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Loads on the Hip Joint Endoprosthesis Due to Impacts

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Abstract
Engineering plays a significant role in the advancement of medicine. One example of this is endoprostheses, which are prostheses implanted inside the body. Hip joint endoprostheses are commonly implanted nowadays, greatly improving physical fitness and the associated quality of life. However, a potential risk in using such endoprostheses is the possibility of dislocation. In this presented work, systems of forces acting on the limb were subjected to analysis, identifying unstable states that increase the risk of dislocation. Most of the analyses are qualitative, presenting rather than solving the problem. Nevertheless, a quantitative approach was presented for the case of dynamic forces generated during kicking a soccer ball. For this purpose, computer simulation was employed, based on an appropriate mathematical model.

Keywords
Hip joint endoprostheses, mechanics of endoprosthesis movement, loads on the hip joint endoptothesis, the Vensim PLE program, innovation.

Introduction

A wide array of technological innovations plays a pivotal role within the field of medical advancement. These innovations emerge through collaboration between medical professionals and engineers, and cover a diverse spectrum of purposes, including diagnostics, treatment methodologies, the replacement or augmentation of impaired body parts, and rehabilitation (Kaczmarska et al., 2021). One specific category within this realm is comprised of endoprostheses, prosthetic devices which, in line with the word's etymology ("endo"), are implanted within the body. Among endoprostheses, a particular subset includes those designed to substitute malfunctioning joints. A special case is hip endoprosthesis, which constitutes an implant characterized by kinematic attributes engineered to replicate natural joint mobility. In contemporary practical implementations, this endoprosthesis typically comprises a specially designed ball joint, forming a kinematic pair with three degrees of freedom (Mrozowski and Awrejcewicz, 2004). It is thus a system subject to the laws of mechanics. Hence, the aim of this study is to analyze the forces affecting the spherical joint of the endoprosthesis concerning conditions that ensure stability, as well as those that may lead to instability, posing a risk of dislocation.

In the solutions commonly used today, an acetabular component (sometimes with an additional shell) is attached to the pelvis, while a stem, connected via a ball-headed neck, is attached to the femur (Fig. 1 and 2) (Serafin, 2003).

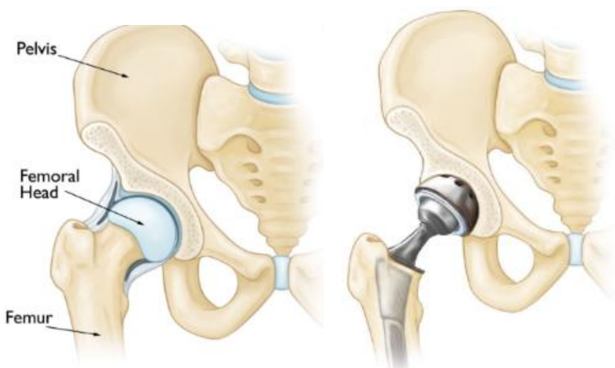

Fig. 1. Endoprosthesis – an example of a contemporary solution (AAOS, 2023)

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The ball-headed acetabular joint employed in the endoprosthesis allows movements around an infinite number of axes, with three primary ones identified as follows (Bober and Zawadzki, 2006):

- bending and straightening with a range of up to 135°, extending to 165° in slight abduction
- abduction and adduction – up to 40° in each direction
- external rotation of up to 15° and internal rotation of up to 35°

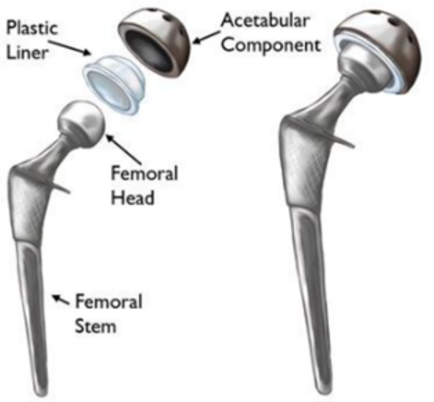


Fig. 2. Components of hip endoprosthesis (Gormack Orthopaedics, 2023)

These movements are initiated and decelerated by muscles and ligaments. Nevertheless, the joint's structural design does not provide protection against the dislocation of the head from the acetabular component. This is an undesired movement (the fourth degree of freedom), which carries the risk of joint dislocation, especially at greater displacement values. In this regard, muscles and ligaments also play a crucial role in restricting the extent of dislocation, acting as a preventive measure. Following surgery, flaccidity of the muscles connecting the ilium to the femur can occasionally lead to dislocation (Karuś, 2010).

