Stress Reduction in the Residual Limb of a Transfemoral Amputee Varying the Coefficient of Friction

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Abstract

A finite element computer model was used to validate the relationship between coefficient of friction (COF) and the shear stresses and pressures obtained at the socket/residual limb interface for four individuals with transfemoral amputation. This model simulated the process of donning the socket and application of loads during the stance phase of gait cycle. The residual limb was modeled as a solid multilayer (skin, fat and muscle) with hyperelastic properties. An experimental design was made to calculate the significance of hairiness, sweating, and texture factors in measuring the COF between polypropylene and skin with the help of a sclerometer; values were obtained from 0.188 to 0.545. It was demonstrated that the surface texture is significant when measuring the COF. On the results of the simulation, a horizontal strip where the greatest stresses and pressures occurred was identified; in this strip, the COF was varied from 0.2 to 0.6 to observe its effect on the distribution of stresses. The average shear decreased up to 25%, whereas the pressures did not vary significantly and the results do not exceed the value of 70 kPa, from which values are potentially harmful to the skin. (*J Prosthet Orthot*. 2014;26:205–211.)

KEY INDEXING TERMS: socket-limb residual interface, finite element analysis, dynamic simulation, stress distribution

The socket is the element of the prosthesis that is in direct contact with the patient. For this reason, its design is fundamental for the patient's perception of comfort with his/her prosthesis. Several authors have demonstrated that shear stresses and excessive pressures in the socket-residual limb interface are responsible for damage to the patient's soft tissue, whereas the normal and tangential forces applied to the residual limb when it interacts with the socket produce circulatory problems.[1–3] However, the worst conditions that lower-limb amputees experience according to Meulenbelt et al.[4] are pressure ulcers, followed by infections and wounds.

The experimental pressure and stress measuring methods in the socket-residual limb interface provide only approximations to the real values because of the complexities associated when implementing these methods and do not account for all of the surfaces in contact.[5,6] The finite element method serves as a complement to the experimental measurements while these validate the method.[7,8] To date, the models used for transfemoral amputees are based on suppositions and simplifications that do not recreate donning the socket and deformation of soft tissues.[9] There are various factors that affect the distribution of stresses on the residual limb, such as variability in the anatomy of persons, skin types, sweating, presence or absence of hair, and others. The aim of this study was to obtain the stress distribution on the residual limb for a transfemoral amputee varying the coefficient of friction (COF) in the socket-residual limb interface to

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1. aid the designer in the design process;
2. decrease stress values and reduce injuries to patients, increasing their quality of life and facilitating their reintegration into society; and
3. assist the prosthetist to an easier adaptation of the patient to his/her prosthesis.

Coefficient of friction was selected as a study variable because it is easily quantifiable and measurable. So far, there have been no studies that aim to answer the question posed in this research.

This study was conducted on four individuals with different physiological characteristics, making a distinction in the physical and mechanical properties of the constitutive layers of the residual limb (skin, fat, and muscle).[10,11] A dynamic explicit simulation was performed to represent the socket donning process and initial support during the gait. The model was divided into three analysis stages, which included the donning phase. This stage accounted for the vertical motion of the socket during donning from a configuration in which there is no contact with the residual limb to its final configuration, in which it contains all of the soft tissue and bone. The relaxation phase is considered as a transition phase between the donning phase and the exertion of the load phase, when the most critical load of the gait phase is applied.

