ELECTROMECHANICAL GENERATOR IMPLANTED IN A HUMAN TOTAL KNEE PROSTHESIS

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ABSTRACT: This paper describes an electromechanical generator, implantable in a human Total Knee Prosthesis (TKP), in which the mechanical energy, related to the movement of the knee joint, is converted into electric energy. This device has been conceived in order to power an autonomous sensor system, integrated into the prosthesis, able to monitor the loads on the articular surfaces and to transmit the values outside the body. This activity is motivated by the necessity to replace the battery, which requires periodical substitution, with a power harvesting system exploiting the mechanical energy naturally available in the measurement environment. Furthermore the possibility of monitoring the working condition, during the life of the prosthesis, has the benefit of supporting the rehabilitation of the patient and of reducing the radiographic examinations.

KEYWORDS: Knee prosthesis, Electromechanical generator, Energy harvesting, Human movement, Biomedical sensors.

1 INTRODUCTION

The implant of the total knee prosthesis (TKP) is grown a common operation, the progress in this orthopedic field is continuously increasing: new materials, new surgical techniques and medical solutions are continuously proposed. The real goal of these efforts are, first of all, the longest life of the implants, its adaptability to the different situations and the better recovery of patient's motor capacity. At the moment, for example, different problems concern the wear of the materials, causing its substitution. The repercussions are both on the integrity of the implants, particularly the damage of the tibial tray, and the state of health of the patient (osteolysis). The constituent material of the tibial tray is the UHMWPE (Ultra High Molecular Weight Polyethylene), a material with excellent characteristics because of its low friction and good biocompatibility, however the femoral component causes a surface damage of the UHMWPE producing debris that induces osteolysis.

The consequences of this situation [1] are an initial detachment of the TKP from the bones, with consequent reduction of precision in the mobility of the knee, and inflammations of the tissues, in this case a revision of the implant becomes inevitable, with different future complications (new surgical ablation of the bones for the new implant, incomplete recovery for the reduction of the joint mobility). In addition the conditions, at which the prosthesis works, are quite variable in the time for a single person and from person to person, so a good design is indispensable for the best adaptability of the implant to the bones and its durability [2].

Those requirements motivate a continuous effort for improving the knowledge of the mechanics of the knee joint with the aim of obtaining the design of a prosthetic implant able to reproduce it as much as possible [3,4].

A correct understanding of the force distribution on the articular surfaces can make their design able to resist to the wear a more long time. Furthermore the possibility of monitoring the working condition, during the life of the prosthesis, has the benefit of supporting the rehabilitation of the patient and of reducing the medical examinations for valuing the working conditions in the time.

Different solutions, oriented to reduce the wear of the tibial tray, have been proposed, and they also could try benefits from a better knowledge of force parameters. Many Authors and companies, for example, have oriented their research towards new materials, for the femoral components, as the pure alumina ceramics against the CoCr femoral material used in the common implants [1]. Different recent papers dealt with the possibility of monitoring *in vivo* the tibia-femur forces with an electronically instrumented prosthesis, in which the measured forces are transmitted in wireless modality [5,6,7]. These studies had an acceptable accuracy in the knowledge of the loads acting on the tibial plate, but in all these cases an external coil induction system is necessary to power the device. A good improvement could be to free the measure system from the external supply, simplifying its portability.

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Many similar study has been conducted in order to value the possibility of harvest the energy from the human movement [8,9], in particular [10] deals with the possibility of energy harvesting from the movement of the knee using an oscillating device, optimized to be implanted in the prosthesis, obtaining comforting results in terms of the produced power.

2 DESCRIPTION OF THE SYSTEM

2.1 WORKING PRINCIPLE

In this paper, an electromechanical generator, implantable into a TKP will be show. It directly converts the relative motion of the tibial and femur inserts in electric energy without the movement of any other device with respect to these ones, advantageously reducing possible failures and simplifying its realization.
The electric conversion of the relative motion is obtained via Faraday's law. Figure 1 shows the sagittal view of a TKP: six magnets are inserted in six prismatic housings, disposed in the thickness of each condyle with the magnetic axis *z* parallel to the surface of the tibial plate.

