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Investigations Into Inductive Propulsion For Centrifugal Blood Pumps

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School of Electrical & Electronic Engineering

A thesis submitted to the Nanyang Technological University
in fulfilment of the requirement for the degree of
Doctor of Philosophy

2007

Statement of Originality

I hereby certify that the work embodied in this thesis is the result of original research and has not been submitted for a higher degree to any other University or Institution.

August 27, 2007

Date

Wu Peng

WU PENG

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SUMMARY

A fair amount of research has been done on the continuous-flow type of blood pumps and several models of these artificial hearts have even been made commercially available. However their usage is not widespread yet as there are still many areas in need of improvements. To overcome some of the current drawbacks, and also to advance the knowledge of centrifugal blood pump design from the electromechanical propulsion aspect, there is a need for research by combining computational electro-magnetics, innovative induction machine design and centrifugal pump principles to examine various new possible designs of shaft-less and compact blood pump devices. This is the overall objective of this project.

A novel concept of multi-phase inductively-driven shaft-less centrifugal pump has been proposed here. Several distinct designs of centrifugal blood pumps are proposed, which can all be classified as of the third-generation type of centrifugal blood pumps. Both radial-flux and axial-flux types of structures have been investigated. The axial-flux structures include single-stator type, dual-stator type and open-impeller type. The unconventional designs all have integrated rotor-impeller structures which combine the motor rotor with a centrifugal pump impeller. This would eliminate the need for a shaft to transmit the torque from motor to impeller. Compared with the earlier generations of blood pumps and even with some recent proposals, these new conceptual designs have advantages where torque transmission method, structure simplicity and compactness and system reliability are concerned.

Mathematical models for the proposed blood pump designs and magnetic bearings have been derived for the analysis and control of these devices. The derived models also aid in deeper understanding of the working principles of the proposed

SUMMARY

devices. Furthermore, certain design optimization and improvement have been facilitated by these mathematical model analyses. They also provided the basis to formulate simple but viable control algorithms to realize both motoring and magnetic suspension actions in future research.

In order to verify the proposed designs and derived mathematical models, Finite Element Method (FEM) analysis has been applied to the proposed inductively-driven blood pump designs and magnetic bearings. By analyzing and comparing the FEM simulation results with the mathematical model calculations, the feasibility and viability of both the proposed blood pumps and magnetic bearings and their derived mathematical models were affirmed. Furthermore, design optimization and structure improvement have been investigated through these FEM analyses and simulations.

An experimental test rig for the axial-flux open-impeller design was set up for furthering the feasibility study using experiments. Several experiments have been conducted and the experimental measurements were compared with FEM simulation results and calculations from derived mathematical model. The obtained experimental data corresponds with both the FEM simulation results and mathematical model calculations reasonably well, which confirmed the feasibility of proposed inductively-driven blood pump.

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LIST OF ABBREVIATIONS

(Alphabet Sequence)

AMB:	Active Magnetic Bearing
CBP:	Centrifugal Blood Pump
CHD:	Coronary Heart Disease
CHF:	Congestive Heart Failure
FEM:	Finite Element Method
IM:	Induction Motor
LVAD:	Left Ventricular Assist Device
MCSS:	Mechanical Circulatory Support Systems
MI:	Myocardial Infarction
PMB:	Passive Magnetic Bearing
PM:	Permanent Magnetic
RVAD:	Right Ventricular Assist Device
TAH:	Total Artificial Heart
TRF:	Torque Ripple Factor
VAD:	Ventricular Assist Device

Chapter One

INTRODUCTION

1.1 Background

Congestive Heart Failure (CHF) and Coronary Heart Disease (CHD) are major health problems with high mortality rate all over the world. Their ultimate therapy, heart transplantation, is limited by the shortage of donor hearts. Since decades ago, researchers have been working to solve this problem by developing Mechanical Circulatory Support Systems (MCSS) that can replace or assist the failing heart [1]. Such MCSS is usually called as artificial heart, which is a term more familiar with the public.

Artificial heart can be classified according to the type of assistance (left, right, or bi-ventricular assist), the nature of generated flow (pulsatile or non-pulsatile), the life-span (short-term, medium-term, or long-term), the core technology (electro-hydraulic or pneumatic), and etc. Another way to categorize artificial heart is based on the need of native heart replacement. Therefore the artificial heart can be categorized into two groups: 1) Total Artificial Heart (TAH); and 2) Ventricular Assist Device (VAD). The TAH is a blood pump meant to replace the function of the native heart totally. It can either be used temporarily as a bridge to transplantation or permanently as a long-term artificial heart. The VAD is also a blood pump but it connects in parallel with the ventricles of the natural heart. The purpose of VAD is to relieve the duties of the natural heart so that the ailing heart would have a chance to recover. The VAD does not require removal of the native heart and therefore can be used as a bridge to transplantation as well as a temporary support system in patients with expected myocardial recovery. The required cardiac replacement is the major difference between TAH and VAD.

1.2 Motivation

From the classification of artificial heart, it is clear that artificial heart, both TAH and VAD, are basically electrically-powered blood pumps with their controllers and power supplies. So the selection and design of suitable type of pump is very important to any artificial heart system. Continuous flow or rotary devices are currently the more popular pump technology for artificial heart. They have several potential advantages over their pulsatile counterparts:

- (1) They are smaller devices;
- (2) They are relatively simple and have fewer moving parts, which mean less possibility of mechanical failure;
- (3) Because of their continuous flow characteristics, there is no need of a compliance chamber in the system, thus the volume of the whole pump system can be reduced significantly;
- (4) They have lower energy requirement [2].

As additional consideration, there are two kinds of rotary pump, centrifugal type and axial flow type, which have major differences compared with each other. Centrifugal flow devices are somewhat larger than axial flow pumps, and can provide comparable output flow rate as that of axial flow pumps, but with much slower rotational speed (about 2,000–4,000 rpm vs. 10,000–20,000 rpm). In contrast to axial flow devices, the slower rotational speed of centrifugal pumps virtually eliminates the potential for air aspiration or tubing disruption. Moreover, the lower speed is less destructive to blood cellular elements and can significantly lower the danger of hemolysis. The very high rotational speed of axial flow pump may also cause thermal removal problem for the artificial heart. Because of the advantages of centrifugal pump over axial flow type, the former is therefore widely regarded as important and viable for long-term TAH or VAD, and was therefore chosen as the focus of this research.

A fair amount of research has been done on the rotary pump type of artificial heart and several models of rotary artificial heart have even been made commercially available like the Gyro PI-710, HeartMate LVAS, the Nikkiso Pump, the CFVAD

CHAPTER ONE: INTRODUCTION

Pumps, the Cleveland CorAide, the MicroMed DeBakey and the Jarvik-2000 [2]. However their usage is not widespread yet as there are still many areas in need of improvements:

- (1) In many designs, the coupling between the torque driver and pump impeller is some form of mechanical shaft and bearing, and their surfaces are a potential source of in-situ thrombus or thromboemboli [2], which is fatal to the patients;
- (2) The shaft seal of the blood pump chamber introduces blood leakage problem and reduces the reliability and durability of the overall artificial heart system;
- (3) Certain improved designs rely on magnetic coupling to transmit the torque from the external motor to the internal impeller, thus still involving rotating parts outside the blood chamber, which is not an optimization of volume;
- (4) Most torque drivers used in centrifugal blood pumps are based on permanent magnetic (PM) type of motors. While PM motors are excellent choice of torque drivers, there are other possible alternatives such as induction type of motors. The operation of Induction Motor (IM) inherently does not require information on the rotor position, unlike that of the PM motor. For the IM, it is more robust and the extra commutation circuit and position sensors are not needed for its control. It is possible to apply the simple but reliable open-loop with slip compensation control strategy based on Voltage/Frequency method to control the speed of the IM. Using induction motor principle may thus bring certain advantages but also disadvantages which have yet to be investigated;
- (5) The concept of producing the torque directly on the blood pump impeller without the need of mechanical or magnetic coupling has not been fully explored.

To overcome some of the drawbacks listed above, and also to advance knowledge of centrifugal blood pump design from the electromechanical propulsion aspect, there is a need for research by combining computational electro-magnetics, innovative induction machine design and centrifugal pump principles to examine various possible designs of shaft-less and compact blood pump device. These lead to the formulation of overall project objective and the detailed sub-objectives to achieve the former. These are described in the next section.

1.3 Objectives

The main objective of this research is **to investigate the feasibility of using induction machine concept and centrifugal pump principle to realize shaft-less and compact blood pump designs through analysis, simulations and experiments.**

This would be achieved through the following sub-objectives:

Sub-objective 1: To produce conceptual designs of inductive propulsion systems for centrifugal blood pumps

A compact motor-pump structure plays a critical role in the entire artificial heart system. The main difficulty is how to combine an induction motor structure and a centrifugal blood pump impeller into one compact unit without compromising their respective functions. Both radial-magnetic-flux and axial-magnetic-flux types of induction motor structures would be investigated. The unconventional designs would have integrated rotor-impeller structures which combine the motor rotor with a centrifugal pump impeller. This would eliminate the need for a shaft to transmit the torque from motor to impeller. Furthermore, how magnetically-assisted or hydro-dynamically-assisted bearings can be incorporated into the device to support and restrain the rotor-impeller structure is another challenge.

Sub-objective 2: To derive the mathematical models of the proposed designs

The mathematical models of the proposed unconventional shaft-less motor-pump designs and magnetic bearings are very important to the analysis or control of these devices. The derived models can also aid in deeper understanding of the working principles of the proposed devices. Furthermore, the design optimization and improvement can be achieved through the mathematical model analyses. The mathematical models of the proposed motor-pump devices and magnetic bearing systems can provide the basis to formulate simple but viable control algorithms to realize both motoring and magnetic suspension actions in further research.

Sub-objective 3: To apply Finite Element Method (FEM) analysis on the proposed designs

FEM analysis is a computer-aided mathematical technique used for obtaining numerical solutions to the abstract equations of calculus that predicts the response of physical systems subjected to external influences. With the advance in computing technology, this method finds more and more applications in the electromagnetic system design and analysis.

The objective here is to understand the nature of torques acting on the rotor-impeller structures by applying two-dimensional and three-dimensional static or time-stepping FEM to the motor-pump designs. By analyzing FEM simulation results and comparing the FEM simulation results with the mathematical model calculations, the feasibility and viability of both the designed motor-pump devices and derived mathematical models would be tested and verified. Furthermore, the design optimization and structure improvement would be achieved through the FEM analysis and simulation. The FEM analysis is also applied to prove the feasibility of designed magnetic bearing system.

Sub-objective 4: To conduct experimental study on the conceptual designs

In order to further verify the feasibility and viability of some of the proposed shaft-less motor-pump designs, experimental test-rigs consisting of engineering prototypes of electromagnetic portion of the proposed motor-pump devices would be set up. The experimental data would be collected from the test-rig and compared with the FEM simulation results and mathematical model calculation results to confirm the validity of the proposed designs and derived mathematical models.

1.4 Contributions

The main contributions of this project can be summarized as the following:

(1) Novel shaft-less motor-pump structure designs based on induction machine concept

The first important contribution of this project is the set of designs of the novel shaft-less motor-pump devices based on induction motor and centrifugal pump principles. After investigation of both radial flux and axial flux induction motor structures, several designs of motor-pump device have been proposed. In details, the structure and geometries of the essential parts of the designs have been obtained successfully.

The novel shaft-less motor-pump designs have the following advantages:

- 1) The novel designs combine the motor rotor and blood pump impeller as an integrated unit, which can develop the torque directly on the impeller magnetically and eliminate the need for mechanical or magnetic coupling between the motor and the impeller. This also means the elimination of blood chamber sealing and less possibilities of thromboemboli. Without the need of any rotating parts outside the blood chamber, the novel designs can be simpler and more reliable.
- 2) In certain designs, the possibility of using passive or active magnetic bearings to constrain the rotor-impeller unit inside the blood pump chamber has also been explored. Some of the active magnetic bearing forces can arise inherently in the motor-pump structure without additional windings. Then, the possibility of thrombus or thromboemboli could be reduced further.
- 3) By using induction motor principles, the electronic commutation circuit, rotor position sensors or sensor-less techniques associated with PM motors are not required. At the same time, the very simple open-loop with slip compensation control strategy based on Voltage/Frequency method can be adopted to control the speed of the induction-based pump. Such simplification can improve the reliability of an artificial heart system.

(2) Mathematical models of the proposed motor-pump designs have been derived.

Unlike critical servo applications which need complex transient models for their control, the proposed motor-pump devices are usually operated in the quasi steady states. As a critical and indispensable basis for the control of induction motor based motor-pump device in steady state, equivalent circuit models of the designed motor-pump devices has been derived and studied, which can eliminate the unnecessary complexity of transient models. Depending on the structure of different proposed designs, parameters of equivalent circuit for radial flux and axial flux type designs are analyzed and calculated, which are directly related to the geometries of the designs. For the designed magnetic bearings in one of the motor-pump devices, the discretizing method and equivalent magnetic circuit method was adopted to obtain their mathematical models respectively. All of these not only give deeper understanding of the working principles of proposed designs but also provide basic foundations for their future research in controller design.

(3) FEM analysis and simulations on the proposed designs

Depending on having axial discontinuities or not, the two-dimensional and three-dimensional time-stepping FEM analysis and simulations have been applied to the designs of motor-pump device, passive magnetic bearing and active magnetic bearing accordingly. From the FEM analysis and simulations, the feasibility and viability of the designs have been tested and verified. The FEM analysis results are also used to compare with mathematical model calculations to confirm the validity of the derived models. Furthermore, the design optimization and improvement has been carried out through the FEM analysis and simulations. Both the rotor and stator structures have been improved to provide more output electromagnetic torque without compromising the compactness of the designs. In addition, the optimized structure of rotor-impeller structure considering the relationship between motor torque ripple and average torque, and also rotor bar number and rotor bar skew angles has been achieved successfully for one design.

(4) Experimental test-rig consisting of one proposed motor-pump device

The set-up which consists of an engineering-prototype of one of the proposed motor-pump devices has been fabricated in the lab. The experimental data collected from the test-rig has been analyzed and compared with the simulation results from FEM analysis and calculation results from derived mathematical model. The accordance of these three types of results verified the feasibility and validity of both the proposed motor-pump concept and derived mathematical model.

1.5 Organization of the Thesis

The summary and organization of the rest of this thesis is introduced here:

In Chapter 2, the review of the development of artificial hearts, the centrifugal type of blood pumps, the basic knowledge about centrifugal pump, an overview of induction motor, and the introduction to magnetic bearings are presented.

Chapter 3 presents a novel design of centrifugal blood pump based on radial flux induction motor. At first, the overall structure of the blood pump system is introduced. Then, a sketch design is obtained to decide the main dimensions of the centrifugal blood pump chamber, which provide a necessary base for the next design stage. This is the design of the torque driver part of the pump. Magnetic bearings are also designed to constrain and support the rotor-impeller unit. The structure of both passive magnetic bearing (PMB) and active magnetic bearing (AMB), and how they restrict the undesired degrees of motion freedom of the rotor-impeller unit are presented. Then, the mathematical modeling of the designed blood pump device and magnetic bearings are presented respectively. Finally, FEM analysis and simulation results of the proposed designs of both the radial flux motor-pump device and the magnetic bearings are presented. The simulation results obtained verified the viability of these designs and the derived mathematical model.

Chapter 4 includes two designs of axial-flux type of centrifugal blood pump, both of which belong to the third-generation centrifugal blood pump design. At first, the design philosophy is presented, which serves as the general guide for the entire design process. The structure and dimensions of the blood pump chamber are similar to that of the radial flux type design introduced in Chapter Three. Based on the obtained structure and dimensions of the blood pump chamber, the design of the axial-flux torque driver carried out is described in details. Then, analysis employing the equivalent circuit model of conventional induction motor is described, concentrating on the motor-pump operation and the produced mechanical torque. The

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last section presents the FEM analysis and simulation results of these new designs as well as of the various options available to improve these initial designs.

Chapter 5 introduces a new concept of multi-phase inductively-driven shaft-less centrifugal blood pump based on the open-impeller structure. The philosophy behind its design is explained. The structure and electromagnetic design of the proposed system is introduced along with its mathematical modeling. Both physical model analysis and FEM analysis have been used to determine the parameters of the model. Time stepping and steady state FEM simulations have been applied to investigate the feasibility and potential of the design. The 3D FEM simulations are also used to check the validity of the mathematical model and the feasibility of the motor-pump, as well as optimizing the rotor structure of this open-impeller. The optimized structure of the rotor considering the relationships between motor ripple torque and average torque, and between rotor bar number and rotor bar skew angles is presented here. Experimental measurement results to further study certain aspects of the design have been obtained and are presented.

Chapter 6 gives the conclusions of this thesis and the recommendations for future work.

Chapter Two

LITERATURE REVIEW

2.1 Introduction

The artificial heart research has experienced over 60 years of progress. This research has achieved very great and exciting breakthroughs during its development. Artificial heart has evolved from a vagarious idea to a clinically applicable device, which has begun saving human lives. However, there still are many problems with present artificial heart systems. These problems prevent currently available artificial hearts from becoming reliable, permanent and totally implantable device which can support the normal daily life activities of a human being for substantial duration. In order to solve these problems, more research efforts are needed, hence the motivation for this project. To set up clear research objectives, it is however necessary to investigate and review thoroughly the development of this technology.

In this review, basic knowledge of human heart and heart diseases, the classification of artificial heart, evolution of different types of artificial hearts and the comparison among them will be examined. Then, the review will focus on the core of this project – the centrifugal blood pumps. The problems faced by this technology and the current stage of centrifugal blood pump development are reviewed thoroughly. After that, the future problems and development trends are summarized.

Because of the interdisciplinary characteristics of artificial heart technology, other background information, including basic concept of general centrifugal pump, review of magnetic bearings and fundamental knowledge of electric motors are also presented to give complementary information, which would be helpful to lead to a comprehensive understanding of the technologies applied to this project.

2.2 Development of artificial heart

2.2.1 Heart structure and function

Fig 2- 1: Structure of human heart [3]

The heart, which is a four-chambered pump that works based on positive displacement, is the most important organ in the human body. The overall structure of the human heart is shown in Fig 2-1. Each of the four pumping chambers is composed of different amount of heart muscle or myocardium. Each chamber has a valve to prevent the backflow of blood. The heart contracts when its muscle is excited by electrical impulses carried through its conduction system. The electrical impulses help control heart rate and rhythm and may affect the strength of its contraction. The energy requirement of the heart is provided by the blood that circulates through a series of blood vessels within the heart known as the coronary circulation. This is responsible for distributing oxygenated blood and other nutrients to the working heart muscle. The heart is connected to the rest of the circulatory system by large veins and arteries [4, 5].

The heart performs two separate pumping functions: (1) the right atrium and the right ventricle are responsible for collecting blood returning from the organs and tissues of the body and pumping this oxygen-depleted blood to the lungs where it is oxygenated; and (2) the left atrium and left ventricle are responsible for receiving oxygenated blood from the lungs and pumping it to the organs and tissues of the body which require oxygenated blood to perform their normal functions. Each pumping cycle of the heart consists of an ejection phase "systole" and a filling phase "diastole". During systole, the muscles comprising the right and left ventricle contract forcefully and cause the blood

to be ejected into the lungs and other parts of the body. The amount of blood pumped through the heart each minute is called the cardiac output [4, 5, 6]. The flow chart of human blood circulation is shown in Fig 2-2.

Fig 2- 2: Flow chart of blood circulation

2.2.2 Congestive Heart Failure (CHF) and Coronary Heart Disease (CHD)

Fig 2- 3: Leading causes of death in the world (WHO 2002)

Like all machines, the heart can malfunction in numerous ways. Cardiovascular disease occurs when the heart becomes “clogged, broken down, or in need of repair” [7]. According to World Health Organization (WHO) statistics, 16.56 million people died

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of heart failure worldwide in 2002, making it the number one cause of death, as shown in Fig 2-3. In U.S., 64.4 million Americans have at least one kind of cardiovascular disease [8]. Heart disease also is the number one cause of death in America. The need to understand how the heart fails and what means are available to fix these problems is extremely important and current [9]. Since many conditions fall under the category of cardiovascular disease, we will focus on the two main causes for heart transplantation and the need for artificial hearts: Congestive Heart Failure (CHF) and Coronary Heart Disease (CHD). Among all kinds of cardiovascular diseases, CHF and CHD are two major heart problems with high mortality rate in most affluent countries.

CHF is a kind of heart failure in which the heart is unable to keep enough blood circulating in the tissues or is unable to pump out the blood returned to it by the veins [10]. The blood starts to fill in the lungs, which leads to congestion. Therefore, the heart must work harder in order to meet the body's oxygen demands. This behavior causes excessive wear to the diseased organ. Initial symptoms of CHF, such as fatigue and swelling of the ankles, are usually so minor that patients are not diagnosed until the condition becomes much more severe. As the disease progresses, patients start to suffer from shortness of breath and palpitations even while remaining stationary [13]. For extremely severe cases, transplantation is the only option. CHF afflicts over 15 million patients worldwide [11]. An estimated over 5 million patients were diagnosed with congestive heart failure in the U.S. in 2001, with approximately 1.5 million of them having moderate-to-severe form [12].

CHD is a general term for both heart attack and angina. About 7 million Americans suffer from it and 500,000 Americans die from it each year [15]. CHD occurs when the arteries that supply blood to the heart muscle (coronary arteries) become hardened and narrowed. This can be due to the buildup of plaque on the inner walls or lining of the arteries (atherosclerosis). Blood flow to the heart is reduced as plaque narrows the coronary arteries. This decreases the oxygen and nourishment supply to the heart muscle [16]. Patients suffering from CHD often exhibit symptoms such as severe chest pain and fatigue due to the lack of oxygenated blood [13]. For severe cases of CHD, the only cure is a heart transplant.

2.2.3 Heart transplantation and mechanical circulatory support systems

There are several treatments for heart disease, which include lifestyle changes, medication, angioplasty, stent placement, bypass surgery and heart transplantation. According to what kind of heart disease and how severe the problem is, different treatments are applied to different patients. However for most end-stage CHF and CHD patients, heart transplantation is the only option. Severe cardiovascular disease is the leading cause for heart transplantation, but other malfunctions such as valve damage and chamber problems also require the need for a new heart.

By definition, heart transplantation, also called cardiac transplantation, is the replacement of a patient's diseased or injured heart with a healthy donor heart [17]. The first heart transplants were performed in the late 1960s. However it was not until the use of anti-rejection medicines in the 1980s that the procedure became an accepted operation [18]. Even though heart transplantation is a cure for end stage heart disease patients, there are many limitations to the procedure. One major limitation is the very strict criteria on cardiac transplantation candidate selection [19, 20]. Many patients that are ineligible for heart transplantation have no option for end stage cardiac disease.

Another main limitation is the serious shortage of donor heart. According to the statistics of WHO, there are eight hundred thousand people worldwide who need heart transplants each year. Unfortunately, only no more than 0.4% of them can get donor hearts. As a consequence, about 99% of these patients do not receive heart transplant. Even in the United States, where the greatest number of heart transplantation are performed, only 2000 out of more than 100000 severe heart disease patients are able to receive heart transplants each year [21, 22].

Therefore, further development is necessary to provide a universal solution for these patients, who have little hope of getting heart transplantation. At the same time, the patients who have cardiac failure after cardiac operations, myocardial infarction (MI) shock or acute cardiomyopathy due to myocarditis or other causes, need assistance of blood circulation because of the heart invalidation. Those who are suitable for transplantation but deteriorate before a donor heart becomes available also require circulation support prior to transplantation. Since decades ago, researchers have been

working to solve these problems by developing Mechanical Circulatory Support Systems (MCSS) that can replace or assist the failing heart [23]. The MCSS is usually called as artificial heart, a term which is more familiar with the public.

2.2.4 History of artificial heart

LeGallois [24] postulated the use of mechanical circulatory support in 1812 and DeBakey proposed a continuous flow blood transfusion instrument using a simple roller pump in 1934 [25]. In 1961, Kolff and Moulopoulos developed the first intra-aortic balloon pump [26], which is deemed as the beginning of artificial heart. The general timeline in Table 2-1 gives a brief introduction on the major development in artificial heart technology during the past 50 years.

Table 2- 1: Milestones in artificial heart technology in past 50 years

1960s	<ul style="list-style-type: none">Engineers invented the first artificial heart
1970s	<ul style="list-style-type: none">Surgeons implanted the first artificial heartEngineers developed the ventricle assist device as an alternative to artificial heartFirst long time artificial heart resulted in poor quality of life
1980s	<ul style="list-style-type: none">Ventricle assist device showed potential for long-term supportTranscutaneous technology eliminated the need for skin-protruding electrical wiring
1990s	<ul style="list-style-type: none">Patients with long-term ventricle assist device recovered from heart failure
2000s	<ul style="list-style-type: none">Results of the Abiocr reflected improved quality of life for patients after implantation

Artificial heart has undergone three generations of development [27]. First generation artificial hearts are those with positive displacement or pulsatile pumps. Examples include CardioWest TAH, Thoratec bi-VAD, Novacor VAD, Pierce Lion heart, TCI HEARTMATE I VAD, etc. Second generation artificial hearts would include the rotary pumps with contact bearings or/and seals. Examples of this generation include MicroMed DeBakey VAD, Jarvik 2000 VAD, TCI HEARTMATE II VAD, etc. Third generation artificial hearts are based on blood pumps without mechanical bearings. This group would include those with magnetic bearings and/or with hydrodynamic bearings. Examples would include Terumo VAD, U. of Pittsburgh Streamliner VAD, MedQuest Heartquest VAD, VentraAssist, Cora Heart, etc. The first two generation artificial hearts are relatively mature and are unlikely to gain further significant clinical

acceptance. Most current research works may be classified as belonging to the third generation.

2.2.5 Classification of artificial heart

Fig 2- 4: Differences between TAH and VAD

Fig 2- 5: Classification of artificial hearts

One basic way to categorize artificial heart is based on the need of native heart replacement. Therefore the artificial heart can be categorized into two groups: (1) Total Artificial Heart (TAH); and (2) Ventricular Assist Device (VAD). The TAH is a complex blood pump system and it can replace the function of native heart totally. Because the device replaces the native heart, it can either be used temporarily as a bridge to transplantation or permanently as a replacement artificial heart. The VAD is also a blood pump that connects in parallel with the ventricles of the natural heart. The purpose of VAD is to relieve the duties of the natural heart so that the diseased heart

would have a chance to recover. The VAD does not require the removal of the native heart and therefore can be used as a bridge to transplantation as well as a temporary support system in patients with expected myocardial recovery. The required cardiac replacement is the major difference between TAH and VAD, as illustrated in Fig 2-4. Further classification of these two main classes is shown in Fig 2-5.