It has been demonstrated that the COF directly affects the magnitude of stresses generated by the socket on the residual limb.[12] For this reason, the COF between the residual limb and the socket was varied from 0.5 to 1.0 to determine the stress and pressure distributions. The stress concentration zone was identified from these values where the COF was altered to determine if the stresses could be reduced. An experimental validation was carried out to determine the influence of socket surface texture on the COF in the socket-residual limb interface to prove the viability of optimizing the socket surface. The factors accounted for in the experimental measurements were

Table 1. Subject information

Code of the subject	Height, m	Body weight, kg	Residual limb length, m	Normal F, N	Shear F, N	Moment, N m
Sub 1	1.53	53.2	0.24	567	207	100
Sub 2	1.75	75.0	0.24	807	295	172
Sub 3	1.65	88.7	0.30	961	352	167
Sub 4	1.63	63.5	0.29	682	249	119

hirsuteness, skin sweatiness, and the surface texture of the polypropylene because their significance has been demonstrated in previous works. Upon completion, it was determined that the COF considerably affects shear stresses because the magnitude was reduced up to 25%.

METHODS

CHARACTERIZATION OF THE SUBJECTS

Participating in the study were four individuals with transfemoral amputation who varied in age, sex, and amputation length. In the absence of a formal institutional review board, the principles outlined in the Declaration of Helsinki were followed. Informed consent was verbally obtained from research subjects before they were involved with the study. The subjects' data are presented in Table 1. Forces and moments were calculated from data obtained from the gait laboratory at Maria Cano University.

MODEL GEOMETRY

To produce the models, a three-dimensional scanner was used for the socket and the residual limb, whereas the bone was modeled from a reconstructed computed tomography (CT) image of each individual. The parameters used for the scan were as follows: Siemens Emotion 6 Scanner, 112 mAs, 130 kV, 512×512 pixel matrix, pixel size 0.758 mm, gantry tilt 0.00, and slice increment 1 mm. The components were assembled according to each subject's biometrical configuration, and the anatomical positions were obtained from the CT. To date, these techniques are being used to design and fit the socket.[13] The patient's residual limb model was produced as a multilayer model including the skin, fat, and muscle. According to Geerligs[10] and Portnoy et al.,[14] the skin and fat have a thickness of 2 and 4 mm, respectively, and the remaining volume corresponds to the muscle.

FINITE ELEMENT MODEL

For solving the model, ABAQUS version 6.9 was used. A dynamic explicit simulation was the solution strategy because of large displacements of the socket and the large strain of skin and fat. Also, the capacities of this type of analysis allow the simulation of the hyperelastic and damping behavior of the soft tissues and define complex contact conditions. A tetrahedron C3D4 element was used to mesh all elements of simulation, and 310320 was the average of elements used; default convergence criteria of the simulation software were used.

The initial considerations taken into account when producing the model included the amputee's donning of the socket, shown in Figure 1A, and its analysis under the most critical loads during gait. The femur is fixed in the acetabulum, as shown in Figure 1B, to restrict movement in any direction and rotation. This attachment occurs at the initial support; there is no relative displacement of the acetabulum with the hip. Owing to action of the muscles on the femur, this restriction of movement makes the transmission of load to the hip joint possible.

