ISSN 0030-5987. Orthopaedics, traumatology and prosthetics. 2026. № 2

УДК 616.718-089.843:685.382]-007.24:612.76](048.8)

DOI: http://dx.doi.org/10.15674/0030-59872026256-63

Computational analysis of the deformation of the biomechanical system «bone – implant – prosthesis»

O. M. Sitenko [1], O. O. Larin [2], D. V. Lavinsky [2], K. A. Barbin [2]

[1] Sytenko Institute of Spine and Joint Pathology National Academy of Medical Sciences of Ukraine, Kharkiv
[2] National Technical University "Kharkiv Polytechnic Institute", Kharkiv, Ukraine

The prospects of the method of endo-exo-prosthetics of lost limbs, the basis of which is the interaction of the patient with the prosthesis through an implant connected to the stump bone, largely depends on ensuring the durability of the functioning of the stump bone, the implant and their connection. This article presents the method and analysis of deformations in the biomechanical system "bone-implant-prosthesis", an assessment of the mechanical condition of the specified objects of study is given. Purpose. To develop an adequate model of the "bone-implant-prosthesis" system and to conduct a computational analysis of deformation in order to predict the strength of the elements. Methods. The finite element method is chosen as the numerical method of analysis. Particular attention is paid to the simulation of contact interaction, taking into account dry friction. The coefficient of friction (adhesion) varied in the calculations, thus simulating different degrees of osseointegration. Results. The obtained distributions of the main characteristics of the deformation process made it possible to find out that under the considered loading conditions, the strength of the bone and implant is preserved. Conclusions. From the point of view of mechanical behavior, both intramedullary and extramedullary implants maintain bone strength and ensure the performance of the prosthesis.

Перспективність методу ендо-екзо-протезування втрачених кінцівок, основою якого є взаємодія пацієнта з протезом через імплантат, з'єднаний з кісткою кукси, значною мірою залежить від забезпечення довговічності функціонування кістки кукси, імплантата та їхнього з'єднання. Наведено методику й аналіз деформацій в біомеханічній системі «кістка – імплантат – протез», вивчено показники механічного стану цих об'єктів дослідження. Мета. Створити розрахункову схему біомеханічної системи «кістка – імплантат – протез» із інтрамедулярним та екстрамедулярним імплантатами та проаналізувати деформування за різних ступенів контакту імплантата й кістки за рахунок варіювання значення коефіцієнта тертя (або адгезії) між ними. Методи. Матеріалом дослідження послужив аналіз релевантної літератури в спеціалізованих джерелах за останні 10 років. Чисельним методом аналізу обрано метод скінченних елементів. Навантаження моделі системи «кістка – імплантат – протез» здійснювали за умови дії сил по її біомеханічній осі. Особливу увагу приділено моделюванню контактної взаємодії з урахуванням сухого тертя, коефіцієнт якого варіювався під час розрахунків і таким чином моделювались різні ступені з'єднання кістки й імплантата. Результати. Отримані характеристики розподілу деформування дозволяють вважати, що з точки зору механічної поведінки як інтрамедулярний, так і екстрамедулярний імплантати за умов навантаження системи без відхилень від біомеханічній осі не змінюють первинну міцність кістки кукси. Висновки. Проаналізовано отримані результати й виявлено, що під час розглянутих умов центрального навантаження кістки кукси вагою тіла людини не має значимих ризиків порушення її міцності, причому для обох типів імплантатів. Ключові слова. Ендо-екзо-протезування, біомеханіка, розрахунковий аналіз, метод скінченних елементів.

Keywords. Endo-ekzo-prosthetics, biomechanics, computational analysis, finite element method

© Sitenko O. M., Larin O. O., Lavinsky D. V., Barbin K. A., 2026

PREVIEW

Introduction

The level of rehabilitation of individuals who have lost limbs depends not only on the biomechanical and biocybernetic properties of prosthetic components but also on how atraumatic and effective the interaction between the patient and the prosthesis is. From this perspective, lower-limb prosthetic methods based on the use of a socket have significant disadvantages: the load is transferred to the musculoskeletal system in an unnatural and traumatic manner — through the soft tissues of the residual limb and other body structures. The residual limb itself is located within a confined, physiologically uncomfortable space, and contact with the socket acts as a kind of "false joint," which results in substantial limitations in prosthesis control. Therefore, proprioceptive and afferent deficits are only minimally compensated even after prolonged training.