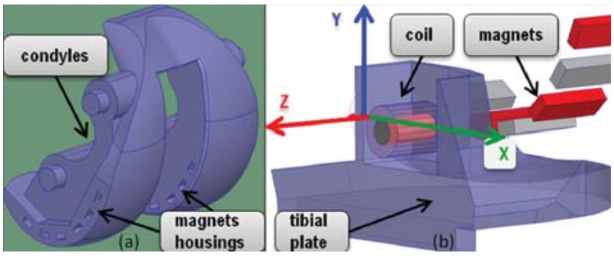

Figure 1: *Condyles with the magnets housings (a); sagittal section of the tibial tray with a coil and the magnets (b).*

The alterations carried out on the prosthesis are very limited, with the clear intent of modifying the minimum possible the biomechanical and structural characteristics of the implant. Between the two condyles, in a prominence of the tibial plate, a cavity is realized, in which two cylindrical copper coils (*A* and *P* in Figure 2) are inserted, with their axis parallel to magnetic axis *z*.
In this way, the relative motion between femur and tibia induces on the coils a potential difference according to Faraday-Newmann-Lenz's law.
The magnets in the same condyle have alternate magnetic axis while two magnets aligned on the same line have the same magnetic axis; the reason is obtaining a more rapid alternation of the induction magnetic field on the coils and a subsequent more velocity in the variation of the induced flux. Furthermore the coils have a core in ferrite with the aim of augment the flux lines going through the coils.
A first positioning of the two coils has been made considering the trajectory of the magnets and their reciprocal position, and choosing, for each coil, a zone in which the magnets have the best induction of the magnetic field. This aspect is strictly related with the efficiency of energy harvesting, and it is quite difficult to approach for the complicated geometry of the system, so a future parametric analysis will be conducted to deal with it. A similar consideration will regard the connection, the number and the shape of the coils.

2.2 NUMERICAL SIMULATION OF THE SYSTEM

In order to obtain a first estimate of the capabilities of the device, a finite elements electromagnetic simulation has been conducted. A discussion of the parameters that affect the relative movement of the tibial and femoral inserts has been necessary for a simulation of the relative motion of the elements of the TKP. A 3D model of a commercial prosthesis has been designed with the aim of using this geometry, both in the simulation, and in a future rapid prototyping realization of the device.
Figure 2 shows the realized model.

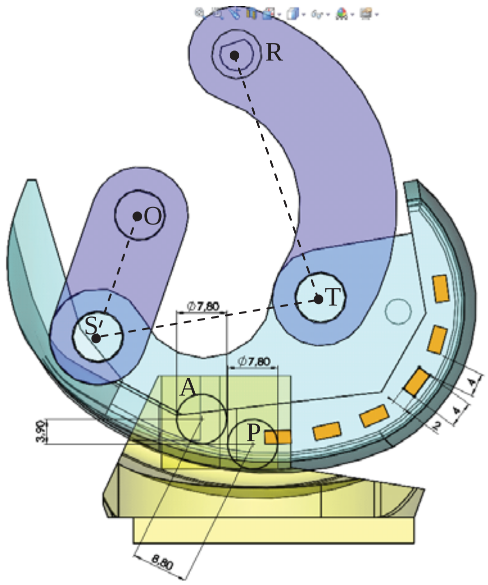

Figure 2: *View of the mechanism that moves the condyles, respect to the tibial tray.*

The reproduction of the correct kinematics of the knee joint, both towards the simulation and towards the future realization of the prototype, has been approximated considering that the exact knowledge of the motion could had led to very little modifications in the efficiency of the electric generation, so it has been considered unneeded for the purpose of this study.
For this reason only the global movement has been reproduced, in particular the real combined motion of rotation and translation has been respected in order to animate the model with a motion, as much as possible, corresponding to the real one. This goal has been obtained with the design of the kinematic chain shown with the dashed lines in Figure 2.
In particular the point *O* is the center of rotation of the crank, directly connected to the drive axle, *R* is the center of rotation of the rocker arm and *OR* is the frame of the four bar linkage.

