Parameter study for the miniaturization of a rotary blood pump drive system

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***Abstract*-- Ventricular Assist Devices (VADs) are a therapy option for patients suffering from cardiac insufficiency. A VAD can unload the diseased heart and yield its recovery or bridge the time until a suitable donor organ is available. Besides the general requirement of establishing a sufficient perfusion of the human body without affecting the blood cells, a minimal invasive implantation technique relying on minimal dimensions of a VAD is essential to minimize the exposure of the patients. This paper focuses on a parameter study for reducing the dimensions of a VAD drive system. The results obtained by Finite Element Analysis (FEA) are evaluated in terms of ohmic losses, torque and forces.**

***Index Terms*—Ventricular Assist Devices (VAD), Drive Design, Drive Optimization, Finite Element Analysis (FEA)**

I. INTRODUCTION

For the therapy of heart failure, the transplantation of donor organs is the gold standard if a previous pharmaceutical therapy fails. However, the number of available and suitable donor organs is limited. In recent years the demand for them is rising while the amount of suitable organs is constant [1]. Therefore, an alternative for heart transplantation is required to narrow that gap.

In current research this alternative is known as Artificial Hearts (AH), which can be divided into Total Artificial Hearts (TAHs) and Ventricular Assist Devices. The mayor difference between these systems is that a TAH replaces the native heart, while a VAD supports it. The choice which system is applied depends on the clinical patterns as well as the physiology of the patient. For example the patient's thorax might be too small for a TAH or the tissue of the heart is necrotic and does not allow for a cannulation, which is required for the connection between heart and VAD. This paper will focus on the drive design of a left Ventricular Assist Device (LVAD) for long term full support. From a clinical point of this requires a system which can guarantee a perfusion of up 7l of blood per minute against a mean systemic pressure of 100 mmHg. Further blood damage in terms of hemolysis, thrombogenecity and denaturation has to be prevented. These aspects will be discussed in a for engineering sufficient way in the drive design section, while other constraints such as biocompatibility, which is related to material science, are not addressed in this paper.
Fig. 1 represents the second generation VAD system HeartMate II (HM2) [2]. The axial flow blood pump is actuated by a rotary drive and has conventional bearings,

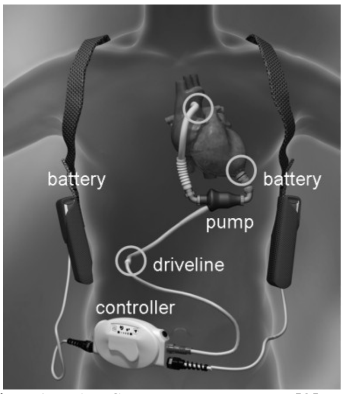

Fig. 1: VAD System HeartMate II [2].

which defines a second generation AH system. Via cannulations the HM2 is connected to the apex of the left ventricle and the ascending aorta. The pump system has an external control and batteries. External and implanted VAD components are connected to each other via a percutanous driveline penetrating the skin. Such VAD systems can restore the perfusion of the human body and unload the diseased heart. This may yield a recovery of the heart, but the main objective is to bridge the time until a donor organ is available.

For a long term support the percutaneous drivelines increase the risk of infections. Therefore, they should be replaced by Transcutaneous Energy Transmission (TET) systems, which are based on the inductive coupling between an external and an implanted coil as described in [4]. A drawback of such systems is the requirement of an additional implanted VAD controller and buffer battery as well as a limitation of the transferred power. In combination with a third generation VAD system, which is defined by a rotary drive system and contactless bearings, a maintenance free long term support is possible. When focusing on the drive, most rotary VADs such as the DuraHeart™ [8] or the Worldheart [9] are operated by axial flux brushless dc drives (BLDC). Their high efficiency and brushless design are beneficial for a VAD operation without blood and tissue damage[6]. Another aspect is their flat design, which is studied in this paper to achieve minimal dimensions to allow for a minimal invasive VAD implantation. In this way the exposure of the patient during the implantation surgery can be reduced yielding an accelerated recovery.