2.2.6 Total Artificial Heart (TAH)

The TAH consists of two blood pumps that are implanted orthotopically, i.e., positioned in the place of the patient's native heart within the pericardium. The first experimental TAH implantation was reported by Akutsu and Kolff in 1958 [28]. Their pneumatically driven TAH was able to keep a dog alive for a period of six hours. In 1969 Cooley introduced the concept of a two-staged cardiac replacement and performed the first clinical TAH implantation as a bridge to transplant [29]. Fourteen years later, in 1983, the first implantation of TAH as a permanent replacement of the native human heart was reported [30]. From 1969 to 1991 a number of 230 TAHs were implanted at 39 centers worldwide, using eleven different devices [31]. Nowadays the use of TAH as a mechanical circulatory support has become an established procedure for bridging patients with cardiomyopathy to cardiac transplantation. In patients with contraindications for heart transplantation, the TAH serves as a permanent heart replacement.

The first generation of TAHs was all pneumatically driven and possessed similar design characteristics. They consisted of two blood pumps [32], each composed of a rigid blood chamber and an air chamber. The membrane, separating the blood from the air, was moved by air derived from an external drive unit. The air tubes that connect the TAH with the driver pass through the skin. Artificial valves, positioned in the inflow and outflow tract, ensure unidirectional flow.

In general, pneumatic assist devices show a high incidence of infections, mainly caused by the percutaneous drivelines, which makes these devices unsuitable as a permanent artificial heart. Therefore, the second generation of TAH was focused on the realization of totally implantable, wireless, electrically powered TAH. The Abiomed/Texas Heart Institute TAH, known as AbioCor. (Abiomed, Inc., MA, USA), uses a high-speed

brushless direct current motor to power a unidirectional centrifugal pump [34, 35]. The Nimbus/ Cleveland Clinic Foundation TAH also uses a high-speed brushless direct current motor and hydraulic actuator to drive two diaphragm-type blood pumps [36, 37]. The Sarns/Pennsylvania State University TAH uses a low-speed high-torque brushless direct current motor and a dual pusher plate roller screw energy converter [38].

2.2.7 Ventricular Assist Device (VAD)

The VAD is mechanical blood pump that can support or replace the function of the failing left ventricle, right ventricle, or both ventricles. The VAD does not require removal of the native heart and therefore can be used as a bridge to transplantation as well as a temporary support system in patients with expected myocardial recovery. Left ventricular assist with a roller pump was introduced by Dennis in 1962 [39]. In 1971 DeBakey reported successful left ventricular assist of patients who could not be weaned from the heart-lung machine [40]. The VAD can be divided into three main groups:

- (1) Left ventricular assist device (LVAD): blood is withdrawn from either the pulmonary veins, left atrium or the apex of the left ventricle and returned to the ascending aorta;
- (2) Right ventricular assist device (RVAD): blood is withdrawn from the right atrium and returned to the main pulmonary artery;
- (3) Biventricular devices (BiVAD): the BiVAD is actually a combination of LVAD and RVAD, and could be seen as a functional heart replacement.

Additionally, the VAD can be categorized into non-pulsatile and pulsatile VAD. The type of ventricular support as well as the device used differs from patient to patient according to the experience and individual preference of the surgical team and the perspective for ventricular recovery.

A. Non-pulsatile ventricular assist devices

According to the mechanical characteristics, the non-pulsatile VAD can be divided into two main categories: centrifugal blood pump and axial flow blood pump.

Centrifugal blood pump:

Centrifugal blood pumps consist of a non-occlusive pump head positioned within a rigid pump housing. The rotating head consists of a various number of impeller blades that generate a non-pulsatile unidirectional flow by creating a vortex. Centrifugal blood pumps do not require artificial valves.

Fig 2- 6: Centrifugal type ventricular assist device configuration [88]

Axial flow blood pump:

The MicroMed DeBakey VAD is an axial flow pump jointly developed by Michael DeBakey and the National Aeronautics and Space Administration (NASA). The pump is 86-mm long, 25-mm wide (about the size of an AA battery), and weighs 95 grams [42]. The pump can produce up to 6 L/minute flow at about 10,000 rpm, and requires less than 10 watts of power. Fig 2-7 shows its structure.

Fig 2- 7: Structure of MicroMed DeBakey axial flow VAD [42]

The Jarvik 2000 Heart, named for its designer Robert Jarvik, is another compact axial flow impeller pump. The pump is 25 mm in diameter, 55 mm long, weighs 85 gram,

has a displacement volume of 25 ml, and provides up to 8 L/minute blood flow at 8,000 to 12,000 rpm.

B. Pulsatile ventricular assist device

All pulsatile ventricular assist pumps are basically membrane pumps in which the membrane is moved by air, liquid or pusher-plate. The driving source (air, water or electricity) reaches the pump via a tube through the skin. The most often use of these devices is bridging to transplantation [44]. The Abiomed Biventricular Support (BVS) 5000 is an external cardiac support system that can be used for short-term left, right or biventricular support. The device consists of an automated driving console and disposable dual-chamber blood pumps. The device operates in asynchronous full-to-empty mode with a maximum stroke volume of 82 ml [45]. Fig 2-8 shows the structure of Abiomed Biventricular Support (BVS) 5000 and its connection with heart. The HeartMate 1000 IP is another totally implantable pneumatically driven blood pump designed for left ventricular support only [46].

Fig 2- 8: Abiomed Biventricular Support (BVS) 5000 pneumatic pulsatile VAD [45]

Fig 2- 9: Novacor N100 electric pulsatile VAD

Novacor N100 is an electromechanical totally implantable pump designed for left ventricular support only. HeartMate 1000 VE is similar to the pneumatically driven version (HeartMate 1000 IP), but the pusher plate is driven by a low-speed torque motor instead of by air.

2.2.8 Comparison of different types of artificial hearts

The emergence of pulsatile type of blood pump was largely due to researchers trying to mimic the working principle of native human heart. They do have many advantages. Firstly, these devices generate a pulsatile flow. The pulsatile flows reduce sympathetic nerve activity and peripheral vascular resistance, thus improving the microcirculation as well as organ function [48]. Secondly, if concerned with heart function recovery, pulsatile blood pump is the best choice which can augment the diastolic pressure so that perfusion pressure through the coronary artery will be increased [49]. Furthermore, the pulsatile VAD causes less blood trauma [50]. However, pulsatile artificial heart also has many drawbacks. Firstly, there is a high incidence of infections, mainly caused by the percutaneous drivelines. This is dangerous and even fatal to patients. Secondly, the membrane structure of this type of artificial heart requires very complex pump design and control method, which decreases the reliability of the whole system. The other major disadvantages of the pulsatile pumps are the high cost due to the required heart valves, large physical volume and low efficiency.

Compared with pulsatile type, there is no membrane structure in continuous-flow type blood pump and its driver is also implanted inside human body. They also rely on rotary mechanical pump as its driver, which is less expensive and more reliable than the membrane structure. Furthermore, rotary pump needs less power than the membrane structure and do not need expensive valves. So, continuous-flow blood pumps can overcome some of the drawbacks of pulsatile blood pumps. Although the pulsatile type has such advantages like less sympathetic nerve activity, less peripheral vascular resistance, better microcirculation and less blood trauma, it is now generally accepted in the artificial heart research community that continuous-flow blood pumps are more viable. So, most current artificial heart research work focuses on the continuous-flow type. The comparison of the main advantages and disadvantages between pulsatile and continuous-flow blood pumps is summarized in Table 2-2.

The continuous-flow blood pumps can be further divided into centrifugal type and axial-flow type. These two sub-groups have differences compared with each other in following aspects: pump size, rotation speed, damage to blood cell and thermal conduction problem. Centrifugal-flow devices are somewhat larger than axial-flow pumps, but the rotational speeds are much slower (about 2,000–4,000 rpm vs. 10,000 – 20,000 rpm). The slower rotational speed of centrifugal pumps virtually eliminates the potential for air aspiration or tubing disruption. Moreover, the centrifugal pumps are less destructive to blood cellular elements and cause significantly lower hemolysis.

Table 2- 2: Comparison between pulsatile and continuous-flow blood pumps

Continuous-Flow Blood Pumps	Pulsatile Blood Pumps
Smaller in size for a specified outflow	Bigger in size for the same specified outflow
Consume less power for a specified outflow	Require more power for the same specified outflow
Valves are not required	Valves are necessary, which are sources of thrombus
Less moving parts	More moving parts, which degrade the device reliability
Does not need flexing plastic chamber	Require flexing plastic chamber which degrade reliability
Possibility of auto-control	More difficult to control
Possibility of higher cell damage if pump chamber is not well designed	Shear stress on blood cells not much greater than normal native hearts
Requires higher output volume	Requires less output volume

Table 2- 3: Comparison between centrifugal and axial-flow blood pumps

	Rotation speed	Pump size	Damage to blood	Thermal conduction problem
Centrifugal type	Low	Big	Less	Easy
Axial flow type	Very high	Small	More	Difficult

For axial flow pumps, a major drawback is the inability to produce high flow rates. Typical flow rates have been 2–4 L/min with rotation rates of up to 25,000 rpm. To produce even higher flow rates, such pumps would need higher motor speeds raising the possibility of increased hemolysis. Also, the very high rotational speed of axial flow pump causes difficult thermal conduction problem inside the artificial heart. Therefore some researchers have favored centrifugal pumps because these larger devices would satisfy the demand for larger flow rates. The larger contacting surfaces as well as the

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potential for more flow have been the advantages of centrifugal pumps over axial ones, particularly as bridge to transplant and total heart replacement devices. The summary of differences between centrifugal type and axial flow type are listed in Table 2-3.

2.3 Review of Centrifugal Blood Pumps (CBP)

As the focus of this project is on centrifugal blood pumps, it is helpful to have a detailed review of the development of CBP as well as of the basic concept of centrifugal pump in general.

2.3.1 Basic concept of a centrifugal pump

The centrifugal pump was firstly developed in Europe in the late 1600's. Its wide spread use, however has occurred only in the last century, largely due to the comparatively recent development of high-speed torque drivers such as electric motors, steam turbines and internal combustion engines. The centrifugal pump is a relatively high-speed machine and the development of high-speed drives has made possible the development of compact, efficient pumps. Nowadays, it is common to find efficiencies of more than 93% for large pumps and better than 50% for small fractional horsepower units.

Centrifugal pump is basically a machine that imparts energy to a fluid, and this energy infusion can cause the fluid to flow, or to rise to a higher level, or both. It is a member of a family known as rotary machines and consists of two basic parts: 1) the rotary element or impeller, and 2) the stationary element or housing. Fig 2-10 shows the cross section of a centrifugal pump.

Fig 2- 10: Cross section of a typical electrically-driven centrifugal pump

As its name implies, a centrifugal pump functions by slinging liquid off its impeller via centrifugal force. Its rotation imparts energy to the liquid causing it to leave the

impeller vanes at a greater velocity than it possessed when it entered the pump. This outward flow reduces the pressure at the inlet suction eye, allowing more liquid to enter. The liquid that leaves the impeller is collected in a part of the housing called the volute, where its increased velocity results in increased pressure [92-99]. The volume of liquid that flows out depends upon the velocity of the rotating impeller. The faster it rotates, the greater the centrifugal force and therefore the greater the volume of liquid discharged. So the flow and head (pressure) developed by a centrifugal pump depends upon the rotational speed, or more precisely, the peripheral velocity of its impeller.

2.3.2 Development of centrifugal blood pump

Centrifugal pumps have been studied sporadically as blood flow devices for many years since 1964 [51]. Although they may be designed in many specific configurations, two main components have constituted the basic configuration among all centrifugal pumps. These are the impeller together with a cauldron-shaped housing with inlet and outlet. The impeller may consist of blades, cones, or disks with a multitude of combinations. Figs 2-11, 2-12 and 2-13 show some examples of CBPs reported in publications [51-84]. Although most CBPs have similar features and operating parameters, most differ in critical ways that make each pump unique. However, major unifying characteristics among most CBPs are the use of some form of electromagnetic and electromechanical means to generate and transmit the torque to the impeller, and some means to support the rotating impeller.

The support of the impeller is perhaps one of the biggest issues in the development of rotary blood pumps. The impeller needs to be supported to constrain the unwanted degrees of freedom when it is rotating. How this is done has great impact on the reliability of the device, the hemolysis issue (destruction of red blood cells) and the thrombosis issue (the formation of fatal blood clots). The quest for the ideal form of bearing support inevitably influences the electromagnetic, mechanical, and fluid dynamic aspects of the pump design philosophy. The various types of bearing support considered so far can be classified into mechanical contact bearings, blood or fluid bearings, magnetic bearings and hydrodynamic bearings.

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The earliest forms of CBP naturally borrow their design from the electric motor driven centrifugal pumps used so successfully in the industries. The torque is generated by an external motor and transmitted by a mechanical shaft to the impeller. The impeller bearing support is based on shaft mounted mechanical contact bearings with seals to prevent leakage of blood from the pump housing. An example of such first generation CBP is the original Blackshear-Medtronic centrifugal blood pump tested in 1979, shown in Fig 2-13(a) [51, 52]. Such pumps cannot be used for more than 2 weeks without servicing because of the tendency of formation of blood clots at the seal/bearing areas. Also, the mechanical bearings and seals occupy additional volume making the pumps unsuitable for implantation. However these designs are inexpensive and mature in terms of technology. Hence they have been successfully applied in cardiopulmonary bypass operations.

The elimination of the shaft and its accompanying shaft-seal was the next step in the advancement of blood pumps. Instead of using a shaft to transmit the developed torque from electric motor to the impeller, magnetic coupling was used. The motor shaft now drives a disk (known as driver disk) containing permanent magnets and is located outside the pump chamber. On the interior side of the pump wall, a similar disk (known as follower disk) with permanent magnets faces the external disk. When the latter is rotated by the electric motor, the follower disk also rotates because of the magnetic coupling between the disks through the pump wall. The follower disk is attached to the impeller or in some cases is actually the shroud of the impeller, and they are together supported primarily by mechanical contact bearings. The magnetic coupling may be radial (Tai-Ta, Fig 2-11c), or axial (Gyro C1E3, Fig 2-11d).

There were several reported designs of the mechanical contact bearings: using single-sided pivot bearing (TMDU, Figs 2-11e and f), using double-sided pivot bearings (Gyro PI-710, Fig 2-11g), or using roller bearings with seals (Nikkiso HPM-15, Fig 2-12a). In some cases, the mechanical bearings are aided by magnetic or hydrodynamic forces. There have been many examples of such second generation CBP and a number of them have been marketed commercially. They have been used for cardiopulmonary bypass, percutaneous cardiopulmonary support, as bridge to transplantation and even as bridge to recovery [53, 54, 55].

The development of a long-term, implantable blood pump for up to 2 years as a bridge-to-transplantation requires a non-mechanical approach such as magnetic or hydrodynamic bearings to provide support of the impeller. To produce a CBP suitable for long-term implantation, the design has also to be more compact and lighter in weight. The usual concept in these so-called third generation CBPs is to integrate the motor design into impeller so that there is no rotating parts outside the blood chamber. In other words, the rotating part of the motor, the rotor, becomes part of the impeller, leaving the stator of the motor outside the chamber (Ibaraki, Figs 2-12c,d; NIAIST, Fig 2-12e). In addition, the stator coils with perhaps some additional coils can also generate magnetic levitation forces on the rotor. For example, in one design, the impeller is supported by a passive radial magnetic bearing acting in tandem with hydrodynamic bearings (Kriton, Fig 2-11h). Torque is transmitted to the impeller by electromagnetic coupling via an integrated radial flux or axial flux motor. Blood lubricated lightly loaded thrust bearings may be required to constrain translation of the rotating assembly in the axis of rotation [53, 54, 55].

The more recent efforts in CBP development can be seen in Fig 2-13. These are the HeartQuest CFVAD III pump, the Cleveland CorAide pump, the VentrAssist pump and the HeartMate III pump, all using permanent magnet rotor-impellers. The main features of all the CBPs reviewed are summarized and compared in Table 2-4.

(a) 1979 Blackshear-Medtronic CBP with shaft

(b) CBP with multiple-disks instead of vanes [61]

(c) Tai-Ta CBP with radial magnetic coupling [64]

(d) Gyro C1E3 CBP with axial magnetic coupling [87]

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(e) Single-sided pivot pumps from TMDU [59, 60] using tripod ceramic balls

(f) Using single ceramic ball as bearing

(g) NEDO Gyro PI-710 CBP with double-sided pivot bearings [67]

(f) Kriton pump with passive magnetic bearing [69]

Fig 2- 11: Examples of experimental centrifugal blood pumps

(a) Nikkiso HPM-15 CBP with axial magnetic coupling [88]

(b) Medtronic Bio-Pump Plus with cone-shaped impeller [89]

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Fig 2- 12: Further examples of experimental centrifugal blood pumps

(c) The VentrAssist pump with PM impeller blades [77]

(d) Exploded view of HeartMate III pump [83]

Fig 2- 13: Recent efforts in CBP development

Table 2- 4: Comparison of Centrifugal Blood Pumps

Name	Organizations	Torque Production and Transmission	Impeller and Support	Development History
Blackshear-Medtronic	Medtronic Inc; U. of Minnesota	Torque produced by external permanent magnet brushless DC motor and transmitted to impeller by mechanical shaft through seal in chamber wall	Bladed impeller on shaft supported by mechanical bearings	One of the early rotary blood pumps. Tested in 1979 on animals. Now superseded by more advanced designs.
Multiple Disk Pump	Virginia Commonwealth U.; Texas A&M U.	Torque produced by external motor and transmitted to impeller by mechanical shaft through seal in chamber wall	Multiple disk impeller on central shaft supported by mechanical bearings	First reported in 1990 [61]. Optimal design studies are on-going [62]
Tai-Ta	National Taiwan University	Torque produced by DC motor and transmitted to impeller by radial flux PM coupler [64]	Unknown, probably mechanical contact type	First reported in 1996 [63]. Currently under going hemodynamic study in 2004 [65]
Gyro C1E3	Baylor College of Medicine	Torque produced by brushless DC motor and transmitted to impeller by axial flux PM coupler [66]	Double sided mechanical contact pivot bearings	First reported in 1994
Gyro PI-710	Baylor College of Medicine	Torque produced by brushless DC motor and transmitted to impeller by axial flux PM coupler [66]	Double sided mechanical contact pivot bearings. Later use single ceramic ball as bearings [60]	Final animal studies and two-year endurance studies on-going in 2004[66]
TMDU	Tokyo Medical and Dental University	Brushless DC motor driving, axial flux PM coupler	Initially use three ceramic balls in tripod configuration [59]. Single sided mechanical contact pivot bearings	First reported in 2000. Awaiting in vivo animal studies in 2001[60]
Nikkiso	NIKKISO CO., LTD	Brushless DC motor driving, mechanical shaft transmits torque to impeller	Use mechanical bearing to support the shaft and impeller	First reported in 1997.

(Table continues next page)

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Table 2-4: Comparison of Centrifugal Blood Pumps (con't)

Name	Organizations	Torque Production and Transmission	Impeller and Support	Development History
Medtronic Bio-Pump	MEDTRONIC, Inc.	Torque produced by external motor and transmitted to impeller by permanent magnetic coupling	Mechanical bearing and sealing	First reported in 1990
Terumo Capiiox SP	Terumo Heart, Inc.	External motor and magnetic coupling	Mechanical bearing and sealing	First reported in 1993
Kriton Pump	Kriton Medical Inc.	Axial permanent magnet	Passive radial magnetic bearing with hydrodynamic axial thrust bearing	First reported in 1998. Long-term animal trials
Ibaraki Pumps	Ibaraki University	Radial flux permanent magnet	Magnetic bearing	First reported in 1999.
Nanyang Pumps	Nanyang Technological University	Axial flux PM rotor-impeller.	Passive radial magnetic bearing and active axial magnetic bearing	First reported in 1999 and several improved designs in progress till now
VentrAssist	University of Technology, Australia	Torque produced on PM impeller blades directly by axial flux stator coils	Unusually shaped thick-bladed impeller promotes levitation by hydrodynamic forces	First report in 1999. Last reported in 2004 on rotor dynamic stability tests [77]
CFVAD / HeartQuest	University of Virginia, University of Utah, MedQuest Products	Axial flux permanent magnet	Active magnetic bearings in five degrees of freedom	First reported in 1997. Last reported in 2000 [78]
NIAIST Pump	National Institute of Advanced Industrial Science and Technology, Japan	Axial flux permanent magnet	Mechanical mono-pivot and passive magnetic bearing [79]	First reported in 1997. Last reported in 2004 under study of thrombus formation [80]
CorAide	Cleveland Clinic Foundation	Torque produced directly on impeller by radial flux bearingless PM motor	Fluid film bearing	First reported in 1994. Animal trial reported in 2001 [91, 126]
HeartMate III	Thermo Cardiosystems Inc., Univ. of Pittsburgh	Bearingless PM motor with direct driven rotor-impeller	Active magnetic bearings in 3 DOFs and passive magnetic bearings in 2 DOFs	First reported in 2000.

The classification of CBP development into first, second and third generations as described above is illustrated in Fig 2-14, and the main features and issues in these generations are summarized in Table 2-5. This way of classification is also in accordance with the generally accepted classification for the wider family of all blood pumps both pulsatile and continuous-flow. The review also reveals that almost all designs are based on PM motors; induction motor concepts have not been considered. The wide variety of designs also indicates that this area is far from mature, and there is

little consensus as to which type of design will succeed eventually. The only certainty is that the third generation feature of integrated rotor-impeller is the current trend.

Table 2- 5: Generations of Centrifugal Blood Pumps

	First Generation	Second Generation	Third Generation
Torque production	External motor	External motor	Electromagnetic interaction between rotor-impeller and stator
Torque transmission	Mechanical shaft	Magnetic coupling	Torque is developed directly on rotor-impeller
Impeller support	Shaft bearing with seals	Contact bearings within bloodstream or shielded by seals	Magnetic and/or hydrodynamic bearings
Volume and weight	Bulky and heavy	Moderate	Compact and light
Lifespan	2 days	2 weeks	2 years or more
Major disadvantages	Thrombus formation at seals	Wearing of bearings and consequent particle discharge	Complex design and control; require multiple sensors
Applications	Bypass operations	Bypass operations and recovery after cardiovascular operations	Bridge to transplant; Permanent implantation

Fig 2- 14: The evolution of CBP

2.3.3 Other issues in CBP Development

Besides the issue of electromechanical propulsion and stabilization of the pump impeller, there exist a host of other problems in the development of blood pumps. These problems and their causes are listed in Table 2-6 [85].

Table 2- 6: Main problems of current artificial heart systems

Problems	Main causes	Possible solution (applied already or under research)
Bleeding	Coagulation system caused by synthetic liver dysfunction; Large and long time surgical procedure; Large contact area of the device surface to the blood;	Meticulous surgical technique ; Intra-operative use of surgical glues; Use of FFP and platelets;
Right ventricle failure	Loss of ventricle septal function due to the left ventricle assist device; Increased pulmonary vascular resistance due to bleeding; Unable to meet increased demand of right ventricular output caused by high left ventricular assist device flows;	Use of nitric oxide and mild hyperventilation; Prevention of surgical bleeding;
Thromboembolic complications	Contact of blood with foreign materials; non-physiologic flow characteristics of the blood;	Use of anti-coagulation drug; Application of “smooth” enough material inside blood pump chamber; Use of magnetic bearing instead of mechanical bearing to support the impeller;
Infection	Contact of driveline with skin when penetrate through skin; Bacterial colonization of the inner device surface ; Thromboembolic events;	Use of antibiotics; Use of induction technology to transfer energy and control signal through skin;
HLA antibodies	Platelets transfusion; Interaction of foreign materials with the immunological components of blood;	Treatment with immunglobulin and cyclophosphamide before transplantation; Treatment with urgent plasmapheresis in an acute setting;
Device failure	Malfunction of electrical power line and signal line; Dysfunction of biologic valve; Dislodgement of device cannulas;	New device structure, new technology and new material to improve the overall reliability of the device;
Poor mobility	Short duration of battery as the power source; High power needed by the device; Heavy and complex monitor devices;	Improvement of battery technology; Better design or new technology to improve the efficiency of the device; Improve of sensor technology; Better and simpler control strategy for the device;

2.4 Review of magnetic bearing technology

As many blood pumps have to rely on some form of bearings, a brief review is given here. Conventional bearings provide support to rotating machinery by allowing relative movement in a plane of rotation. They allow rotation and provide support in either radial or axial planes of rotation. The most common types of bearings are rolling element bearings, and fluid film bearings. Conventional bearings have been used for hundreds of years. It has evolved into a well-engineered product. However, its frictional loss and wear make it not so suitable for very high-speed operations (such as $>10000\text{rpm}$) or when mechanical contacts are to be avoided as in the case of blood pumps.

Magnetic bearing systems represent a completely different approach to the support of rotating equipment. This is a non-contacting technology, which means negligible frictional loss and wear, and higher reliability. It also enables previously unachievable surface speeds to be attained. Lubrication is eliminated, which means that these bearings can be incorporated into processes that are sensitive to contamination, such as the vacuum chambers or biomedical devices. Generally, there are two kinds of magnetic bearing: active magnetic bearing (AMB) and passive magnetic bearing (PMB).

2.4.1 Active magnetic bearing

Fig 2-15 shows the schematic of the working principle of a general AMB. The application of AMB is based on the principle that an electromagnet will attract ferromagnetic material. AMB provides attractive electromagnetic suspension by application of electric current to ferromagnetic materials used in both the stationary and rotating parts (the stator and rotor, respectively) of the magnetic bearing actuator. This creates a flux path that includes both parts, and the air gap separating them, through which non-contact operation is made possible [100]. As the air gap between these two parts changes, the attractive forces vary. Therefore these electromagnets are inherently unstable. A control system is thus needed to regulate the current and provide stability of the forces, and therefore, position of the rotor.

Fig 2- 15: Electromagnetic suspension principle of AMB

The control process relies on measurement of the rotor position with a position sensor. The signal from this device is received by the control electronics, which compares it to the reference position. Any difference between these two signals results in calculation of the force necessary to pull the rotor back to the desired position. This is translated into a command to the power amplifier connected to the magnet actuator. The current is changed accordingly, causing a variation both in magnetic flux and in the forces between the rotating and stationary components, and finally movements of the rotor toward the stator along the axis of control.

2.4.2 Passive magnetic bearing

While, complete suspension with AMB can be truly mechanically frictionless; however, the power requirement for active magnetic suspension can be substantial. The electromagnets used for AMB can add significant volume and weight, making these pumps bulky and heavy. In addition, active suspension requires position sensors and servo circuits, which increase the system complexity and may compromise reliability. PMB do not require this hardware; thereby they have the potential to increase system efficiency and reliability. Another advantage of PMB is that larger rotor-stator gaps may be possible since the current for active magnetic bearings is proportional to the square of the rotor-stator gap [102]. Fig 2-16 shows the sectional view of an industrial PMB assembly.

Fig 2- 16: Sectional view of a PMB assembly.