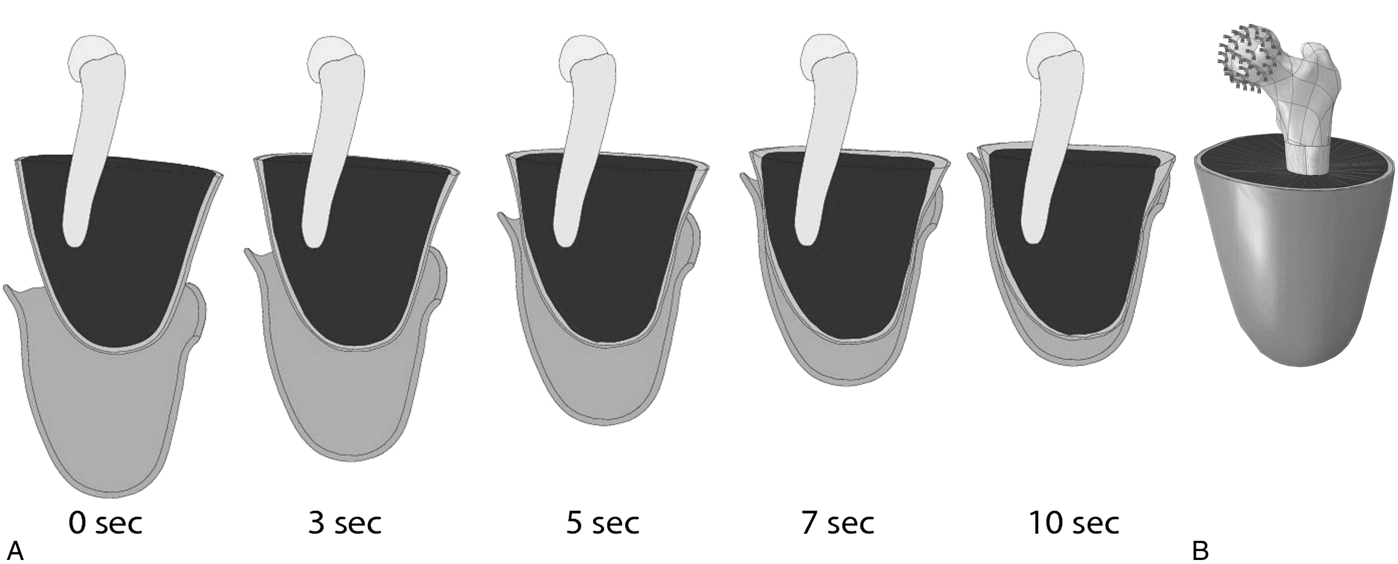

Figure 1. A, Fitting of the socket; B, fixing of the acetabulum.

The donning procedure was divided into three phases to produce an accurate model of the procedure.

DONNING

In this step, the amputees facilitate the fitting (sliding) of the socket using a thin sock or a liner that reduces the friction coefficient and therefore allows for smoother donning. This stage was considered as an independent stage because it has specific characteristics differing from the other two stages. Moreover, the greatest relative socket movement with respect to the residual limb occurs during this stage.

During the donning step, the soft tissue must deform considerably to be contained by the socket, which has a specific geometry. Figure 1 shows the socket fitting procedure.

RELAXATION

In this step, a 3-second interval was established to allow the soft tissue to adjust to the socket's geometry and to dissipate the changes occurring as waves or creases in the skin because of the considerable deformations that occurred during the motion step.

LOAD

This load is located on the rigid part of the socket, where it joins the rod of the thigh. It can reach magnitudes of up to 120% of the individual's body weight and is produced at the beginning of the initial support phase and at the end of the initial and final support phases, with a horizontal load representing 7% to 10% of the individual's body weight. This step is applied after the relaxation phase and lasts for 0.1 seconds because it is the average duration of this stance phase. In this phase, the relative motion between the socket and residual limb is not considered.

The main consideration accounted for when this investigation was performed was the fact that soft tissues, such as skin, fat, and muscle, were treated as hyperelastic, homogeneous, and isotropic materials. Using the Mooney-Rivlin generalized deformation (Equation 1), the constitutive parameters C_{10}, C_{11}, and D_1, shown in Table 2, were introduced into the ABAQUS version 6.9.2 software.

$$w = C_{10}(I_1-3) + C_{11}(I_1-3)(I_2-3) - \frac{1}{D_1}(J-1)^2. \quad (1)$$

The bone and the polypropylene of the socket were assumed as lineal, homogeneous, and isotropic materials. The complete properties of the materials used in the simulation are shown in Table 2.[12,14–17]

In addition, reference models were produced for each individual by varying the COF from 0.5 to 1.0 to determine the zones where stresses were concentrated. Once the area of interest was identified (Figure 2), the socket was modified to allow for the change of the COF on a 10-cm-wide horizontal strip located 2 cm below the ischial support. This change was made to observe the effect of varying COF over stress distribution. The rest of the model had a global COF set to 0.9.[14]

