Validation of the electromagnetic design of a Total Artificial Heart under physical load conditions

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Abstract- 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. This paper introduces the design of the RWTH's Total Artificial Heart ReinHeart and how the physiological constraints have been considered for its design. The drive design is based on electromagnetic Finite Element Simulations. During this process the copper losses and a force displacement characteristic have been determined. For evaluation of the design a test bench with a mock loop is set up. Measurements validated the drive design.

I. Introduction

Risk of infections and limited durability are the main obstructions preventing the implantation of Total Artificial Hearts (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 [1] have to be replaced by electrical ones, as electrical power can be transmitted into the body wirelessly. Such transcutaneous energy transmission systems (TET-systems) have been described before [2]. 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 depicted in Fig. 1. Previous work indicates that for destination therapy a minimum durability of five years is required [3] but difficult to be achieved in gear based drives as maintenance is not possible [4]. 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.

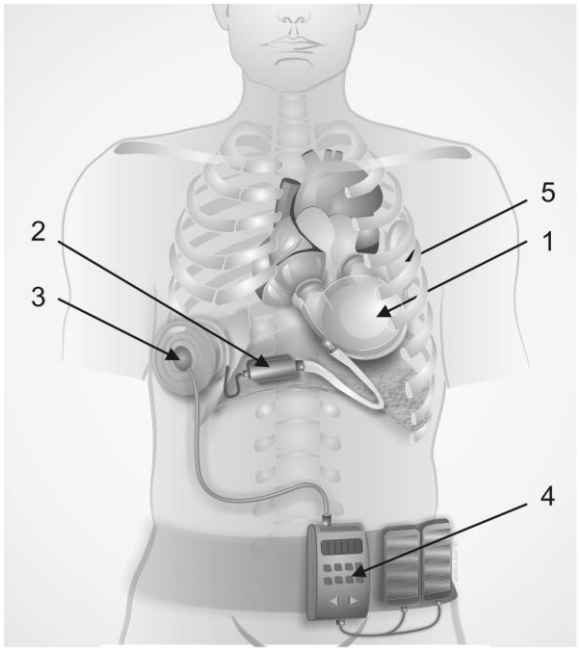

Fig. 1. TAH system, consisting of pump unit (1), motor controller(2), energy transmission system (3), external controller(4), compliance chamber (5).

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

a) 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.
b) The total weight of the entire pump unit should be less than 800 g requiring a maximum drive weight of 500 g.
c) 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.
d) Electrical losses have to be less than 20 W to avoid blood and tissue damage due to excessive heat.
e) The drive has to generate a force of up to 60 N [5].

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 b). This problem has been addressed by a redesign of the magnetic circuit and presented in [4].

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

The verification of the remaining constraints c) to e) is presented in this paper. Fig. 2 illustrates the assembly of the pump unit. The drive is located between left and 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.

II. Simulation and Experimental Procedure

A. Drive Design

The CAD drawing of the drive is shown in Fig. 3. It is excited by a stationary inner and outer neodymium iron boron (NdFeB) permanent magnet ring. At room temperature the remanence of the inner and outer rings are 1.43 T 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%. Numerical electromagnetic field computation assisted in dimensioning the drive.

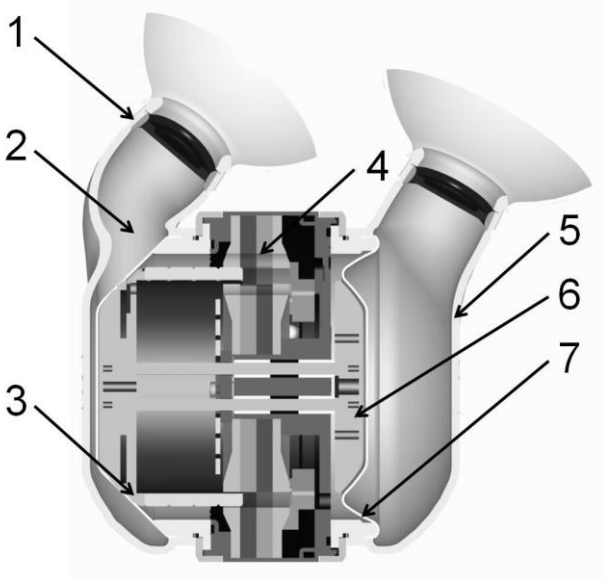

Fig. 2. Cross section of pump unit, consisting of valves (1), right pump chamber (2), coil assembly (3), stator (4), left pump chamber (5), pusher plate (6) and membrane (7).

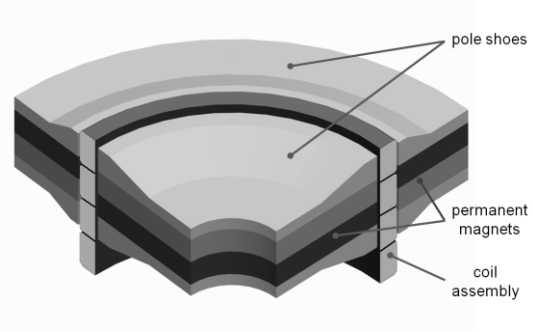

Fig. 3. CAD drawing of the drive.