II. DRIVE DESIGN

The parameter study for the miniaturization of a VAD drive is based on the BLDC drive design given in Fig. 2.

Its rotor consists of a solid ferromagnetic back-iron (9S20) and 16 alternating axially magnetized neodymium iron boron (NdFeB) permanent magnets, which have a remanent induction B_r of 1.4 T at 20°C. The stator is realized by a solid stator yoke (9S20) having 18 coils. Each coil is hand wounded with rectangular shaped copper wire. In this way a copper fill factor of 60% is achieved. According to [3] attracting forces act between the permanent magnets in the rotor and the ferromagnetic iron of the stator teeth. These forces are compensated with a hydrodynamic spiral groove bearing. It is relying on a thin layer of blood (up to 50 µm) and can generate axial repelling forces of about 10 N. A spiral groove bearing can be regarded as an active and noncontact bearing. When designing a drive system for VAD system it has to meet three mayor requirements. Obviously the drive should be stable for the predefined operation range. Otherwise high shear forces are imposed from the spiral groove bearing to the blood cells resulting in hemolysis. As the hydrodynamic bearing has to be designed for a fixed speed and pressure characteristic due to its operation principle, the spiral groove bearing sets the limits for the range of operation. In this design the operation speed n_N is 2500 rpm, while the bearing can compensate 10 N of axial forces. At this speed the pump system requires a torque T_{out} of 12 mNm to provide a sufficient perfusion of the body. According to

$$P_{out} = \frac{T_{out}}{2 \cdot \pi \cdot n_N} \qquad (10)$$

the required output power P_{out} of the drive is given. The drive system has to provide this output power without damaging the blood cells. As the main factors causing hemolysis and thrombogenicty are high shear forces and blood flow stagnation, they have to be avoided by a well designed hydrodynamic bearing and pump system. But denaturation caused by overheating can be directly linked to the electric losses of the drive, which are transformed into heat. Based of a required drive efficiency of at least 40% the copper and iron losses must not exceed 1.3 Watts. Although the hard limits of the drive system are determined by the axial attracting forces and electric losses, the drive efficiency should be as high as possible with respect to the power range and performance of the peripheral components such as the TET system or the implanted battery and control system.

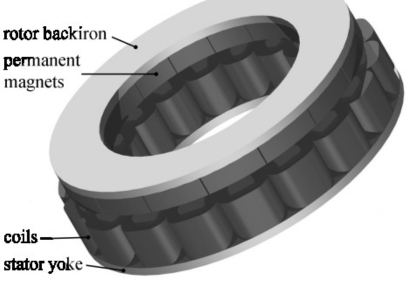

Fig. 2: BLDC drive of a VAD.

III. METHOD

As in [7] the objective of the parameter study is to establish a hierarchic order of parameters, which could be applied to influence the drive system characteristic in a desired way. For a single sided BLDC VAD drive system applied in a VAD system balancing the attractive axial force while reducing the ouput torque as less as possible is essential. For the parameter variation the following simulation chain is applied. Obviously the resulting axial forces and driving torques have to be calculated. The axial force $\vec{F}_{axial}$ is given by

$$\vec{F}_{axial} = \frac{B^2 \cdot A}{2 \cdot \mu_0} \cdot \vec{e}_{axial} \,. \qquad (2)$$

It depends on the magnetic induction B, the cross section area A of the stator teeth and the permeability of the vacuum μ_0. According to

$$\vec{F}_{rad} = I \cdot (\vec{l} \times \vec{B}) \cdot \vec{e}_{rad} \qquad (3)$$

the radial force $\vec{F}_{rad}$ can be determined by the Lorentz force equation. The second term is the cross product of the magnetic induction B and active wire length l. When multiplying the result with the coil current I the radial force is obtained. The output torque is linked to the radial force by the average rotor diameter. Equations 2 and 3 both rely on the magnetic induction. Due to a large air gap, saturation effects in the rotor backiron as well as in the tips of the stator teeth, a numerical Finite Element Analysis has to be applied for the determination of the distribution of the magnetic induction. For these simulations a transient solver from the iMoose solver package [5], developed at the Institute of Electrical Machines of the RWTH Aachen University, was applied, considering the quasi transient movement of the rotor. Besides the coil currents all other variables of equations 2 and 3 are defined by the drive design and can therefore be considered to be constant. Therefore the required coil supply is a function of the torque requirement. By this way, the torque can be directly linked to the copper losses. The calculation of the copper losses requires three steps including two FEM simulations. First, the cogging torque has to be