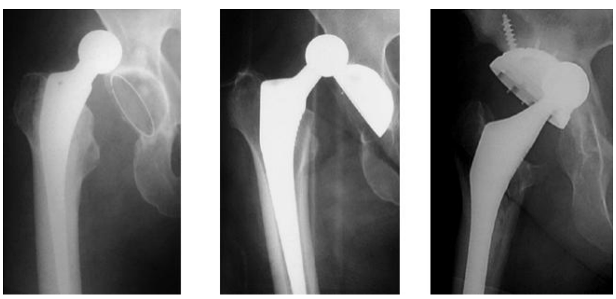


Fig. 3. Hip endoprosthesis dislocation – sample roentgenograms (Karuś, 2010)

The reason for implanting an endoprosthesis is various joint degenerations, which have been recognized by the WHO as a lifestyle disease. It is estimated that one out of every three elderly individuals will require hip replacement surgery. According to estimates, about 400,000 hip replacement surgeries are performed annually in Europe. More than 45,000 surgeries are performed annually in Poland, falling short of meeting the demand and resulting in prolonged waiting periods for many patients awaiting the procedure (Care Experts, 2023).

The implantation of a hip endoprosthesis, as well as other joints (such as the knee), significantly improves physical fitness and the associated quality of life.

Mechanics of endoprosthesis movement

The hip endoprosthesis is a mechanical system designed to facilitate movement and transfer forces. Therefore, in selected analyses, methods of mechanics can be applied in the fields of statics, kinematics, and dynamics.

In a stable state, the distribution of forces ensures the prevention of excessive dislocation of the head from the acetabular component, ensuring the proper functioning of the kinematic pair. Hence, correct walking, sitting, and lying down with the prosthesis should become a habit for individuals with an endoprosthesis.

There are a number of precautionary recommendations to prevent dislocation, which must be initially or permanently observed. Such a basic set of recommendations includes:

- avoiding excessive bending of the hip joint
- sitting in a way that the angle between the legs and the torso does not exceed 90 degrees
- sitting down and standing up by extending the leg forward and distributing body weight onto the arms
- avoiding excessive internal rotation (refraining from excessive torso twisting when standing up)
- avoiding excessive adduction: in the hospital, the patient lies on their back with a special wedge placed between the thighs. At home, it is also recommended to rest or sleep on one's back, with the operated leg slightly tilted at approximately 30 degrees and turned outward. In the subsequent months following hip replacement, it is allowed to sleep on one's side with a pillow placed under the limb or between the thighs
- avoiding bending
- avoiding walking with exaggerated knee lift
- refraining from crossing the legs or sitting with one leg over the other

- avoiding sitting on the edge of chairs or very soft, “sinking” armchairs.

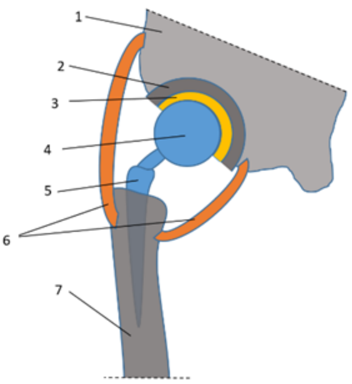
Strict adherence to these recommendations during the postoperative period will help patients develop new habits upon returning to daily routines, substantially reducing the risk of hip joint dislocation Care Experts (2023).