B. Simulation

For the prediction of the generated forces, the flux, coupling the coils, was determined by static, three dimensional, non linear Finite Elements (FE) simulations. The flux density distribution in Fig. 4 was obtained by the FE solver iMoose.stat3d [6]. 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,

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

was applied to calculate the resulting axial forces 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 [5]. 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 can be obtained from FE simulations. In this way the flux coupling the coils can be determined. The flux density at each coil 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 $k_{I,n}(x)$:

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

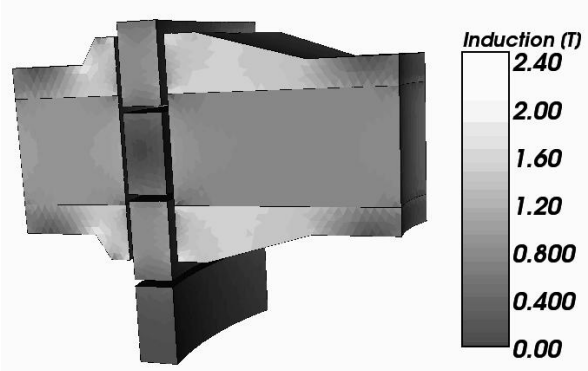

Fig. 4. Flux Density Distribution.

According to [5] electric losses are minimized when the relation of the current in each coil follows equation 3.

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

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% water and 40% 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 Fig. 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 160bpm measurement during the oscillations (position 18.5 to 8 mm) and the maximum forces for the remaining stroke. Although the deduced profile does not 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 presented in Fig. 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.

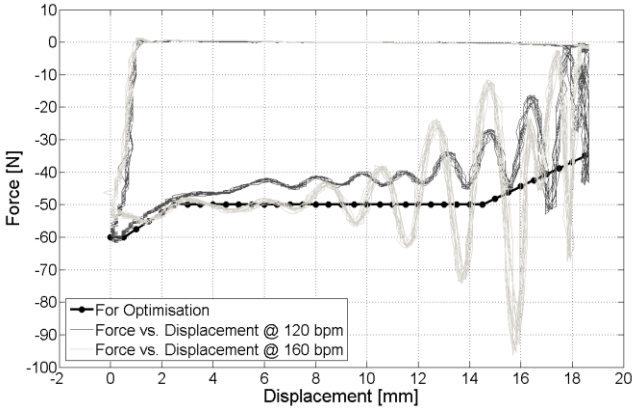

Fig. 5: Forced vs. displacement obtained from dynamic mock loop test.

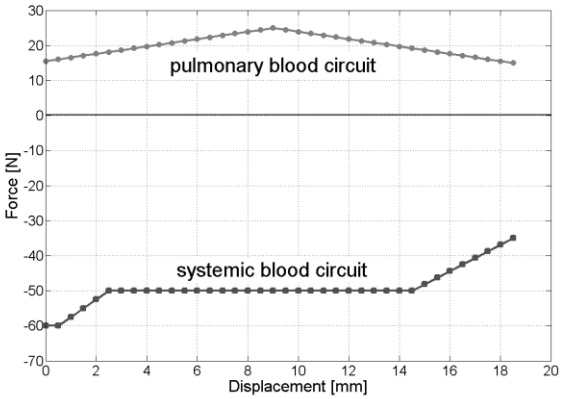

Fig. 6: Force vs. Displacement characteristic for drive design.

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 $F_r(x)$.

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

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.

C. Validation

The designed magnetic components were constructively integrated into a drive with housing, bearing and accessories as seen in Fig. 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 Fig. 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. Fig. 7 gives 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.

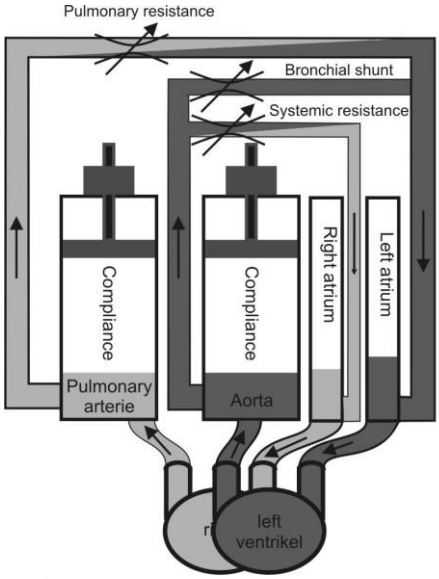

Fig. 7. Schematic of the mock circulation loop employed as physiologic load for the TAH.

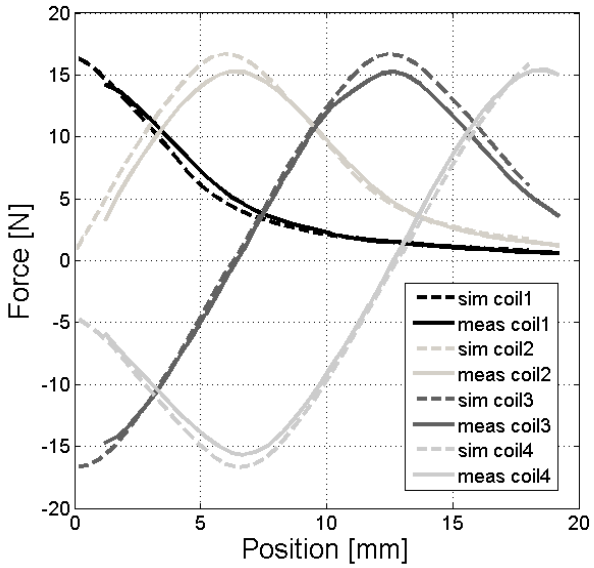

Fig. 8. Simulated (sim) vs. measured (meas) force generated by each coil over coil position.

III. Results

Fig. 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 to 9 % 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 applying equation 2,

$$P_{cu} = I^2_{rms} \cdot R, \qquad (2)$$

resulting in 8.3 W in for 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).

When integrating the force profile of Fig. 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 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, http://www.beikimco.com/).

IV. Conclusions

Based on the clinical demand for a durable drive system for a TAH a new linear motor concept has been developed. FEM simulations 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.

V. References

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