methodology/approach – The force characteristics of the drive have been simulated in a finite element (FE) approach. Additionally the coppler losses within the motor coils have been predicted based on the FE-simulation. Both results are compared to laboratory measurements of a prototype to validate the design.

Findings – The presented results show a good correlation between simulation and measurement and proof the applicability of the new design drive system.

Research limitations/implications – The used hydraulic models of the cardiovasular system used as a load for the device are not fully validated with data from living organisms. Therefore, further *in vivo* trials are needed.

Originality/value – The high force density of the drive allows its integration into a fully implantable, total artificial heart, in order to significantly improve durability. This hopefully will extend the indication for artificial hearts as alternatives to transplantation.

Keywords Artificial organs, Design, Electromagnetism, Heart, Validation of finite elements simulations, Linear drive, Total artificial heart

Paper type Research paper

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Introduction

Total artificial hearts (TAH) are required to address the problem of reduced oxygen supply associated with terminal heart insufficiency. It is a severe occurrence of cardiovascular diseases, the major cause of death worldwide, where drug based therapies fail. Due to the insufficient number of available donor hearts, the standard therapy, heart transplantation (THX), can often not be applied. If TAHs meet the physiological and anatomical constraints of the human body they can be used as alternative for THX. Risk of infections and limited durability are the main obstructions preventing the implantation of available TAH as an alternative to heart transplantation (destination therapy). A reduction of the incidence of infections can be achieved, if the skin is not penetrated by drivelines. Therefore, present pneumatic actuation principles (Körfer *et al.*, 2007) have to be replaced by electrical ones, as electrical power can be transmitted into the body wirelessly. Such transcutaneous energy transmission systems have been described before (Miura *et al.*, 2006). They rely on high frequency inductive coupling between an external primary and an implanted secondary coil. The complete TAH system consists of the implanted components pump unit, motorcontroller, volume compliance chamber and secondary coil as well as the external components primary coil, external controller and battery supply. An overview of the position of the components within the human body is shown in Figure 1. Previous work indicates that for destination therapy a minimum durability of five years is required (Kwant *et al.*, 2007) but difficult to be achieved in gear based drives as maintenance is not possible (Kwant, 2007). Thus, we propose a linear direct drive to actuate the TAH as it comprises a minimum of wear prone components.

Available drives do not meet the requirements posed by this application, as especially force density is too low. This paper focuses on the design of a novel linear drive with a maximized force density, meeting the TAH specifications.

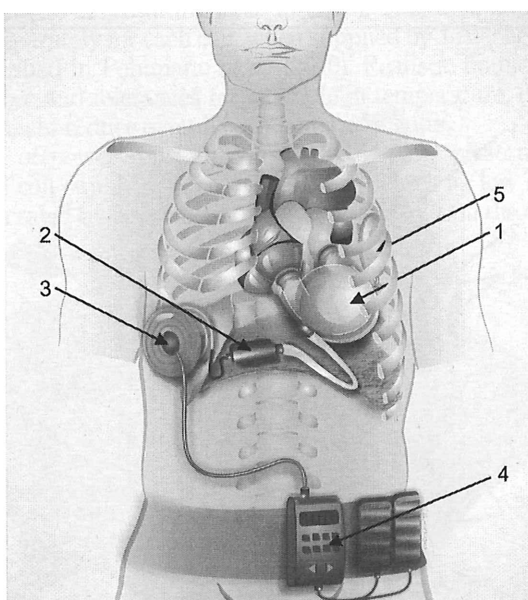

Figure 1.
TAH system, consisting of pump unit (1), motor controller (2), energy transmission system (3), external controller (4) and compliance chamber (5)

For designing the drive the following anatomical and physiological constraints have been considered:

(1) Due to the limited space in the human thorax, the dimensions of the pump unit should not exceed 85 mm in diameter and 95 mm in length, allowing a motor height of 36 mm.

(2) The total weight of the entire pump unit should be less than 800 g requiring a maximum drive weight of 500 g.

(3) The average blood flow should amount up to 5 l/min against a medium aortic pressure of 95 mmHg and provide an additional overload capacity.

(4) Electrical losses have to be less than 20 W to avoid blood and tissue damage due to excessive heat.

(5) The drive has to generate a force of up to 60 N (Pohlmann *et al.*, 2010).

The key specifications of the current version of the linear drive are listed in Table I. While the dimensions exactly meet the requirements, its weight exceeds requirement (2). This problem has been addressed by a redesign of the magnetic circuit and presented in Pohlmann *et al.* (2010).