Adhering to the provided recommendations reduces the likelihood of forces that could lead to dislocation (Olszówka and Niścior, 2015). However, this reference pertains solely to static forces, i.e. not arising from abrupt movements or impacts. Furthermore, these recommendations stem from long-standing practical experience rather than in-depth analyses grounded in the principles of mechanics.

In the following section, an analysis of favorable and unfavorable force configurations for ensuring the stable function of the endoprosthesis was carried out. With regard to dynamic loads, simulation studies were performed to investigate the forces generated by typical impacts.

In an engineering approach, the analysis of mechanical systems relies on models that act as substitutes for reality. When constructing such a model, it is crucial to define its purpose and intended use, as the model’s complexity depends on these factors. The model thus represents a simplification of reality and should be complex enough to account for important study features, yet simple enough to allow for analysis using available methods and tools (Gierulski, 2016; Nowak, 2009).

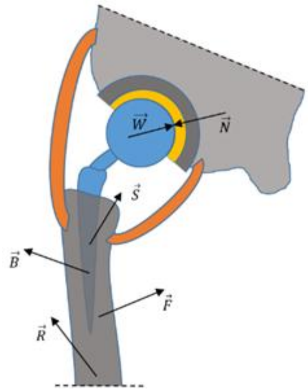
In the case under consideration, analysis and testing were carried out using the endoprosthesis model shown in Figure 4.



1. Pelvis
2. Acetabular component
3. Plastic Liner
4. Fermoral head (ball)
5. Fermoral stem
6. Muscles and tendons
7. Thightbone

Fig. 4. The model used in the analysis
Source: Own elaboration

Figure 5 shows the forces acting on the limb with the endoprosthesis in the model used. The points of application, directions and values of forces are included for explanatory purposes.



$\overrightarrow{R} = \sum R_i$ – ground reaction forces
$\overrightarrow{F} = \sum F_i$ – external forces
$\overrightarrow{S} = \sum S_i$ – forces from muscles, tendons and ligaments
$\overrightarrow{B} = \sum B_i$ – fictitious forces
$\overrightarrow{N} = \sum N_i$ – pressure force on the acetabular component
$\overrightarrow{W}$ – resultant force

Fig. 5. Forces within the analyzed model
Source: Own elaboration

The individual forces (R, F, S, B, N) are the resultant of the component forces (denoted by the subscript “i”). According to the principles of mechanics, the condition for force equilibrium in the system is (Niezgodzinski, 2007):

$$\overrightarrow{F} + \overrightarrow{S} + \overrightarrow{R} + \overrightarrow{B} + \overrightarrow{N} = 0 \tag{1}$$

or, after denotation:

$$\overrightarrow{W} = \overrightarrow{F} + \overrightarrow{S} + \overrightarrow{R} + \overrightarrow{B} \tag{2}$$

This condition can be expressed as:

$$\overrightarrow{W} + \overrightarrow{N} = 0 \tag{3}$$

Hence, the pressure force of the acetabular component on the head of the ball joint N is equal to the resultant of the other forces W, that is, the force applied by the head onto the acetabular component. These forces act in opposite directions. The second equilibrium condition is that the moment with respect to the center of the spherical joint head must be zero. The moment is not marked in the figure because the kinematic pair adjusts itself through rotation to satisfy this condition. This corresponds to the correct rotation of the limb. In the absence of inertia force $(B = 0)$, equilibrium pertains to the system in a static state. In the opposite case $(B > 0)$, force equilibrium pertains to motion (dynamics).

When the force W has a direction encompassing the contact surface between the acetabular component and the ball joint, the situation is stable, and there is no cause for concern. However, in certain cases, the direction of the force W does not meet this condition and is deviated by an angle α from the extreme position determined by the last point of contact (Fig. 6). In such a scenario, the force N balances only the normal component of the force W with a value:

$$W_n = W \cdot \cos \alpha \tag{4}$$

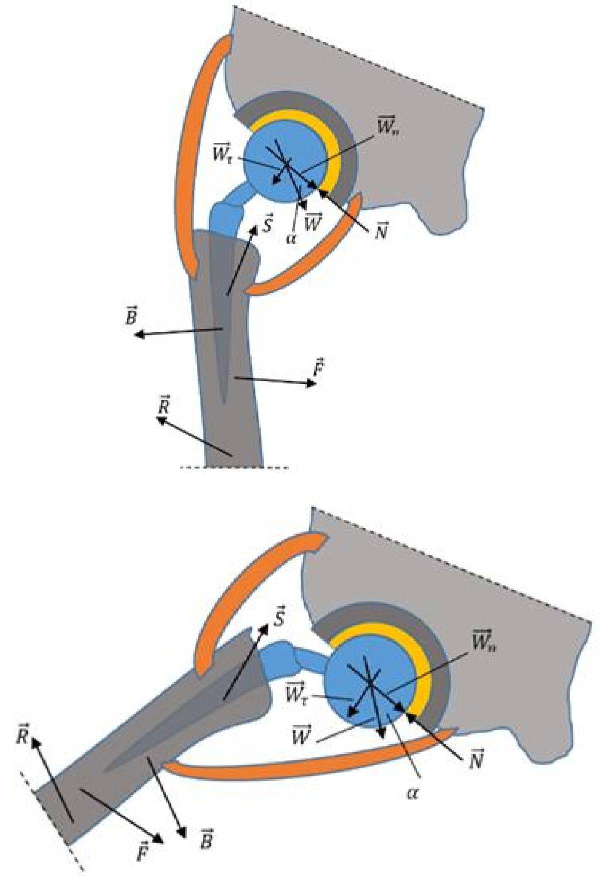


Fig. 6. Forces in an unstable system – various limb positions
Source: Own elaboration

In contrast, the tangential component with a value of:

$$W_\tau = W \cdot \sin \alpha \tag{5}$$

is not balanced and causes the ball of the joint to slide out of the acetabulum component, posing a risk of dislocation (Fig. 7).

The presented analysis is purely qualitative in nature. It outlines the problem and the necessary conditions to be met to prevent the force system from posing a risk of dislocation. No studies have been conducted to measure the actual forces generated if the indicated recommendations are not followed. Therefore, this remains an open research topic, and the presented analyses can be valuable in planning future experiments.

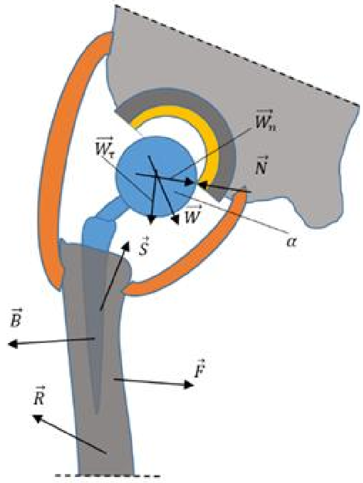


Fig. 7. Mechanics of dislocation
Source: Own elaboration

Impact – simulation model

An impact provides an illustration of the generation of dynamic forces. These forces are of short duration, variable in intensity, and can reach significant values. Impacts can happen either accidentally, like when a foot collides with an object on the ground during walking, or deliberately, for instance, when kicking a ball.

The study of dynamic processes can be effectively conducted through the use of simulation models (Nowak, 2009). For this purpose, a mathematical model was constructed and implemented using an appropriate computer program.

To construct a mathematical model of the ball-kicking process, a simplified diagram (Figure 9) was employed, in which the following components were single out:

1. leg
2. ball
3. deformation zone
4. fixed component of the joint (the acetabulum component of the endoprosthesis).

The ball can be kicked with the leg in different positions, either vertically (Figure 8a) or inclined away from the vertical (Figure 8b). This affects the direction of force N, thereby increasing the risk of dislocation in certain scenarios.