The use of endo-exo prostheses (EE prostheses), which are connected and interact with the human body through an implant directly placed either inside the residual bone canal (intramedullary method) or externally on the bone (extramedullary method), makes it possible to avoid these problems. However, long-term experience with EE prosthetic systems has shown [1–5] that adverse mechanical events occur relatively frequently, including fractures of the residual bone or implant, loosening of their fixation, and other complications.

The practice of lower-limb EE prosthetics and the early development of its theoretical foundations indicate the need for data on the distribution of mechanical stresses in the residual bone and implant that occur under loading conditions, as well as an understanding and consideration of their dependence on various factors, such as prosthetic design parameters, rotational moments, operating conditions, and others. This information is also required for further improvement of the "implant – bone" system, for defining requirements for EE prosthesis designs, and for developing prosthetic technologies in general. Direct measurement of stresses using technical methods in this case is extremely difficult, if practically feasible at all. However, such data can be obtained through mathematical modelling methods by performing an appropriate computational analysis of bone–implant interaction under loading conditions. One of its important aspects is the investigation of bone deformation to assess its strength depending on the degree of contact with the implant and other factors.

Purpose: To develop a computational model of the biomechanical "bone–implant–prosthesis" system with intramedullary and extramedullary implants and to analyze deformation under different degrees of implant–bone contact by varying the value of the friction (or adhesion) coefficient between them.

Materials and Methods

The study material consisted of an analysis of relevant literature from specialized sources published over the past 10 years.

The application of computational modelling methods aims to clarify the essence of the term "osseointegration" from the perspective of understanding the mechanism of contact interaction between metal and living bone tissue, since in some sources this phenomenon is interpreted as fusion of the implant with the residual bone. It should be noted that precision in terminology and its correspondence to the essence of phenomena has always been recognized as a necessary condition for solving scientific and other problems, beginning from the stage of their emergence. In our opinion, the widespread use of the term "osseointegrative prosthetics" limits the understanding of the scope of problems by reducing them mainly to medical aspects associated with implant placement. Practical experience has already demonstrated the consequences of neglecting biomechanical, technical, and other important aspects of this method, which should also be considered fundamental to its scientific justification and the development of technologies for practical implementation.

Attention should be paid to the fact that the very concept of "integration," regardless of the field of application, implies the formation of new structures or the functional unification of existing elements into coordinated activity determined by their common existence. However, no data describing such phenomena in biology or physiology, in the sense of integrative processes between non-living and living objects, have been found in the literature.

For example, an oxide film exists on the surface of a titanium intramedullary implant. After the medullary canal is cleared and the implant is placed, osteoblasts form in the microgaps between the bone and implant, actively filling this space with new bone tissue. The new bone tissue (as a part of the original bone) gradually becomes stable and begins to directly and tightly contact the implant surface. This process plays a key role in the formation of adhesion between bone cells and the oxide film of the titanium implant (physical intermolecular interaction between contacting substances). Thus, the metallic implant and living bone do not form any common biological structures that should be considered during model-

PREVIEW

ing, and the concepts of osseointegration or fusion, particularly in relation to this study, can be regarded as conditional. This makes it possible to apply established methods of contact mechanics when modeling the "bone – stump – implant" system.

Implant surface roughness may also enhance mechanical interaction, as surface irregularities create a microrelief that becomes filled with newly formed bone cells.

The vast majority of approaches to mathematical modeling and computational analysis are based on numerical schemes of the finite element method (FEM) [6]. In orthopedics and prosthetics, such studies involve modeling the deformation of human bones under extreme conditions (diseases, fractures, application of limb distraction devices, etc.) [6–13]. It should be noted that published studies concerning EE prostheses [7, 8] consider models that include only the bone and the implant, while the influence of force interaction with the prosthesis is substantially simplified. In addition, insufficient attention has been paid to studying the conditions of contact interaction between the implant and bone, the degree of which is determined by the coefficient of friction or adhesion.

The motivation for selecting the research directions at this stage using mathematical modeling methods and computational analysis of deformation of the biomechanical "bone – implant – prosthesis" system is based on several considerations.