EXPERIMENTAL DESIGN

A 2^k factorial type experiment was performed with three factors (hirsuteness, sweatiness, and texture) in two levels (high and low). To test the high hirsuteness level, the subject's arm was not altered; for the low hirsuteness level, the test area was shaved. The high sweatiness level was tested with an artificial sweat solution prepared according to the NTC 5221 standard; a spray in the area of interest was applied. For the low level, the skin was tested in dry conditions. Finally, to test the high texture levels, the polypropylene coatings of the indenters were molded with a nylon stocking, simulating the socket fabrication process; for the low level, the surface of the coatings was left smooth. The experiment design results required eight measurements, with a replica (16 measurements total) to increase the reliability of the results.

The primary roughness profiles of the coatings for the indenters used in the experiment were obtained with a Mitutoyo SV-3000 M surface roughness tester, distinguishing the surface topography of the coatings with high and low texture. The measurements where performed at a speed of 0.1 mm/s, with a 0.5 μm pitch and an 8 mm test length.

RESULTS

SIGNIFICANCE OF THE FACTORS

To determine the COF between the polypropylene and the skin, pertinent measurements were taken with a sclerometer on the same individual's moisture levels (with no clinical conditions on the skin) to prevent skin tone, body mass index, physical activity, and all other factors inherent in the individual

Table 2. Properties of the materials

Material		Young modulus, GPa	Poisson coefficient	C_{10}, kPa	C_{11}, kPa	D_1, MPa^{-1}
Elastic	Polypropylene socket[17,24]	1.5	0.3	NA	NA	NA
	Cortical bone[15]	15	0.3	NA	NA	NA
Hyperelastic	Skin[14,16]	NA	NA	9.400	82	0
	Fat[14,16]	NA	NA	0.143	0	70.2
	Muscle[14,16]	NA	NA	8.075	0	1.243

NA indicates not applicable.

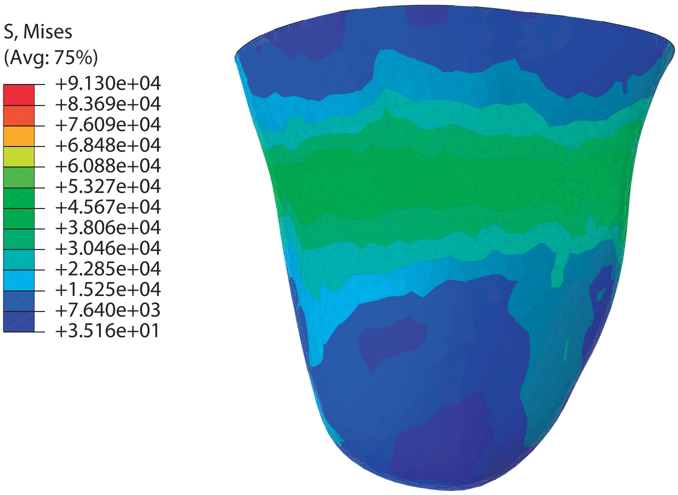

Figure 2. Residual limb Von Mises surface stress distribution.

from affecting the results. Other factors that could not be controlled included relative humidity and temperature; because of these, the tests were performed on the same day. Also, the equipment was calibrated before use to ensure that the data were as precise as possible.

A total of 16 measurements were taken, and the value of the COF was calculated for each one. These are presented in Table 3.

A significance analysis of the studied factors was accomplished with MINITAB 16 software. For a value to be significant, its p value must be below 0.05; for this experiment, p values of all the factors were less than 0.05.

REFERENCE MODEL STRESSES

The shear stresses presented a high concentration gradient that is not desirable because it increases the occurrence of soft-tissue injuries. The highest stress concentrations occurred on a 10 cm-wide strip located 2 cm below the ischial support. It must be noted that this stress distribution was consistent for all the study subjects.