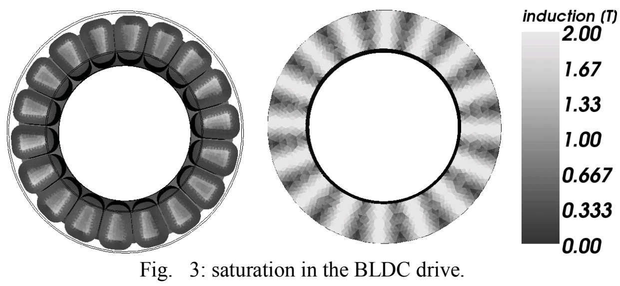

Fig. 3: saturation in the BLDC drive.

determined, which is dependent on the relative position between the permanent magnets in the rotor and the stator teeths. In the next simulation, the coils are supplied with a constant current linkage θ of 50A. In order to determine the torque, generated by the current linkage, the cogging torque has to be substracted from the simulated torque. As a six step commutation is applied for the control of the drive, the following equation applies:

$$T_{out} \propto \theta_R \theta_S \sin(\varepsilon) \tag{4}$$

In this equation the generated torque T is a function of the rotor θ_R and stator θ_S current linkages and the sine of the angle ε between them. Due to its permanent magnet excitation, θ_R is constant and ε is always 90 degree. Therefore, the required stator current linkage for generating the required driving torque of 12mNm can be obtained by down- or upscaling the torque generated for a current linkage of 50A. In the next step, the resulting copper losses are calculated according to

$$P_{co} = \frac{2}{3} \cdot n_c \cdot \theta^2 \cdot \rho_{cp} \cdot \frac{l}{A \cdot CF} \cdot \left(1 + \alpha_{co} \cdot (T - T_{20})\right) \tag{5}$$

In this equation the factor $2/3$ considers the six step commutation control of the drive, where only two out of three phases are supplied simultanously. The copper fill factor CF is assumed to be 60% and θ represents the current linkage in the coils supplying two third of the n_c coils of the drive. The resistance of one coil is calculated by its resitivity ρ_{cp} at room temperature, its wire length l and its cross sectional area A. In order to obtain the resistance at an assumed stator and rotor backiron temperature T of 45˚C, the last term of equation 6, containing the temperature coefficent α_{co} for copper and the room temperature T_{20}, is required.

For the final design it is important to know the absolute values for example for the attracting force in order to meet the design constraints. However, on the way to the final design it is important to be able to evaluate design parameters and their effects on the object functions. In terms of a parameter study, the variations can be regarded as members of the n elements of a Euclidian plane. The average Euclidian distance can be calculated for the attracting forces and losses by

$$F_n(i) = \frac{F(i)}{\sqrt{\sum_{i=1}^{n} (F(i))^2}}, \quad P_n(i) = \frac{P(i)}{\sqrt{\sum_{i=1}^{n} (P(i))^2}}. \tag{6}$$

The lowest sum of the average Euclidian distance of the axial forces and losses indicates the optimum design point for each parameter variation.

IV. Parameter Study

The initial drive was designed with the help of the Differential Evolution Algorithm in [10]. In order to reduce the drive dimensions the outer diameter has been reduced from 38 to 28mm. Simultaneously the inner diameter has been reduced from 24 to 16 mm. This is beneficial for a better anatomical fitting as well as a less invasive implantation procedure. Table 1 collects the design data for the initial design and the one, which is studied in this paper. The objective of the parameter study is to balance the axial attracting forces and to keep the ohmic losses of the drive as low as possible. Altogether five parameters are studied. Initially the variation of the air gap between the bottom of the rotor magnets and the top of the stator teeth is applied for showing the starting of the parameter variation in Fig. 4. According to the design force constraint this drive could be operated for air gaps larger than 1.5 mm. This would yield an increase of the axial drive dimensions when compared to the initial design. As the torque is only linear dependent on the magnetic induction it decreases nearly linear with the rising air gap. But the attracting force is dependent to the square of the magnetic induction and therefore decreases faster than the driving torque.