According to the practice of EE prosthetics [3, 5], fractures of both the stump bone and the metallic intramedullary implant (I-implant) may occur in up to 10 % of cases. This is one of the reasons indicating that modeling should provide:

– a general assessment of the safety margin of the "bone–implant" system for different prosthesis designs, separately for each of its components;

– characteristics of the dependence of the strength of the "bone – implant" system on various factors, including the length and area of contact, the type of prosthesis fixation within the bone canal, non-central loading relative to the biomechanical axis, and other factors.

Considering that surgical procedures based on the use of the I-implant are currently the predominant method in clinical practice, these data will allow refinement of indications for this approach based on whether the stump length is sufficient or insufficient, as well as determination of critically acceptable amputation levels at which such prosthetic treatment can be recommended.

The E-implant is used less frequently, although, unlike the I-implant, it does not require removal of the bone marrow. Therefore, in this case, centrifugal blood circulation is preserved along the entire length of the stump bone, which is a necessary condition for preventing degradation of bone tissue. For the design of an E-implant, it is necessary to determine the optimal implant – bone contact area from the standpoint of reliability. This will make it possible, during preoperative planning, to determine the locations of such contacts on the bone surface, as well as individual prosthesis parameters for each patient.

Furthermore, data on the mechanical stresses arising within the system will provide a basis for establishing requirements regarding the accuracy of prosthesis design parameters, developing approaches for optimizing the prosthetic configuration, and defining the necessary instrumentation and technical support for this process.

The study of mechanical stresses associated with destructive risks is also necessary during the design of those EE-prosthesis components that are intended to protect the "implant–bone" system. Examples include a load-relieving module or a rotational mechanism with its control system. Mathematical modeling and computational analysis of deformation will play a decisive role in these developments.

The first stage of computational analysis is the creation of an adequate 3D model of the "bone – implant – prosthesis" system.

The I-implant is a rod (material — titanium oxide) with a supporting head (material — carbon–carbon composite), which is in contact with the resected end of the femoral bone. The E-implant, in terms of its shape, can be conditionally described as a "cup" (material — titanium oxide), into which the stump bone is inserted; its end surface also contacts the supporting head (material — carbon–carbon composite).

The prosthesis model consists of three elements combined according to the principle of functional performance during the support phase: a femoral load-bearing module with a knee joint, a tibial load-bearing module, and an artificial foot. The load-bearing modules are metal tubes, while the foot is a curved leaf spring. A hinged connection between the prosthesis elements was modeled. The geometric 3D models of the "bone – implant – prosthesis" system for the I- and E-implants are presented in Fig. 1.

This model makes it possible to study deformation during all major types of human locomotion; however, this study considers only quasi-static modeling. The difference between this work and, for example, [14], is the introduction of three mutually

PREVIEW

perpendicular forces (Fig. 2), which simulate possible effects of limb bending and torsion.

Within the computational model, several contact zones were identified. The main focus was placed on the interaction zones between the rod of the I-implant and the internal surface of the femoral stump bone, as well as between the E-implant and the external surface of the femoral bone. In these areas, contact was modeled taking into account dry sliding friction with a variable coefficient of friction (or adhesion). Its value varied within the range from 0.2 to 6. Another contact zone occurs at the interface between the femoral resection surface and the supporting head of the implant; however, in this case, the coefficient was constant and equal to 0.5.

The following geometric dimensions were considered in the calculations: femoral segment length — 316 mm; I-implant rod length — 110 mm, rod diameter — 12 mm; upper prosthetic tube length — 100 mm, lower tube length — 400 mm, with internal tube diameters of 30 and 26 mm, respectively; "foot" dimensions — 80 × 200 mm, plate thickness — 7 mm.

The physical and mechanical properties of the model components are presented in the table [14].

The FEM was used as the numerical approach. A tetrahedral element with 48 degrees of freedom was selected as the basic finite element. Mesh refinement was performed in areas with geometric irregularities.

The occurrence or absence of a dangerous situation from the standpoint of structural strength was assessed using failure criteria. For the implant, the von Mises yield criterion was applied. A situation was considered critical when the equivalent stress (stress intensity) reached the yield strength of the material. For bone, the Mohr failure criterion was selected, which is recommended for materials characterized by brittle fracture [15]. The equivalent stress according to the Mohr criterion is determined by the formula:

$$\sigma_{\text{екв}}^{\text{Мор}} = \sigma_1 - \frac{\sigma_p}{\sigma_c}\sigma_3,$$

where σ1 and σ_3 are the first and third principal stresses, σ_p is the tensile strength limit, and σ_c is the compressive strength limit. A situation was considered critical when the equivalent stress (stress intensity) reached the tensile strength limit of the material.