The verification of the remaining constraints (3)-(4) is presented in this paper. Figure 2 shows the assembly of the pump unit. The drive is located between left and

Table I.
Key specifications of the linear drive

Parameter	Value
Outer diameter	85 mm
Motor height	36 mm
Weight of primary and secondary drive components	620 g
Stroke	18.5 mm

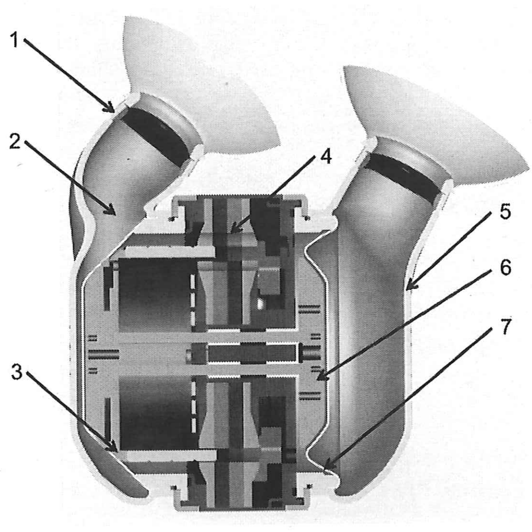

Figure 2.
Cross section of pump unit, consisting of valves (1), right pump chamber (2), coil assembly (3), stator (4), left pump chamber (5), pusher plate (6) and membrane (7)

right pump chamber, and pumps blood in an alternating mode to the lungs and the body. Mechanical heart valves at in- and outlets direct the blood flow in the appropriate direction.

Simulation and experimental procedure

Drive design

The CAD drawing of the drive is shown in Figure 3. It is excited by stationary inner and outer neodymium iron boron (NdFeB) permanent magnet rings. At room temperature the remanence of the inner and outer rings are 1.43 and 1.38 T, respectively. Pole shoes, made of an iron vanadium cobalt alloy with a saturation induction of 2.35 T, above and below the magnets concentrate the magnetic induction in the air gap. The coils inside the air gap are moving according to the Lorentz force equation in dependency of their current supply. In order to enable an efficient TAH the coil arrangement is divided into four separately suppliable coils. Additionally the coils are made of rectangular shaped enameled copper wire, achieving a copper fill factor of about 75 percent. Numerical electromagnetic field computation assisted in dimensioning the drive.

Simulation

For the prediction of the generated forces, the flux, coupling the coils, was determined by static, three dimensional, non linear finite elements (FEs) simulations. The flux density distribution in Figure 4 was obtained by the FE solver iMoose.stat3d (www.iem.rwth-aachen.de, accessed February 2011). As the radial component of the flux density (B) as well as the active wire length (l) of the coils is known, the Lorentz Force equation:

$$\vec{F} = I \cdot (\vec{l} \times \vec{B}) \tag{1}$$

was applied to calculate the resulting axial force (F). The calculations were performed across the stroke separately for each coil, when supplied by 1 A. The copper losses were calculated as described in Pohlmann *et al.* (2010). Realistic boundary conditions are crucial as differences and tolerances for example in temperature, dimensions, current and force measurement reduce overall correlation of results.

The evaluation of power consumption of the drive is performed in three steps. First an optimized coil supply is necessary to keep losses as low as possible. As the flux density is generated by permanent magnets, it depends on the drive geometry and