The simulations regarded a particular fraction of the knee movement during the gait cycle, i.e. the swing phase, in which a complete relative rotation (about 60°) of tibial and femoral axis occurs, also in subjects with TKP.
The common commercial software applications for finite elements electromagnetic simulation are able to simulate only a purely rotational or translational motion. Because in the case in object the magnets move off with respect to the coils with a composed motion, an alternative path has been followed. For recreating a situation, as much as possible corresponding to the real one, using the possibility of motion simulation offered by the 3D-CAD software, the motion of the system was been subdivided in a sequence of configurations reproduced by the kinematic chain described before.
In order to obtain a first estimate of the simulation, the full rotation has been divided in small steps of 5 degrees each (for an improvement of the approximation a more tight subdivision can be made). The electromagnetic induction on the two solenoids has been carried out considering the magnetic induction field generated by the magnets in each configuration, being evident that this field, evaluated in magnetostatic conditions, is coincident with the instantaneous value of the field in dynamic conditions, i.e. when the system is getting through this configuration.
In this initial phase, the copper coils used in this simulation have five hundred turns and the diameter of the wire is 0.1 mm. The geometrical parameters of the magnets and the coils are reported in the Table 1. The magnets, made of NdFeB (Neodymium N45), have the characteristics reported in the Table 2.

Table 1: *Geometrical data of magnets and coils*

Component	Dimensions	
	Parameter [units]	*Value*
Magnets	(length, height, width) [*mm*]	(4,2,13)
Coil A	(diameter, length) [*mm*]	(7.80,13.6)
Coil P	(diameter, length) [*mm*]	(7.80,17.9)

Table 2: *Physical data of the NdFeB magnets*

Parameter	Value [unit]
residual induction	1320÷1380 [mT]
coercive force	923 [kAm^{-1}]
intrinsic coercive force	955 [kAm^{-1}]
energy product BH max	342÷366 [kJm^{-3}]
max operating temperature	80 [°C]

In order to estimate the variation of the magnetic induction field along the coils, for a first understanding, the flux in the solenoids has been evaluated only on three particular sections. Figure 3 shows, in particular, the variation of the flux induced in the sections of the coil *A*: due to reverse polarity of two following couples of the magnets, the flux has an oscillating variation. This figure shows, also, that if the exam had be limited only to the two extreme cross sections, one could had supposed the stray flux on the solenoid *A* very little, in fact the values of the flux on the two sections *S_(versor±**k**)* are practically coincident.
A particular situation was observed in the configuration relative to the position 10°, in fact in this configuration the flux induced in the section *S_(x-y)* is twice respect to flux on the surfaces *S_(versor **k**)* and *S_(versor **-k**)*.

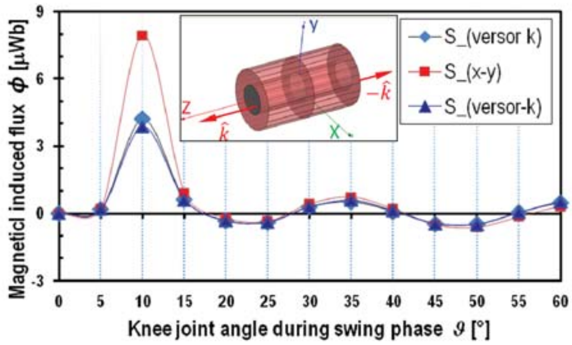

Figure 3: *Magnetic induced flux through the cross sections S_(versor ±k) and S_(x-y) of the coil A as a function of the rotation ϑ of the femoral insert respect to the tibial tray during the swing phase of the gait cycle.*

A 3D representation of the magnetostatic induction field, reported in Figure 4, gives a better comprehension of this result, in fact, limiting the attention to the plane *(y-z)* of the coil *A*, a more intensive induction field in the medial cross section (plane *(x-y)*), with respect to the z_1 and z_2 sections, is visible.

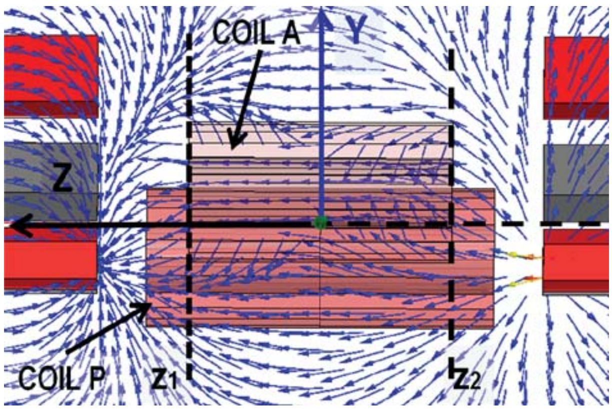

Figure 4: *Flux lines of the magnetic induction field in the plane (y-z) of the coil A. The flux lines have the greater density in the medial cross section.*