Results

The conducted computational studies showed that under loading along the biomechanical axis, deformation of the structural components occurs in both the intramedullary and extramedullary implant systems. Figures 3 and 4 present the distributions of equivalent stresses in the bone and in the two types of implants for a friction coefficient of 0.2 (corresponding to the initial stage of the adhesion process).

The presented illustrations demonstrate that the most highly loaded regions of both the bone and the implant are the areas of their mutual contact. Therefore, in such computational studies, accurate modeling of the contact interaction between the bone and the implant is an essential step.

When using an I-implant, the most highly loaded area is the region of the bone surrounding the implant canal. In the case of an E-implant, the highest loads occur on the external surface of the bone. It should be noted that, for the considered loading conditions, the stress intensity in the implants does not exceed the yield strength, and the equivalent stress in the bone according to the Mohr criterion also does not reach critical values.

Figure 5 compares the maximum equivalent stress values according to the Mohr criterion as a function of the adhesion coefficient for the intramedullary and extramedullary implants.

Discussion

The obtained results demonstrate that in all analyzed cases, the equivalent stresses in the residual bone do not reach the tensile strength limit. This indicates that under central loading along the biomechanical axis, the bone maintains its structural integrity both with intramedullary and extramedullary implants.

Table

Mechanical properties of the elements of the computational model

Element	Young's modulus, GPa	Poisson's ratio	Yield strength, GPa	Tensile strength, GPa	Compressive strength, GPa
Femur	10	0.35	—	0.80	1.20
Implant rod (titanium oxide)	96	0.36	0.93	—	—
Implant head (carbon–carbon composite material)	180	0.19	—	1	1.40
Prosthesis components (steel)	200	0.30	0.25	—	—

PREVIEW

This conclusion is consistent with findings from other researchers who, using finite element analysis, demonstrated that under moderate physiological loading corresponding to body weight, there is no immediate risk of periprosthetic bone failure; however, the safety factor remains relatively low [16, 17]. In particular, in study [16], the authors compared screw-type and press-fit intramedullary implants for transfemoral amputation under normal gait conditions and found that neither design caused direct bone damage in patients weighing up to 100 kg. At the same time, the patterns of stress distribution and strain energy density differed substantially between the two designs.

A. Mirulla et al. analyzed different implant types and found that peak stress values in the bone and implant were similar for both designs; however, the press-fit type produced larger areas of elevated stress in the distal femoral region [17]. In our study, the fundamental similarity in the mechanical behavior of intramedullary and extramedullary implants under central loading was demonstrated (Figs. 3 and 4). However, a different pattern of dependence of peak equivalent stresses on the degree of bone – implant contact was identified (Fig. 5).

A key aspect of the computational analysis was the modeling of bone–implant contact interaction using Coulomb's dry friction law with a variable friction (adhesion) coefficient. This approach is widely used in biomechanical studies of cementless and osseointegrated implants [16, 18, 19].

In most studies involving the modeling of transfemoral osseointegrated systems, a fixed friction coefficient within the range of 0.3–0.5 is assumed [16, 18], which corresponds to experimental data for the contact between trabecular bone and

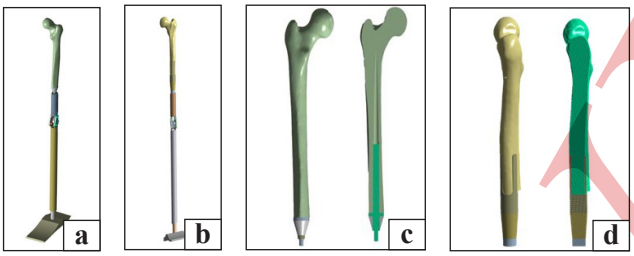


Fig. 1. Geometric models: a) "bone–implant–prosthesis" with an I-implant; b) "bone – implant – prosthesis" with an E-implant; c) model of the I-implant; d) model of the E-implant.