Table 3. Values of the friction coefficient

Measurement	Friction coefficient	Replica	Friction coefficient	Factors		
				Hirsuteness	Sweatiness	Texture
1	0.361	9	0.240	Low	High	Low
2	0.522	10	0.545	Low	Low	Low
3	0.349	11	0.345	High	Low	Low
4	0.229	12	0.232	High	High	High
5	0.188	13	0.186	High	High	Low
6	0.266	14	0.396	Low	Low	High
7	0.242	15	0.270	High	Low	High
8	0.261	16	0.250	Low	High	High

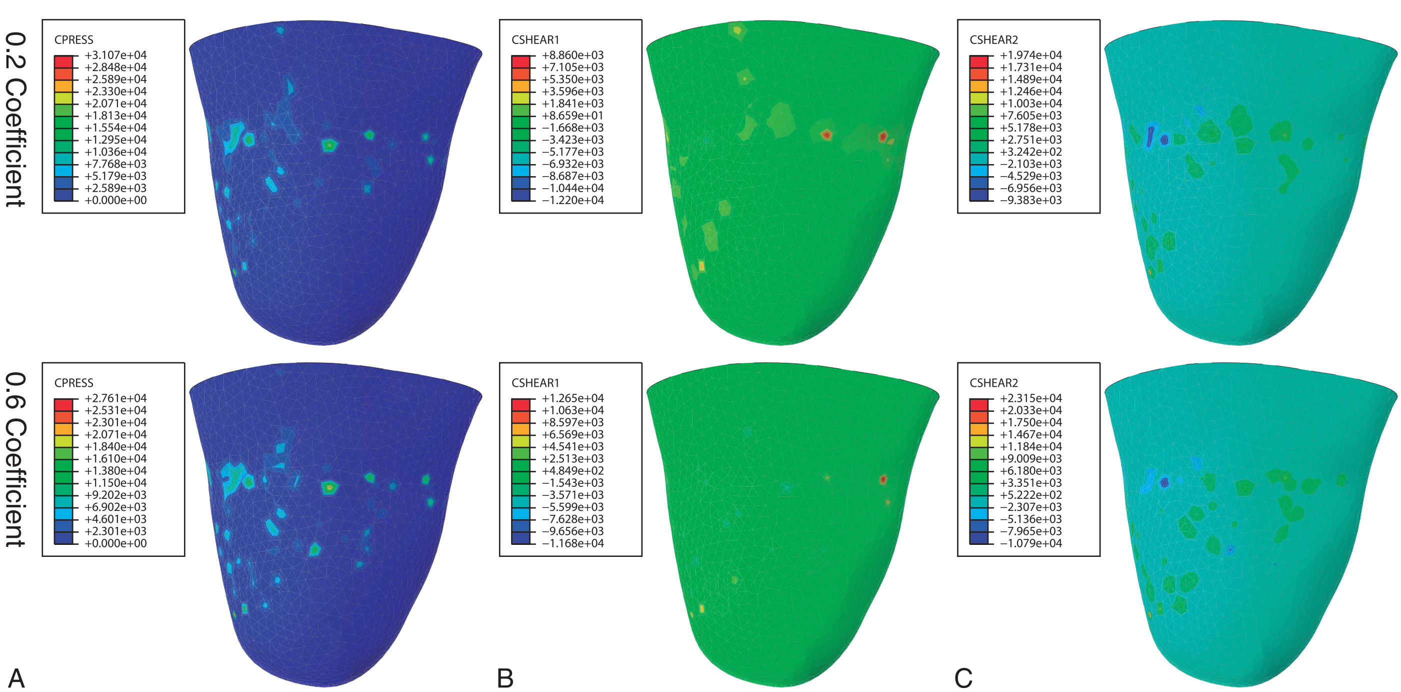

Figure 3. Stress distribution on the reference model for subject.