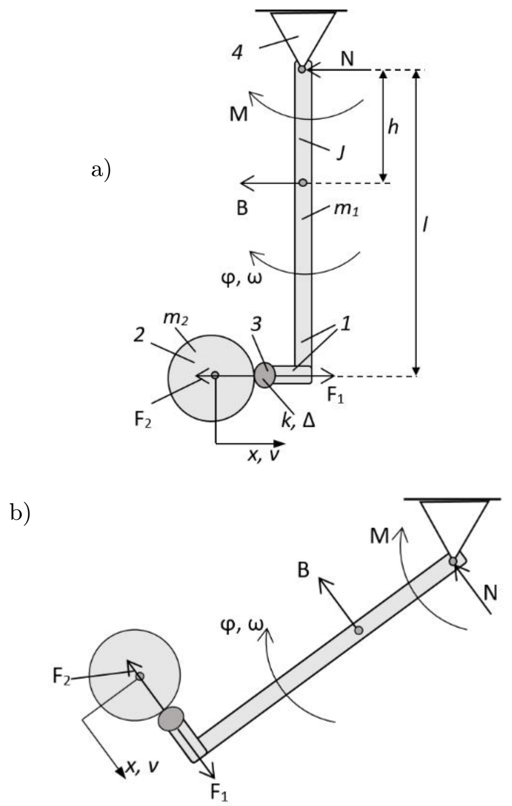


Fig. 8. a, b. Simplifications in model construction
Source: Own elaboration

Other symbols in the diagram:

φ	–	angle of pivot (rotation) of the leg
ω	–	angular velocity of leg rotation
x	–	ball displacement
v	–	ball speed
m_1	–	mass of the leg
J	–	moment of inertia of the leg
m_2	–	mass of the ball
h	–	position of the leg's center of mass
l	–	leg length
k	–	elastic modulus in the deformation zone
Δ	–	the amount of deformation
F_1, F_2	–	force of interaction between the ball and the leg, and the leg and the ball
N	–	pressure force of the acetabular component
B	–	fictitious force
M	–	external force moment acting on the leg originating from muscle force.

Taking into account the principles of mechanics, a mathematical model describing the impact phenomenon has been constructed. The principles of dynamics require the use of differential equations for the description. This model only pertains to the time when there is direct contact between the leg and the ball. During this time, both the ball and the leg undergo deformation in the contact zone ($\Delta \geq 0$). The following mathematical expressions constitute a mathematical model describing the contact between the leg and the ball in the considered example (Niezgodziński, 2007).

$$\frac{\mathrm{d}\varphi}{\mathrm{d}t} = \omega \tag{6}$$

$$\frac{\mathrm{d}\omega}{\mathrm{d}t} = \epsilon \tag{7}$$

$$\epsilon = -\frac{1}{J} \cdot (F_1 \cdot l - M) \tag{8}$$

$$\frac{\mathrm{d}x}{\mathrm{d}t} = v \tag{9}$$

$$\frac{\mathrm{d}v}{\mathrm{d}t} = \alpha \tag{10}$$

$$\alpha = -\frac{1}{m_2} \cdot F_2 \tag{11}$$

$$F_1 = F_2 = F \tag{12}$$

$$F = k \cdot \Delta \tag{13}$$

$$\Delta = x + \varphi \cdot l \tag{14}$$

$$B = m_1 \cdot \epsilon \cdot h \tag{15}$$

$$N = F - B \tag{16}$$

The mathematical model was implemented in the Vensim PLE (Personal Learning Edition) program. This is a simulation package with a graphical module designed to support the modeling process in the system dynamics convention. It allows for the easy construction of system dynamics models in the form of a structural diagram. Gierulski (2016) and Krupa (2008) shows such a diagram for the analyzed issue (Fig. 9).