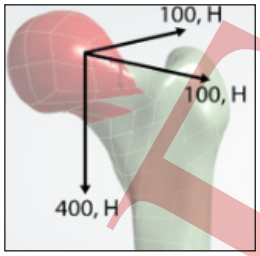


Fig. 2. Bone loading scheme.

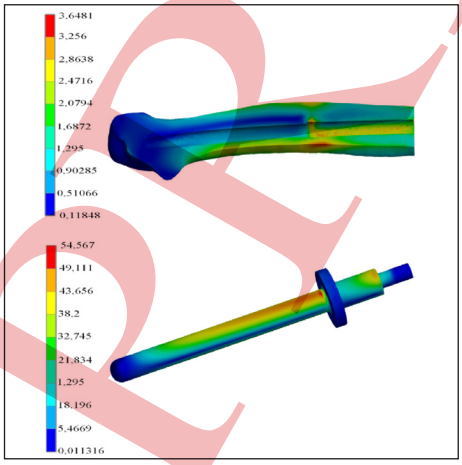


Fig. 3. Distribution of equivalent stresses (MPa) in the stump bone and the I-implant.

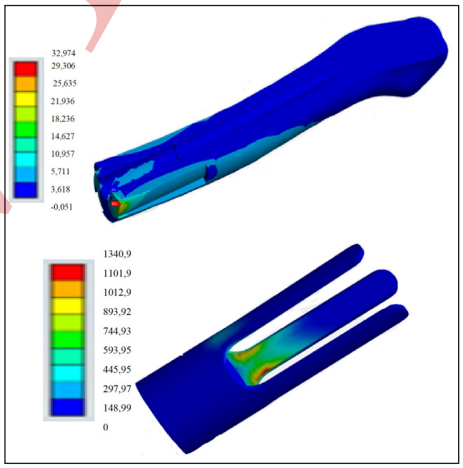


Fig. 4. Distribution of equivalent stresses (MPa) in the stump bone and the E-implant.

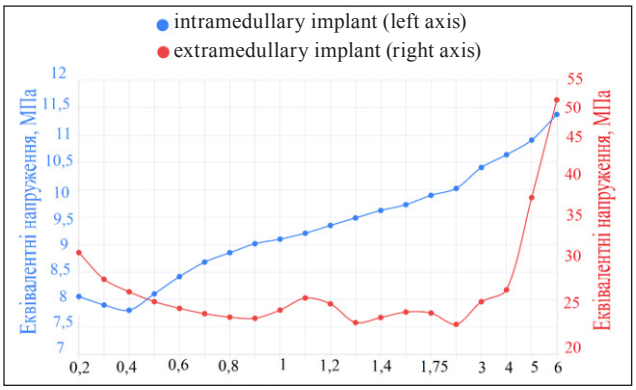


Fig. 5. Comparison of maximum equivalent stresses according to the Mohr criterion depending on the value of the adhesion coefficient in the bone when using intramedullary (blue) and extramedullary (red) implants.

PREPRINT

porous-coated metal surfaces. However, the actual bone – implant interface has a complex and dynamic nature: the degree of contact changes over time due to osseointegration processes, and the classical Coulomb friction law does not fully reflect the nonlinear behavior of the interface before and after completion of osseointegration [19]. Therefore, varying the friction coefficient in our study within the range of 0.2 to 6 allowed simulation of different stages of contact interaction (from the early postoperative period to a state of complete osseointegration) and assessment of the sensitivity of the system's mechanical response to this parameter.

For the I-implant, the graph of maximum equivalent stresses according to the Mohr criterion demonstrates a pronounced minimum at an adhesion (friction) coefficient of 0.4, followed by an almost monotonic increase. For the E-implant, the curve remains nearly constant up to a coefficient value of 4, with only minor fluctuations (up to 0.5 MPa), whereas a sharp increase occurs only beyond this value. The slight increase in stresses at the initial friction coefficient values may be associated with substantial relative displacements between the bone and implant, whereas the effect of a sharp increase at high coefficient values requires further investigation and possibly the application of alternative interaction models under conditions of significant adhesion. This finding is consistent with the conclusions of the authors of study [19], who proposed a modified Coulomb law incorporating normal cohesion and tangential interface debonding and demonstrated that interface behavior under partial osseointegration differs substantially from classical assumptions of either complete bonding or complete sliding.