Table 4. Stress reduction percentage

Code of the subject	Average pressure, %	Average shear, %
Sub 1	4.75	20.69
Sub 2	4.58	18.67
Sub 3	7.39	14.65
Sub 4	2.68	25.61

OPTIMIZED MODEL STRESSES

After modifying the socket in the previously identified strip, the simulations were solved with a fixed (0.9) global COF between the skin and the polypropylene[14] and a variable strip COF ranging from 0.2 to 0.6. The shear stress distributions obtained on the skin for a strip COF of 0.2 was similar to that shown in Figure 3.

The results for the optimized model are presented in Figure 4A; there is a small variation in the average pressures after modifying the COF.

The optimized model was considered as the definitive component (with a strip COF of 0.2 and a global COF of 0.9) because it produced lower values for the resulting shear stress. A comparison of the average pressure and shear stress for the optimized model and the reference model without the strip with a global COF of 0.9 in the entire socket-residual limb interface is presented in Table 4.

DISCUSSION

SIGNIFICANCE OF FACTORS

The data analysis served to identify the significance of the factors in the experiment. Particularly, the surface roughness increased the apparent contact area and therefore reduced the maximum pressure and shear stress in the material.[18] The three factors studied (hirsuteness, sweatiness, and texture) account for 90.56% of the data (R^2). This study presents the statistical analysis that determines the influence of the socket's surface texture on the COF between the polypropylene and the skin, which had not been previously ascertained.

REFERENCE MODEL STRESSES

Shear stress values tend to be linear and proportional to the COF. This behavior is similar to the one reported by Li et al.[19] and Zhang et al.[9] for a transtibial amputee and Zhang et al.[12] for a transfemoral amputee.

The average pressures did not portray a similar behavior to that of the shear stresses. These show a more stable tendency with no significant change when modifying the COF. Zhang et al.[9] reported a different behavior from the one presented in this work. However, it must be noted that the model used in that investigation was simpler; soft tissues were treated as elastic and there have been no further studies that validate the tendency or the results. The shear stresses presented a high concentration gradient, which may be attributed to the use of hyperelastic materials in the simulation that better accommodate the socket's geometry. It has been previously shown that skin breakdown associated with formation of skin lesions occurs when applying shear stresses higher than approximately 70 kPa on the skin surface[20]; the results obtained are under this rank.

OPTIMIZED MODEL STRESSES

The optimized model shows a similar tendency to that of the reference model, with a shear stress and pressure concentration in the same location forming a horizontal strip below the ischial support. Figure 4B presents the same behavior of the reference model, which corroborates the concept that the COF directly affects the value of the resulting shear stresses.

When using the optimized model with the strip, considerable reductions were obtained for the stresses and pressures on the skin in the area in contact with the socket. The 25%