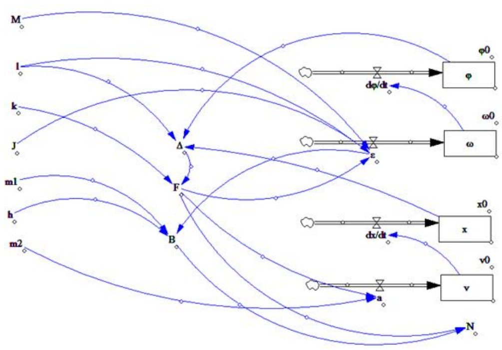


Fig. 9. Structural diagram in Vensim program
Source: Own elaboration

The lettered symbols shown on the diagram correspond to the variable $(\phi, \omega, x, v, \varepsilon, a, \Delta, F, B, N)$ and constant parameters $(M, l, k, J, m_1, h, m_2)$ of the mathematical model. Arrows indicate the relationship between them. The relationships linking variables and constants are included according to the mathematical model, but their forms are not visible on the structural diagram. Due to the integration operations performed, initial values $(\phi0, \omega0, x0, v0)$ must be provided. After entering the values of constants and initial values, it is possible to solve the system of equations of the mathematical model for successive moments of time, that is, to provide a computer simulation. The results are obtained in numerical form, and can also be presented as graphs.

Impacts – calculation results

Using the constructed mathematical model and the Vensim application sample experiments were conducted by simulating various data values (Nowak, 2009). Changes in parameters were limited to the elastic modulus k, mass m_2 and initial velocity v0. These values were determined on the basis of data from the literature (Bober and Zawadzki, 2006), own measurements and analysis, and some of them were adopted without practical justification for the presentation of potentially possible cases. The times of contact between the leg and the ball are very short, which poses certain challenges in numerical calculations.

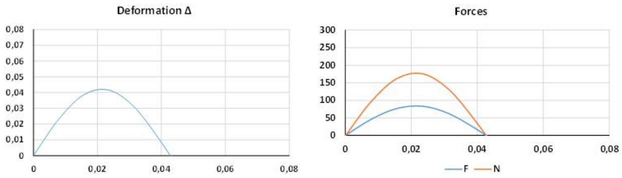
In the calculation process using the Vensim program, an integration step of 0.0078125 s was adopted, which often does not allow for precise determination of result values. The figures presented, as seen in the results given in numerical form, should therefore be considered approximate.

Experiment 1

The contact time between the leg and the ball is slightly greater than 0.04 s (Fig. 10). The maximum deformation in the elastic zone is 0.042 m, while the maximum forces are: $F_{\max} = 83.22$ N, and $N_{\max} = 175.45$ N.

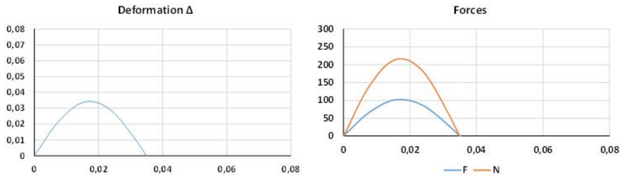
Experiment 2

The elastic modulus has been significantly increased, resulting in a decrease in the time of contact between the leg and the ball (between 0.03 and 0.04 s) (Fig. 11) and a maximum deformation of about 0.034 m. Consequently, the forces also increase: $F_{\max} = 101.80$ N, and $F_{\max} = 214.83$ N.



Parameter	M	l	k	J	m_1	h	m_2	ω0	v0	φ0	x0
Value	0,7	0,8	2000	3	12	0,35	0,4	2	1,5	0	0
Unit	*Nm*	*m*	*N/m*	*kgm²*	*kg*	*m*	*kg*	*1/s*	*m/s*	*rad*	*m*

Fig. 10. Simulation results – experiment 1
Own elaboration

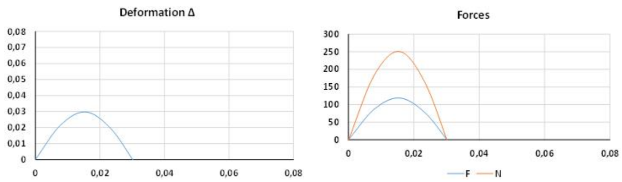


Parameter	M	l	k	J	m_1	h	m_2	ω0	v0	φ0	x0
Value	0,7	0,8	3000	3	12	0,35	0,4	2	1,5	0	0
Unit	*Nm*	*m*	*N/m*	*kgm²*	*kg*	*m*	*kg*	*1/s*	*m/s*	*rad*	*m*

Fig. 11. Simulation results – experiment 2
Own elaboration

Experiment 3

Again, the elastic modulus $k = 4,000$ N/m was increased (Fig. 12). The contact time between the leg and the ball is less than 0.03 s, and the maximum deformation is about 0.03 m. The forces have increased and now are as follows: $F_{\max} = 118.90$ N, and $N_{\max} = 251.08$ N.