The disadvantages of PMB are that they typically have lower stiffness and lower damping than similarly sized active magnetic bearings. The stiffness and damping are not adjustable “on-the-fly”, therefore active vibration control cannot be used with PMB. PMB systems must have sufficient stiffness and damping built-in to allow them to perform over their entire operating range [101]. According to Earnshaw’s Theorem, it is impossible to have a completely three dimensionally stable PMB using only the forces of static fields of permanent magnets. So PMB may have to be used together with AMB,

2.4.3 Applications in blood pump

In general, biomedical devices in contact with living tissue must be biocompatible and sterile. These devices should also be non-traumatic, highly reliable, robust and durable. Magnetic levitation technology offers several features to address these issues. By contact-free actuation of moving parts through the walls of a housing, hermetically sealed systems become feasible. The absence of seals further limits the potential entry of microbial organisms into the system. Wide gaps in such pumps also help to reduce blood-trauma (hemolysis). Friction-less operation renders such devices durable and robust, while permitting a wider selection of biocompatible materials. For instance, the blood contacting surfaces of the magnetically levitated blood pumps can be made of textured titanium. For rotary blood pump devices, the magnetic bearing technology has a lot of potential.

CHAPTER TWO: LITERATURE REVIEW

However, although much effort has been put into the research of magnetic bearing technology for the blood pump impeller, there are also reservations as the use of magnetic suspension creates its own problems. Additional energy is needed to power the electromagnets, making these devices less energy efficient. At the same time, extra sensor and electronics are needed for the control of magnetic levitation. In addition, there are stabilization issues that cause magnetically suspended impellers to wobble, creating regions of hemolysis or stagnation. Thus, the benefits of magnetic suspension versus contact bearing systems remain uncertain.

2.5 Review of electromagnetic propulsion technology

From the review of CBP in Section 2.3, it is clear that many different designs of CBP are now based on PM propulsion. However, there has been no report of CBP based on inductive propulsion. In industrial motor drives, the advantages and disadvantages of PM motors and induction motors (IM) are well-known. It is however not so clear if inductive propulsion without mechanical coupling can be successfully implemented for CBP. Hence this is the motivation behind the research described in this thesis. A brief review of these two types of electromagnetic propulsion technology is given here.

2.5.1 Comparison between permanent magnetic and inductive propulsion

Most torque drivers used in many proposed centrifugal blood pumps are based on permanent magnetic direct current (PMDC) type of motors. The basic theory of PMDC motor is based on Lorentz force which can be briefly explained as follow. If an electron is moving through an electric field, it experiences a force that is perpendicular to both the magnetic field and the direction it is moving. Furthermore, if there is a current passing through a wire in a magnetic field, the wire experiences a force whose direction can be easily determined using the right hand rule. If this single wire is replaced with a loop of wire between the magnetic poles, this looks like two wires with current flowing in opposite directions. Then, the forces on the wires cause the loop to rotate. This basic theory of PMDC motor is shown in Fig 2-17.

Fig 2- 17: Basic theory of PMDC motor

PMDC motors have high power factor and high efficiency. They have relatively simple control method. While PM motors are excellent choice of torque drivers, there many challenges in applying them to blood pump applications. The most crucial one is that

the control of PMDC motor needs extra commutation circuit and position sensors, which will greatly reduce the reliability of the system. The commutation may be accomplished mechanically, or more commonly, electronically.

There are other possible alternative drivers such as the induction type of motors. IMs are very common in the industry. In the radial flux squirrel-cage IMs, which constitute a vast majority of IMs, the rotor is inaccessible. No moving contacts, such as commutator and brushes in PMDC machines or slip rings and brushes in AC synchronous motors, are needed. Rotor position sensors are also not normally required. This arrangement greatly increases reliability of induction motors. An additional degree of ruggedness is provided by the rotor winding being consisted of metal bars forming the “squirrel-cage” that gives the name to the motor. Such a robust rotor can run at high speeds and withstand heavy mechanical and electrical overloads. At the same time, it is possible to apply the simple but reliable open-loop with slip compensation control strategy based on Voltage/Frequency method to control the speed of the IM. Axial flux version of the squirrel-cage IM has a spoked wheel structure rotor, but has same operation and control principles as its radial flux counterpart. Both radial and axial flux IM principles have yet to be applied to designs of CBP.

2.5.2 Basic theory of induction motor [102, 103]

Fig 2- 18: Two-pole stator of induction motor

Induction motors employ a simple but ingenious scheme of electromechanical energy conversion. The three-phase stator winding produces a revolving magnetic field, which

constitutes an important property of induction motors. Generation of the revolving magnetic field by the stationary phase windings of the stator is explained in Fig 2-18. A simplified arrangement of the windings, each consisting of a one-loop single-wire coil, is depicted in Fig 2-18. The coils are displaced in space by 120 degrees from each other. Fig 2-19 shows the waveforms of currents i_{as} , i_{bs} and i_{cs} in individual phase windings. The stator currents are given by

$$i_{as} = I_{sm} \cos \omega t \quad (2.1)$$

$$i_{bs} = I_{sm} \cos(\omega t - \frac{2\pi}{3}) \quad (2.2)$$

$$i_{cs} = I_{sm} \cos(\omega t - \frac{4\pi}{3}) \quad (2.3)$$

Fig 2- 19: Waveforms of stator currents

where I_{sm} denotes the peak value of currents and ω is the supply radian frequency. They are mutually displaced in phase by the same 120 degrees. The phasor diagrams of stator current, at the instant of $t=0$ and $t=T/6$, are shown in Fig 2-20 with the corresponding distribution of currents in the stator winding. T denotes the period of stator voltage, that is, a reciprocal of the supply frequency, f . Also shown are vectors of the magnetomotive forces (MMFs), F_{sa} , F_{sb} and F_{sc} , produced by the phase currents. These, added together, yield the vector, F_s , of the total MMF of the stator, whose magnitude is 1.5 times greater than that of the maximum value of phase MMFs. The two half-circular loops represent the pattern of the resultant magnetic field, that is, lines of the magnetic flux, ϕ_s , of stator.

Fig 2- 20: Phasor diagram of stator currents and the resultant magnetic field

The stator in Fig 2-20 is called a two-pole stator because the magnetic field, which is generated by the total MMF and which closes through the iron of the stator and rotor, acquires the same shape as that produced by two revolving physical magnetic poles. In general, the synchronous speed of the rotating magnetic field in revolutions per minute (rpm) is

$$n_{syn} = 60f / p \quad (2.4)$$

where p is the number of pole pairs. According to the well-known right-hand rule, this rotating magnetic field induces an electromotive force. This electromotive force will produce a current in the conductor of the rotor. The interaction of the current and the rotating magnetic field results in an electrodynamic force on the conductor. The left-hand rule determines direction of the force. The force acts on the conduction in the same direction as that of the field motion. In other words, the stator field pulls conductors of rotor, which move with a lower speed than that the rotating magnetic field. The developed torque is a product of the rotor radius and sum of electrodynamic

forces generated in individual rotor conductors. Clearly, according to Faraday’s Law, no electromotive force would be induced if the speed of the conductor of rotor and speed of the field were equal. The difference of these velocities, called a slip velocity is

$$\omega_s = \omega_{syn} - \omega_r \tag{2. 5}$$

where ω_{syn} is the field velocity and ω_r is the rotor speed. Dividing the slip velocity by ω_{syn} yields the so-called slip, s , of the motor, defined as

$$s = \omega_s / \omega_{syn} = 1 - \omega_r / \omega_{syn} \tag{2. 6}$$

2.5.3 Steady-state equivalent circuit model of induction motor [102, 103]

Fig 2- 21: Equivalent circuit model of induction motor

Unlike critical servo applications which need complex transient model for their control, the devices proposed in this project usually operate in the steady state and the equivalent circuit model shown in Fig 2-21 is therefore sufficient for its operation control, which can eliminate the unnecessary complexity of transient model. All characteristic algebraic equations for the steady-state power, torque, and efficiency of the induction motor can be derived from the per-phase equivalent circuit model of Fig 2-21. The volt/hertz method of speed control can also be derived from the circuit model.

2.6 Concluding remarks

In order to set up clear and practical objectives and to avoid duplicated effort or to embark on something that is not possible or has little practical value, a thorough literature review of major important background information relevant to this project was carried out and summarized in this chapter. This literature review covered the development of artificial heart in general, the development of centrifugal blood pumps, the magnetic bearing technology and the principle of electromagnetic propulsion.

From the review of the history, classification and comparison of artificial heart systems, important facts emerged regarding the two major types of artificial heart devices: the pulsatile type and the continuous-flow type. Compared with the pulsatile type, the continuous-flow blood pumps have more advantages in terms of size, power consumption, reliability and complexity, despite the early concern about the absence of pulsatile flow. So most current artificial heart research work focuses on the two major types of continuous-flow blood pumps: the centrifugal type and the axial flow type. The larger contact surfaces as well as the potential for more flow rate with lower rotation speed are the advantages of centrifugal pumps over axial flow type, particularly as bridges to transplant and as total heart replacement devices. So, medical experts consider the centrifugal blood pumps to be more viable for longer-term applications.

The review of the development of the centrifugal type blood pumps has shown that there have been three major stages of evolution, with the latest developments aiming to achieve high durability through magnetic and/or hydrodynamic bearings, high reliability through minimum of sensors, moving parts and simple controls, and compactness through integration of rotor and impeller. So far, most of these efforts have been based on applications of permanent magnets. Electromagnetic induction principle has not been considered. There are advantages in utilizing the simple structure of induction motor, centrifugal pump principles, and magnetic bearing technology to produce compact and reliable integrated rotor-impeller designs for shaft-less pump applications. Of course there would also be disadvantages in such approach, which have to be studied carefully in this project to arrive at a fair assessment of the feasibility and viability of inductively-driven centrifugal blood pumps.

Chapter Three

RADIAL-MAGNETIC-FLUX INDUCTIVE PROPULSION

3.1 Introduction

This chapter presents the investigation of a novel concept of CBP based on radial-flux induction motor. The proposed device can be classified as a third-generation CBP, which integrates the motor design with the impeller so that there are no rotating parts outside the blood chamber. In other words, the rotating part of the motor, the rotor, becomes part of the impeller, leaving only the stator of the motor outside the chamber. The electromagnetic torque produced by the blood pump driver can therefore act on the pump impeller directly. This rotor-impeller structure leads to a more compact and simple CBP design. Furthermore, this novel design uses a combination of PMB and AMB to support and constrain the rotor-impeller, thereby eliminating the blood stagnation and sealing problems caused by conventional mechanical shaft and bearing.

This chapter firstly presents the overall concept of the proposed blood pump system. Although the detailed design of the CBP chamber is out of the scope of this project, a sketch design was undertaken to decide the main dimensions of the chamber. These main dimensions then provided the necessary specifications for the subsequent detailed design of the inductive blood pump driver and the magnetic bearing system.

Then, the mathematical modeling of the designed blood pump device and the magnetic bearings are presented respectively. The mathematical model of the radial-flux inductive pump driver is introduced and its major parameters are defined. Furthermore, how these important parameters may be determined are also explained. The equations to calculate the electromagnetic torque acting on the rotor are derived. The proposed magnetic bearing system is then analyzed mathematically, using equivalent magnetic charges method and magnetic circuit method. Lastly, the proposed radial-flux inductive blood pump design and its magnetic bearing system are then analyzed using Finite Element Method (FEM) simulations to assess their viability.

3.2 Overall concept and structure

In conventional pump applications, the entire motor consisting of stator and rotor are located outside the pump chamber. A mechanical shaft is needed to transmit the torque from the rotor, through the pump wall, to the impeller inside the pump chamber. The shaft passes through the pump wall and sealing is required. In the proposed concept here, the induction motor stator is outside the pump chamber while the induction motor rotor is inside the pump chamber. The rotor is mechanically integrated with the pump impeller. Thus there is no mechanical shaft passing through the pump chamber wall, and therefore no need for blood chamber sealing.

Fig 3- 1: Functional diagram of radial-flux inductive CBP system

The functional diagram of entire proposed blood pump system is shown in Fig 3-1. It has two main parts: (1) the blood pump chamber; and (2) the blood pump driver. Through the blood pump chamber wall, the magnetic flux produced by the driver (the stator of an induction motor) magnetically couples with the rotor which is integrated with an impeller. A combination of PMB and AMB is adopted to support and constrain the impeller against the undesired degrees of motion freedom, under the active control of the AMB controller. The overall operation of the blood pump driver is controlled by its controller, which is

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made up of power supply, power converter and control electronics. The output speed and torque of the blood pump driver must be controlled to meet the requirement of blood pump output head and flow rate. In other words, the magnitude and frequency of input three-phase voltage of the stator windings must be controlled to enable the squirrel-cage rotor-impeller to produce the necessary electromagnetic torque and therefore the rotational speed of the impeller.

According to this proposed general concept, a novel radial-flux type of CBP is designed. The 3-D geometries of the design are shown in Fig 3-2. In order to show the structure clearly, the magnetic bearings, position sensors and rotor core are not included in this figure.

Fig 3- 2: Three-dimensional view of CBP design

The impeller shroud is elongated to accommodate a cylindrical iron rotor core embedded with a copper or aluminum squirrel-cage. The outer cylindrical surface of the stator is surrounded by the rotor and is separated from the latter by part of the pump wall, as shown in Fig 3-2. Outside this pump wall is the dry compartment containing the motor stator core and windings, and the active magnetic bearing winding. Within the pump wall is the wet compartment where the fluid (blood) is centrifugally pumped by the impeller

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attached to the motor rotor. This form of motor structure is known as the inner-stator/outer-rotor type. The blood inlet is at the centre and the fluid leaves the chamber at high pressure through the outlet which is tangential to the circular pump wall.

Fig 3-3 shows the 2-D cross-sectional structure of the proposed CBP, which includes the proposed magnetic bearings and its position sensors. Its components are listed in Table 3-1.

Fig 3- 3: 2D cross-sectional structure of CBP

Table 3- 1: Components of proposed CBP

1	Pump inlet	8	Stator PMs of AMB
2	Pump outlet	9	Stator of IM
3	Pump impeller	10	Windings of IM
4	Rotor of AMB	11	Stator of PMB
5	Stator of AMB	12	Rotor of PMB
6	Rotor PMs of AMB	13	Squirrel-cage rotor of IM
7	Coil of AMB	14	Position sensors

Items #1, #2 and #3 are the pump inlet, outlet and impeller respectively. The impeller is physically attached to the squirrel-cage IM rotor, which is marked as item #13. Items #9

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and #10 are the stator core and windings of the IM respectively, which are located outside the pump wall. The PMB consists of two permanent magnetic rings: item #12 which is fixed to the impeller and therefore rotates with it, and item #11, which is fixed to the exterior of the pump wall and therefore is stationary. Items #4 and #5 are the movable and stationary cores of the AMB respectively. Items #6 and #8 are the biasing permanent magnets of the AMB, while item #7 is the winding of the AMB.

3.3 Pump chamber design

Pump chamber and impeller structures have significant impact on the performance of centrifugal pump. So, sufficient attention should be paid to their designs. The design of hydraulic machinery in general and of centrifugal pump in particular has been, and still is, essentially empirical. In spite of extensive theoretical knowledge of fluid dynamics and mechanical design, the design process for centrifugal pump still relies heavily upon experimental data [111]. This is a very complex design process. One of the reasons is the great variety of types, sizes and services demanded from them which, in turn, require a great multiplicity of parameters to define the design problem in hand. Two separate groups of parameters can be distinguished: the first describing the hydraulic (and thermodynamic) factors related to fluid motion, and the second representing the physical size and shape of the units' components. Both groups are three dimensional which, of course, contributes to the complexity of the design process. At the same time, due to the close interactions that take place between the geometry of the passageways and the hydraulic (and thermodynamic) factors, the design process become iterative, thus adding further to the complexity of the design task. Moreover, the sizing of the unit, the shaping of its components, as well as the quality of the execution, has strong impact, not only on the overall performance, but also on the cost of the device. Therefore, the practical aspects reflecting the cost, service reliability, and so on, represent additional factors which the designer cannot ignore. Because of these all, the detailed CBP design process, which is out of the scope of this project, cannot be conducted here; only a sketch design was done to decide the main dimensions of the blood pump chamber. The obtained main dimensions will provide an indispensable precondition for further design process of blood pump driver and magnetic bearings.

At first, the variables used in the design procedure will be defined. Based on the typical outputs of human heart, the target outputs of the proposed CBP are analyzed as the following: the typical cardiac output in the resting adult human is about $5.6L/min$. However, during vigorous exercise, the body may demand 30 or more L/min . The natural heart meets this demand both by beating faster and by increasing its stroke volume. Hence, a replacement blood pump must be capable of variable flow upon demand. The heart,

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actually the left ventricle, has to push blood into the aorta where the average pressure is over 100mmHg above atmospheric pressure. The pressure of the blood that returns to the right atrium is much lower, about 10mmHg above atmospheric pressure. So, the typical head to be produced by the replacement blood pump is about 110mmHg, which is an average value of outgoing blood and is set a little bit high value to ensure the adequate head pressure produced. The typical rotation speed of the pump impeller is prescribed by the selected blood pump driver, which is 2000 to 3000 rpm here. The safe speed of blood pump is directly related with the shape, configuration, dimensions and material of pump impeller and pump chamber. However, compared to the more than 10 thousand rpm rotation speed of axial flow blood pump, this 2000 to 3000 rpm rotation speed is much slower, which means much less damage to the blood cell. Then, the input variables of pump design are summarized in Table 3-2.

Table 3- 2: Pump design input variables

Variable	Definition	Value
Q	pump flow rate	5.6 L/min
H	pump head pressure	110 mmHg
N	impeller rotation speed	2500 rpm

The rest of variables used in the pump design procedure are listed in Table 3-3.

Table 3- 3: Pump design variables

Variable	Definition	Variable	Definition
D_1	impeller inlet diameter	b_1	chamber inlet diameter
D_2	impeller diameter	b_2	outlet diameter
D	pump chamber diameter	h	blood pump chamber height
t	chamber wall thickness	N_s	pump specific speed
U_2	tip speed	ψ	head coefficient
η	pump efficiency		

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After defining all the related variables, the main dimensions of the blood pump chamber, including inlet diameter, outlet diameter, chamber diameter, height and thickness, can be decided by the following process [111, 112].

It is well known that specific speed is a reference number that describes the hydraulic features of a centrifugal pump. Pump specific speed (N_s) as it is applied to centrifugal pump is defined in U.S. units as:

$$N_s = \frac{NQ^{1/2}}{H^{3/4}} \quad (3.1)$$

where N (rpm) is the rotation speed of pump impeller, Q (gpm) is the flow rate and H (ft) is the head. By converting the metric units to U.S. units and substituting the values, the specific speed of this proposed centrifugal pump is calculated as 1032.71m/s.

The impeller inlet diameter D_1 is usually selected to minimize the inlet relative velocity. Low velocity favors low loss. As a start, the optimum inlet diameter should be calculated from

$$D_1 = 1.533 \left(\frac{Q}{\omega} \right)^{1/3} \quad (3.2)$$

as 1.1cm. Then the chamber inlet diameter b_1 is chosen as 1.5cm.

According to several empirical correlations recommended in the centrifugal design literature [111, 112, 113] and the specific values in this design, the head coefficient ψ is chosen as 0.75, pump efficiency η is chosen as 25%. Then the tip speed U_2 is calculated from

$$U_2 = \sqrt{\frac{gH}{\eta\psi}} \quad (3.3)$$

as 8.85m/s.

Then the impeller diameter D_2 is calculated from

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$$D_2 = \frac{60U_2}{\pi N} \quad (3.4)$$

as 6cm . The radial gap between the impeller and the casing is about 10% of the impeller diameter. Then the pump chamber diameter D is chosen as 7cm .

The outlet diameter b_2 is calculated from

$$b_2 \approx 2\sqrt{\frac{2Q}{\omega D\pi}} \quad (3.5)$$

as 0.4cm . Because the velocity of the fluid at the outlet is smaller than the rotation speed of the impeller, the outlet diameter should be adjusted a little bigger to provide enough area. So, the diameter of blood pump chamber outlet is chosen as 1cm .

The height of designed blood pump chamber h is closely related with the structure and dimensions of the pump impeller, the design of which is another complex process out of the scope of this project. Since this dimension does not affect the blood pump driver part of the design, it is chosen as 3cm in this initial design stage. The thickness of chamber wall t is roughly selected to be 1-2mm, where the actual value should be decided according to the different material used in final design.

As mentioned in the beginning of this section, above presented design procedure is just a first trial design of the blood pump chamber to obtain the rough mechanical dimensions of pump chamber for further propulsion and magnetic bearing designs. There is much scope for further optimization for this blood pump chamber design from the fluid dynamic aspect, such as the operating range of Reynolds number, the degree of laminar or turbulence flow inside the pump, the considerations of recirculation, stagnation and high stress regions under the designed range of flow rate, the levels of viscosity and the shear stress within the pump for assessment of blood cell damage. These will be an important topic for future work.

According to the above analysis and calculations, the main dimensions of the target blood pump chamber can be summarized in Table 3-4 and Fig 3-4.

CHAPTER THREE: RADIAL-MAGNETIC-FLUX INDUCTIVE PROPULSION

Table 3- 4: Designed dimensions of blood pump chamber

Variable	Designed value	Variable	Designed value
b_1	1.5 cm	b_2	1 cm
D	7 cm	h	3 cm
t	1-2 mm

Fig 3- 4: Main dimensions of the blood pump chamber

3.4 Inductive torque driver design

In order to produce enough output from the blood pump to meet the requirement of human body, the blood pump impeller must have the necessary rotational speed and torque, all of which have to be provided by the blood pump driver. After getting the main dimensions of the blood pump chamber, the design of the driver can be carried out. In general, there are two structures in electric motor topologies, radial-flux and axial-flux. In this chapter, the conventional radial-flux structure is presented. As introduced in Chapter Two, the three-phase induction motor could be a viable candidate for the blood pump driver. In biomedical applications, devices are normally powered from low voltage dc power supplies for safety reasons. Since an inverter is already required to drive the brushless ac pump motor, it is definitely preferable to adopt a three-phase design rather a single-phase design. Even though the power requirement is small, a three-phase design provides better torque performance and controllability. All existing PM-driven CBPs are also based on three-phase stator designs. The performance and reliability of single-phase machines are not adequate for high-performance applications such as blood pumps and data disk drives.

A three-phase IM consists essentially of two major parts, the stator and the rotor. The construction of each is basically a laminated core provided with slots which house the windings (for squirrel-cage induction motor, the rotor is in the form of shorting rings and rotor bars within the core). When the stator windings are excited by a three-phase AC power source, a rotating magnetic field is set up in the air gap between rotor and stator. This field produces EMF in the rotor windings by inductive action which in turn produces currents in the rotor windings if they are short-circuited. The currents flowing in the rotor windings interact with the air-gap flux to generate a torque which is responsible for the rotation of rotor. A major difference between the inductive pump driver proposed here and the conventional IM is that now the rotor is separated from the stator by a pump chamber wall which effectively implies an unconventionally large air-gap.

3.4.1 Design specifications

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From the relative position of rotor and stator point of view, the radial-flux type of motor or also known as cylindrical motor can be divided into two types: inner rotor (with outer stator) and outer rotor (with inner stator). They are similar in working principle and design process. In this blood pump driver application, the outer rotor type of structure with concentrated windings is chosen to produce a compact design which would be compatible with the proposed magnetic bearing system.

For the rotor windings, they can be of the squirrel-cage type or the wound-rotor type. For small motors, the squirrel-cage rotor is the standard configuration because of its many advantages over the wound-rotor type. Hence a squirrel-cage outer rotor would be adopted in the CBP design. The top view of this proposed radial-flux outer-rotor squirrel-cage inductively-driven driver is shown in Fig 3-5.

Fig 3- 5: Top view of the proposed radial-flux outer-rotor squirrel-cage induction motor driver (pump wall between stator and rotor is not shown)

To provide adequate propulsion to achieve enough cardiac output rate and pressure, the CBP driver must satisfy the requirements of output torque and speed. So based on all these requirements and empirical induction motor data [114, 115], the nominal values for the initial motor structure design are selected as following:

- (1) Rated input power: $P_n = 10W$
- (2) Rated input line voltage: $V = 10V$

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- (3) Rated power frequency: $f = 50\text{Hz}$
- (4) Number of phases: $m = 3$
- (5) Power factor at rated load: $\cos \varphi = 0.65$
- (6) Expected efficiency: $\eta = 75\% - 80\%$
- (7) Rated rotation speed: $n = 2000 - 3000\text{rpm}$

The pole pair number p can be calculated from the rated rotation speed n and rated power frequency f .

$$(8) \quad n = \frac{60f}{p} \Rightarrow p = \frac{60f}{n} = \frac{3000}{3000} = 1$$

3.4.2 Stator of the radial-flux motor driver

The cylindrical stator is designed to accommodate concentrated windings, so as to minimize the copper winding overhang volume. Radial-flux motors with axial length shorter than its diameter would have relatively poorer volume utilization if other forms of winding configurations are used. Radial-flux motor cores are usually composed from sheets of silicon steel laminations to reduce iron losses. For the proposed design with short axial length, the number of laminations would be small as it is not easy to source for laminations thinner than the minimum standard thickness. It may also be possible to construct the stator core from ferrite materials. Anyway for the present investigations at a conceptual stage, the consideration of core losses is not accorded a high priority.

The overall dimensions of an IM can be estimated using the output coefficient, D^2L , approach [103, 114, 115]. Applying this approach, the input power of this CBP driver is first considered via

$$P_n = \sqrt{3}VI \cos \varphi \tag{3. 6}$$

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where I is the input phase current value. In order to make use of this formula in determining the dimensions of the motor, certain substitutions need to be made. The quantity V may be expressed as

$$V = 4\sqrt{3} f K_w N \phi \quad (3.7)$$

where K_w [103] is the winding factor, which is 0.4–0.6, N is the winding number per phase and ϕ is the flux per pole, which can be further expressed as

$$\phi = B_g \frac{\pi D L}{2p} \quad (3.8)$$

where B_g is the average flux density in air gap, which lies in the range 0.3–0.6 Wb/m², D is outer diameter of the stator and L is the axial length of induction motor.

Therefore,

$$V = 2\sqrt{3} f K_w N B_g \frac{\pi D L}{p} \quad (3.9)$$

Another important parameter common in electrical machine design is the specific loading q , which is defined as ampere-conductors per meter of air-gap periphery, which varies between 120 A/cm to 500 A/cm.

$$q = \frac{3NI}{\pi D} \quad (3.10)$$

Then, the input phase current I can be expressed as

$$I = \frac{q\pi D}{3N} \quad (3.11)$$

By substituting equation (3.9) and equation (3.11) into equation (3.6), the relation between motor dimensions and input power can be derived as

$$D^2 L = \frac{P_n \cdot p}{2\pi^2 f K_w B_g q \cos \varphi} \quad (3.12)$$

It is not possible to lay down hard-and-fast rules regarding the relation of the diameter of the stator to its axial length. Considering the geometry of designed blood pump chamber,

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the diameter of this motor driver must not differ too much from that of the blood pump chamber. Otherwise, it would not be possible to integrate them to produce a compact design. Furthermore, the rotor of this motor driver is supposed to be integrated with the pump impeller, all of which should be located within the blood pump chamber. Considering the outer rotor structure of this motor driver, the diameter of stator must be smaller than the blood pump chamber to leave enough space for the rotor to be accommodated inside the blood pump chamber. Moreover, the axial length of the target motor driver has to be sufficient to provide enough electromagnetic torque. Taking all these considerations into account, the main dimensions of proposed radial-flux induction motor driver is selected as: $D = 50mm$ and $L = 12mm$.