Analysis of the stress distribution demonstrated that the most highly loaded regions in both cases are the areas of direct bone–implant contact. For the I-implant, this corresponds to the bone region surrounding the medullary canal, whereas for the E-implant it is the external surface of the femur within the area enclosed by the "cup". Similar localization of peak stresses has been reported by other authors for osseointegrated transfemoral implants: stress concentrations occur near the proximal end of the intramedullary stem and in the region of osteotomy [16, 18]. This confirms the importance of accurate modeling of contact interactions in such analyses.

In addition to the acute risk of fracture, long-term alterations in local stress distribution within the periprosthetic region may induce bone remodeling processes due to the "stress shielding" effect, clinically manifested as osteopenia, cortical bone thinning, and an increased risk of periprosthetic fracture [12, 13, 20]. Study [20] described a clinical case of periprosthetic fracture in a patient with transfemoral amputation and a custom femoral implant, in which extensive bone resorption caused by stress shielding was identified as one of the main causes of structural failure. In our study, central loading along the biomechanical axis did not reveal signs of substantial stress redistribution that could lead to such phenomena; however, further modeling of eccentric and dynamic loading conditions is required for an adequate assessment of this risk.

From a methodological perspective, an important feature of the present study is the application of the Mohr criterion for assessing the strength of bone tissue as a brittle material [15] and the von Mises yield criterion for evaluating the mechanical integrity of the metallic implant, which corresponds to current recommendations for modeling tissues and materials with fundamentally different mechanical properties. Furthermore, unlike most published studies [7, 8, 16–18], which analyze either only the "implant–bone" system or simplify the mechanical interaction with the prosthesis, our model incorporates the complete "bone–implant–prosthesis" assembly with three mutually perpendicular forces, allowing simulation of bending and torsional effects of the limb. This provides a methodological foundation for further expansion of the computational model to dynamic loading conditions and different human locomotion patterns.

A systematic review by the authors of study [21], which included 17 *in vitro* and *in silico* investigations of osseointegrated transfemoral prostheses, indicates the absence of standardized methodological approaches for evaluating primary stability, stress shielding, and stress concentration. This limits the ability to directly compare quantitative results between different studies and supports the need for developing individualized models with clearly defined assumptions and boundary conditions. In this context, the methodological value of our approach lies in the possibility of using individualized computational analysis results to justify indications and implantation parameters for specific patients, which is particularly relevant for the clinical implementation of endo-exo prosthetic systems in the context of post-war rehabilitation in Ukraine [4, 5].

The limitations of this stage of the study include the quasi-static nature of the analysis, consideration of only central loading along the biomechanical axis, and the assumption of an idealized geometry of the residual bone. Further research should focus

PREPRINT

on real-life prosthesis operating conditions, including the incorporation of dynamic factors, force moments, and non-standard loading conditions (such as falls and stumbling) [17], as well as modeling different stages of contact interaction using nonlinear friction and cohesion models [19].

Conclusions

Analysis of available literature regarding the current level of development of the scientific foundations of the EE-prosthetic method and its clinical application indicates the need for further investigation, justification of its advantages, and mitigation of associated risks.

In this study, 3D models of the biomechanical systems "bone–implant–prosthesis" were developed for both intramedullary and extramedullary implant configurations. At this stage of the research, a computational analysis of deformation of the "bone – implant – prosthesis" biomechanical system was performed under idealized loading conditions, namely along the biomechanical axis. The finite element method was used as the numerical approach. The conditions of contact interaction between the bone and implant were defined based on a dry friction model. A series of simulations was performed, in which different stages of contact interaction were modeled by varying the adhesion coefficient.

The principal conclusion derived from the obtained results is that, under the considered conditions of central loading corresponding to body weight, no significant risks of loss of residual bone strength were identified for either type of implant.

The significance of the developed analytical methodology also lies in the possibility of justifying decisions regarding implant application in individual patients based on patient-specific results of corresponding computational analyses.

Further research should be directed toward realistic prosthesis operating conditions, including consideration of dynamic loading factors affecting the biomechanical EE-prosthesis system (primarily moments of force associated with prosthesis geometry and rotational moments in the horizontal plane, as well as non-standard loading conditions that may occur during use). Further studies should also examine alternative models and stages of contact interaction within the "implant–residual bone" system. Mathematical models may provide the basis for the development and implementation technologies of such prosthetic components as load-relief modules, rotational mechanisms with control systems, and other structural elements.