When operating the drive at an air gap larger than 1 mm the axial dimensions of the drive are increased in comparison to the initial design. Therefore, the next studied parameter is the stator teeth height in order to avoid increased axial dimensions. The height of the coils is kept constant at 5 mm, while the stator teeth iron is reduced in steps of 1 mm. This yields coils which are partially filled with air and partially filled with iron. The force characteristic in Fig 5 reveals that a force of 10 N is reached for a stator tooth height of 4.5 mm. This equals a distance of 1.5 mm between the bottom of the rotor magnets and the top of the stator teeth. The torque amounts to 17.5 mNm at this distance. Therefore increasing the air gap or reducing the stator tooth height has the same effect on the electromagnetic characteristic. This shows that stray flux can not significantly contribute to the torque generation for the studied air gap ranges. But in terms of dimensions reducing the stator tooth has to be preferred. Fig. 6 represents the normalized losses and forces according to equation 6. According to this characteristic the optimum design point would be at a stator tooth height of 3.5 mm. But as the resulting force is far below 10 N and the losses are decreasing for higher stator tooth heights, the optimum design point will be

Table 1: Drive Design Parameters.

design objectives	Prototype	Mini Design
nominal speed	2500 rpm	2500 rpm
nominal torque	12 mNm	12 mNm
max. axial force	10 N	10 N
stator height	7mm	≤ 7 mm
outer diameter	38 mm	28 mm
inner diameter	24	16 mm

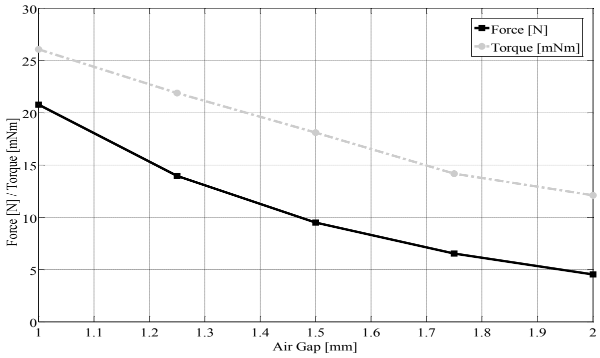

Fig. 4: Force and Torque vs. Air Gap.

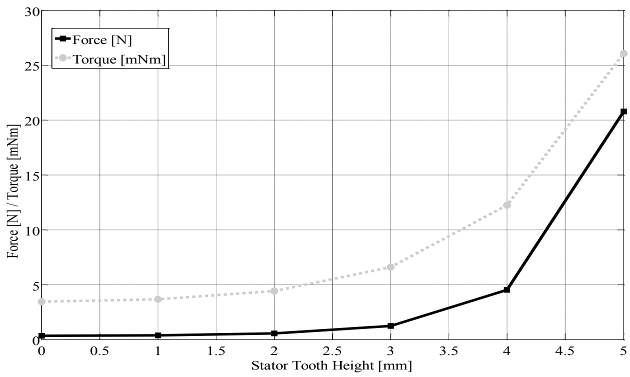

Fig. 5: Force and Torque vs. Stator Tooth Height.

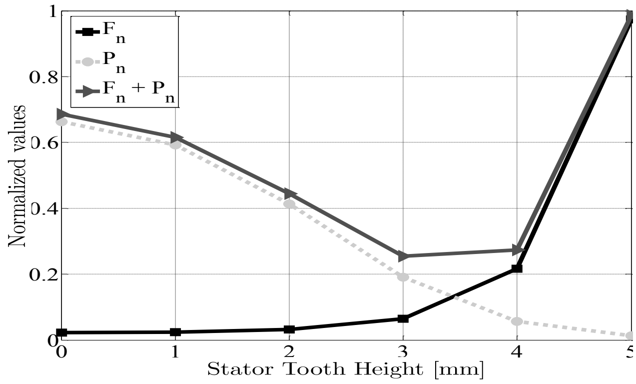

Fig.6: Euclidean distance of Force and Losses vs. Stator Tooth Height.