Then, the dimensions of the stator tooth can be decided. The stator tooth height can be expressed as

$$h_s = \frac{A_l}{\frac{B_g}{B_{ts}} j_{con} K_{fill}} \quad (3.13)$$

where A_l is the stator current load in the range $150A/cm \leq A_l \leq 500A/cm$. B_g is the magnetic flux density in air gap and B_{ts} is the tooth flux density. Due to magnetic saturation, the nominal value of the latter is chosen as $B_{ts} = 1.2T$. j_{con} is the conductor current density in the range of $3.5 \times 10^6 A/m^2 \leq j_{con} \leq 8 \times 10^6 A/m^2$. K_{fill} is the slot fill factor: $0.4 \leq K_{fill} \leq 0.6$. Then the stator slot length is calculated and selected as $h_s = 15mm$.

After determining the main dimensions of stator, the stator winding and stator slot dimensions can be designed in the following process:

The stator winding number can be obtained from equation (3.9):

$$N = \frac{V / \sqrt{3}}{4 f K_f K_w \phi} \quad (3.14)$$

where N is the number of turns per phase, V is input line voltage, f is frequency of power supply, K_f is saturation factor, $1.11 \geq K_f \geq 1.02$, K_w is winding factor, ϕ is pole

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flux, $\phi = \frac{\pi D}{2 \rho m} L B_g$. From this calculation above, the stator winding number per phase is selected as $N = 150$. According to the current density, the stator coil diameter $d_l = 0.45\text{mm}$ is chosen from standard wire gauges and the diameter inclusive of insulation is 0.4835mm .

Fig 3- 6: Dimensions for stator geometry including tooth and slot

The dimensions of the stator and stator slots can be summarized in Table 3-5, all of which are also depicted in Fig 3-6.

Table 3- 5: Dimensions of stator and stator slot

Item	Value	Item	Value
Stator outer diameter (D_1)	50mm	Stator slot top height (h_1)	3.8mm
Stator inner diameter (D_2)	20mm	Stator slot height (h_2)	11.2mm
Stator tooth height (h_s)	15mm	Stator slot top width (w_t)	6mm
Stator tooth width (w)	8mm	Stator axial length (L)	12mm

Then the stator winding resistance per phase can be calculated as:

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$$R_1 = \rho_{25^\circ} \frac{W_1 2(w + L)}{S} \quad (3.15)$$

where ρ_{25° is the resistivity of copper under 25°C and S is the cross section area of stator coil wire. The calculation result is $R_1 = 0.5\Omega$.

3.4.3 Air gap of the radial-flux motor driver

The air gap length of induction motor is another important issue which affects its overall performance significantly. The MMF required to maintain a particular air gap flux density is proportional to the product of the flux density and the length of the air gap. Even with very small densities, the MMF required for air gap is much more than that for the rest of the magnetic circuit. Therefore, it is the length of air gap that primarily determines the magnetizing current drawn by the motor. Bigger air gap length means bigger magnetizing current drawn in the stator coil. This causes bigger phase angle difference between stator voltage and stator current, which produce a smaller power factor. From above, it is clear that the length of air gap in an induction motor should be as small as mechanically possible in order to keep the magnetizing current small and to improve the power factor. This is a major consideration. However in this proposal, the rotor of inductive driver is to be integrated with the pump impeller inside the blood pump chamber to eliminate the need of mechanical shaft and sealing. As the stator is outside the blood pump chamber, this means that the air gap must be big enough to accommodate the blood pump wall thickness, 1-2mm for this blood pump chamber. So, the air gap length is chosen as $g = 3\text{mm}$.

Although, this big air gap motor driver may have disadvantages such as low power factor and high magnetizing current, there may be other mitigating features such as, less pulsation loss, less eccentricity impact on performance, better cooling and smaller noise caused by zigzag leakage flux [114].

3.4.4 Rotor of the radial-flux motor driver

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The rotor is of the squirrel-cage type, which is composed of several rotor bars and two end rings, as shown in Fig 3-2. The rotor bars are short-circuited by the end rings at the both ends.

As a first step, the number of rotor bars, which is equal to rotor slots, should be determined. This is important because with certain combination of numbers of poles and slots of the stator and rotor, poor performance may result. For example, poor starting torque, sub-synchronous operation or torque pulsations may occur for certain inappropriate combination. In order to avoid these problems, the number of rotor slots may be selected based on the following rules:

- (1) For avoiding cogging and crawling: Difference between number of stator and rotor slots should not be equal to $6p$, where p is the number of pole pairs. In this case, $S_s = 6$ and $p = 1$. So, $|S_s - S_r| \neq 6p = 6 \Rightarrow S_r \neq 12$;
- (2) For avoiding “synchronous cusps” in the torque-speed characteristics: Difference between stator and rotor slots should not be equal to $2p$, $4p$ and $10p$. In this case, $|S_s - S_r| \neq 2p, 4p, 10p = 2, 4, 10 \Rightarrow S_r \neq 2, 4, 8, 10, 16$;
- (3) For avoiding noisy operation: Difference between the stator and rotor slots should not be equal to 1, 2, $(2p+1)$ or $(2p+2)$. So, in this case $|S_s - S_r| \neq 1, 2, (2p+1), (2p+2) \Rightarrow S_r \neq 2, 3, 4, 5, 7, 8, 9, 10$;
- (4) The number of rotor slots should not be equal to stator slots, otherwise motor may not even be able to start because of insufficient starting torque. So, $S_r \neq S_s = 6$.

From the analysis above, $S_r \neq 2, 3, 4, 5, 6, 7, 8, 9, 10, 12, 16$. Also, the rotor slot pitch should not be too small to accommodate the rotor bar and the rotor bar number should be as small as possible to simplify the fabrication. So, the number of rotor bars is chosen as $S_r = 11$.

The performance of an induction motor is greatly influenced by the resistance of the rotor cage. A motor designed with high rotor resistance has the advantage that it has high starting torque. However, a rotor with substantial resistance has the disadvantage of higher copper loss and therefore lower efficiency under running condition. It has to have a

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compromise between a high resistance rotor which gives a good starting torque and a low resistance rotor which gives a high value of efficiency under running conditions.

Depending on the input power, input line voltage and acceptable power factor, the stator input current I_1 can be calculated as $I_1 = \frac{P_n}{3 \cos \phi V / \sqrt{3}} = 0.89 A$. Approximately, equivalent rotor current is $I_2' = 0.85 I_1 = 0.76 A$. Then, the current per rotor bar can be calculated as

$$I_2 = \frac{K_{w_1} \times S_s \times Z_1}{K_{w_2} \times S_r \times Z_2} I_2' \quad (3.16)$$

where $K_{w_1} = 0.6$ (assumed) is the winding factor for the stator winding, $K_{w_2} = 1$ is the winding factor for the rotor winding, $S_s = 6$ is the number of stator slots, $S_r = 11$ is the number of rotor slots, $Z_1 = 75$ is the number of conductors per stator slot and $Z_2 = 1$ is number of conductors per rotor slot.

Substituting these values, current per rotor bar is obtained as $I_2 = 18.65 A$. Thus the cross-sectional area of the rotor bar, $a_2 = \frac{I_2}{\delta_2}$ where δ_2 , the current density of the rotor bar, may be taken as between 4 to 7 A/mm^2 . So, the cross-sectional area of the rotor bar is chosen as $a_2 = 4.66 mm^2$. Accordingly, diameter of rotor bar can be gotten as $D_b = 2.5 mm$. The height of rotor bar h_b , is chosen the same as the axial length of stator, which is 15mm.

End ring current can be determined as, $I_e = W I_2 / \pi$, where W is the number of rotor bars per pole, which is $W = S_r / p = 5.5$. So, I_e can be calculated as 32.67A. Thus the cross-section area of rotor end rings, $a_e = I_e / \delta_e$, is designed as $8.2 mm^2$. In order to provide enough space for the end ring to accommodate the rotor bars, the width and height of the end ring are chosen as 5mm and 2mm, respectively.

The selected dimensions of the rotor bar and end ring are summarized in Table 3-6, and are illustrated in Fig 3-7.

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The resistance of each rotor bar is given by $R_{bar} = \rho h_b / S_{bar} = \rho h_b / \pi (D_b / 2)^2$. According to the rotor bar dimensions listed in Table 3-3, $R_{bar} = 0.515 \times 10^{-4} \Omega$. Mean length of end ring is $C_1 = \pi (d_1 + d_2) / 2 \approx 0.174m$. Then, the resistance of end ring can be calculated as $R_{end_ring} = \rho C_1 / (h_e \times W_e) = 2.9 \times 10^{-4} \Omega$. This resistance can be converted to an equivalent bar resistance by $R'_{end_ring} = (R_{end_ring} / G) / 4 \sin^2(\pi p / G)$ where G is the number of rotor bars. In this design, number of rotor bars is $G = 11$. So, $R'_{end_ring} = 0.83 \times 10^{-4} \Omega$.

Then, the equivalent rotor resistance reflected to the stator side can be obtained by $R'_2 = 4m(k_w W_1)^2 (R_{bar} + 2R'_{end_ring}) / G = 5.92 \Omega$ where m is the phase number and W_1 is the number of turns of stator winding per phase.

Table 3- 6: Dimensions of rotor bar and end ring

Item	Value	Item	Value
End ring outer diameter (d_1)	58mm	End ring width (w_e)	5mm
End ring inner diameter (d_2)	53mm	Rotor bar diameter (D_b)	2.5mm
End ring height (h_e)	2mm	Rotor bar height(h_b)	15mm

(a) Dimensions of rotor bar

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(b) Dimensions of end ring

Fig 3- 7: Dimensions of rotor bar and end ring for squirrel-cage rotor

3.5 Mathematical modeling and analysis of inductive driver

As the torque driver of the proposed CBP is a special form of radial-flux induction motor, this section provides a discussion on how a mathematical model of the torque driver can be formulated based on the conventional equivalent circuit model of induction motor.

3.5.1 General equivalent circuit model of induction motor

When balanced sinusoidal voltages U_A U_B U_C are applied across the terminals of the stator phase windings of an induction motor, an input-voltage space vector $\vec{U}_1$ is obtained, where:

$$U_A = \sqrt{2}U_1 \cos(\omega_1 t) \quad (3.17)$$

$$U_B = \sqrt{2}U_1 \cos(\omega_1 t - 2\pi/3) \quad (3.18)$$

$$U_C = \sqrt{2}U_1 \cos(\omega_1 t - 4\pi/3) \quad (3.19)$$

$$\vec{U}_1 = U_A + U_B e^{j\frac{2\pi}{3}} + U_C e^{j\frac{4\pi}{3}} = \frac{3\sqrt{2}}{2} U_1 e^{j\omega_1 t} \quad (3.20)$$

U_1 is the constant rms value of the applied phase voltage and ω_1 is its constant angular frequency. The voltages generate a set of balanced phase currents which can be written as:

$$I_A = \sqrt{2}I_1 \cos(\omega_1 t - \varphi_1) \quad (3.21)$$

$$I_B = \sqrt{2}I_1 \cos(\omega_1 t - \varphi_1 - 2\pi/3) \quad (3.22)$$

$$I_C = \sqrt{2}I_1 \cos(\omega_1 t - \varphi_1 - 4\pi/3) \quad (3.23)$$

Thus, the space vector of the stator currents is obtained as

$$\vec{I}_1 = \frac{3\sqrt{2}}{2} I_1 e^{j(\omega_1 t - \varphi_1)} \quad (3.24)$$

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in which I_1 is the constant rms value of the stator phase currents; φ_1 represents the current phase angle with respect to the voltage. Similarly, the rotor-current space vector referred to the stator side is written as:

$$\vec{I}_2 = \frac{3\sqrt{2}}{2} I_2 e^{j(\omega_1 t - \varphi_2)} \quad (3.25)$$

where I_2 is the constant rms value of the rotor phase currents; φ_2 represents the rotor current phase angle with respect to the stator voltage.

$$\vec{I}_m = \vec{I}_1 + \vec{I}_2 \quad (3.26)$$

Then the general equivalent circuit model of the squirrel-cage induction motor can be obtained as shown in Fig 3-8.

Fig 3- 8: General equivalent circuit model of the squirrel-cage induction motor

The general equivalent circuit model consists of five main parameters: stator and rotor resistances, R_1 and R_2 respectively, stator and rotor phase leakage inductances, $L_{\sigma 1}$ and $L_{\sigma 2}$ respectively, and the principle inductance of the stator phases, L_p . An auxiliary parameter related to the principle inductance, namely the magnetizing inductance, L_0 , may be used in the general equivalent circuit model. For a given temperature and frequency, the values of R_1 and R_2 can be conveniently measured or calculated. Special attention should be paid to determining the parameters L_p and L_0 .

3.5.2 Inductance calculation

The proposed radial-flux induction motor model is shown in Fig 3-9. Let the concentrated winding of the stator phase-*A* be fed by a constant current I , assuming that all remaining stator and rotor phases are open-circuited. The magnitude of the fundamental component of the magneto-motive force, $\hat{F}$, would be

$$\hat{F} = \frac{4}{\pi} \frac{K_w N}{p} I \quad (3.27)$$

in which $K_w N$ is the effective number of turns winding and p is the number of pole-pairs. The factor $4/\pi$ in equation (3.27) accounts for the fundamental magnitude of an ideal square waveform.

Fig 3- 9: Radial-flux induction motor for CBP

The peak value of the flux-density within the air-gap is calculated as

$$\hat{B} = \mu_0 \frac{\hat{F}}{2\delta} = \frac{4}{\pi} \mu_0 \frac{K_w N}{2\delta p} I \quad (3.28)$$

in which δ is the air gap length. The average flux-density per pole is obtained as

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$$B_{av} = \frac{2}{\pi} \hat{B} = \frac{8}{\pi^2} \mu_0 \frac{K_w N}{2\delta p} I \quad (3.29)$$

The principle flux per pole is

$$\Phi = B_{av} \frac{\pi DL}{2p} \quad (3.30)$$

in which D and L are respectively the width of stator bar and stack length. The principle flux linkage is thus

$$\Psi = K_w N \Phi = \frac{4\mu_0 (K_w N)^2 DL}{\pi p^2} I \quad (3.31)$$

Finally, the principle inductance is obtained as

$$L_p = \frac{\Psi}{I} = \frac{2\mu_0 (K_w N)^2 DL}{\pi p^2 \delta} \quad (3.32)$$

Because of the air-gap increase due to the slots in both rotor and stator and the magnetic saturation occurring to some extent in all real machines, there is a difference between the value of principle inductance calculated according to equation (3.32) and its measured value. The two coefficients: K_δ , air gap coefficient and K_s , saturation coefficient are introduced to account for these effects. Then, the principle inductance is therefore more accurately expressed as

$$L_p = \frac{2\mu_0 (K_w N)^2 DL}{\pi p^2 K_\delta K_s \delta} \quad (3.33)$$

A commonly used parameter in the general equivalent circuit model is the magnetizing inductance, L_o , which, in the general case of an m -phase machine, is related to principle inductance by

$$L_o = \frac{3}{2} L_p \quad (3.34)$$

Taking into account equation (3.33), the magnetizing inductance in this radial flux three-phase motor-pump system is calculated as

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$$L_o = \frac{3\mu_o}{\pi p^2} \frac{(K_w N)^2 DL}{K_\delta K_s \delta} \quad (3.35)$$

The leakage inductance of the stator and rotor phases are respectively labeled as $L_{\sigma 1}$ and $L_{\sigma 2}$.

3.5.3 Magnetic energy and electromagnetic torque [104]

A. Leakage magnetic energy

The first component of the magnetic energy corresponds to the leakage magnetic flux, which links only with the stator or rotor windings. Taking into account equation (3.24), the magnetic energy related to the stator leakage inductance is

$$W_{\sigma 1} = \frac{1}{2} L_{\sigma 1} (I_A^2 + I_B^2 + I_C^2) = \frac{1}{3} L_{\sigma 1} \bar{I}_1^2 \quad (3.36)$$

Similarly, the magnetic energy related to the rotor leakage inductance is calculated as

$$W_{\sigma 2} = \frac{1}{3} L_{\sigma 2} \bar{I}_2^2 \quad (3.37)$$

B. Magnetizing energy

The second component of the magnetic energy corresponds to the common magnetic flux, which links with both the stator and rotor windings. This component is known as magnetizing energy. The magnetizing energy results from the action of the magnetomotive forces created by both the stator and rotor windings. In order to calculate the magnetizing energy, the principle inductances as well as the mutual inductances between various stator phases must be determined. The principle inductances of the stator phases A, B and C are all equal and can be expressed as

$$L_{AA} = L_{BB} = L_{CC} = L_p \quad (3.38)$$

where L_p is the principle inductance of each phase, which can be calculated by equation (3.33) for radial-flux geometry.

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As there is no overlap between adjacent stator concentrated windings, the mutual inductances between the stator phase windings A, B and C are negligible. Then, the magnetizing energy is calculated as

$$W_{\sigma} = \frac{1}{2} L_p \bar{I}_m^2 \quad (3.39)$$

Taking into account equation (3.34), the magnetizing energy can also be written as

$$W_{\sigma} = \frac{1}{3} L_o \bar{I}_m^2 \quad (3.40)$$

C. Total magnetic energy

The total magnetic energy stored within the system is

$$W_{mag} = \frac{1}{3} L_o \bar{I}_m^2 + \frac{1}{3} L_{\sigma 1} \bar{I}_1^2 + \frac{1}{3} L_{\sigma 2} \bar{I}_2^2 \quad (3.41)$$

D. Electromagnetic torque

Electromagnetic torque can be computed as the partial derivative of the magnetizing energy versus the mechanical angular deviation of the rotor with respect to the stator. The derivative is calculated assuming that the rms values of currents in both the stator and rotor windings remain unchanged. Thus, by definition, the electromagnetic torque can be written as

$$T = \left(\frac{\partial W_{mag}}{\partial \lambda} \right)_{\bar{I}_1, \bar{I}_2 = cst} = \frac{2}{3} L_o \text{Im} \{ \bar{I}_1 \bar{I}_2^* \} \quad (3.42)$$

where Im stands for the imaginary part and the asterisk stands for the conjugate of a complex number. According to the model of Fig 3-8,

$$\bar{I}_2 = - \frac{j\omega_1 L_o}{R_2/s + j\omega_1 (L_o + L_{\sigma 2})} \bar{I}_1 = A_i \bar{I}_1 \quad (3.43)$$

where A_i is therefore the current ratio $\bar{I}_2$ over $\bar{I}_1$,

$$A_i = - \frac{j\omega_1 L_o}{R_2/s + j\omega_1 L_2} \quad (3.44)$$

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and $L_1 = L_o + L_{\sigma 1}$ and $L_2 = L_o + L_{\sigma 2}$. The open-loop-rotor stator time-constant and the open-loop-stator rotor time-constant are respectively defined as

$$\tau_1 = \frac{L_1}{R_1} \quad (3.45)$$

$$\tau_2 = \frac{L_2}{R_2} \quad (3.46)$$

The stator, rotor and total dispersion coefficients are respectively defined as

$$\sigma_1 = \frac{L_{\sigma 1}}{L_o} \quad (3.47)$$

$$\sigma_2 = \frac{L_{\sigma 2}}{L_o} \quad (3.48)$$

$$\sigma = 1 - \frac{1}{(1 + \sigma_1)(1 + \sigma_2)} \quad (3.49)$$

The input impedance of the circuit as seen from the stator terminals is obtained as

$$Z_1 = R_1 + j\omega_1 L_1 \frac{1 + j\omega_1 \sigma s \tau_2}{1 + j\omega_1 s \tau_2} \quad (3.50)$$

Then the electromagnetic torque from equation 3.42 can be expressed as

$$T = \frac{2}{3} L_o \operatorname{Im} \left\{ \frac{\bar{U}_1}{Z_1} (A_i \frac{\bar{U}_1}{Z_1})^* \right\} \quad (3.51)$$

By expanding equation 3.51, we get

$$T = 3(1 - \sigma) L_1 \frac{\bar{U}_1^2}{(L_1 \omega_1)^2} \frac{\hat{s}}{1 + \sigma^2 \hat{s}^2} \quad (3.52)$$

where $\hat{s} = \tau_2 s \omega_1$ is termed as the normalized slip

3.6 Magnetic bearing system for proposed CBP

The conceptual design, theoretical analysis and dimension selection of magnetic bearings for the proposed radial-flux inductively-driven CBP are described in this section. The effects of fringing and leakage flux have been neglected in the analysis.

3.6.1 Six degrees of freedom of the motion of the impeller

In a conventional motor, there are mechanical bearings to support the rotor. For blood pump applications, it would be advantageous if a magnetic system can be designed to support it. The function of the magnetic bearings is to support the impeller in a desired position so that it can be rotated freely. In Fig 3-10, it can be seen that a freely suspended rotor-impeller of the type proposed in the previous sections has altogether six degrees of motion freedom [104]. For pump application, we only desire the rotation about the z-axis and the other 5 freedoms could be restricted by means of passive and active magnetic bearings [105].

Fig 3- 10: The six degrees of freedom of a pump impeller

Numerous advantages such as long life, high reliability, frictionless nature, lubrication-free operation, low losses, are offered by magnetic bearing system consisting of permanent magnets and controlled electromagnets. They have become indispensable for use in applications such as high-speed turbo-pumps, space applications, vacuum and clean

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room applications and flywheel energy storage systems. The satisfactory operation of this type of magnetic bearing is strongly dependent on the characteristics of the permanent magnet and its configuration in the bearing system. This project needs to investigate a set of PMB and AMB to control the 1st, 2nd, 3rd, 4th and 5th degrees of freedom of the blood pump impeller. To simplify the bearing design, passive repulsive type of magnetic bearing is proposed to constrain the 1st, 2nd, 4th, and 5th degrees of freedom shown in Fig 3-8. Since the magnetizing directions of the PM rings mounted on outer cores of both rotor and stator of the PMB are parallel, the repulsive magnetic coupling between these two PM rings can produce forces to constrain the translations and rotations along x-axis and y-axis. The chosen dimensions of these two PM rings will determine the constrain forces. However, the 3rd degree of freedom has to be controlled by an active magnetic bearing. The design of these two kinds of the bearings is explained in the next two sections.

3.6.2 Conceptual design of PMB

Fig 3- 11: Structure of proposed PMB

The proposed PMB is a kind of passive repulsive type of magnetic bearing. Its structure is very simple, consisting of two permanent magnetic (PM) concentric rings, as shown in Fig 3-11. The inner PM ring is to be integrated with the impeller and the outer one is to be mounted on the pump wall. The working principle of this PMB is that when the impeller displaces from central point along x-axis or y-axis, the PM rings would magnetically react with each other to produce a force to push the impeller back to the central position. Then the first and second degrees of freedom of the impeller can be controlled by this PMB.

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Similarly, if there is rotation about x and y axes, the net increase in repulsive forces will return the impeller back to equilibrium position. So the 4th and 5th degrees of freedom are also constrained by the PMB.

The drawback of this kind of bearing is that it cannot be controlled automatically by the system so the control of the degrees of freedom is only passive. Fig 3-10 shows the proposed incorporation of this repulsive type magnetic bearing in the sketch design of the CBP. In order to show the PMB clearly and to simplify the drawing, only the impeller, the pump wall and the PMB are included in Fig 3-12.

Fig 3- 12: Axial cross-section of pump chamber showing PM rings of proposed PMB

3.6.3 Theoretical analysis of proposed PMB

A. Field theory [119-122]

There are numerous techniques for analyzing permanent magnetic devices. The method used here is to model the permanent magnet as distributions of equivalent magnetic charge and then compute the force between the two charge distributions. The basic formulae of the magnetic field for current-free regions are

$$\nabla \times H = 0 \quad (3.53)$$

$$\nabla \cdot B = 0 \quad (3.54)$$

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where H is the magnetic field strength and B is the magnetic flux density. In magnetic materials the two fields H and B are related to the physical magnetization density M ,

$$B = \mu_0 (H + M) \quad (3.55)$$

It is well known that if the magnetization $M(\vec{x})$ is confined to a given volume V and falls abruptly to zero outside this volume (as in the case with permanent magnets), then the magnetic field can be written in the following integral form:

$$B(\vec{x}) = \frac{\mu_0}{4\pi} \int_V \frac{\rho_M(\vec{x}')(\vec{x} - \vec{x}')}{|\vec{x} - \vec{x}'|^3} dV' + \frac{\mu_0}{4\pi} \int_S \frac{\sigma_M(\vec{x}')(\vec{x} - \vec{x}')}{|\vec{x} - \vec{x}'|^3} dS', \quad (3.56)$$

where S defines the surface about bounds V (the surface of magnet), and

$$\rho_M(\vec{x}') = -\nabla' \cdot M(\vec{x}') \quad (3.57)$$

$$\sigma_M(\vec{x}') = M(\vec{x}') \cdot \hat{n} \quad (3.58)$$

are the volume and surface charge densities, respectively.

These integrals can be evaluated numerically by discretizing the volume and the surface into a mesh of indexed elements (denoted by ΔV_j and ΔA_k) and summing the contribution from each element:

$$B(\vec{x}_i) = \frac{\mu_0}{4\pi} \sum_j \frac{\rho_M(\vec{x}_j)(\vec{x}_i - \vec{x}_j)}{|\vec{x}_i - \vec{x}_j|^3} \Delta V_j + \frac{\mu_0}{4\pi} \sum_k \frac{\sigma_M(\vec{x}_k)(\vec{x}_i - \vec{x}_k)}{|\vec{x}_i - \vec{x}_k|^3} \Delta A_k \quad (3.59)$$

It follows from above that the field at a point $\vec{x}_1$ due to a point source $Q_M(\vec{x}_2)$ at $\vec{x}_2$ is

$$B(\vec{x}_1) = \frac{\mu_0}{4\pi} \frac{Q_M(\vec{x}_2)(\vec{x}_1 - \vec{x}_2)}{|\vec{x}_1 - \vec{x}_2|^3} \quad (3.60)$$

where, if the point source is due to a volume charge,

$$Q_M(\vec{x}_2) = \rho_M(\vec{x}_2) \Delta V_2 \quad (3.61)$$

or, if it is due to a surface charge,

$$Q_M(\vec{x}_2) = \sigma_M(\vec{x}_2) \Delta A_2 \quad (3.62)$$

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For that the sources in (3.59) are given in terms of equivalent charges, the field can be computed by first reducing the magnet to an equivalent charge distribution of volume and surface in accordance with (3.57) and (3.58), and applying (3.59).