In our opinion, it is also necessary to investigate the potential risk of adhesion failure caused by temperature-related effects transmitted through the metallic implant (for example, during exposure of a prosthesis user to cold environments).

Conflict of interest. The authors declare no conflict of interest.

Prospects for further research. Future studies should focus on several aspects, including incorporation of the dynamic component of loading into the "bone – implant – prosthesis" system, as well as analysis of the influence of thermal deformations on the strength and integrity of the bone–implant connection (adhesion).

Funding information. The study received no external funding.

Author contributions. Sitenko O. M. — general formulation of the research problem, collection and analysis of information sources, participation in discussion and interpretation of findings, manuscript preparation; Larin O. O. — participation in the development of mathematical models, analysis of computational results, critical revision and editing of the manuscript; Lavinskyi D. V. — critical analysis of information sources, participation in the development of mathematical models, analysis of computational results, manuscript preparation and editing; Barbin K. A. — performance of computational studies and analysis of computational results.

References

1. Hoyt, B. W., Walsh, S. A., &Forsberg, J. A. (2020). Osseointegrated prostheses for the rehabilitation of amputees (OPRA): results and clinical perspective. *Expert review of medical devices, 17*(1), 17–25. https://doi.org/10.1080/17434440.2020.1704623
2. Evans, A. R., Tetsworth, K., Quinnan, S., & Wixted, J. J. (2024). Transcutaneous osseointegration for amputees. *OTA International, 7*(2S). https://doi.org/10.1097/oi9.0000000000000326
3. Sreedharan, S., Gray, S., & Bruscino-Raiola, F. (2021). Osseointegrated prostheses for lower limb amputees: a review of complications. *Australasian journal of plastic surgery, 4*(1), 55–61. https://doi.org/10.34239/ajops.v4n1.199
4. Haluzynskyi, O., Linenko, O., Bondarenko, S., Nikolov, M., & Maltseva, V. (2024). Osseointegrative prosthetics: Opportunities, challenges, and prospects for its application in the rehabilitation of patients with amputated limbs (Literature review). *Orthopaedics traumatology and prosthetics,* (3), 86–97. https://doi.org/10.15674/0030-59872024386-97
5. Sitenko, O., & Sitenko, T. (2024). Biomechanical aspects of Endo-exo-prosthetics of the lower limbs. *Orthopaedics traumatology and prosthetics,* (2), 69–75. https://doi.org/10.15674/0030-59872024269-75
6. Du, X., Zhao, X., Lei, J., & Zhang, G. (2024). Finite element modeling and injury criteria investigation for the lower leg of the Chinese human body under impact loads. *Theoretical and applied mechanics letters, 14*(4), 100547. https://doi.org/10.1016/j.taml.2024.100547
7. Tuninetti, V., Fuentes, G., Oñate, A., Narayan, S., Celentano, D., García-Herrera, C., ... & Valle, R. (2024). computational shape design optimization of femoral implants: towards efficient forging manufacturing. *Applied sciences, 14*(18), 8289. https://doi.org/10.3390/app14188289
8. Kumar, P., & Jain, N. K. (2021). Finite element analysis of femoral prosthesis using ti-6al-4 V alloy and TiNbZrTaFe high entropy alloy. Materials today: *Proceedings,* 44, 1195–1201. https://doi.org/10.1016/j.matpr.2020.11.239
9. Johnson, J. E., Brouillette, M. J., Miller, B. J., & Goetz, J. E. (2023). Finite element model-computed mechanical behavior