ignored here. But the characteristic shows the sensitivity of the stator tooth height starting at a height of 4 mm to the electromagnetic drive characteristics. Small deviations from the desired height result in high deviations in force and torque. The effects of these parameters are based on the air gap dependency of the magnetic inductions. Further parameters to reduce the magnetic induction are the remanent induction and the height and therefore the volume of the magnets. Although a reduced remanent induction yields a lower attracting force it has no impact on axial dimensions such as a reduced magnet height. But it offers the opportunity to balance forces in limited range without design changes. In the studied range between 1.2 and 1.4 T the axial force can be adjusted between 13 and 21 N for an operating air gap of 1mm (s. Fig. 7). The maximum torque deviation amounts to 5 mNm. According to Fig. 8 the optimum

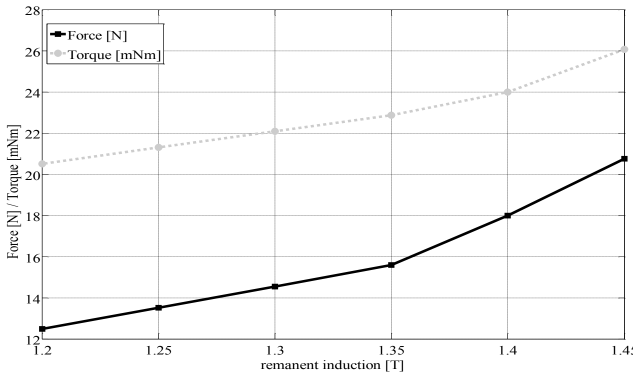

Fig. 7: Force and Torque vs. Remanend Induction.

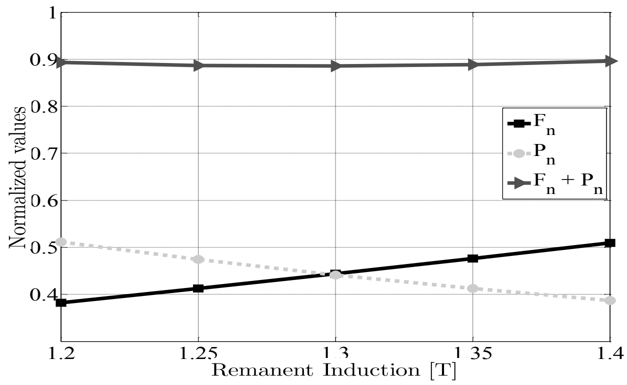

Fig. 8: Euclidean distance of Force and Losses vs. Induction.

operation point is achieved for a remanent induction of 1.3T and an axial force of about 16 N and a torque of 23 mNm. The variation of the permanent magnet thicknes is limited by manufacturing and the axial dimensions, which should not increase. Fig. 9 shows the flat force and torque characteristic, with its optimum at a magnet thickness of 1.5 mm (s. Fig. 10). While the force varies between 14 and 21 N, the torque amounts between 22 and 26 mNm. According to the flat characteristic representing the minimum of the sum of normalized force and torque deviations from the optimum design points have only a slight effect on the design. Therefore this parameter is well suited for a fine adjustment of the electromagnetic drive characteristics. In the studied range the same applies for a variation of the remanent magnetic induction. But in theory the flux density could be reduced to 0T, allowing for adjustments in a larger force range.

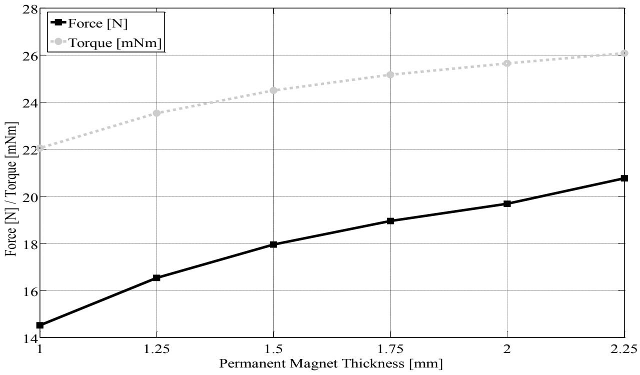

Fig. 9: Force and Torque vs. PM Thickness.