The force between two magnets can be computed using the following integral expression:

$$F = \int_V \rho_M(\vec{x}) B_{ext}(\vec{x}) dV + \int_S \sigma_M(\vec{x}) B_{ext}(\vec{x}) dS \quad (3.63)$$

where $B_{ext}(\vec{x})$ is the field due to one of the magnets and $\rho_M(\vec{x})$, $\sigma_M(\vec{x})$ are the charge densities defining the other magnet. Once again, the integration can be evaluated numerically by discretizing V and S :

$$F = \sum_{V_n} \rho_M(\vec{x}_n) B_{ext}(\vec{x}_n) \Delta V_n + \sum_{A_m} \sigma_M(\vec{x}_m) B_{ext}(\vec{x}_m) \Delta A_m \quad (3.64)$$

where ΔV_n and ΔA_m denote the indexed volume and area elements. Combining this expression with (3.59) gives the force in terms of the interaction of two separated charge distributions:

$$F = \frac{\mu_0}{4\pi} \sum_{V_n} \left\{ \sum_{V_j} \frac{\rho_M(\vec{x}_n) \rho_M(\vec{x}_j) (\vec{x}_n - \vec{x}_j)}{|\vec{x}_n - \vec{x}_j|^3} \Delta V_j \right\} \Delta V_n + \frac{\mu_0}{4\pi} \sum_{V_n} \left\{ \sum_{A_k} \frac{\rho_M(\vec{x}_n) \sigma_M(\vec{x}_k) (\vec{x}_n - \vec{x}_k)}{|\vec{x}_n - \vec{x}_k|^3} \Delta A_k \right\} \Delta V_n + \frac{\mu_0}{4\pi} \sum_{A_m} \left\{ \sum_{V_j} \frac{\sigma_M(\vec{x}_m) \rho_M(\vec{x}_j) (\vec{x}_m - \vec{x}_j)}{|\vec{x}_m - \vec{x}_j|^3} \Delta V_j \right\} \Delta A_m + \frac{\mu_0}{4\pi} \sum_{A_m} \left\{ \sum_{A_k} \frac{\sigma_M(\vec{x}_m) \sigma_M(\vec{x}_k) (\vec{x}_m - \vec{x}_k)}{|\vec{x}_m - \vec{x}_k|^3} \Delta A_k \right\} \Delta A_m \quad (3.65)$$

where the indices j and k label the discretization of one magnet while n and m label that of the other. In other words, once two magnets have been reduced to discretized charge distributions, (3.65) can be used to compute the force between them. In addition, it follows that the force between two discrete point charges $Q_M(\vec{x}_1)$ at $\vec{x}_1$ and $Q_M(\vec{x}_2)$ at $\vec{x}_2$ is:

$$F = \frac{\mu_0}{4\pi} \frac{Q_M(\vec{x}_1) Q_M(\vec{x}_2) (\vec{x}_1 - \vec{x}_2)}{|\vec{x}_1 - \vec{x}_2|^3} \quad (3.66)$$

where the charge Q_M is computed using (3.61) or (3.62) depending on whether volume charge or surface charge is considered. Equations (3.65) and (3.66) are the basis for the derivation of the force equations which will be presented next.

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B. Interactive force

In this section, a model is derived for the radial force between the rotor and the stator of the passive magnetic bearing when the rotor (inner ring) deviates from the central point.

It is assumed that the magnetization is constant throughout the magnets, this means:

$$M(\vec{x}) = M\hat{z} \quad (3.67)$$

It follows from (3.57), (3.58) that

$$\rho_M = -\nabla \cdot M = 0 \quad (3.68)$$

$$\sigma_M = M \cdot \hat{n} = M \quad (3.69)$$

Thus, the volume charge density in both magnets is zero and the surface charge density is constant. Therefore, the volume terms in (3.65) vanish and the force between the two magnets reduces to

$$F = \frac{\mu_0}{4\pi} \sum_{A_m} \left\{ \sum_{A_k} \frac{\sigma_M(\vec{x}_m) \sigma_M(\vec{x}_k) (\vec{x}_m - \vec{x}_k)}{|\vec{x}_m - \vec{x}_k|^3} \Delta A_k \right\} \Delta A_m \quad (3.70)$$

Instead of evaluating (3.70) directly, it is easier to derive the force in the following steps: First, we get the equation of the force between two point sources. Then, the force between two faces is obtained. Lastly, the total force can be figured out by summing the contributions from all faces. And considering the symmetric properties of the PMB, the radial force between the rotor and stator is only the sum of the force between the two upper faces and that between the two lower faces when the inner ring deviate from the central position. Also, these two forces are equal both in value and direction. So, the net radial force produced by the PMB can be calculated as twice that of any one of the forces.

The geometrical model for the analysis of interactive force of PMB is shown in Fig 3-13. Since the geometry is cylindrical, it is more convenient to express it in the cylindrical coordinates (r, θ, z) . Without any loss of generality, we can set $Q_M(\vec{x}_1)$ at $(r_1, \theta_1, 0)$ and the other $Q_M(\vec{x}_2)$ at $(r_2, \theta_2, 0)$, which are expressed in their own coordinates respectively. And according to the relation between these two coordinates, $Q_M(\vec{x}_2)$ at $(r_2, \theta_2, 0)$ can be converted to the same coordinate of $Q_M(\vec{x}_1)$ as $Q_M(\vec{x}_2)$ at $(r_2', \theta_2', 0)$, where

$$r_2' = \sqrt{d^2 + r_2^2 - 2dr_2 \cos \theta_2} \quad (3.71)$$

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$$\theta_2' = \arcsin \frac{r_2 \sin \theta_2}{r_2'} = \arcsin \frac{r_2 \sin \theta_2}{\sqrt{d^2 + r_2^2 - 2dr_2 \cos \theta_2}} \quad (3.72)$$

Fig 3- 13: Analysis of radial forces produced by the proposed PMB

The interactive force between these two point sources follows from (3.65),

$$F_{point-point} = \frac{\mu_0}{4\pi} \frac{Q_M(r_1, \theta_1, 0) Q_M(r_2', \theta_2', 0)}{[r_1^2 + (r_2')^2 - 2r_1 r_2' \cos(\theta_1 - \theta_2')]^{3/2}} \quad (3.73)$$

The next step is to compute the radial force exerted by one surface on the face of the other magnet. The method is to reduce the two magnets into a mesh of subsections, define a point charge at the middle point of each element in accordance with (3.58), then compute the force exerted by one surface on the other by summing the contributions from the individual point charges.

The first step is to discretize the surface area of each magnet into a set of indexed elements. Let R_1 and R_2 denote the inner radius and outer radius of inside magnet respectively. At same time, R_3 and R_4 are the inner radius and outer radius of outside magnet. Since the area of the surface is spanned by the two coordinate variables r and θ , two indices (i, j) are needed. Let N_r and N_θ be the number of mesh divisions of these two indices. Then each element of the inside magnet surface will have the same area ΔA_1 given by

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$$\Delta A_1 = \frac{\pi(R_2^2 - R_1^2)}{N_r N_\theta} \quad (3.74)$$

And applying the same method to the outside magnet, we get

$$\Delta A_2 = \frac{\pi(R_4^2 - R_3^2)}{N_r N_\theta} \quad (3.75)$$

It is obvious that the angular measure for both magnets can be expressed in

$$\Delta\theta = \frac{2\pi}{N_\theta} \quad (3.76)$$

To the inside magnet, Let (r_i, θ_j) denote the middle point of the element ΔA_1 and let $R(i+1)$ and $R(i)$ denote its radial boundaries. Then

$$r_i = \frac{R(i+1) + R(i)}{2} \quad (i = 1, 2, \dots, N_r) \quad (3.77)$$

$$\theta_j = (j-1)\Delta\theta + \frac{\Delta\theta}{2} \quad (j = 1, 2, \dots, N_\theta) \quad (3.78)$$

The end points are computed using the following equation:

$$R(i+1) = \sqrt{R(i)^2 + \frac{R_2^2 - R_1^2}{N_r}} \quad (i = 1, 2, \dots, N_r) \quad (3.79)$$

where $R(1) = R_1$. Equation (3.79) is used to generate all end point values from which all the middle point values of the inside magnet are obtained.

At the same time, applying the above strategy to the outside magnet, we get:

$$r_{i'} = \frac{R(i'+1) + R(i')}{2} \quad (i' = 1, 2, \dots, N_r) \quad (3.80)$$

$$\theta_{j'} = (j'-1)\Delta\theta + \frac{\Delta\theta}{2} \quad (j' = 1, 2, \dots, N_\theta) \quad (3.81)$$

$$R(i'+1) = \sqrt{R(i')^2 + \frac{R_4^2 - R_3^2}{N_r}} \quad (i' = 1, 2, \dots, N_r) \quad (3.82)$$

Therefore, (3.77)-(3.82) define a scheme for discretizing the surface of the two magnets into elements and assigning values of middle point of each element.

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Next, consider the magnetic interaction between each element of the inside magnet's upper surface with that of the outside magnet's upper surface. Let the coordinates of the middle point of the two elements be (r_i, θ_j, H) and (r_i', θ_j', H) . The magnetic charge associated with each element is

$$\Delta Q(r_i, \theta_j, H) = \sigma(r_i, \theta_j, H) \Delta A_1 \quad (3.83)$$

$$\Delta Q(r_i', \theta_j', H) = \sigma(r_i', \theta_j', H) \Delta A_2 \quad (3.84)$$

And (r_i', θ_j', H) can be obtained from (r_i, θ_j, H) by (3.71) and (3.72).

From (3.69), (3.74) and (3.75) these charges reduce to

$$\Delta Q(r_i, \theta_j, H) = M \frac{\pi(R_2^2 - R_1^2)}{N_r N_\theta} \quad (3.85)$$

$$\Delta Q(r_i', \theta_j', H) = M \frac{\pi(R_4^2 - R_3^2)}{N_r N_\theta} \quad (3.86)$$

Then, the force is obtained by summing the contribution S of all the elements in accordance with (3.74) and (3.75). By exploiting the symmetry of the magnets, the force reduces to:

$$F_{upperfaces} = \frac{\mu_0}{4\pi} M^2 \frac{\pi^2 (R_2^2 - R_1^2)(R_4^2 - R_3^2)}{(N_r N_\theta)^2} N_\theta \sum_{i=1}^{N_r} \sum_{i'=1}^{N_r} \sum_{j=1}^{N_\theta} \frac{1}{[r_i^2 + (r_i')^2 - 2r_i r_i' \cos(\theta_j - \theta_j')]^{3/2}} \quad (3.87)$$

Lastly, applying two times of the force of upper faces, we get the total radial force:

$$F_{total} = 2F_{upperfaces} \quad (3.88)$$

Based on the above analysis and the blood pump chamber dimensions calculated in Section 3.3, the dimensions of the inner and outer PM rings are selected as listed in Table 3-7.

Table 3- 7: Dimensions of PMB components

Height of inner and outer PM rings	5mm
Inner radius of inner PM ring	29mm
Outer radius of inner PM ring	32mm
Inner radius of outer PM ring	35mm
Outer radius of outer PM ring	40mm

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3.6.4 Conceptual design of AMB

Earnshaw’s Theorem (Earnshaw, 1842) states that if inverse-square-law forces govern a group of charged particles, the latter cannot be in stable equilibrium. The theorem is based on the consequence that inverse-square-law forces are subjected to Laplace partial differential equation. Since the solution of this equation does not have any local maxima or minima and only saddle-type equilibrium points exist, there can be no equilibrium. The theorem usually applies to charged particles and magnetic dipoles but can also be extended to solid magnets. Thus, this theorem explains why it is impossible to have a completely stable PMB using only the forces of static fields of permanent magnets. Active feedback control of electromagnetic field is the most commonly used method to circumvent Earnshaw’s Theorem.

Fig 3- 14: Half sectional view of AMB

Fig 3- 15: The location of AMB in the pump chamber

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Fig 3-14 shows the structure of the proposed active magnetic bearing to complement the PMB. As it is symmetric around z axis, only half sectional view is shown here. The AMB has two major parts: 1) rotor; 2) stator. The rotor is integrated with impeller of the blood pump, both of which are located inside the blood pump housing. There is copper coil wound around the central core of stator, which is outside of the pump wall. Permanent magnets 1 and 2 are mounted on the central core's surface of rotor and stator, respectively. They have the same magnetizing direction along y-axis. Permanent magnets 3 and 4 are mounted on the outer core's surface of rotor and stator. However they have the opposite magnetizing direction along y-axis. So, the magnetic coupling between permanent magnets 1 and 2 produces an attraction force on the rotor. Conversely, a repelling force is acting on the impeller by the magnetic coupling between magnets 3 and 4. The sum of these two forces is the net force acting on the impeller in y-axis direction. As the available area in the CBP is limited, the area covered by PMs 3 and 4 is limited. With PMs 1 and 2, the permanent magnets cover a greater area and thus a more stable AMB can be designed. The dimensions used in Fig 3-14 are summarized in Table 3-8.

Table 3- 8: Parameters used in Fig 3-14

h_1	Height of permanent magnet on the surface of rotor inner core (PM 1 in Fig 3-16)
h_2	Height of permanent magnet on the surface of stator inner core (PM 2 in Fig 3-16)
h_3	Height of permanent magnet on the surface of stator inner core (PM 3 in Fig 3-16)
h_4	Height of permanent magnet on the surface of stator inner core (PM 4 in Fig 3-16)
g_1	Length of air gap between the permanent magnets on the surfaces of stator inner core and rotor inner core (air gap between PM 1 and PM 2 in Fig 3-16)
g_2	Length of air gap between the permanent magnets on the surfaces of stator outer core and rotor outer core (air gap between PM 3 and PM 4 in Fig 3-16)
r_1	Radius of inner core both for stator and rotor
r_2	Inner radius of outer core both for stator and rotor
r_3	Outer radius of outer core both for stator and rotor

It is obvious that the electromagnetic field produced by the AMB coil can strengthen or weaken the coupling between magnets 1 and 2, according to the polarity and magnitude of

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coil current. So the net force acting on impeller in z-axis direction can be controlled by the current of the coil.

3.6.5 Theoretical analysis of proposed AMB

Fig 3- 16: Equivalent magnetic circuit of proposed AMB

In order to derive a mathematical model of the proposed AMB, equivalent magnetic circuit method is adopted here. According to the structure of the AMB and the magnetization directions of its permanent magnets, the magnetic circuit is obtained and shown in Fig 3-16. Because of the very high magnetic permeability of the stator and rotor cores, their magnetic reluctances are neglected here. The parameters used in Fig 3-16 are explained in Table 3-9.

Table 3- 9: Parameters used in magnetic circuit analysis of AMB

F_{m1}	Magnetomotive force produced by the permanent magnet on the surface of rotor inner core (PM 1 in Fig 3-16)	R_{m1}	Magnetic reluctance produced by the permanent magnet on the surface of rotor inner core
F_{m2}	Magnetomotive force produced by the permanent magnet on the surface of stator inner core (PM 2 in Fig 3-16)	R_{m2}	Magnetic reluctance produced by the permanent magnet on the surface of stator inner core

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F_{m3}	Magnetomotive force produced by the permanent magnet on the surface of rotor outer core (PM 3 in Fig 3-16)	R_{m3}	Magnetic reluctance produced by the permanent magnet on the surface of rotor outer core
F_{m4}	Magnetomotive force produced by the permanent magnet on the surface of stator outer core (PM 4 in Fig 3-16)	R_{m4}	Magnetic reluctance produced by the permanent magnet on the surface of stator outer core
R_{g1}	Magnetic reluctance produced by the air gap between the permanent magnets on the surfaces of stator inner core and rotor inner core (air gap between PM 1 and PM 2 in Fig 3-16)	R_{g2}	Magnetic reluctance produced by the air gap between the permanent magnets on the surfaces of stator outer core and rotor outer core (air gap between PM 3 and PM 4 in Fig 3-16)
NI	Magnetomotive force produced by the stator coil current		

According to the geometry of the proposed AMB, the magnetic reluctance produced by the permanent magnets can be calculated by the following formulas

$$R_{m1} = \frac{h_1}{\mu\pi r_1^2} \quad (3.89)$$

$$R_{m2} = \frac{h_2}{\mu\pi r_1^2} \quad (3.90)$$

$$R_{m3} = \frac{h_3}{\mu\pi(r_3^2 - r_2^2)} \quad (3.91)$$

$$R_{m4} = \frac{h_4}{\mu\pi(r_3^2 - r_2^2)} \quad (3.92)$$

where μ is the permeability of used permanent magnet material.

Similarly, the magnetic reluctance produced by the air gaps can be calculated by the following formulas

$$R_{g1} = \frac{g_1}{\mu_0\pi r_1^2} \quad (3.93)$$

$$R_{g2} = \frac{g_2}{\mu_0\pi(r_3^2 - r_2^2)} \quad (3.94)$$

where μ_0 is the permeability of air.

The magnetic flux ϕ in this magnetic circuit can be derived as

$$\phi = \frac{F_{m1} + F_{m2} + F_{m3} - F_{m4} + NI}{R_{m1} + R_{m2} + R_{m3} + R_{m4} + R_{g1} + R_{g2}} \quad (3.95)$$

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so, the magnetic flux density in the air gap between the permanent magnets on the surfaces of stator inner core and rotor inner core B_{g1} can be calculated as

$$B_{g1} = \frac{\phi}{\pi r_1^2} \quad (3.96)$$

the energy density in this air gap W_{g1} is

$$W_{g1} = \frac{\mu_0 B_{g1}^2}{2} \quad (3.97)$$

the attracting force acting on the rotor produced by the magnetic coupling within this air gap F_{g1} is

$$F_{g1} = \frac{dW_{g1}}{dg_1} \quad (3.98)$$

Similarly for the air gap between the permanent magnets on the surfaces of stator outer core and rotor outer core, the magnetic flux density in this air gap B_{g2} can be calculated as

$$B_{g2} = \frac{\phi}{\pi(r_3^2 - r_2^2)} \quad (3.99)$$

the energy density in this air gap W_{g2} is

$$W_{g2} = \frac{\mu_0 B_{g2}^2}{2} \quad (3.100)$$

the repelling force acting on the rotor produced by the magnetic coupling within this air gap F_{g2} is

$$F_{g2} = \frac{dW_{g2}}{dg_2} \quad (3.101)$$

Then the total force acting on the rotor of AMB along the positive direction of axis Z F_{total} can be derived as

$$F_{total} = F_{g2} - F_{g1} \quad (3.102)$$

Using computational hydrodynamics or experimentally, it is possible to estimate the axial force acting on the impeller, which is to be balanced by the force produced by the AMB in

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Equation 3.102. From here and with the designed dimensions of the components of the AMB, the product NI can be obtained from Equation 3.95. The optimization of the values of N and I shall be a future work which studies the control and the design of such AMB concept for CBP. The equations presented here are more suited for analysis than for design optimization, which should preferably be done using 3-D FEM.

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3.7 FEM analysis and simulation

3.7.1 Electromagnetic theory

Magnetic fields in electric machines are always associated with transient excitations and nonlinear magnetic materials. The following three Maxwell equations are relevant to transient applications.

$$\nabla \times \vec{H} = \vec{J} = \sigma \vec{E} \quad (3.103)$$

$$\nabla \times \vec{E} = -\partial \vec{B} / \partial t \quad (3.104)$$

$$\nabla \cdot \vec{B} = 0 \quad (3.105)$$

where $\vec{H}$ denotes the magnetic field density, $\vec{J}$ is the electric current density, σ is the conductivity of the dielectric, and $\vec{E}$ is the electric field intensity.

From equations (3.103) and (3.105), we can obtain

$$\nabla \times \frac{1}{\sigma} \nabla \times \vec{H} + \frac{\partial \vec{B}}{\partial t} = 0 \quad (3.106)$$

The force and torque can be calculated as the derivative of the stored magnetic co-energy with respect to a small displacement. The co-energy can be written as

$$W' = \int_V \left[\int_0^B \vec{H} \cdot d\vec{B} \right] dV \quad (3.107)$$

Then the component of instantaneous force F in the direction of the displacement s is

$$F = \frac{dW'}{ds} \quad (3.108)$$

The instantaneous torque T with a small angular rotation displacement θ is represented by

$$T = \frac{\partial W'}{\partial \theta} \quad (3.109)$$

3.7.2 FEM analysis

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FEM analysis and simulation have been applied to the proposed inductive torque driver and magnetic bearing system designs. The Maxwell Simulator from Ansoft Corporation is chosen as the simulation software. Though direct involvement with the underlying mathematics is not essential, however, the concepts and related terminology that are germane to our problem need to be understood. Basically the key point is a numerical solution to the governing equations and loading conditions that characterize the behavior of the systems under study.

There are four basic concepts, the system, the domain, the governing equations and loading conditions. The system includes the electromagnetic system of the proposed blood pump torque driver and magnetic bearings, which are composed of various material and empty space. The domain of the problem is typically the region of space occupied. The governing equations are differential equations expressing a conservation or balance of the physical properties such as flux, torque and speed. They also include constitutive equations, which describe material behavior and contain physical properties such as the permeability of the materials. Loading conditions are externally originating currents or voltages that interact with the systems causing the state of the systems to change. Partitioning the domain of the problem into smaller regions called elements is a salient feature in every FEM analysis. The shapes of the elements are intentionally made as simple as possible such as triangles in 2D domain and tetrahedrons in 3D domain. The process of partitioning a domain into a mesh of elements is mesh generation. This procedure is automated by computer programs called preprocessors.

According to the different structure character of every design, 2-D field simulations [106] were chosen to analyze the electromagnetic fields in the proposed torque driver and AMB. Three-dimensional field simulations [107] were used to simulate the proposed PMB. 3-D FEM simulations are used only when the object does not possess axis of symmetry or when the axial length is very short. This is the case for PMB especially when its rotor is displaced from the central axis. If there is axis of symmetry or considerable axial length such as the radial-flux induction motor, 2-D FEM simulations are sufficient. 2-D simulations have far lesser computational overheads compared to 3-D simulations. As the motor, AMB and PMB are magnetically decoupled, it sufficiently accurate for a first order

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conceptual design to ignore any mutual effects. Considering all magnetic systems at the same time is possible but computationally very intensive.

The simulation procedures were carried out in five steps:

(1) Step 1 –Specification of design geometry

Maxwell simulators provide interactive graphical user interface to specify the designs' geometries. Sophisticated drawing tools available in the simulator allow fine geometrical details to be accurately defined. So in the drawing panel, simulation models can be created, which includes the stator core, stator yoke, air-gap, coil, rotor cover, rotor yoke, rotor base and the cylindrical permanent magnets.

(2) Step 2 –Specification of material properties

Table 3- 10: Material assignment of motor-pump design

Component	Material
Stator	Cold rolled steel
Rotor shorting rings and rotor bars	Copper
Rotor core	Silicon steel
Coils	Copper
Air-gap and environment	Air

Table 3- 11: Material assignment of magnetic bearings

Component	Material
Stator of AMB	Silicon steel
Rotor of AMB	Silicon steel
Coils	Copper
Stator of PMB	NdFe35
Rotor of PMB	NdFe35
Impeller	Rubber
Pump wall	Glass
Air-gap and environment	Air

The properties of the material type for various parts of the proposed designs are either already available in the simulator's material database, or user-specified by inputting experimental data. Different materials can be assigned to different parts of the simulation

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object. The assignments of material type for the different parts of the designed motor-pump systems and magnetic bearings are indicated in Table 3-10 and 3-11, respectively.

(3) Step 3 –Specification of boundary conditions and excitation sources

The behaviors of the magnetic fields at the borders of problem region and at object interfaces are specified. The boundary condition in the simulation is balloon and the excitation sources are the three phase sinusoidal voltages of the motor coils and direct current of the magnetic bearing coils.

Fig 3- 17: Adaptive mesh refinement

(4) Step 4 –Generating the solution

After steps 1-3, an initial coarse finite element mesh of the defined model was automatically created at the beginning of the solution process. The mesh was refined adaptively according to error energy analysis as the simulation progresses. Through executing repeatedly, the simulation will continue until the desired energy error has been achieved. The adaptive mesh refinement flow diagram is shown in Fig 3-17.

(5) Step 5 –Graphical post-processing and analysis

A powerful post-processor is provided by the simulator to manipulate and display the field and variable solution. The post-processor includes a powerful measurement tool and a mathematical analyzer. Various field quantities can then be measured and calculated from data extracted from the simulations.

3.7.3 Magnetic bearings

A. AMB

2-D simulations were used to test whether the axial movement along z-axis can be controlled by the proposed AMB. Figs 3-18 and 3-19 are the mesh plot and flux plot of the AMB, respectively.

Fig 3- 18: Mesh of magnetic bearings

Fig 3- 19: Flux plot of magnetic bearings

Fig 3-20 shows the FEM analysis simulation results of the AMB. The force is acting on the rotor of AMB, impeller and inner PM ring of PMB as a unit. The curves in this figure show the force versus coil current for different positions of the impeller away from nominal position. The changes in position are in steps of 0.1mm, up to 0.5mm away from nominal position in either direction. It can be seen that the force acting on the impeller at any position can be adjusted by changing the coil current of AMB stator. In other words,

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the impeller's displacement along z-axis can be controlled by the coil current. Furthermore, the current to keep impeller at nominal position is very small. So, the power consumption of the AMB is small.

Fig 3- 20: AMB’s Force VS Ampere-Turn at different impeller position

B. PMB

Fig 3- 21: Simulation model of PMB

3D FEM simulations were used to verify whether PMB can keep the impeller at the central position of the chamber of heart pump. In other words, it finds out whether PMB can constrain the radial distance shift along x-axis and y-axis. Fig 3-21 shows the simulation model of PMB.

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In the simulation, the stator of PMB was kept quiescent and the central point of rotor of PMB moves along y-axis. Then we got the repulsive force acting on rotor of PMB along y-axis during the movement. Fig 3-22 shows these repulsive force characteristics along Y axis and the calculated stiffness characteristics.

Fig 3- 22: PMB FEM simulation results

From Fig 3-22, it can be found that the PMB has the ability to produce a force, which can prevent the rotor of PMB from moving along y-axis. And the designed PMB can provide enough high stiffness along y-axis. Considering that impeller is symmetric along z-axis, which means that it is the same case with x-axis. So, it is verified that PMB can prevent the impeller from moving away from the central position of the chamber of heart pump. In other words, it can control the radial displacement along x-axis and y-axis.

3.7.4 Radial flux motor driver

The Maxwell 2D simulator is adopted to do the transient simulation of the designed radial flux motor-pump driver. The source is three phases sinusoidal voltages supplied to the driver stator coils. The mesh of this novel motor-pump driver is shown in Fig 3-23. And Fig 3-24 is the flux plot of it.

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Fig 3- 23: 2D mesh of radial flux motor-pump driver

Fig 3- 24: 2D flux plot of radial flux motor-pump driver

The simulation results of this radial flux motor-pump driver under no applied load are shown in Fig 3-25. Fig 3-25 (a) shows the electromagnetic torque acting on the rotor. The electromagnetic torque enters a steady state after the transient starting period at the beginning. The simulation result of the rotation speed of this driver’s rotor is presented in Fig 3-25 (b). The rotation speed keeps rising during the starting period and also enters a steady state after transient period. Fig 3-25 (c) shows the position of the rotating rotor which continues rotating from 0 degree to 360 degree.