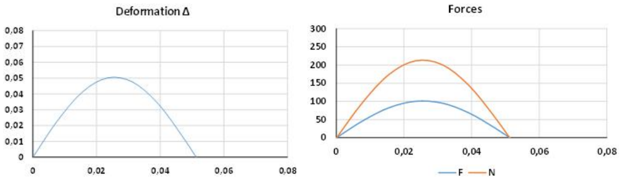


Parameter	M	l	k	J	m_1	h	m_2	ω0	v0	φ0	x0
Value	0,7	0,8	4000	3	12	0,35	0,4	2	1,5	0	0
Unit	*Nm*	*m*	*N/m*	*kgm²*	*kg*	*m*	*kg*	*1/s*	*m/s*	*rad*	*m*

Fig. 12. Simulation results – experiment 3
Own elaboration

Experiment 4

In this experiment (Fig. 13), the value of the elastic modulus was the same as that from Experiment 1 ($k = 2,000$ N/m), while the mass of the ball was increased ($m_2 = 0.6$ kg). This results in an increase in the time of contact between the leg and the ball: about 0.05 s, with the forces reaching the following values: $F_{\max} = 100.29$ N, and $N_{\max} = 211.64$ N.

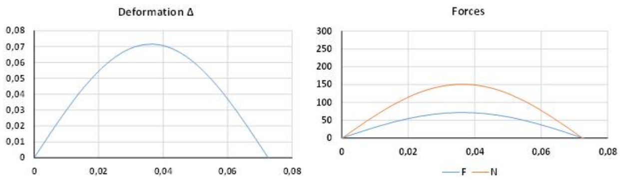


Parameter	M	l	k	J	m_1	h	m_2	ω0	v0	φ0	x0
Value	0,7	0,8	2000	3	12	0,35	0,6	2	1,5	0	0
Unit	*Nm*	*m*	*N/m*	*kgm²*	*kg*	*m*	*kg*	*1/s*	*m/s*	*rad*	*m*

Fig. 13. Simulation results – experiment 4
Own elaboration

Experiment 5

The mass of the ball was the same as in the previous experiment, while the elastic modulus was significantly reduced ($k = 1,000$ N/m, $m_2 = 0.6$ kg). This corresponds to kicking a heavy and soft ball (Fig. 14). The time of contact between the leg and the ball increased (over 0.07 s) while the forces decreased ($F_{\max} = 71.07$ N, and $N_{\max} = 149.69$ N).

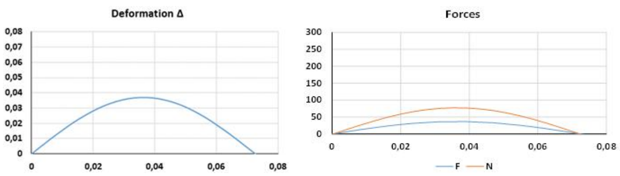


Parameter	M	l	k	J	m_1	h	m_2	ω0	v0	φ0	x0
Value	0,7	0,8	1000	3	12	0,35	0,6	2	1,5	0	0
Unit	*Nm*	*m*	*N/m*	*kgm²*	*kg*	*m*	*kg*	*1/s*	*m/s*	*rad*	*m*

Fig. 14. Simulation results – experiment 5
Own elaboration

Experiment 6

In this experiment (Fig. 15), kicking a soft stationary ball was simulated ($k = 1,000$ N/m, $m_2 = 0.6$ kg, $v_0 = 0$ m/s). The time of contact between the leg and the ball remained similar to the previous experiment (over 0.07 s) while the forces were further reduced ($F_{\max} = 36.72$ N, and $N_{\max} = 76.89$ N).