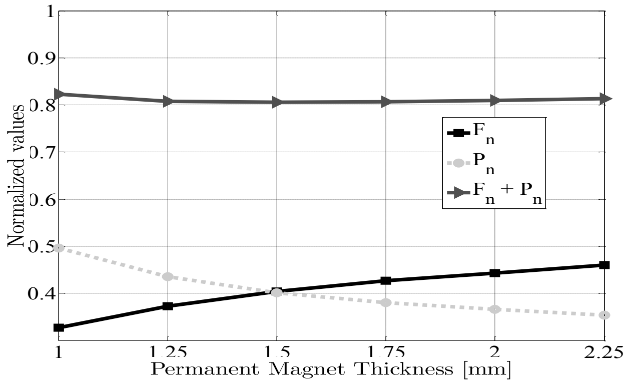

Fig. 10: Euclidean distance of Force and Losses vs. PM Thickness.

V. RESULTS

Table 2 collects the simulation results of the drive design which is miniaturized when compared to the drive Prototype. For the miniaturized design partially air filled coils have been applied. The iron stator tooth height amounts to 4.5 mm. In order to reduce the resulting force of about 12 N, permanent magnets with a remanent induction of 1.3 T have been applied. This has also the advantage of a reduced risk of demagnetization when compared to the magnet with a remanent magnetic induction of 1.4 T. For this design the copper losses sum up to 953 mW, which is about 25% higher when compared to the prototype. However, when considering the maximum allowable losses of 1.3 W, the losses are well with the predefined limit. In this study the axial drive thickness has only been considered for adjusting the axial forces. But as the axial force is mostly dependent on the air gap between the top of the stator teeth and the bottom of the magnets in the rotor, the coil and therefore the stator height can be reduced. The limit of this reduction is again given be the allowable losses of 1.3 W. When reducing the coil height the winding resistance decreases as well as the active wire length. Therefore a higher current supply is required to generate the nominal torque. In this way the axial drive height could be reduced further.

VI. CONCLUSIONS

Single sided BLDC drives have a flat and compact design. This makes them an optimal drive system for VAD systems. In combination with a contactless bearing system a maintenance free operation is possible for a long term assist. The mayor objective for a VAD support is to provide a sufficient perfusion without damaging the blood and cell structure. For the drive design these objectives are transferred to drive specifications in terms of torque, speed and maximum allowable losses. As single sided BLDC motors have the drawback of high attracting forces, the maximum forces, which can be compensated by the bearing, must be added to the drive specifications. In order to speed up the design process of a minituarized drive system, the design of an existing system has been shrinked in radial dimensions. Afterwards a parameter study is applied to adjust the electromagnetic drive characteristics. It turned out that parameters such as increasing the operation air or reducing the stator teeth height are sensitive in terms of force and torque. Nevertheless they have to be applied to roughly adjust the axial attracting force. Then varying the remanent induction by changing the magnet material or reducing the volume of the rotor magnets can be applied for a fine adjustment of the attracting forces.

In order to simplify design decisions the losses and attracting forces have been normalized by applying the Euclidian vector. When calculating the sum of the normalized values in dependency of the variation parameter, the minimum reveals the optimum design point. During parameter variation this optimum design point had to be ignored for some parameters. On the other hand the characteristic helped to make design decisions. For this reason the method itself works, if applying a suited object function. In this study losses and forces were equally weighted although the correct object function would have been to adjust the axial attracting forces to 10 N and keep the losses as low as possible.

Table 2: Prototype and miniaturized drive design.

design objectives	Prototype	Mini Design
nominal speed	2500 rpm	2500 rpm
nominal torque	12 mNm	12 mNm
axial force	10 N	10 N
copper losses	712 mW	953 mW

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