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Fig 3- 25: 2D FEM simulation results for radial flux design without applied load

(c) Rotor position

Fig 3- 26: 2D FEM simulation results for radial flux design without applied load

In order to test the ability of designed motor-pump device to produce output torque and power, the proposed device were also simulated under loaded conditions. Fig 3-26 shows its simulation results under maximum load torque (0.0075Nm) condition.

Fig 3- 27: Torque-Speed curves comparison

And the Torque VS Rotor Speed comparison between curves obtained from mathematical model calculations derived in section 3.6 and FEM simulations results is shown in Fig 3-27. The comparison shows good accordance between these two results, which support the viability of the derived mathematical model.

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3.7.5 Assessment on viability of proposed CBP

As introduced in the blood pump chamber design, depending on the dimensions of blood pump chamber, the proposed blood pump usually operates at the speed between 2000rpm to 3000rpm to provide enough cardiac output for patients. As shown in Fig 3-27, the design device only can produce no more than 0.002Nm output torque when working at this speed segment, which is able to deliver 0.25W power to the pumping blood. However, the human cardiac power at rest is around 0.9-1.2W [118], which cannot be met by this proposed device. The key problem here is the low electromagnetic torque produced by this design. After reviewing the driver design, it was judged that this is an inherent disadvantage when applying radial flux design to disc type driver which has small ratio of active axial length to outer diameter. In this situation, the application of axial flux design may overcome this obstacle.

Besides the technological and manufacturing differences, many references have studied the comparison between axial flux motors and radial flux motors to understand when and where the axial flux motors show potential advantages [123-128]. A general comparison of axial flux motors versus radial flux motors is not possible. Then, comparison between electric motors is often performed using the “sizing equations” which link the motor electromagnetic torque to the active motor length and to the motor reference diameter. For the radial flux motors the most frequently encountered sizing equation is in the form

$$T_e = \xi_r \cdot D_r^2 \cdot L \quad (3.110)$$

where D_r (m) is the air-gap diameter, and L (m) is the active axial length of the stator core. The coefficient ξ_r ($N \cdot m / m^3$) depends on the air-gap flux density and on the chosen electric loading (air-gap current linear density). A comparison between the AF PM synchronous motors and the RF ones, based on (3.110), is reported in [125]. Equation (3.110) does not take into account the actual flux and current densities that are present in the different motor parts; thus, the electric and the magnetic loads need to be chosen by the designer on the basis of experience. Other forms of sizing equations (see [126], [127], and [128]), such as, for instance,

$$T_e = \xi_e \cdot D_e^3 \cdot L \quad (3.111)$$

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where D_e (m) is the external motor diameter and the coefficient ξ_e ($N \cdot m / m^4$) depends on the flux density in the stator yoke, in the stator tooth, and on the current density in the conductors. Applying equation (3.110) or equation (3.111) to the proposed design with corresponding diameter parameters and coefficient parameters, it is clear that the produced electromagnetic torque is directly decided by the active axial length. With certain corresponding diameter parameters and coefficient parameters, the higher output electromagnetic torque only can be achieved by increasing the active axial length. Unfortunately, the increasing of active axial length is not practical in the proposed device; otherwise the compact blood pump device structure can not be achieved.

The studies in [123, 124 and 125] also reveal that compared to radial flux motors, the axial flux motors are attractive for flat geometries ($\lambda = L / D_e < 0.3$). Under this condition, the axial flux motors can provide both a higher electromagnetic torque and a higher torque density than the radial flux motors. In this proposed design, its $\lambda = L / D_e = 0.24 < 0.3$ just falls into the foresaid category. So, it was judged that the axial flux motor design may be more suitable for this disc-type of motor driver device, which is expected to be able to produce more electromagnetic torque with similar overall size and volume.

3.8 Concluding remarks

This chapter presents the proposal and investigation of the concept of a radial-flux inductively-driven CBP, which may be classified as belonging to the third-generation of CBP designs. Compared with the first two generations of CBP and even with other third generation CBP, this new concept has some potential attractive features when the torque transmission method, the structure simplicity and compactness and the system reliability are all concerned.

Firstly, the proposed design integrates the motor design into the impeller so that there is no rotating part outside the blood pump chamber. It means that the rotating part of the motor, the rotor, becomes part of the impeller, leaving the stator of the motor outside the pump chamber. This rotor-impeller structure leads to a very compact and simple design of the blood pump, which can very suitable for implantation application. Furthermore, the electromagnetic torque produced by the blood pump driver can act on the pump impeller directly. There is no need of mechanical shaft for the transmission of torque, which can eliminate the problem of blood clot on the shaft and the sealing problem of blood pump chamber.

Secondly, a system of magnetic bearings, instead of mechanical bearing, is proposed to support and constrain the rotor-impeller in the CBP. Biomedical devices in contact with living tissue must be biocompatible and sterile. Medical devices should also be non-traumatic, highly reliable, robust and durable. Magnetic levitation technology offers several features to address these issues. By contact-free actuation of moving parts through the walls of a housing, hermetically sealed systems become feasible. The absence of seals further limits the potential entry of microbial organisms into the system. Wide gaps in such pumps also help to reduce blood-trauma (hemolysis). Friction-less operation renders such devices durable and robust, while permitting a wider selection of biocompatible materials.

Thirdly, the proposed torque driver is based on the familiar squirrel-cage induction machine. Compared with the PM motor used in other blood pump designs, electronic

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commutation circuits, rotor position sensors or sensorless algorithms are not needed here. A simple open-loop control strategy based on the Voltage/Frequency method can be adopted to control the rotation of the IM. All of these can improve the simplicity and reliability of the whole blood pump system.

As presented in this chapter, the sketch design of the blood pump chamber determines the overall dimensions of the entire pump. The detailed structure and dimensions of the inductive torque driver have been designed. Then, mathematical modeling and analysis of the proposed motor-pump systems are carried out based on the general equivalent circuit model of squirrel-cage induction motor. The magnetizing energy and developed electromagnetic torque are analyzed using the magnetizing inductance equations. The proposed magnetic bearings are also analyzed and designed through equivalent magnetic charges method and magnetic circuit method, which showed that it is possible to constrain the undesired degrees of freedom of the rotor-impeller. All these give a more detailed and deeper understanding of the operation principles of the proposed concept. FEM analysis and simulations of the proposed magnetic bearings and radial-flux inductively-driven motor-pump device have been conducted to assess the feasibility of the entire concept.

Although the FEM simulations of the proposed motor-pump design and dimensions have shown that it is largely functional, its electromagnetic torque performance is insufficient. For a more practical design, the dimensions need to be re-sized and the geometry needs to be optimized, which is beyond the scope of this project. The analysis has also suggested that there is an inherent structural disadvantage in radial-flux motor when applied to flat geometries having a small ratio of active axial length to outer diameter. There is thus a need to study the axial-flux design, which will be presented in subsequent chapters.

Chapter Four

AXIAL-MAGNETIC-FLUX INDUCTIVE PROPULSION

4.1 Introduction

Although the radial-flux inductive torque driver design introduced in Chapter Three shows good potential for CBP application, its surface area utilization may not be the optimum. As the pump impeller has a disk-type geometry, it is intuitive to investigate the axial-flux version of induction motor as the torque driver, as it also has the disk-type geometry. This may offer a better match in terms of surface area utilization, thus providing a higher torque to volume ratio. This chapter thus presents the concept of axial-flux inductive propulsion for centrifugal blood pumps. The axial-flux rotor is to be integrated with the impeller so that there is no rotating part outside the blood chamber. The ultimate goal is to achieve a compact and simple shaft-less CBP design in the future. The current objective in this part of the project is to investigate the feasibility of axial-flux inductive propulsion from the angle of electromagnetism application.

The first part of this chapter presents the design philosophy, which serves as the general guide for the whole design process. The structure and dimensions of the blood pump chamber are similar to that of the radial-flux design introduced in Chapter Three. Based on these structure and dimensions, the conceptual designs of the axial-flux inductive torque driver were obtained. Two possible designs were proposed: the single-stator version and the dual-stator version. The designs were subsequently investigated and analyzed by equivalent circuit modeling and finite element method of simulation. The latter also served as a tool for the improvement of the designs.

4.2 Design philosophy and proposed structures

In pace with the application of new materials, innovation in manufacturing technology and improvements in cooling techniques, further increase in the power density (output power per mass or volume) of electrical machines has been made possible. However, there is an inherent limit to this increase for conventional radial-flux machines because of [116]:

- (1) The bottle-necks in the flux paths at certain parts of the core of the stator and/or rotor due to the radial curvature;
- (2) Parts of the stator or rotor core away from the air gap may not be fully utilized as magnetic circuits;
- (3) Heat removal from the stator windings can be poor as in the case of exterior rotor machines.

These limitations associated with radial-flux structures cannot be easily overcome unless a new topology is adopted. The axial-flux machine is one possible alternative. For some applications, it has certain properties which are considered advantageous compared to radial-flux machine, such as [116]:

- (1) Axial-flux machine can have much larger diameter-to-length ratio than radial-flux machine, which means a more compact structure;
- (2) It has a planar and somewhat adjustable air gap;
- (3) It can be designed to possess a higher power density with some saving in core material;
- (4) The larger the outer diameter of the core, the higher the number of poles that can be accommodated, making the axial-flux machine a suitable choice for high electrical frequency or low mechanical speed operations.

Because of these advantages, it was proposed that the axial-flux induction motor should be investigated for the torque driver application of CBP. Two possible axial-flux designs of shaft-less CBP systems have thus been proposed. Their design philosophy is based on the following guidelines:

- (1) In order to do away with the need for a motor shaft connecting the driving motor rotor to the pump impeller, the motor rotor shall be integrated with the pump impellor within the pump housing;
- (2) In order not to compromise the pump efficacy, the basic geometry of the pump impeller must be maintained. Pump impellers usually come in the form of semi-open

or close geometry, as shown in Fig 4-1. The design of the rotor shall complement the disk-type geometry of the impeller. The semi-open impeller can be integrated with a single rotor while the close impeller can be integrated with a dual-rotor structure;

- (3) The stator or stators shall be located outside the pump housing. The surface area of the stator pole shoes may form part of the pump housing walls. The stator windings shall be of the three-phase type to produce smooth torque;
- (4) Traditional induction motors work well with small air gap between the rotor and the stator. The rotor-stator gaps are now occupied by the blood fluid and pump housing wall and therefore may be significantly larger than conventional dimensions. Motor torque and power production are expected to be worse off and this aspect should therefore be investigated;
- (5) The impeller shall be supported or balanced by one or a combination of more than one of the following:
 - (i) mechanical bearings within the pump housing (difficult and not desirable);
 - (ii) hydrodynamic forces developed during pump operation;
 - (iii) passive magnetic bearings;
 - (iv) active magnetic bearings;
 - (v) magnetic forces developed between the rotor and stator across the pump housing wall.

(a) Semi-open configuration

(b) Close configuration

Fig 4- 1: Types of pump impeller

4.2.1 Axial-flux single-stator design

(stator and rotor cores not shown)

Fig 4- 2: Axial-flux single-stator CBP concept

The concept of axial-flux single-stator inductively-driven CBP is illustrated in Fig 4-2. Here, the impeller shroud is combined with a disk-shape rotor core made of ferromagnetic material such as silicon steel, amorphous iron ribbon or ferrite. Within the rotor core are a number of copper spokes joined at their ends by two circular copper rings. These form the rotor windings. The axially oriented stator windings and stator core are located outside the pump housing. The rotor and stator cores are not shown in Fig 4-2 for clarity. The rotor and impeller unit inside the pump chamber can be supported by hydraulic bearing or pivot bearing. Theoretically, it is also possible to operate this pump without any mechanical bearings, by relying on the balance of the axial thrust force on the impeller created by the low pressure system at the inlet port, and the magnetic force between the rotor and the stator. Using the theory of direct-quadrature (d-q) axes transformation of a three-phase system, it can be shown that the speed of rotation is a function of the quadrature-axis current. By controlling the direct-axis and quadrature-axis currents, it is possible to control both the rotation torque and axial thrust independently. Thus this axial-flux CBP has an inherent magnetic self-bearing feature without requiring additional magnetic bearing component.

4.2.2 Axial-flux dual-stator design

Fig 4- 3: Axial-flux dual-stator CBP concept

This concept is suitable for the closed-type impeller shown in Fig 4-3, where both impeller shrouds are integrated with disk rotors. Both single suction inlet and double suction inlet may be used. Double suction has the advantage that the axial thrusts are present on both sides of the impeller and can balance out, removing the need for any thrust bearings. Any unbalance in the thrust forces can be controlled by independent control of the stator currents in each of the stators. This dual-stator design can be regarded as two single-stator designs combined together, which is therefore expected to produce more electromagnetic output torque per unit volume.

4.3 Sketch design of axial-flux inductively-driven CBP

The structure and dimensions of the blood pump chamber will be similar to that of the radial-flux design introduced in Chapter Three and are therefore not repeated here.

4.3.1 Design specifications

The designed specifications for a sketch design of the axial-flux driver are similar to those for radial-flux driver, which are listed below. As explained in Chapter Three, these specifications arise from practical considerations of what a typical CBP would possess. The lack of any prior experience for the unconventional design meant that the initial sketch design may be far from optimized and may have low and impractical efficiency. However the objective here is to investigate the working principle of the proposed concept.

- (1) Rated input power: $P_n = 10W$
- (2) Rated input line voltage: $V = 10\text{ V}$
- (3) Rated power frequency: $f = 50Hz$
- (4) Number of phases: $m = 3$
- (5) Power factor at rated load: $\cos \varphi = 0.65$
- (6) Rated rotation speed: $n = 2000 - 3000rpm$
- (7) Poles pair number: $p = \frac{60f}{n} = 1$

4.3.2 Stator design

As the stator has a simple disk-type geometry as shown in Fig 4-5, it is convenient to adopt the concentrated winding structure as shown in Fig 4-8. The main dimensions of the stator can be determined by a process similar to that presented in Chapter Three for the radial-flux design.

Suppose the input power to a star-connected three-phase induction motor is

$$P_n = \sqrt{3}VI \cos \varphi \quad (4.1)$$

where I is the input line current which is also the phase current. The input line voltage V would be related to the air-gap magnetic flux by

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$$V = 4\sqrt{3}fK_w N\phi \quad (4.2)$$

where K_w is the winding factor, which is typically around 0.4 to 0.6, N is the number of winding turns per phase and ϕ is the magnetic flux per pole. The latter can be further expressed in terms of the axial-flux stator dimensions as

$$\phi = B_g \frac{\pi[(D_1/2)^2 - (D_2/2)^2]}{2p} \quad (4.3)$$

where B_g is the average flux density in the air gap, typically would be around 0.3 to 0.6T, D_1 is the outer diameter of the stator and D_2 is its inner diameter.

Therefore,

$$V = 2\sqrt{3}fK_w NB_g \frac{\pi[(D_1/2)^2 - (D_2/2)^2]}{p} \quad (4.4)$$

The specific loading q , which is defined as the ampere-conductors per meter of air-gap periphery, is equivalent to the current density at the average radius of the stator. For the axial-flux stator, this is expressed as

$$q = \frac{3NI}{\pi(D_1 + D_2)/4} \quad (4.5)$$

Then, the input phase current I can be expressed as

$$I = \frac{q\pi(D_1 + D_2)}{12N} \quad (4.6)$$

By substituting equations (4.4) and (4.6) into (4.1), the relation between motor dimensions and input power can be derived as

$$D_1 = \sqrt[3]{\frac{8P_n \cdot p(1 - k_d^2)(1 + k_d)}{\pi^2 f K_w B_g q \cos \varphi}} \quad (4.7)$$

where $k_d = D_2 / D_1$ is the ratio of the stator inner diameter to its outer diameter, which is typically chosen to be 0.25 in axial-flux machine to produce enough electromagnetic torque. Then the outer and inner diameters of the proposed axial-flux motor driver are obtained as $D_1 = 58mm$ and $D_2 = 16mm$ respectively.

After determining the main dimensions of the stator, the stator winding and slot dimensions can then be systematically designed. From equation (4.4), the number of winding turns per phase can be expressed as:

$$N = \frac{V_p}{2\sqrt{3}fK_w B_g \pi [(D_1/2)^2 - (D_2/2)^2]} \quad (4.8)$$

For the single-stator design, the required number of turns is $N = 140$.

The rated stator winding current can be calculated as $I_1 = P_n / (\sqrt{3}V \cos \varphi) = 0.9A$. If the current density for the stator winding is not to significantly exceed 4 A/mm^2 , then the cross-sectional area of the stator winding conductor needs to be at least 0.22 mm^2 . So, the diameter of conductor is $d_1 = 2\sqrt{A_1/\pi} \approx 0.51 \text{ mm}$. Then from the standard copper wire data, the AWG-24 gauge is chosen, which for light insulation has a diameter of 0.544 mm . The stator windings around one stator tooth are 70 turns and they can be arranged as shown in Fig 4-4.

The stator slot width along the radial direction of the stator is constant, which means that the width of stator teeth would change linearly along radial direction. So, the stator slots and teeth dimensions at both inner and outer radial surfaces of the stator are enough to define the geometries of the stator slots and teeth.

Fig 4- 4: Stator winding coil arrangement

At the stator inner radial surface, with the windings arrangement in Fig 4-4, the width and depth of the slot are calculated as follows:

Width of the slot:

- (1) Horizontal space occupied by the conductors, $14 \times 0.544 \text{ mm} = 7.62 \text{ mm}$
- (2) Space between two sets of coils = 1 mm
- (3) Slot lining, $2 \times 0.5 \text{ mm} = 1 \text{ mm}$
- (4) Slackness and roughness, 0.55 mm

The total width of the slot is thus chosen to be 10 mm.

Depth of the slot:

- (1) Vertical space occupied by the conductors, $10 \times 0.544 \text{ mm} = 5.44 \text{ mm}$
- (2) Slot lining, $2 \times 0.5 \text{ mm} = 1 \text{ mm}$
- (3) Slackness and roughness, 0.55 mm

The total depth of the slot is thus designed to be 7 mm. Thus, the area of the slot is $10 \times 7 \text{ mm}^2$.

So, according to the analysis, design and calculation for the stator of the proposed axial-flux inductively-driven CBP, the obtained dimensions of the stator core are illustrated in Fig 4-5.

Fig 4- 5: Dimensions of stator core of axial-flux inductively-driven CBP

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The width of each stator tooth and slot at inner radial surface is $\frac{2\pi R_1}{6} \approx 15.7\text{mm}$. So, the width of the stator tooth alone is $15.7 - 10 = 5.7\text{mm}$. The geometry of the stator inner radial surface is shown in Fig 4-6.

Fig 4- 6: Stator inner radial surface dimensions

At the stator outer radial surface, the width of the each stator tooth and slot is $\frac{2\pi R_2}{6} \approx 31.4\text{mm}$. The width of the stator slot at outer surface is same as that at inner surface, which is 10mm. Then the width of the stator tooth at outer surface is $31.4\text{mm} - 10\text{mm} = 21.4\text{mm}$. The thickness of the stator outer core and tooth should be the same as the inner side. The geometry of the stator outer radial surface is shown in the following Fig 4-7.

Fig 4- 7: Stator outer radial surface dimensions

The approximate length for one coil of stator winding is given by the following formula:

$$L = (\pi D_1 / 6 + \pi D_2 / 6) + (D_1 - D_2) \approx 7 \text{ cm} \quad (4.9)$$

Then, the total length of an entire stator winding is NL , which is 9.8m. The total resistance of stator winding is $R_{\text{statorwinding}} = \rho \frac{NL}{\pi(d_1/2)^2}$, which is 0.65Ω .

Fig 4- 8: Stator winding configuration of axial-flux CBP

4.3.3 Air gap of the axial-flux CBP

The air gap length in an induction motor is an important issue which can significantly affect the overall performance. The MMF required to maintain the flux in the air gap is proportional to the product of the flux density and the air gap length. Even with very small flux densities, the MMF required for the air gap is much more than that for the rest of the magnetic circuit. Therefore, it is the length of the air gap that primarily determines the magnetizing current drawn by the motor. With bigger air gap, the larger magnetizing current drawn will cause bigger phase angle difference between the stator voltage and the stator current, resulting in a smaller power factor. Therefore the air gap length in an induction motor should be as small as mechanically possible in order to reduce the magnetizing current and to improve the power factor.

For the proposed axial-flux CBP, the air gap must be big enough to accommodate the pump wall thickness and to leave sufficient fluid gap to minimize hemolysis. Hence the air gap length may be as large as 3 mm depending on the material for the pump wall, and low power factor and high magnetizing current may be expected. However, there are positive qualities such as less pulsation loss, less eccentricity impact on performance, better cooling

and smaller noise caused by zigzag leakage flux which should also be noted [114]. It is proposed here that air gap of 2mm to 4mm be investigated for the sketch design.

4.3.4 Rotor of the axial-flux CBP

The rotor of the proposed axial-flux CBP has a disk-type structure whose radial cross-sectional view is shown in Fig 4-10. It consists of a magnetically permeable core with an electrically conducting rotor winding in the form of a “wheel”, as shown in Fig 4-9. The rotor wheel is embedded on one side of the core. The other side of the core is to be integrated with the impeller structure. The core material may be made from amorphous iron or ferrites. The rotor wheel may be made from copper or aluminum.

Though the axial-flux rotor wheel may look very different from the radial-flux rotor squirrel-cage, they are actually similar in terms of electromagnetic and electromechanical characteristics. Hence the selection of number of rotor bars would follow a similar process as described in Section 3.4.4. Using this process, the possible number of rotor bars is $S_r = 11, 13, 14, 15$. For simplicity, $S_r = 11$ is selected for the sketch design.

The rms current per rotor bar can be calculated as

$$I_2 = \frac{K_{w1} \times S_s \times Z_1}{K_{w2} \times S_r \times Z_2} I_1' \quad (4.10)$$

where $K_{w1} = 0.6$ (assumed) is the winding factor for the stator winding, $K_{w2} = 1$ is the winding factor for the rotor winding, $S_s = 6$ is number of stator slots, $S_r = 11$ is the number of rotor slots, $Z_1 = 70$ is number of conductors per stator slot and $Z_2 = 1$ is the number of conductors per rotor slot.

Substituting these values, the current per rotor bar is obtained as $I_2 = 17.33A$. The cross-sectional area of the rotor bar is $a_2 = I_2 / \delta_2$ where δ_2 , the current density of the rotor bar, may be taken as between 4 to 7 A/mm^2 . So, the cross-sectional area of rotor bar is chosen as $a_2 = 4.33mm^2$. Accordingly, the diameter of the rotor bar can be obtained as $D_b = 2.35mm$. The length of rotor bar L_b , is chosen as $(D_1/2 - D_2/2)$, which is 21mm.

The end ring current can be determined as, $I_e = WI_2 / \pi$, where W is the number of rotor bars per pole, which is $W = S_r / p = 5.5$. So, I_e can be calculated as 30.36A. Thus the cross-sectional area of rotor end rings, $a_e = I_e / \delta_e$, is designed as 7.6 mm^2 . In order to provide enough space for the end ring to accommodate the rotor bars, the width and height of the end ring are chosen as 2mm and 5mm respectively.

From the analysis, design and calculation, the designed rotor structure dimensions can be summarized in Table 4-1, all of which are also shown in Fig 4-9.

Fig 4- 9: Dimensions and structure of rotor wheel for axial-flux inductively-driven CBP

Table 4- 1: Dimensions of rotor bars and end rings

Item	Value	Item	Value
Outer diameter of outer end ring (D_{o1})	60mm	Inner diameter of outer end ring (D_{o2})	56mm
Outer diameter of inner end ring (D_{in1})	16mm	Inner diameter of inner end ring (D_{in2})	12mm
Rotor bar diameter (d_b)	2.35mm	End ring height (h_e)	5mm

The resistance of one rotor bar is given by

$$R_{bar} = \rho(D_{o2} - D_{in1}) / S_{bar} = \rho(D_{o2} - D_{in1}) / \pi(d_b / 2)^2$$

The mean length of outer end ring is $C_1 = \pi(D_{o1} - D_{o2}) / 2$ and the mean length of inner end ring is $C_2 = \pi(D_{in1} - D_{in2}) / 2$. Then, the resistance of the inner and outer end rings can be obtained by $R_{outerring} = \rho C_1 / [(D_{o1} - D_{o2}) \times h_e]$ and $R_{innerring} = \rho C_2 / [(D_{in1} - D_{in2}) \times h_e]$. The resistance of ending ring can be converted to the bar by $R'_{outerring} = (R_{outerring} / G) / 4 \sin^2(\pi p / G)$ and $R'_{innerring} = (R_{innerring} / G) / 4 \sin^2(\pi p / G)$ where G is the number of rotor bar. Then, we can convert the rotor resistance to the stator side by $R'_2 = 4m(k_w W)^2 (R_{bar} + R'_{innerring} + R'_{outerring}) / G$. By substituting the dimension values listed in Table 4-1, the rotor resistance converted to the stator side is calculated to be $R'_2 \approx 7.86\Omega$ for this design.

Fig 4- 10: Radial sectional view of rotor through rotor bars

Depending on the dimensions of rotor bars and end rings, the structure and dimensions of the rotor core can be decided accordingly. The cross-sectional view of designed end rings, rotor bars, and rotor core is shown in Fig 4-10. The overall diameter of rotor core is 65mm and the total height is 8mm.

4.3.5 Magnetizing inductance

Similar to the inductance in radial flux induction motor, one phase winding can be characterized by its magnetic inductance, L_o , and its leakage inductance, L_σ . Let the concentrated windings of one stator phase be fed by constant current I , assuming that all remaining stator phases are open. The magnitude of the first harmonic of the magnetomotive force, $\hat{F}$, would then be

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$$\hat{F} = \frac{2k_w N}{\pi p} I \quad (4.11)$$

in which $k_w N$ is the number of turns per phase of the concentrated winding and p the number of pair poles. For the stator winding distribution in this design depicted as Fig 4-8, the pitch factor $k_p = \sin(\pi/6) = 1/2$, the distribution factor $k_d = 1$, so the winding factor $k_w = k_p \times k_d = 1/2$.

The peak value of the flux-density within the air gap is calculated as

$$\hat{B} = \mu_0 \frac{\hat{F}}{2\delta} = \frac{\mu_0 k_w N}{\pi p \delta} I \quad (4.12)$$

In which δ is the air gap length. The average flux-density per pole is obtained as

$$B_{av} = \frac{2}{\pi} \hat{B} = \frac{2\mu_0 k_w N}{\pi^2 p \delta} I \quad (4.13)$$

The principle flux per pole is

$$\phi = B_{av} \frac{\pi(r_1^2 - r_2^2)}{2p} \quad (4.14)$$

in which r_1 and r_2 are respectively the stator outer radius and inner radius.

The principle flux linkage is thus

$$\psi = k_w N \phi \quad (4.15)$$

Finally, the magnetic inductance per phase is obtained as

$$L_0 = \frac{\psi}{I} = \frac{2\mu_0 (k_w N)^2 (r_1^2 - r_2^2)}{\pi p^2 \delta} \quad (4.16)$$

Compared with the derived magnetic inductance for radial flux design in equation (3.35), the magnetic inductance for axial-flux design depends on the stator outer and inner radius, which is different from the radial flux design.