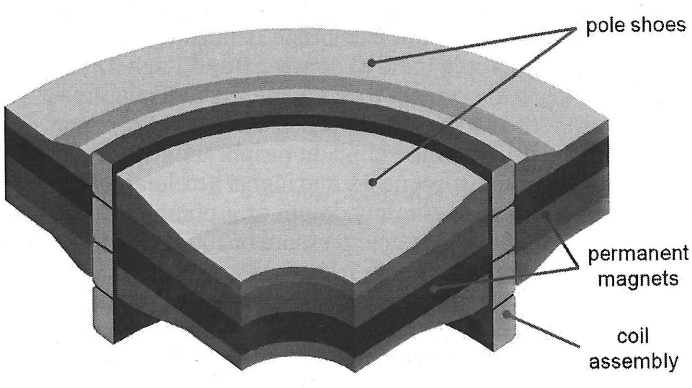

Figure 3. CAD drawing of the drive

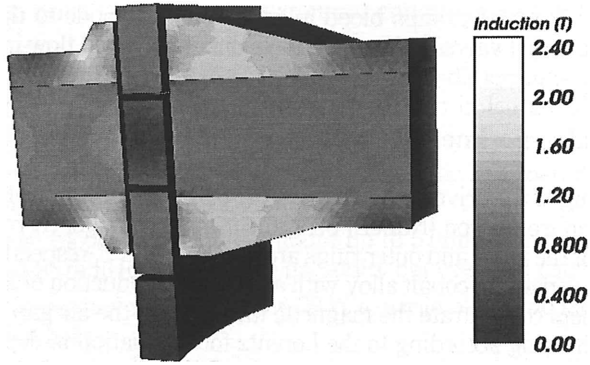

Figure 4.
Flux density distribution

can be obtained from FE simulations. In this way the flux coupling in the coils can be determined in every position (x) of the coil assembly. The flux density at each coil (B_n) can be put into relation of the flux density of all coils according to equation (2) and is referred to as the axial position dependent current factor ($k_{I,n}(x)$):

$$k_{I,n}(x) = \frac{\bar{B}_{r,n}(x)}{\sum_{n=1}^{4} \left| \bar{B}_{r,n}(x) \right|} \tag{2}$$

According to Pohlmann *et al.* (2010) electric losses are minimized when the relation of the current in each coil follows equation (3), where (I_{sum}) represents the sum of the current in all coils and (I_n) represents the sum of the current in coil n:

$$I_n(x) = I_{sum}(x) \cdot k_{I,n}(x) \tag{3}$$

In a second step the required force profile is determined. Therefore, a pump chamber was connected to a physiologic hydraulic load and actuated by a servo-hydraulic testing machine (Zwick HC5), which replaced the linear drive system and measured force and position of the pusher plate. As fluid a mixture of 60 percent water and 40 percent glycerol was used to simulate the viscosity of human blood.

The measurement results at pumping frequencies of 120 and 160 bpm as well as the deduced characteristic, applied for the drive design, for the systemic blood circuit are shown in Figure 5. According to the TAH assembly the left blood chamber is completely filled and evacuated at a displacement of 18.5 and 0 mm, respectively. At position 0 mm the maximum absolute force sums up to 60 N for a frequency of 120 bpm. When the chamber starts to fill again, first the membrane is pushing against the pusher plate, resulting in a reduction of force in the area between 0 and 1 mm. Due to the inertia of the fluid the membrane loses the contact to the plate. At a displacement of 18.5 mm the moving direction reverses and the plate touches the membrane again. At this position oscillations are initiated between the compliant membrane and the inertia in the outflow tract. Due to the higher pumping frequency and higher acceleration of the membrane, the magnitude of the oscillations has its maximum at a pumping frequency of 160 bpm. When pumping against a mean systemic pressure of 100 mmHg a flow rate of 5.5 l/min and 7.2 l/min were achieved at pumping frequencies of 120 and 160 bpm, respectively. Therefore, the force profile, used for the drive design and its optimization, considers the mean force of the 160 bpm measurement during the oscillations (position 18.5 to 8 mm) and the maximum forces for the remaining stroke. Although the deduced profile does not

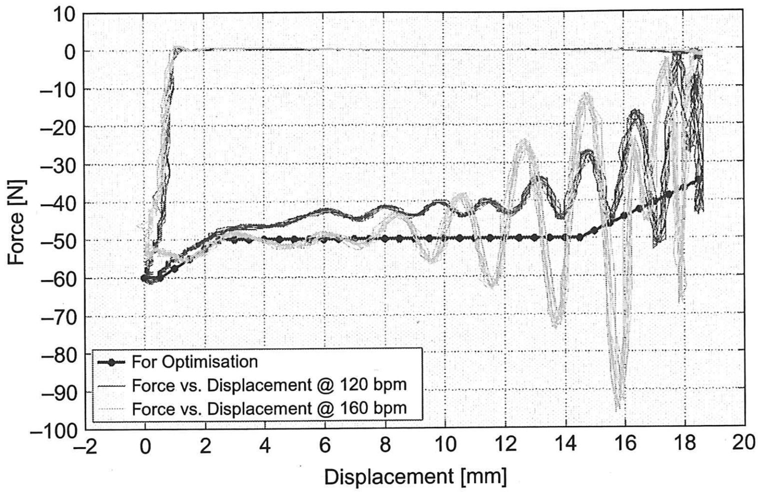

Figure 5.
Forced vs displacement obtained from dynamic mock loop test

cover the peak forces, it is assumed that a corresponding actuation will empty the pump chambers and only result in insignificant position errors.