PREPRINT

of femurs with metastatic disease varies between physiologic and idealized loading simulations. *Biomedical engineering and computational biology,* 14, 11795972231166240. https://doi.org/11795972231166240
10. Rostamian, R., Silani, M., Ziaei-Rad, S., Busse, B., Qwamizadeh, M., & Rabczuk, T. (2022). A finite element study on femoral locking compression plate design using genetic optimization method. *Journal of the mechanical behavior of biomedical materials,* 131, 105202. https://doi.org/10.1016/j.jmbbm.2022.105202
11. Ceddia, M., Solarino, G., Cassano, G. D., & Trentadue, B. (2023). Finite element study on stability in the femoral neck and head connection to varying geometric parameters with the relates implications on the effect of wear. *Journal of composites science, 7*(9), 387. https://doi.org/10.3390/jcs7090387
12. Xiao, Z., Wu, L., Wu, W., Tang, R., Dai, J., & Zhu, D. (2023). Multi-scale topology optimization of femoral stem structure subject to stress shielding reduce. *Materials, 16*(8), 3151. https://doi.org/10.3390/ma16083151
13. Sun, X., Curreli, C., &Viceconti, M. (2024). Finite element models to predict the risk of aseptic loosening in cementless femoral stems: a literature review. *Applied sciences, 14*(8), 3200. https://doi.org/10.3390/app14083200
14. Lavinsky, D., Larin, O., Sitenko, O., Tomashevskyi, R., & Barbin, K. (2025). Biomechanical analysis of bone-implant interaction in a femoral prosthetic reconstruction. *IFMBE Proceedings,* 483–491. https://doi.org/10.1007/978-3-032-06494-3_48
15. Li, S. (2024). Rational implementation of the Mohr criterion in its general form. *International Journal of mechanical sciences,* 279, 109449. https://doi.org/10.1016/j.ijmecsci.2024.109449
16. Tomaszewski, P. K., Verdonschot, N., Bulstra, S. K., & Verkerke, G. J. (2010). A comparative finite-element analysis of bone failure and load transfer of osseointegrated prostheses fixations. *Annals of biomedical engineering, 38*(7), 2418–2427. https://doi.org/10.1007/s10439-010-9966-9
17. Mirulla, A. I., Di Paolo, S., Di Simone, F., Ingrassia, T., Nigrelli, V., Zaffagnini, S., & Bragonzoni, L. (2020). Biomechanical analysis of two types of osseointegrated transfemoral prosthesis. *Applied Sciences, 10*(22), 8263. https://doi.org/10.3390/app10228263
18. Xu, W., Xu, D. H., & Crocombe, A. D. (2006). Three-dimensional finite element stress and strain analysis of a transfemoral osseointegration implant. *Proceedings of the institution of mechanical engineers, Part H: journal of engineering in medicine, 220*(6), 661–670. https://doi.org/10.1243/09544119JEIM84
19. Immel, K., Nguyen, V. H., Haïat, G., & Sauer, R. A. (2022). Modeling the debonding process of osseointegrated implants due to coupled adhesion and friction. *Biomechanics and modeling in mechanobiology, 22*(1), 133. https://doi.org/10.1007/s10237-022-01637-7
20. Wik, T. S., Foss, O. A., Havik, S., Persen, L., Aamodt, A., & Witsø, E. (2010). Periprosthetic fracture caused by stress shielding after implantation of a femoral condyle endoprosthesis in a transfemoral amputee — a case report. *Acta orthopaedica, 81*(6), 765–767. https://doi.org/10.3109/17453674.2010.533937
21. Galteri, G., & Cristofolini, L. (2023). In vitro and in silico methods for the biomechanical assessment of osseointegrated transfemoral prostheses: a systematic review. *Frontiers in bioengineering and biotechnology,* 11, 1237919. https://doi.org/10.3389/fbioe.2023.1237919

The article has been sent to the editors	Received after review	Accepted for printing
22.04.2026	04.05.2026	05.05.2026

COMPUTATIONAL ANALYSIS OF THE DEFORMATION OF THE BIOMECHANICAL SYSTEM «BONE – IMPLANT – PROSTHESIS»

O. M. Sitenko [1], O. O. Larin [2], D. V. Lavinsky [2], K. A. Barbin [2]

[1] Sytenko Institute of Spine and Joint Pathology National Academy of Medical Sciences of Ukraine, Kharkiv

[2] National Technical University "Kharkiv Polytechnic Institute", Kharkiv, Ukraine

✉ Oleksandr Sitenko, PhD in Biomechanics and Prosthetics: sitenko@meta.ua

✉ Oleksiy Larin, Prof. in Tech. Sci.: Oleksiy.Larin@khpi.edu.ua; https://orcid.org/0000-0002-5721-4400

✉ Denis Lavinsky, Prof. in Tech. Sci.: Denys.Lavinskiy@khpi.edu.ua; https://orcid.org/0000-0002-1380-3131

✉ Kostiantyn Barbin, PhD-student: Kostiantyn.Barbin@infiz.khpi.edu.ua; https://orcid.org/0009-0003-1152-4950

PREVIEW