4.4 FEM analyses and simulations

This section presents the results of investigation by finite element method into the viability of the proposed inductively-driven axial-flux CBPs. Both the single-stator and dual-stator versions have been analyzed by three-dimensional time-stepping FEM simulations using the Maxwell 3D software. Simulation techniques and difficulties have described in Chapter Three and are not repeated here.

4.4.1 Single-stator design

The FEM simulation model of the single-stator design is shown in Fig 4-11. In order to show the structure clearly, the rotor core is not included in this diagram although it was present in the simulation. The copper rotor wheel is depicted as dark brown color. The six wedge-shaped rings in red color are the six stator coils wound around the six teeth of the stator. The stator teeth and yoke are shown in blue color. The 3-phase sinusoidal voltage sources are supplied to the stator coils to produce a rotating magnetic field in the air gap between rotor and stator. This rotating magnetic field interacts with the rotor to produce electromagnetic torque on it.

Fig 4- 11: Simulation model of axial-flux single-stator induction motor

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Fig 4-12 shows the flux distribution in the air gap. The two lighter color regions represent the instantaneous positions of the two poles. Figs 4-13 and 4-14 shows the transient simulation results obtained with 50-Hz stator voltage supply under no applied load and loaded conditions respectively. It is obvious that the flux linkage, induced voltage and phase current all are quasi-sinusoidal. The transient behavior of the developed electromagnetic torque on the impeller and its rotational speed is therefore consistent with the expected transient response of a conventional induction motor. The rotor speed differs from the synchronous speed because of damping losses due to eddy currents in stator and rotor cores, which have yet to be optimized to minimize the eddy currents. The rated torque of the design is 0.01Nm. The applied load torque in Fig 4-14 is 0.009Nm, or 90% of the rated torque.

Fig 4- 12: Simulated flux distribution in the air gap

(a) Phase current

(b) Flux linkage

Fig 4- 13: FEM simulation results for axial-flux single-stator design with 50Hz sinusoidal voltage supply under no-load condition (air gap length=2mm)

In the blood pump application, the load can be simulated as consisting of two major parts: viscous friction (T_v) and windage (T_w). Viscous friction is the component of friction proportional to speed.

$$T_v = K_1 \omega_{rotor} \quad (4. 17)$$

Windage is the component of friction proportional to square of the speed. This is mainly due to liquid movement arising from rotation. It is the dominant component in the blood pump application.

$$T_w = K_2 \omega_{rotor}^2 \quad (4. 18)$$

Fig 4- 14: FEM simulation results for axial-flux single-stator design with 50Hz sinusoidal voltage supply under loaded condition (air gap length=2mm)

In the loaded case, these two kinds of load torque should be included in the simulation setup. However, both K_1 and K_2 are related with the structure and geometry of the pump chamber as well as of the pump impeller. Computation of these two parameters involves detailed fluid dynamic knowledge which is out of the scope of this project. So, several arbitrary values are chosen as the load torque acting on the impeller during the loaded condition simulations. After the starting period, the electromagnetic torque acting on the rotor balances the load torque of the rotor, which keeps the rotation speed of rotor constant. The results in Fig 4-13 and Fig 4-14 indicate that the concept of this axial magnetic flux inductive propulsion for CBP is viable.

4.4.2 Dual-stator design

The simulation model of the axial-flux dual-stator design is shown in Fig 4-15. The rotor core integrated with the impeller is not shown here. This design can be regarded as two symmetrical axial-flux single-stator designs. So, the design of each stator and rotor is the same as those of single-stator design, which are introduced in Section 4.3.

Similar to the simulations of the single-stator design, Fig 4-16 shows the flux distribution in the two air gaps of the dual-stator design. Fig 4-17 and Fig 4-18 show the transient simulation results obtained using 3D time-stepping FEM analysis, both of which also

support the viability of this double-stator axial-flux design. The applied frequency is 50 Hz and the air-gap is 2mm. The applied load in Fig 4-18 results is 0.02 Nm.

Fig 4- 15: Simulation model of double-stator design

Fig 4- 16: Flux distribution in the two air gaps

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(c) Electromagnetic torque

(d) Impeller rotation speed

Fig 4- 17: FEM simulation results for axial-flux dual-stator design with 50Hz sinusoidal voltage supply under no-load condition (air gap length=2mm)

(a) Phase current

(b) Flux linkage

(c) Electromagnetic torque

(d) Impeller rotation speed

Fig 4- 18: FEM simulation results for axial-flux dual-stator design with 50Hz sinusoidal voltage supply under loaded condition (air gap length=2mm)

4.4.3 Verification of mathematical model using FEM Simulation

Though the inductively-driven CBP here is of the axial-flux type, it is expected that the equivalent circuit model and its mathematical equations presented in Chapter Three for the radial-flux design should still be applicable to certain extent. Hence it is necessary to investigate this aspect. In order to verify the feasibility and viability of the mathematical model, FEM analysis was applied to the axial-flux single-stator design and axial-flux dual-stator design. The parameters used in the FEM analysis simulation and mathematical model calculation are listed in Table 4-2.

Table 4- 2: Parameters in FEM simulation and mathematical model calculation

Phase winding turns N_s	280
Inertia moment of rotation J Single-sided design Double-sided design	$7 \times 10^{-6} \text{ kg} \cdot \text{m}^2$ $1.4 \times 10^{-5} \text{ kg} \cdot \text{m}^2$
Air gap length g Single-sided design Double-sided design	$2\text{mm}, 3\text{mm}, 4\text{mm}$ $2\text{mm}, 3\text{mm}, 4\text{mm}$
Stator phase winding resistance R_s	2Ω
Rotor phase resistance R_r	8Ω
Input phase voltage V	10V
Leakage inductance L_σ (from FEM) Single-sided design Double-sided design	$0.13\text{mL}(2\text{mm}), 0.17\text{mL}(3\text{mm}), 0.2\text{mL}(4\text{mm})$ $0.31\text{mL}(2\text{mm}), 0.36\text{mL}(3\text{mm}), 0.4\text{mL}(4\text{mm})$

A. Single-stator design

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(c) Air gap=4 mm

Fig 4- 19: Torque-Speed curves comparison between FEM analysis results and mathematical model calculations for single-stator design

(a) Air gap=2 mm

(b) Air gap=3 mm

(c) Air gap=4 mm

Fig 4- 20: Current-Speed curves comparison between FEM analysis results and mathematical model calculations for single-stator design

Figs 4-19 and 4-20 show the comparisons of the torque-speed and stator phase current-speed curves obtained from FEM analysis and from the mathematical model using different air gap values. The lines with “+” markers are the FEM analysis results and the lines without markers are the mathematical model calculations. The comparisons show the similarity of FEM analysis results and mathematical model calculations. The differences between them are caused by the variation of leakage inductances and rotor resistance

according to different load conditions. Also, Fig 4-19(a) indicates that the designed motor-pump device can produce around 1W of output power when it operates between 2000rpm and 2500rpm. Since the human cardiac power at rest is around 0.9-1.2W [118], this can be met by the proposed CBP even though this is an initial sketch design far from optimized.

B. Dual-stator design

Similar to the analysis of axial-flux single-stator design, Figs 4-21 and 4-22 shows the comparisons of the torque-speed and the stator phase current-speed curves obtained from FEM analysis and from using the derived mathematical model using different air gap values for the dual-stator design. The lines with “+” markers are the FEM analysis results and the lines without markers are the mathematical model calculations. The comparisons in Fig 4-21 and Fig 4-22 again show the strong similarity of FEM analysis results and mathematical model calculations for different air gap values, which further support the validity of the equivalent circuit model.

Fig 4- 21: Torque-Speed curves comparison between FEM analysis results and mathematical model calculations for dual-stator design

Fig 4- 22: Current-Speed curves comparison between FEM analysis results and mathematical model calculations for dual-stator design

From the simulated torque-speed curves in these two designs, it can be observed that the drivers have relatively large slips in the torque-speed range where they can output sufficient power to meet the blood pump power requirement. The main reason is due to the high rotor resistance in these designs, caused by the relatively small cross-sectional areas of rotor bars and ending rings in these initial first-cut designs. The large effective air-gap to accommodate the pump wall also makes the magnetizing inductance very small, causing the ratio between magnetizing inductance and rotor resistance to be much smaller than that in conventional motor design. These inductive drivers have linear torque-speed characteristics similar to that of the industrial NEMA D type of induction motors which also have typically high slip values at their rated working points. This large slip value brings both advantages and disadvantages to the application. The major advantages are the high starting torque and minimum risk of stalling. These features are very important for the reliability required of blood pump application. The main drawback is the lower power efficiency under the rated working condition. Hence there is need to improve the torque-speed characteristics by lowering the rotor resistance, as described in next Section 4.5.

4.5 Design enhancements

In order to improve the proposed axial-flux motor-pump device designs, especially to increase the electromagnetic torque which is very essential for blood pump application without compromising its compact size and structure, several studies has been implemented using FEM simulations and mathematical analyses on the stator and rotor structures of the axial-flux single-stator design. Because of their similarities, the results of these studies would also applicable to the axial-flux dual-stator design.

4.5.1 Stator structure

The electromagnetic torque produced by the proposed axial-flux single-stator design can be calculated using equation (3.52). From this equation, it can be seen that with the same rotor resistance, slip and synchronous speed, if the rotor current (reflected to stator side) can be increased, the output electromagnetic torque will also be increased accordingly. The general equivalent circuit shown in Fig 3-8 shows that if the magnetic inductance L_o increases, the impedance of magnetic inductance branch paralleled with the rotor branch will become bigger. Then, the back EMF of the design will also increase, which cause the rise of rotor current and accordingly, the output electromagnetic torque.

Fig 4- 23: Impact of stator inner radius on developed electromagnetic torque

In the magnetizing inductance in equation (4.16), the parameters as pole number, air gap length, stator winding factor and stator winding turns cannot be changed easily. At the

same time, the stator outer radius is also fixed according to the blood pump chamber design. So, the stator inner radius was changed step by step to study its impact on the produced output electromagnetic torque. This study was implemented using both FEM analysis and mathematical calculations.

Fig 4-23 shows the model calculated and FEM simulation results by keeping the stator outer radius as 29mm and decreasing the inner radius from 18mm to 8mm by 5mm steps. The torque-speed curves obtained from both mathematical model calculations and FEM simulations show that the decreasing of stator inner radius, which increases the active stator radial length, does not help much to improve the output electromagnetic torque. The reason is that when the stator active radial length is increased, the stator leakage inductance will also increase accordingly. Then, both the back EMF and rotor current will not increase very much to cause the increase of output electromagnetic torque. So, keeping the overall size and volume of stator fixed in this proposed design, the output electromagnetic torque will not be improved much by changing only the stator geometries.

4.5.2 Rotor structure

Fig 4- 24: Torque-Speed curves obtained by changing rotor resistance

The axial-flux single-stator design has relatively high starting torque, but the output electromagnetic torque in its operation speed range (2000rpm-2500rpm) decreases substantially, which is shown in the obtained torque-speed curves of Fig 4-21(a). This also means very low efficiency of the device during its operation. Judging from the parameters of this proposed design, this phenomenon is mainly caused by the high rotor resistance value. By only changing the rotor resistance (reflected to the stator side) values, a series of torque-speed curves of this designed device can be obtained using the equations of the equivalent circuit model, which are shown in Fig 4-24

From Fig 4-24, it can be seen that the output torque of the proposed design between 2000rpm and 2500rpm rotation speed can be increased by reducing the rotor resistance (reflected to stator side) value from 8Ω to 0.1Ω . Then, FEM analysis and simulations were carried out to study the effect of rotor resistance acting on the developed electromagnetic torque of the device. During the FEM analysis, only the cross sectional areas of rotor end rings and rotor bars are enlarged compared with their original design introduced in section 4.3. All the other geometries and structures are kept the same as their original design. The torque-speed curves obtained from FEM simulations of the new rotor geometries are compared with that of the original design, which are shown in Fig 4-25.

Fig 4- 25: Torque-Speed curves comparison of different rotors

The rotor-1 is the original rotor design with dimensions introduced in section 4.3.4. The cross-sectional area of rotor bars and end rings in rotor-2 and rotor-3 are respectively twice

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and four times those of the values in rotor-1. From the comparison of their torque-speed curves in Fig 4-25, it is clear that the decrease of rotor resistance value caused by the increase of the cross-sectional area of the rotor bars and end rings can improve the electromagnetic torque significantly. The output electromagnetic torque values between 2000rpm to 2500rpm of rotor 2 are almost twice those of rotor 1. This improvement is very important to the proposed design because the more electromagnetic torque produced means more head pressure can be produced by the blood pump. This would in turn enable the proposed blood pump design meet the requirement for more acute human body activities. The pump efficiency would also be increased due to lower rotor losses. However, the comparison between rotor-2 and rotor-3 shows that further increase of rotor bars' and end rings' cross-sectional areas cannot increase the output electromagnetic much. This is probably due to the skin effect of rotor bars and shorting rings. Although the physical cross-sectional areas are increased, the effective cross-section areas utilized for current conduction are not increased in proportion due to skin effect. This means that the actual rotor resistance cannot be decreased much further by simply increasing the physical cross-sectional area of the rotor bars and end rings. Hence the improvement in output electromagnetic torque will eventually saturate with the increase in physical cross-sectional areas of the rotor.

4.6 Concluding remarks

In an effort to get around the inherent conflict between producing a compact structure and providing enough electromagnetic torque from the radial-flux type of CBP introduced in Chapter Three, it was decided that the axial-flux type of inductively-driven CBP should be investigated as well. Two designs of axial-flux type of CBP have been proposed and are presented in this chapter. Compared with the first two generations of blood pumps, these new third-generation CBP designs have various attractive features where the torque transmission method, the structure simplicity and compactness and the system reliability are all concerned. Compared with the radial-flux design in Chapter Three, they have the potential to produce more electromagnetic torque without compromising the compactness.

The axial-flux designs share the same structure and main dimensions of the blood pump chamber with the radial-flux design of Chapter Three. Using the structure and dimensions of the blood pump chamber, the detailed structure and dimensions of the proposed axial-flux inductively-driven CBP were designed. The two designs were termed as the single-stator design and the dual-stator design.

In order to test the feasibility of the proposed axial-flux designs and also the validity of known equivalent circuit model for such designs, the proposed blood pump designs were analyzed using time-stepping FEM simulations under both no-load and loaded conditions. The torque-speed curves and stator-current-speed curves for different air gap values were obtained and compared. The transient FEM simulation results supported the feasibility of both the single-stator and the dual-stator designs. The FEM analysis results also verified the validity of the equivalent circuit model for axial-flux designs. The equivalent circuit model and its mathematical equations can thus be used as the basis for the study of system control of such motor-pumps in future. In addition, the study of design improvement of both the stator and the rotor has been conducted, again using mathematical model analysis and FEM simulations. In this study, it was shown that there are various design options available to increase the developed output electromagnetic torque.

Chapter Five

OPEN-IMPELLER INDUCTIVE PROPULSION

5.1 Introduction

A new concept of three-phase inductively-driven shaft-less CBP based on the open-impeller, shown in Fig 5-1, is introduced in this chapter. This is an extension of the axial flux inductively-driven CBP concept introduced in Chapter Four. Instead of having one or two axial flux induction motor rotor integrated with the pump impeller, the idea here was to explore the possibility of the impeller itself functioning as the inductive rotor wheel. This may yield a very compact design for small shaft-less pump application.

Fig 5- 1: Structure of open type of pump impeller

The philosophy behind this concept is explained in this chapter. The structure and electromagnetic design of the proposed system is introduced along with its mathematical model and parameter estimation. Both circuit model analysis and FEM analysis have been used to investigate the feasibility and potential of the design. FEM analysis and simulations have also been used to optimize the rotor structure of this open-impeller. The optimum structure of the rotor-impeller in terms of the relationship of ripple torque and average torque with both rotor bar number and rotor bar skew angles is presented. An experimental test-rig has also been fabricated to further study certain aspects of the design. Experimental results have been obtained and are presented to support the proposed concept.

5.2 Concept and sketch design

Fig 5- 2: Open-impeller type of centrifugal blood pump design

Fig 5- 3: Electromagnetic relationship of proposed open-impeller CBP

The open-impeller CBP shares the same design philosophy of the axial flux type of CBP introduced in Chapter Four, which is therefore not repeated here. According to the design philosophy, an axial flux open-impeller inductively driven CBP concept is proposed and its structure is shown in Fig 5-2. In order to show the structure clearly, Fig 5-2 is not drawn to scale and the dimensions of the air gap are drawn exaggerated. Also for this purpose, certain components such as stator cores are not shown. The impeller of the proposed device consists

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of a number of spiral vanes made from electrically conducting material such as copper. The vanes are not webbed by a shroud but instead are connected by an inner end ring and an outer end ring, both made from the same material as the vanes. This design utilizes the spiral vanes of the impeller as the rotor induction bars. This results in a very light impeller. However in order not to affect impeller efficiency much, special attention should be paid to the shape and geometries of the impeller. The stator magnetic flux passes across the pump chamber from one stator to the other. Fig 5-3 shows the electromagnetic relationship model of the pump. In general, a three-phase inverter, normally working under PWM condition, shall drive the two sets of three-phase windings located outside the pump chamber.

In operation, the impeller is balanced by the developed axial thrust or constrained by suitable bearing systems within the chamber. The pump chamber has dual inlets at the centre of the impeller and an outlet near to the rim. Six copper windings arranged as double-sided three-phase systems located outside the pump chamber are fed with three-phase ac voltages. These supplied voltages produce three-phase ac currents in the copper windings, which are 120 electrical degrees apart in phase relationship. The applied currents generate a rotating magnetic field in the pump chamber. The rotating magnetic field induces circulating currents in the short-circuited impeller vanes, which react with the magnetic field to produce an initial torque on the impeller. In effect, the impeller vanes with the end rings are a form of core-less rotor windings while the copper windings outside the pump chamber are the stator windings. The produced torque will cause the impeller to spin at increasing speed until steady-state conditions are met, where the electromagnetically induced torque balances the load torque on the impeller arising from the fluid dynamics.

To produce an initial sketch design, the blood chamber and the stator structure would be based on the axial-flux dual-stator design presented in Chapter Four. Hence the stator core and windings were designed with similar dimensions and geometries as those from Chapter Four and are summarized in Table 5-1 for convenience. As the objective of this sketch design is to investigate its feasibility in electromagnetic torque production rather than the effectiveness of the rotor-impeller in pushing the liquid to be pumped, the impeller vanes shall simply be straight bars rather spiral blades. Therefore the rotor wheel design presented in Chapter Four shall form the basis of the open impeller-rotor, without any rotor core. The dimensions of the rotor-impeller are summarized in Table 5-2. Fig 5-4 shows the structure of the stator core, windings and the open rotor-impeller.

Table 5- 1: Dimensions of stator core and windings

Stator inner radius	8mm
Stator outer radius	30mm
Number of slots	6
Slot width	6mm
Slot depth	12mm
Stator yoke thickness	4mm
No. of winding turns per phase	280
Diameter of winding	0.5mm

Table 5- 2: Dimensions of open-impeller rotor

Inner shorting ring diameter	5mm
Inner shorting ring thickness	2mm
Outer shorting ring diameter	32mm
Outer shorting ring thickness	2mm
Rotor shorting ring height	2mm
Rotor bar height	2mm
Rotor bar width	1mm

(a) 3-D view of one of the dual stator cores

(b) Arrangement of stator windings

(c) Structure of open-impeller rotor

Fig 5- 4: Sketch design of open-impeller CBP

5.3 Modeling and analysis

5.3.1 Equivalent circuit model

The device described in the previous section is basically a form of axial-flux induction motor where the axial rotor is coreless. It is therefore expected that the general per phase equivalent circuit model will still be applicable for analysis, simulation and control. However, the determination of the parameters of this equivalent circuit model would be challenging, as the very large effective air gap of the coreless rotor implies very large leakage fluxes. Some of the parameters can be determined from physical dimensions, while others have to be determined from FEM analysis.

5.3.2 Parameter estimation from physical dimensions

In this analysis, the stator core is assumed to be made of infinitely magnetically-permeable material and there is no overlap between adjacent concentrated windings in the stator. Each phase winding can be characterized by its magnetizing inductance, L_p , and its leakage inductance, L_σ . If one phase of the proposed motor-pump system is fed by a constant current I , while the remaining stator phases are open-circuited, then the magnitude of the fundamental component of the magneto-motive force, $\hat{F}$, would be

$$\hat{F} = \frac{2k_w N}{\pi p} I \quad (5.1)$$

in which p is the number of pole-pairs and N is the number of turns per phase for each stator winding (which consists of four stator coils in this design). For the concentrated stator winding distribution in this design as depicted in Fig 5-4, the pitch factor $k_p = \sin(\pi/6) = 1/2$, the distribution factor $k_d = 1$, so the winding factor $k_w = k_p \times k_d = 1/2$. The peak value of the flux-density within the air gap is calculated as

$$\hat{B} = \mu_0 \frac{\hat{F}}{2\delta} = \frac{\mu_0 k_w N}{\pi p \delta} I \quad (5.2)$$

in which δ is the total effective air gap length. The average flux-density per pole is obtained as

$$B_{av} = \frac{2}{\pi} \hat{B} = \frac{2\mu_0 k_w N}{\pi^2 p \delta} I \quad (5.3)$$

The magnetizing flux per pole is

$$\phi = B_{av} \frac{\pi(r_1^2 - r_2^2)}{2p} \quad (5.4)$$

in which r_1 and r_2 are respectively the stator outer radius and inner radius. The magnetizing flux linkage is thus

$$\psi = k_w N \phi \quad (5.5)$$

Finally, the magnetizing inductance per phase is obtained as

$$L_p = \frac{\psi}{I} = \frac{\mu_0 (k_w N)^2 (r_1^2 - r_2^2)}{\pi p^2 \delta} \quad (5.6)$$

The per phase resistance of stator winding is given by

$$R_1 = \rho \frac{NL}{\pi(d_1/2)^2} \quad (5.7)$$

where L is the mean length of one turn of stator winding coil, d_1 is the diameter of stator wire cross-section and ρ is the resistivity of stator winding material.

The per phase rotor winding resistance transformed to stator side, R_2' can be calculated as:

$$R_2 = \frac{12(k_w N)^2}{S_r} (R_{bar} + \frac{R_{outerring} / S_r}{4 \sin^2 \frac{\pi p}{S_r}} + \frac{R_{innerring} / S_r}{4 \sin^2 \frac{\pi p}{S_r}}) \quad (5.8)$$

where R_{bar} , $R_{outerring}$ and $R_{innerring}$ are resistances of rotor bar, outer shorting ring of rotor and inner shorting ring of rotor respectively. S_r is the number of rotor bars.

Because of the large effective air gap, it is difficult to determine the stator and rotor leakage inductances from the physical dimensions. The empirical formulae published in literature are only valid for conventional radial-flux induction motors with small air gaps. Hence it is necessary to determine the leakage inductances using 3-D finite element analysis, as presented in next section.

5.3.3 Estimation of leakage inductances using FEM

To estimate the leakage inductances, three-dimensional FEM simulation is required. The technique is illustrated in Fig 5-5. The 3-D geometries of the designed stator and rotor of Fig 5-4 are created in the Maxwell simulator. Each phase winding consists of 4 coils. One phase, say phase-*A*, is chosen and simulated with a constant current *I*. Because of the leakage flux, the flux density in coil-1, B_{avg1} , is greater than the flux density in the rotor-impeller, B_{avg2} . By carefully measuring the simulated fluxes in coil-1 and in the rotor-impeller, the difference in their values can be computed, which is the leakage flux, $\psi_{\sigma1}$. The stator leakage inductance of the phase winding, $L_{\sigma1}$, can be computed from:

$$L_{\sigma1} = \frac{N\psi_{\sigma1}}{I} \tag{5.9}$$

where *N* is the total number of turns in the phase winding. One advantage of this technique is that the variance of leakage inductances with phase current can be obtained by repeating the computations with different values of excitation *I*.

Fig 5- 5: Estimation of leakage inductance from FEM-computed fluxes

Table 5- 3: Parameters of equivalent circuit model for air gap *g* = 4mm

Magnetizing inductance L_p	1.64mH
Stator leakage inductance $L_{\sigma1}$	0.37mH
Stator winding resistance R_1	1.67Ω
Rotor leakage inductance $L_{\sigma2}'$ (reflected to stator side)	0.37mH
Rotor winding resistance R_2' (reflected to stator side)	7.86Ω

The rotor leakage inductance transformed to the stator side, $L_{\sigma2}'$, can be assumed to be equal to $L_{\sigma1}$. By setting the physical air gap to values of 4mm, 5mm and 6mm, the leakage inductances were found to be 0.37mH, 0.40mH and 0.46mH respectively. Using the physical dimensions listed in Table 5-1 and Table 5-2, the other parameters for the per phase

equivalent circuit model are calculated and the complete set of nominal parameters for physical air gap $g = 4\text{mm}$ are listed in Table 5-3.

5.3.4 Parametric analysis using FEM

Table 5- 4: FEM simulation conditions

Number of turns per phase of stator winding N	280
Stator phase voltage (rms value) V_1	10V
Rotor-impeller moment of inertia J	$5 \times 10^{-6} \text{ kg} \cdot \text{m}^2$
Air gap length δ	4mm

To analyze the variation of model parameters of the proposed CBP with different loads, a series of parametric FEM simulations were run with the conditions given in Table 5-4. Fig 5-6 shows the magnified views of the rotor current distributions at the surfaces in rotor bar, inner end ring and outer end ring with 0.019Nm of applied load. It can be seen that the surface currents in the bar and end rings are not uniform due to skin effect. Hence rotor resistance is expected to vary with loading. Fig 5-7 shows the flux density distributions of stator with no applied load and with 0.019Nm of applied load respectively. The flux distribution of the stator concentrates at the surface when the load is applied. As a result, the certain parts of the stator core are highly saturated at 0.019Nm of applied load.

(a) Rotor-impeller

(b) Cross section A-A' (rotor bar)

(c) Cross section B-B' (inner end ring) (d) Cross section C-C' (outer end ring)

Fig 5- 6: Rotor current distributions with 0.019Nm of applied load

(a) Under no applied load

(b) Under 0.019Nm applied load

Fig 5- 7: Flux density distributions in stator

(a) Stator and rotor leakage inductances

(b) Magnetizing inductance

(c) Rotor resistance referred to stator side

Fig 5- 8: Variations of model parameters with load simulated by FEM analysis

The variations of equivalent circuit model parameters with load conditions are shown in Fig 5-8. The analysis using FEM can calculate the parameters at different load conditions while the physical model method provide only the nominal values, which does not include the magnetic saturation and skin effect. The magnetizing inductance decreases because of the magnetic saturation effect shown in Fig 5-7. On the other hand, the rotor resistance increases with load because of the skin effect as shown in Fig 5-6. The stator and the rotor leakage inductances also show load dependencies, which are caused by both the skin effect and the magnetic saturation.