The final force vs displacement characteristic for both blood circuits are shown in Figure 6. As pulmonary resistance is lower than systemic resistance, the force needed to eject the right pump chamber is significantly lower. Its maximum of 25 N is found at mid stroke. The reason is that at the lower absolute pressure, effects of valve resistance and inertial forces are more noticeable.

As the required force is known, the resulting losses are calculated. Therefore, equation (1) is modified to determine the position dependent overall current ($I_{sum}(x)$) needed to generate the required force ($F_r(x)$):

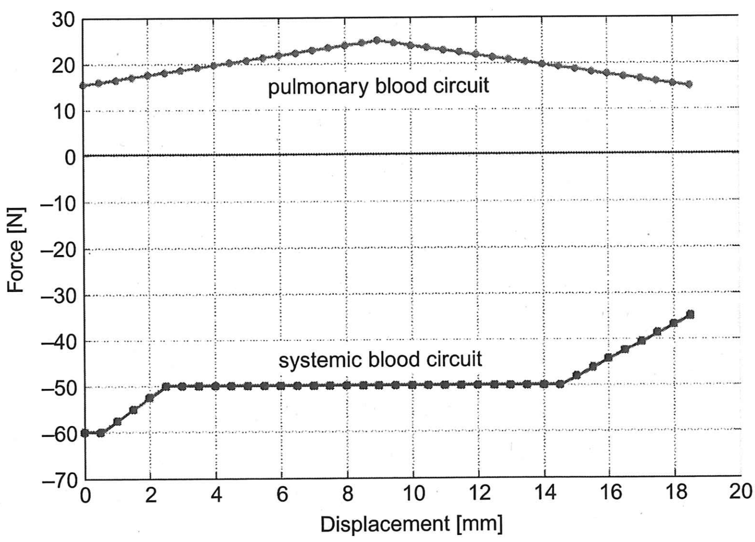

Figure 6.
Force vs displacement characteristic for drive design

$$I_{sum}(x) = \frac{F_r(x)}{\sum_{n=1}^{4} \bar{B}_{r,n}(x) \cdot k_{I,n}(x) \cdot l} \quad (4)$$

equations (3) and (4) yield the position dependent losses by applying Ohm's law.

To calculate the average losses of the prototype, the position dependent losses were integrated over a whole cycle.

Validation

The designed magnetic components were constructively integrated into a drive with housing, bearing and accessories as seen in Figure 2. Thereafter prototypes were manufactured. For validation of the new drive concept the forces of each coil at a constant current of 1 A was measured over the whole stroke and compared to simulation results. In order to achieve accurate results the previously described servo-hydraulic testing machine was applied for force and position measurements. Additionally, the influence of friction was eliminated by subtracting forces taken from a zero current measurement.

After validation of the static performance we tested the drive in a dynamic test setup. Therefore, pump chambers have been attached to the drive as shown in Figure 2. The TAH has been operated by a specially designed drive controller, which measures the position of the mover and calculates the position dependent currents in all coils. Commercially available servo amplifiers have been used to supply the coils. To apply a load similar to physiologic demands of future patients, the TAH was connected to a mockup loop of the human circulatory system that provided adjustable hydraulic compliances (electrical analog: capacitance) and resistances. The compliances imitate the elasticity of the arterial vessels, while the resistances represent the capillaries where the mayor pressure drop takes place. Figure 7 shows an overview of the interconnections between the components. In this setup the hydraulic performance of the TAH along with its power requirement and copper losses was measured in order to validate the whole drive concept under physiologic load conditions.

Results

Figure 8 shows a comparison of static forces of each coil over the stroke from measurement and simulation. Due to mechanical restrictions, measurements could not be performed at positions between 0 and 1.5 mm. Hence measurement data for these positions is missing. The difference in force between simulation and experiment is in the range of 6-9 percent and can be explained by uncertainties for example in current measurements and exact material properties as well as manufacturing tolerances.

During the operation in a dynamic test setup, the active power input as well as the effective coil currents were measured with a power meter (Yokogawa PZ 4000). After the resistances of the lead cables and coils were determined, the copper losses have been calculated by multiplying the root mean square current (I_{rms}) with the relevant resistance (R) (equation (5)):

$$P_{cu} = I_{rms}^2 \cdot R \quad (5)$$

Total measured copper losses are 8.3 W in the coils and 3.9 W in the lead cables. By subtracting these losses from the total active power, the mechanical output power remains 3.5 W.