Parameter	M	l	k	J	m_1	h	m_2	ω0	v0	φ0	x0
Value	0,7	0,8	1000	3	12	0,35	0,6	2	0	0	0
Unit	*Nm*	*m*	*N/m*	*kgm²*	*kg*	*m*	*kg*	*1/s*	*m/s*	*rad*	*m*

Fig. 15. Simulation results – experiment 6
Own elaboration

The parameter values in the six experiments presented should be considered as examples.

An appropriate program tool was presented, allowing for the investigation of impact phenomenon in terms of generating forces in the endoprosthesis. Simulations also enable tracking changes in other variables within the model, such as velocities and displacements. However, since the research was focused on analyzing forces, other variables were not shown. The applied model only considers elasticity in the contact zone between the leg and the ball. Damping, which dissipates energy, was not taken into account. Nevertheless, for very short contact times between the leg and the ball, the amount of energy dissipation is minimal, justifying the omission of damping properties.

Conclusions

In line with the research objective, analyses were conducted on the forces within the endoprosthesis, ensuring both stable and unstable states that pose a risk of dislocation. The quantitative analyses considered dynamic forces resulting from impacts. The conducted analyses demonstrate the forces generated as a result of kicking a ball. These forces can reach significant magnitudes, and their direction can be highly unfavorable for maintaining system stability. Stability requires balancing of forces, which is achieved by obtaining a sufficiently large value of force S (muscles, tendons, and ligaments). This requires good physical condition, which can often be problematic for older individuals who undergo endoprosthesis implantation. An additional challenge in generating such a force is the very short duration of dynamic forces resulting from the impact. Simply put, muscles may not have enough time to generate the necessary force within a fraction of a second.

Another way to counter dislocations in such cases could be the modification of the endoprosthesis design. An example of such a modification involves extending the spherical part of the acetabular component and plastic liner (Fig. 16a) with a cylindrical part (Fig. 16b) of length λ (Fig. 16). This allows for the displacement of the spherical part by λ (Fig. 16c) without the risk of dislocation. From a mechanical perspective, this transforms a system with three degrees of freedom into a system with four degrees of freedom (three rotations and displacement). (Gierulski (2020) and (Kaczmarska and Gierulski (2021-2).

Such displacement requires muscle elongation (stretching), which significantly increases the acting forces and reduces the likelihood of dislocation. Ad-

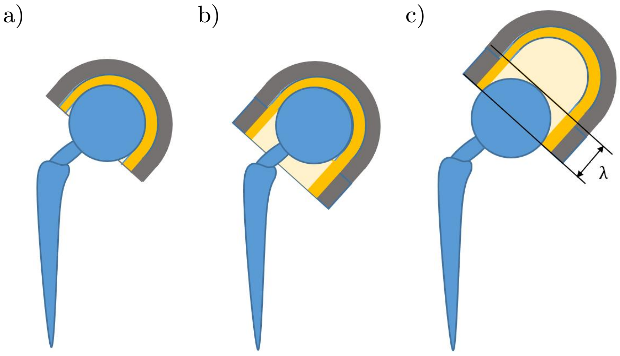


Fig. 16. Modification of the endoprosthesis design
Source: Own elaboration

ditionally, it provides some extra time for the muscle system to generate the necessary force. The change in the design with this innovative solution is minimal, which can have a positive impact on its widespread adoption. This modification does not alter the entire construction of the endoprosthesis but only its components. However, widespread adoption requires the commercialization process (Gierulski et al., 2020). The decision to initiate the commercialization process should be preceded by an analysis of the implementation potential (Kaczmarska and Gierulski, 2021-1), including a valuation of this innovative technology (Terpilowska et al., 2021). This will help estimate the necessary costs and seek appropriate sources of financing (Janasz et al., 2020).

This paper does not exhaust the subject of engineering analysis of the functioning of endoprostheses. It demonstrates a feasible research methodology and the results of sample simulation experiments using an efficient tool. Thus, the paper reinforces the belief in the need for joint efforts in the fields of medicine and technical sciences.

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