5.3.5 FEM simulation of motor-pump performance

To further study the viability of the proposed open-impeller design, this novel motor-pump device was simulated with time-stepping FEM.

Fig 5- 9: Magnetic flux density distribution in air gap

By simulating the proposed device with 50Hz power supply and without applied load, Fig 5-9 shows the magnetic flux density in the air gap at a particular instant. There are two regions with very light shade in the air gap, which indicates higher magnetic flux density. These are the instantaneous magnetic pole positions of the rotating air gap flux.

The transient FEM simulation results with 50Hz power supply and effective air-gap of 4mm under no applied load condition are presented in Fig 5-10. It can be seen that the flux linkage, induced voltage and phase current all are quasi-sinusoidal. The transient behavior of the

developed electromagnetic torque on the impeller and its rotational speed are also consistent with the expected transient response of an induction motor without applied load.

Fig 5- 10: FEM simulation results under no applied load condition

In order to test the sketch design under loaded condition, different values of load torque have been applied to the impeller of the motor-pump device. The transient FEM simulation results under 0.01 Nm load are shown in Fig 5-11, which shows the tenability of the design under

loaded condition. The results in Fig 5-10 and Fig 5-11 indicate the concept of this motor-pump device to be viable.

Fig 5- 11: FEM simulation results under loaded condition

5.3.6 Torque-speed curves comparison

FEM simulation has also been used to verify and compare the accuracy of the equivalent circuit model parameters derived from physical model and from FEM analysis. According to the torque equation (3.52), the torque-speed curve has been obtained using the equivalent circuit with parameters derived from physical model. Similarly, the equivalent circuit with parameters derived from FEM analysis was also used to calculate the torque-speed curve using equation (3.52). Fig 5-12 shows the comparison of these two torque-speed curves with the result obtained from direct FEM simulation. It is clear from Fig 5-12 that the equivalent circuit model calculations correspond with the direct FEM simulation results very well. The equivalent circuit model with parameters derived through FEM analysis provide more closer results than those derived using parameters from physical model. The difference between them is mainly caused by the overestimation of rotor resistance R_2 by the physical model listed in Table 5-3. At the same time, Fig 5-12 indicates that the sketch design can produce around 1W output power when it operates between 2000rpm and 2500rpm. Hence the human cardiac power requirement at rest of around 0.9-1.2W [118] can just be met by the proposed sketch design. It also has relatively high slip at this operating point, due to non-optimized rotor structure dimensions, as discussed in Chapter Four. Optimization of rotor design is presented in the next Section 5.4.

Fig 5- 12: Torque-speed curves comparison

5.4 Rotor design optimization

As the inductively-driven open-impeller has low rotor-impeller inertia, it can be susceptible to torque ripple. Hence the optimization of the number of rotor bars and their skew angle to minimize the torque ripple is very important. Fig 5-13 shows definition of rotor bar skew angle

Fig 5- 13: Definition of skew angle

5.4.1 Rotor bar number and skew angle optimization

According to the torque ripple assessment in [108-110], if stator leakage inductance and resistance are neglected, at any instant t , the general instantaneous electromagnetic torque for an electrical machine can be expressed as

$$T_{em}(t) = \frac{\sum_{j=1}^m e_j(t) i_j(t)}{\omega_m} \quad (5.10)$$

where j is the order of machine phase, m is the number of machine phases, $e_j(t)$ and $i_j(t)$ are the back-EMF and current respectively in phase j , and ω_m is the rotor angular speed.

In the electromagnetic and ripple torque analyses, it is assumed that the motor is unsaturated, the armature reaction is negligible and the fundamental components of the currents and the corresponding back-EMF's are maintained in phase. For the Y-connected three-phase stator winding, the back-EMF in phase a can be written as

$$e_a = E_1 \sin \omega t + E_3 \sin 3\omega t + E_5 \sin 5\omega t + E_7 \sin 7\omega t + \dots \quad (5.11)$$

and the current in phase a can be written as

$$i_a = I_1 \sin \omega t + I_5 \sin 5\omega t + I_7 \sin 7\omega t + I_{11} \sin 11\omega t + \dots \quad (5.12)$$

where E_n is the n^{th} time harmonic peak value of EMF, which is produced by n^{th} space harmonic of the air-gap magnetic flux density B_{gn} , I_n is the n^{th} time harmonic peak value of current, which depends on current waveform. The product $e_a i_a$ consists of an average component and even-order harmonics for phase a

$$e_a i_a = P_0 + P_2 \cos 2\omega t + P_4 \cos 4\omega t + P_6 \cos 6\omega t + \dots \quad (5.13)$$

The instantaneous electromagnetic torque is produced by phase a , b , and c . That is,

$$T_{em}(t) = \frac{[e_a(t)i_a(t) + e_b(t)i_b(t) + e_c(t)i_c(t)]}{\omega_m} \quad (5.14)$$

Since the phase shifts between $e_a i_a$ and $e_b i_b$ and between $e_a i_a$ and $e_c i_c$ are $-2\pi/3$ and $2\pi/3$, respectively, the sum $(e_a i_a + e_b i_b + e_c i_c)$ will contain an average component and harmonics of order of six. The other harmonics are eliminated. Therefore, the instantaneous electromagnetic torque can be written as

$$T_{em}(t) = T_0 + \sum_{n=1}^{\infty} T_{6n} \cos n6\omega t \quad (5.15)$$

where T_0 is the average torque, and T_{6n} are the harmonic torque components. The torque-ripple factor (TRF) can be defined as the ratio of peak-to-peak ripple torque to average torque, which has the following form

$$TRF = \frac{T_{pp}}{T_0} \quad (5.16)$$

where T_0 is the average torque and T_{pp} is the peak-to-peak ripple torque which is given as

$$T_{pp} = 2\sqrt{T_6^2 + T_{12}^2 + T_{18}^2 + \dots} \quad (5.17)$$

In order to study the relationship between torque ripple and both the rotor bar number and rotor bar skewness, FEM simulations with different rotor bar number and different rotor bar skew angles were carried out. From the discussion in Chapter Three, it has been noted that the possible and practical rotor bar number should be chosen as 11, 13, 14 and 15. For each rotor bar number, FEM simulations were carried out with skew angles varying from 0-degree to 40-degree in 5 degree-steps.

Fig 5-14 and Fig 5-15 shows the average torque and TRF as a function of rotor bar number and skew angle under loaded condition respectively. The maximum average torque occurs at a skew angle of 35-degree. The TRF has a trend of decrease with the increase of rotor bar skew angle and the minimum value occurs at a skew angle of 35-degree. Also, the optimum rotor bar number appears to be 11 in terms of average torque produced. .

With the analysis and comparison of the 3D FEM simulation results in loaded conditions, the most optimized structure of rotor-impeller considering the relationship between motor ripple torque and average torque with rotor bar number and skew angles is 11 for rotor bar number and 35-degree for rotor bar skew angle.

Fig 5- 14: Average torque vs rotor bar number and skew angle

Fig 5- 15: TRF vs rotor bar number and skew angle

5.4.2 Rotor design improvement

In order to improve the proposed pump-motor design, especially to improve its ability to produce electromagnetic torque which is very essential for this blood pump application, without compromising to its compact size and structure, the impact of rotor resistance (reflected to stator side) on output electromagnetic torque has been studied through FEM simulations and mathematical analysis. By changing the rotor resistance (reflected to the stator side) values with a 4mm air gap length, a series of torque-speed curves of this sketch design can be obtained from equivalent circuit model calculations, which are shown in Fig 5-16. It is clear that the output torque of the proposed design between 2000rpm and 2500rpm rotation speed can be increased by reducing the rotor resistance value.

Fig 5- 16: Rotor resistance impact on output electromagnetic torque obtained from mathematical model

Fig 5- 17: Rotor resistance impact on output electromagnetic torque obtained from FEM analysis

In order to reduce the rotor bar resistance, two methods can be used: one is to change the material, while the other is to change the dimensions. For changing the material, it is difficult to improve in this aspect as copper is already a very good conductor. For changing the dimensions, increasing cross-section area can reduce the rotor bar resistance. However, this would eventually be limited by the size of rotor. Also, since the rotor bars are to function as the impeller blades in this open-impeller concept, the effects of the rotor bar size and geometry will impact on the fluid dynamics performance. Hence in the future phase of feasibility study, the study of rotor design has to be done in a holistic manner by considering electromagnetic performance, fluid dynamics and mechanical construction, all simultaneously.

In this current project, the study is only on electromagnetic aspect. Hence electromagnetic FEM analysis and simulations have been carried out to further assess the effect of reducing rotor bar resistance. In the FEM analysis, only the cross sectional areas of rotor shorting rings and rotor bars are enlarged to two times (rotor-2) and three times (rotor-3) of their original design (rotor-1). According to equation (5.8), the modified rotor resistance values should be half for rotor-2 and one-third for rotor-3, of the rotor resistance value in the original design (rotor-1). All the other geometries and structures are kept same as their original design. The torque-speed curves obtained from FEM simulations of the modified rotors are compared with that of the original design, as shown in Fig 5-17. The results show similar rotor resistance impact on output electromagnetic torque as Fig 5-16. The output electromagnetic torque of rotor-2 is almost twice of that of original design between 2000rpm and 2500rpm. This enables the output power of the device during operation to exceed 2W to meet the power requirement of heart pump application.

The differences between Fig 5-16 and Fig 5-17 are caused mainly by the skin effect of the rotor which makes the current density in rotor bars and shorting rings non-uniform. The mathematical model is based on a set of analytical equations which use lumped parameters and does not account for distributed effects. Hence skin effect of electric currents is one major effect neglected. Other effects excluded are the magnetic flux fringing, eddy currents and magnetic hysteresis behaviour. The FEM analysis is able to include all these effects and hence is more accurate. Hence the rotor resistance values obtained from FEM analysis are bigger than those calculated from mathematical model which does not account for the rotor skin effect. This effect has been illustrated in Fig 5-6.

5.5 Experimental investigations

As part of the project objective is to conduct the feasibility study using experiments, an instrumentation test rig was set up for this purpose. This section presents the implementation of the hardware platform for the investigation of axial-flux designs. The design of the test-rig was based on the intention of providing as much flexibility as possible to accommodate a wide range of experiments to test the feasibility of the proposed concepts.

Due to time constraints, only the open-impeller rotor was investigated. The obtained results was compared with FEM analysis results and mathematical model calculation results to verify the design and validate the mathematical model.

5.5.1 Test-rig construction

The components of this experimental test-rig are listed in Table 5-5. As shown in Fig 5-18, the two stators, #6 and #7, are fixed to their respective supports, #2 and #3. Each stator core is fabricated with silicon steel and has three-phase windings wound around its teeth. The windings are made up of 0.5mm insulated copper wire and the winding turn is 70 turns per tooth. The rotor, #8, is sandwiched in between these two stators. As this experiment was to investigate the electromagnetic induction phenomenon, the rotor is supported by a shaft and mechanical ball bearings. In an actual blood pump, the rotor-impeller would be supported by hydrodynamic and axial electromagnetic forces. The shaft with two ball bearings at each end goes through the centers of the two stators and their supports. The two ball bearings are supported by two rotor supports, #5 and #11. Two plastic rings are also used to insulate the ball bearings electrically from the rest of the rig. The two stator supports can slide on a sliding axis, #4, so that the air gap length of the motor can be adjusted. A micrometer screw gauge, #10, can change and measure the air gap length very precisely. The base, #9, gives solid support of the entire test-bed. For a clear view of this experimental set-up, detailed photographs of one of the two stators and the rotor-impeller are given in Figs 5-19 and 5-20 respectively.

Fig 5- 18: Photograph of the entire experimental test-rig

Table 5- 5: Components of experimental test-rig

Component number	Components
1	Power supply (Phase A B C and Neutral)
2	Stator support 1
3	Stator support 2
4	Sliding axis
5	Rotor support 1
6	Stator 1
7	Stator 2
8	Rotor
9	Base
10	Micrometer screw gauge
11	Rotor support 2 (not shown in Fig 5-17)

Fig 5- 19: Photograph of fabricated stator

Fig 5- 20: Photograph of fabricated rotor

5.5.2 Experimental measurements and data comparisons

A. With different frequency input

Fig 5- 21: Rotor speed with different frequency power supplies

During the experimentation, three phases of 10-volt rms phase voltage are applied to the stator windings. The phase voltages are produced by HP power source 6834B whose output can be adjusted in both the voltage magnitude and voltage frequency. Firstly, the phase voltage magnitude is kept at 10V and the voltage frequency is changed from 50 Hz to 200 Hz by 50-Hz frequency step. The air gap between the two stators is kept constant at 6mm. At every

power supply frequency, the motor-pump system rotation speed is measured and is compared with FEM simulation results as shown in Fig 5-21. The difference between the two sets of data is probably caused by the unknown windage and friction of the test-rig which is difficult to characterize.

B. With different air gap length

In order to validate the mathematical model with parameters derived from physical dimensions, experimental measurements have been obtained to compare with mathematical model calculations as well as FEM simulation results. As the equipment necessary to measure developed torque accurately from this set-up is not available, the stator phase currents and rotor speeds were measured instead to make a comparison with the FEM analysis results and mathematical model calculations, as shown in Fig 5-22.

(a) Air gap=4 mm

(b) Air gap=5 mm

Fig 5- 22: Stator Phase Current-Speed comparison among experimental measurements, FEM analysis results and mathematical model calculations

In the measurements, the power supply is kept constant at 10V and 50 Hz. The air gap distance is set to be 4mm, 5mm, and 6mm respectively. The comparisons show the similarity of experimental measurements, FEM analysis results and mathematical model calculations with different air gap values, which support both the feasibility of the design and the viability of the derived mathematical model.

5.6 Concluding remarks

A new concept of open-impeller centrifugal blood pump is proposed here, where the pump impeller is driven by torque produced from electromagnetic induction. The induction arises from a high speed rotating magnetic field generated by currents in multi-phase coils located outside the pump chamber, and controlled by a power electronic conversion circuit. Circulatory currents induced in the impeller vanes react with the magnetic field to produce the torque. The equivalent circuit model with parameters derived from physical model and FEM analysis has been obtained successfully. The theoretical and FEM analyses presented point to the feasibility of the proposed inductively driven open-impeller motor-pump designs.

Besides testing the feasibility of the motor-pump design under no-load and loaded conditions, FEM analysis and simulations are also used to study the optimization of the rotor structure of the proposed open-impeller design. The improved rotor design would be capable of producing more output electromagnetic torque. In addition, the optimization study also consider the relationship of motor ripple torque and average torque with rotor bar number and skew angles.

To further investigate its feasibility and the viability of the proposed design, an experimental setup was implemented. Several experiments were carried out and experimental data obtained were compared with FEM simulation results as well as calculations from the equivalent circuit model. The experiment data corresponded with both the FEM simulation results and mathematical model calculations, which prove the feasibility of the designed open-impeller motor-pump device and the validity of the mathematical equations derived from the circuit model and their FEM-generated parameters.

Chapter Six

CONCLUSIONS AND RECOMMENDATIONS

6.1 Conclusions

The feasibility of the novel concept of multi-phase inductively-driven shaft-less centrifugal blood pump, which utilizes induction machine theory and centrifugal pump principle to realize compact designs of shaft-less centrifugal blood pump, has been investigated. To achieve this goal, the following four tasks were addressed separately: (i) producing conceptual designs of integrated motor-pump device based on squirrel-cage induction motor and centrifugal pump; (ii) deriving mathematical models of the proposed motor-pump designs and their magnetic bearings; (iii) applying FEM analysis to the proposed designs to assess their feasibility; (iv) fabricating hardware prototype of at least one of the designs to study the feasibility further.

To achieve the **first task**, several novel motor-pump structures and magnetic bearing systems have been designed. Based on induction motor theory and centrifugal pump principle, four conceptual designs of centrifugal blood pump were proposed: one is based on radial magnetic flux induction motor and the other three are based on axial magnetic flux induction motor. All these four designs belong to the third-generation type of centrifugal blood pumps. Compared with the first two generations of blood pumps and even with other third-generation blood pumps, these new CBP designs have certain advantageous features where the torque transmission method, the structure simplicity and compactness and the system reliability are all concerned. Firstly, these four blood pump designs integrate the motor rotor with the impeller so that there is no rotating part outside the blood pump chamber. It means that the rotating part of the motor, the rotor, becomes part of the impeller, leaving the stator of the motor outside the pump chamber. In addition, the stator coils with perhaps some additional coils can also generate magnetic levitation forces on the rotor-impeller. Because of the rotor-impeller structure, the electromagnetic torque produced can act

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on the pump impeller directly. Then, there is no need of mechanical shaft for the transmission of torque, which can eliminate the problem of blood clotting on the shaft and sealing problem of blood pump chamber. Secondly, this rotor-impeller structure leads to a very compact and simple design of the blood pump, which is very suitable for the implantation application. Thirdly, variants of the squirrel-cage induction machine structure are applied to the propulsion of these four centrifugal blood pumps. The operation of induction machine inherently does not require information on the rotor position, unlike that of the PMDC motors used in other blood pump designs. At the same time, it is more robust and the extra commutation circuit and position sensors are not needed for its control. Furthermore, it is possible to apply the simple but reliable open-loop with slip compensation control strategy based on Voltage/Frequency method to control the speed of the blood pump. All of these will improve the simplicity and reliability of the whole blood pump system.

A magnetic bearing system has been designed for one of the proposed blood pump devices, which includes both passive and active magnetic bearings. The analysis of the degrees of motion freedom of the rotor-impeller structure provides the control objective for the proposed magnetic bearing system. After analyzing the structure and the working principle of both the passive and active magnetic bearings, the feasibility of the designed magnetic bearing to restrict undesired degrees of motion freedom of the rotor-impeller has been assessed positively.

For the **second task**, mathematical models for the proposed motor-pump designs and magnetic bearings have been derived. For the proposed motor-pump structures, the general equivalent circuit of induction machine can be adopted to perform the mathematical analysis. This model is thus adapted for the proposed inductive propulsion components of the blood pumps and the model parameters are expressed in terms of geometrical dimensions. Because of the different structures of radial magnetic flux and axial magnetic flux designs, the parameters are analyzed mathematically according to the corresponding geometries of the different designs. From the mathematical analysis, equations of the magnetizing inductance for the different designs were derived. Then, expressions for the magnetizing energy and the

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developed electromagnetic torque were obtained from the magnetizing inductance equations. All these give a more detailed and deeper understanding of the operation and electromagnetic torque production process of the designed motor-pump devices. The derived mathematical models can be used as the basis for the development of the control techniques of these devices in future studies. Furthermore, the derived mathematical models have been used to improve the initial sketch designs to produce more output electromagnetic torque.

For the design of magnetic bearing system in the radial flux blood pump, by applying discretization strategy to the magnetic rings of the PMB part, the calculation of the radial interactive force between the inner and outer magnetic rings is simplified. The equation for this radial interactive force was obtained by this discretization technique, and the computed value supported the feasibility of this proposed PMB. The designed AMB part of the bearing system was analyzed using basic Maxwell equations and simplified magnetic circuit method. The equation for the developed force of the AMB was derived accordingly. These mathematical analyses also supported the tenability of the design of this AMB.

For the **third task**, FEM analysis was applied to the proposed motor-pump and magnetic bearing designs. Depending on the different structures of the designs, two-dimensional or three-dimensional FEM was used for the analyses. The simulations of the blood pumps focused on the electromagnetic torque acting on the rotor and the rotation speed of the rotor. For the radial flux motor-pump design, the Maxwell 2D simulator was sufficient to study its performance under both loaded and unloaded conditions. Maxwell 3D was however required to simulate the axial flux single-stator design, the axial flux dual-stator design and the axial flux open impeller motor-pump design. In order to simplify the simulation model and not to incur excessive computational overheads, the analyses were restricted to electromagnetic and electromechanical aspects, without considerations on the fluid dynamics. The FEM simulation results of radial flux design have shown that this design produce less output electromagnetic torque compared to the axial flux designs.

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Besides testing the feasibility of the motor-pump designs, FEM simulation results were used to optimize their structures. The optimum structure of the axial flux rotor wheel considering the relationship between the motor ripple torque and average torque with the rotor bar number and skew angle has been obtained for the open-impeller design. The improved structures of the axial flux designs are capable of producing more output electromagnetic torque.

The FEM analysis has also been used to test the validity of the derived mathematical models of the proposed motor-pump devices. The FEM simulation results have been compared with the mathematical model calculation results. In all the four proposed designs, the FEM simulation results have shown close correspondence with the mathematical model calculations, thereby validating the application of general equivalent circuit model of induction machine to the proposed inductively-driven blood pumps. To improve the accuracy of the circuit-based mathematical model, FEM analysis and simulations were used to determine the parameters which vary significantly with operating conditions.

In order to test whether the proposed magnetic bearings for radial flux design can constrain the undesired degrees of motion freedom analyzed in this thesis, Maxwell 2D was applied to the simulation of the AMB, while Maxwell 3D was used for the simulation of the PMB. From the simulation results of the AMB, it can be seen that the force acting on the impeller at any position can be adjusted by changing the coil current of the AMB stator. In other words, the displacement of the impeller along z-axis can be controlled by the coil current. Furthermore, the current to keep impeller at nominal position is very small and so is the power consumption. From the simulation results of the PMB, it was found that the PMB has the ability to produce a restraining force to prevent the rotor of PMB from moving in the radial axes. So the proposed PMB can prevent the impeller from moving away from the central position of the blood pump chamber.

For the **last task**, an experimental test rig was set up for conducting the feasibility study experimentally. Because of the limitation of time and expense, only the open-

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impeller design was implemented experimentally in the lab. The design of the test-rig was based on the intention of providing as much flexibility as possible to accommodate a wide range of experiments to test the feasibility of axial flux inductively-driven blood pumps. The experimental measurements were compared with the FEM simulation results as well as calculations from the mathematical model. In all the comparisons, the obtained experiment data corresponds well with both the FEM simulation results and the mathematical model calculations, which further prove the feasibility of inductively-driven blood pumps.

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6.2 Recommendations

The aspect of power losses and energy efficiency have not been studied and hence not presented in this thesis. The research presented in this thesis focuses on the first-stage feasibility study of the inductive propulsion for CBP. So, the most important target investigated is whether it can provide enough output power to replace the human heart and to meet the requirement of blood circulation. Therefore, the torque-speed characteristics of the various designs considered were presented to demonstrate the viability of the inductive concept. Now that this concept has been determined to be viable for CBP, the future feasibility studies shall examine how power losses can be reduced for better energy efficiency. The studies in this thesis have also implicitly pointed out that there are issues in pump control, energy storage and supply, thermal effects and blood cell damage. So, based on the results obtained from this project, possible further investigations are recommended in the following areas

6.2.1 Structural optimization for power efficiency and levitation

The sketch designs introduced in this thesis are meant to evaluate the feasibility of inductively-driven blood pumps and are far from optimized. With the aid of FEM simulations and analyses, the proposed structures of the stators, rotor-impeller and magnetic bearings can be optimized in terms of power losses, energy efficiency, and levitation control. For power efficiency, the focus would be on reducing copper losses in the rotor bars and iron losses in the stator and rotor cores. Using the derived mathematical model in Section 5.3.2, the rotor could be further optimized to reduce its resistance. Lowering rotor resistance to shift the profile of torque-speed characteristics towards NEMA B type can lead to lower slip operation. For levitation, there are several ways to ensure the rotor is positioned correctly in both axial and radial directions and maintained so when subjected to load fluctuations and other disturbances. One way is to use passive magnetic bearings consisting of permanent magnet rings as proposed in Section 3.6.2. This can take care of motions in the radial directions. For translational motion in the axial direction, this can be taken care of by variations of active magnetic bearing proposed in Section 3.6.4, or by controlling the relative strength of the stator currents in the dual-stator design. Yet another way is to

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balance the magnetic forces produced by the stator windings against the hydrodynamic forces produced by the flowing fluids.

6.2.2 Study of control strategy for the propulsion and magnetic levitation of pump impeller

The mathematical models for the motor-pump and magnetic bearing designs have been derived successfully, which provide a base for the further study of control strategy for the proposed designs. The following aspects are recommended for the study and design of the control strategies: (i) understanding the process and translating the dynamic performance requirements into time, frequency, or pole-zero specifications; (ii) selection of appropriate sensors considering the technology, functional performance, physical properties, reliability, durability, maintainability and cost; (iii) design based on PID control and mathematical models; (iv) pole-placement design based on optimal control; (v) modeling and simulation of the performance of the design, including the effects of nonlinearities, noise and parameter variations; (vi) experimental prototyping for study of electromagnetic and electromechanical performance.

6.2.3 Study of rotor-impeller design on fluid dynamics and blood cell damage

This is important for the maximization of the pump flow-rate and the minimization of back-flow and blood stagnation which may lead to clot formation. The shear stress on the blood cells leading to cell damage is another aspect which needs to be investigated.

6.2.4 Study of thermal effects arising from CBP power losses on blood cells and body circulation

The power loss and energy efficiency of any CBP are very important issues in actual clinical applications. This is not so much about energy costs, as saving and prolonging human life over-ride all running costs considerations. More importantly, this is about the portable energy supply constraint and the potential damaging effects on blood cells subjected to prolong heating. If the energy efficiency of the CBP is low, it will place a strain on the portable energy pack, whether the latter is implanted in the

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body or transmitting energy transcutaneously through the skin. The power loss in the electromagnetic propulsion component of the CBP results in heat which cannot be dissipated easily because the pump is implanted in the human body. This may increase the overall body temperature. The blood cells coming into contact with the impeller blades if the latter also serve as induction rotor bars in the case of open-impeller concept, may receive direct thermal effects. Thus future feasibility studies should not only focus on improving the energy efficiency of the electromagnetic propulsion design, but also on thermal effects on blood cells and body circulation.

6.2.5 Study of energy storage system and transcutaneous power transmission for CBP

The energy efficiency of CBP has very important bearing on the energy storage element. There are two aspects to it: reliability and portability. Even for the most efficient CBP currently available, an implantable battery pack will not last long before requiring an energy charge. Hence for the patient to have a mobile lifestyle, he needs to have an internal battery pack small enough to fit into the human body, and a bigger external battery pack which transmit power to the CBP transcutaneously through the skin. Most of the time, the power should come from the external battery, which is more easily replaceable when its life-span is over. All batteries have limited life-spans which are shortened by deep charge-discharge cycles. Hence the internal battery should normally not be subjected to deep discharges. Internal battery is necessary should the external battery fail or the transcutaneous transmission gets interrupted. Hence the energy efficiency of CBP propulsion will determine how the future technology of energy storage and power transmission for CBP evolves, and they all have to be studied together holistically.

6.2.6 Implementation of entire blood pump for animal trial tests

When all the problems associated with electromagnetic, electromechanical, bearing, control, fluid dynamics, transcutaneous power supply, and medical compatibility have been solved, animal trials may be carried out as the penultimate stage in the long process of developing successful artificial hearts. The ultimate goal would be the implantation of the blood pumps in human bodies.

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