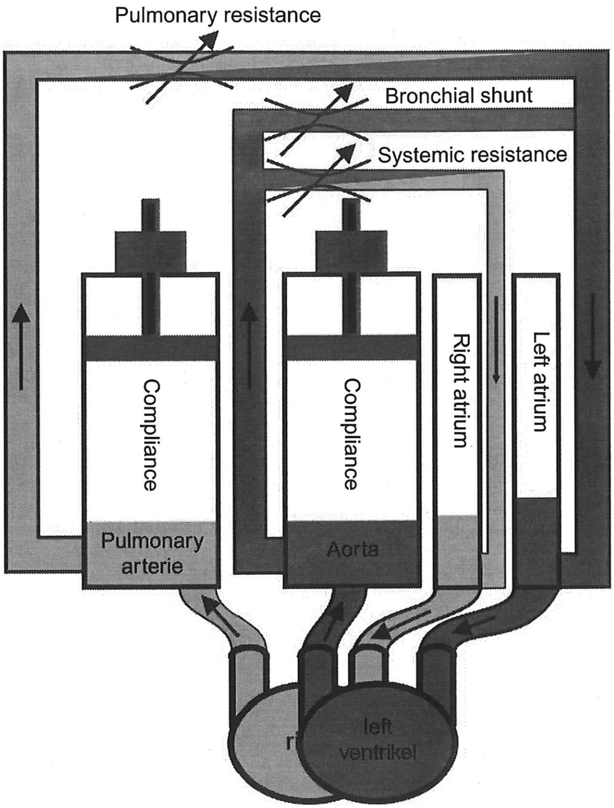

Figure 7. Schematic of the mock circulation loop employed as physiologic load for the TAH

When integrating the force profile of Figure 6 the resulting mechanical power is 3 W close to measurement results. The difference is contributed to the neglected friction and uncertainties of the measurements. The copper losses in the coils, have been precisely predicted to be 8.0 W.

Overall the linear drive can generate up to 116 N/kg at losses of 20 W. This is a significant improvement, compared to commercially available drives (e.g. 45 N/kg, LA24-20, kimco voice coil actuator, www.beikimco.com/).

Conclusion

Based on the clinical demand for a durable drive system for a TAH a new linear motor concept has been developed. FEM simulation of the flux density has been applied to determine the main dimensions of the drive and to predict the position dependent force generation and the resulting copper losses. After integrating the designed magnetic components into a TAH, the pump unit was connected to a mock circulation loop in order to apply a physiologic load. The comparison of simulation and measurement results has shown a good agreement, proving the excellent force density of the new drive concept. Initial short term *in vivo* test of the linear driven TAH have shown the proof of concept of the drive system. Further studies are required to evaluate the thermal distribution within the drive and to prove the durability of the entire system.

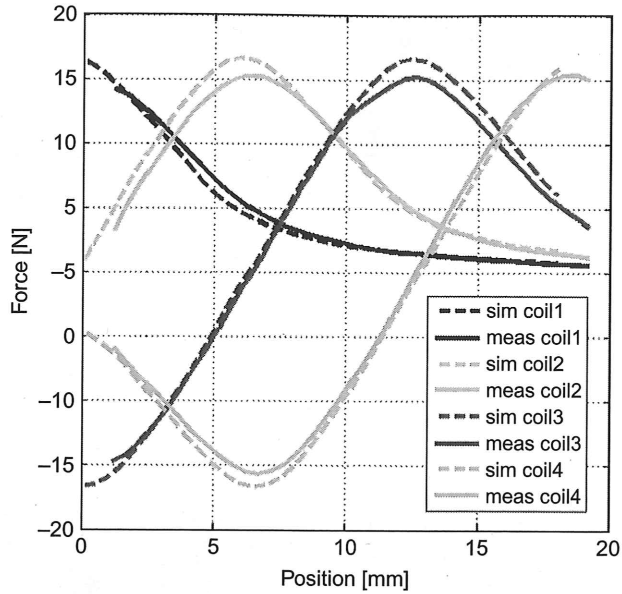

Figure 8.
Simulated (sim) vs measured (meas) force generated by each coil over coil position

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Corresponding author
Thomas Finocchiaro can be contacted at: Finocchiaro@hia.rwth-aachen.de