The method used to calculate the total induced flux in the different windings of the coils followed the following reasoning. The distribution per unit of length of the turns of the winding, *n(z)* [m^{-1}], and the cross section of the coils, *A(z)* [m^2], are supposed constant on the length *l*

[m] of each coil (i.e. for the coil A, the interval $[z_1, z_2]$ on *z-axis* in Figure 4). The total flux, $\phi_{solenoid}(\vartheta)$ [Wb], induced in a single solenoid in each single angular configuration ϑ, is calculable using Equation (1). In this expression, the evaluation of the dot product is carried out considering that only the B_z component is necessary, as shown by the Figure 4.

$$\begin{aligned}\phi_{solenoid}(\vartheta) &= \int_{z_1}^{z_2} n(z)\,dz \iint_{A(z)} \overrightarrow{B(x,y,z;\vartheta)} \bullet \overrightarrow{dA} = \\ &= n \cdot \int_{z_1}^{z_2} dz \iint_A B_z(x,y,z;\vartheta)\,dA = \qquad (1) \\ &= n \cdot \iiint_{solenoid} B_z(x,y,z;\vartheta)\,dA\,dz\end{aligned}$$

The next step is the evaluation of the induced electromotive force, V_i, [V], expressed by the Faraday-Newmann-Lenz's law of electromagnetic induction:

$$V_i = -\frac{\partial \phi_{solenoid}}{\partial t} \qquad (2)$$

In order to obtain a relationship of the flux as a function of the time t, it is necessary to know the variation of the knee joint angle ϑ during the considered rotation.

Near to swing phase the considered rotation of sixty degrees proximately occurs in the 25% of the gait cycle; assuming a duration of the gait cycle of about 1.5 s, an average angular velocity ω_{avg} of 160 °/s results. Consequently the Equation (2) becomes:

$$\begin{aligned}V_i &= -\frac{\partial \phi_{solenoid}}{\partial t} = -\frac{\partial \phi_{solenoid}}{\partial \vartheta} \cdot \frac{\partial \vartheta}{\partial t} = \\ &\cong -\frac{\partial \phi_{solenoid}}{\partial \vartheta} \cdot \omega_{avg}\end{aligned} \qquad (3)$$

where $\phi_{solenoid}(\vartheta)$ is provided by Equation (1).

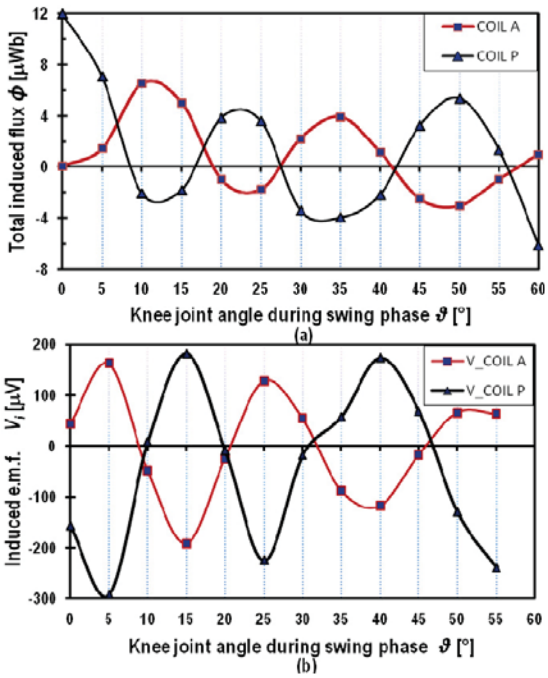

Figure 5: *Induced flux (a) and voltage (b) on the coils A and P.*

Figures 5(a) and 5(b) show, respectively, the trend of the total induced flux $\phi_{solenoid}$ and the trend of the induced electromotive force V_i as function of the knee joint angle ϑ during the sixty degrees swing.

3 CONCLUSIONS

In this article an energy harvesting system for the power supply of a force sensor, implantable in human knee prostheses, has been studied. This activity has been conducted designing the system and performing electromagnetic finite element simulations for the evaluation of the electromotive force generated by the designed system. The next steps will be the experimental validation of the simulation results and the analysis of an improved simulation model finalized to the optimization of the real system with the aim to increase the electrical power produced by modifying the geometry and magneto-electric parameters. A preliminary prototype of the system has already been made and the first measurements of the electromotive force confirm qualitatively the data obtained from the simulation. At the same time a first design of an electronic circuit able to manage the produced power is under development.

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