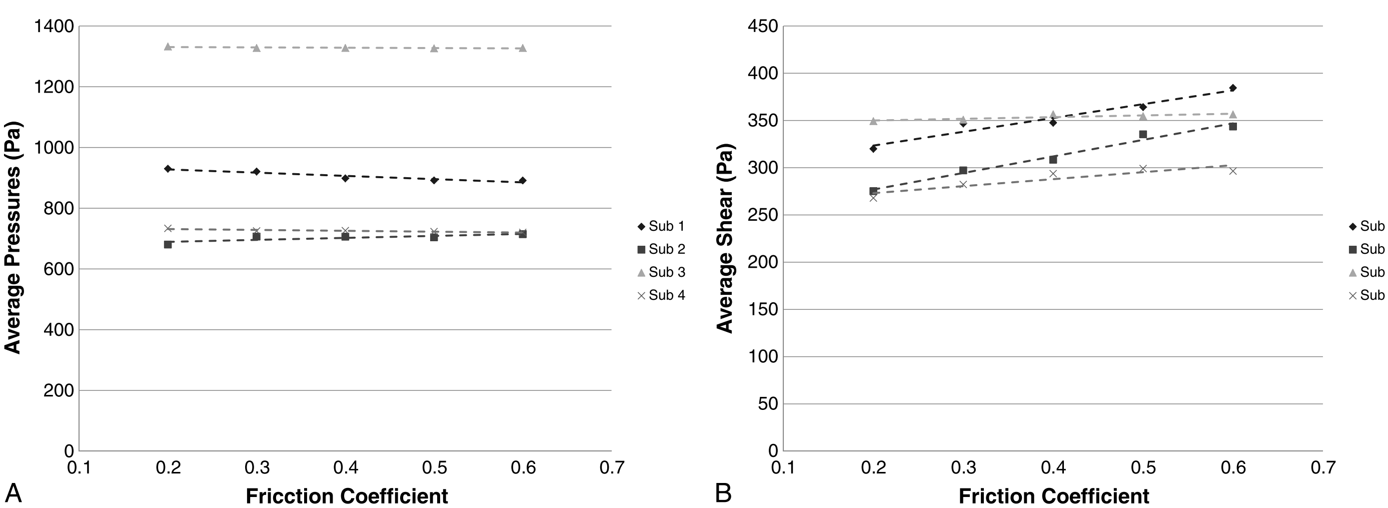

Figure 4. A, Average pressures; B, average shear stress versus friction coefficient (reference model).

reduction is expected to diminish injuries to patients and discomfort, thus increasing their level of comfort.

One of the main limitations in this type of study is obtaining an adequate characterization of the tissues composing the residual limb because existing models involve simplifications with behaviors veering off from real ones.[21] Moreover, it is recommended to include a larger number of subjects to perform a more adequate statistical analysis of the data.

The most recent international studies on transfemoral amputees date back to the 1990s. Because of recent development of computational resources and the evolution of soft tissue models,[21–23] new simulations involving these advancements are needed to better understand the socket-residual limb interaction and propose optimized designs to improve amputees' perception of comfort.

CONCLUSIONS

The selection of soft tissues (skin, fat and muscle) as hyperelastic materials proved to be a valid procedure, as these behaved like the real materials. The results are strictly comparable with those found in similar works.[2,23,24] Also the deformation process of the real tissue is physically compatible to that observed in the simulation.

A horizontal strip was identified below the ischial support, where the pressures and shear stresses concentrate. These large stress concentrations occurred because the soft tissues were treated as hyperelastic materials because these more effectively duplicate the socket geometry when they deform.

It was demonstrated that the COF between the polypropylene and the skin directly affects the value of the shear stresses, upon which the socket was modified by reducing the COF in the proposed horizontal strip. When the shear stresses are decreased, the risk of soft tissue damage is reduced. These results are consistent with investigations.[8,9,12,24]

The multilayer model is the most adequate to mimic stress distribution in the residual limb with respect to the entire model because the physical properties of the soft tissues are differentiated. This allows the simulation to deform in a way similar to the real phenomenon.

The COF between the polypropylene and the skin was modified by varying the surface texture of the coatings. From the analysis of the results, it can be concluded that the texture indeed affects the average value of the COF. Therefore, a detailed study is proposed which deals with the influence of the tribological parameters Ra, Rq, HSC, and others measured on the internal surface of the socket on the COF.

The average shear stresses were reduced up to 25% after varying the COF on the strip. This proves that when changing the surface texture on the proposed strip, a considerable improvement of the stress distribution can be obtained.

The models developed serve as a basis for future studies using more complex models that account for the interaction between the muscle and the bone and simulate stress distribution in the residual limb when negative pressure is applied to the sockets with a suction valve.

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REFERENCES

1. Bennett L, Kavner D, Lee BK, Trainor FA. Shear vs pressure as causative factors in skin blood flow occlusion. *Arch Phys Med Rehabil* 1979;60:309–314.
2. Li W, Liu XD, Cai ZB, et al. Effect of prosthetic socks on the frictional properties of residual limb skin. *Wear* 2011;271:2804–2811.
3. Zhang M, Turner-Smith AR, Roberts VC, Tanner A. Frictional action at lower limb/prosthetic socket interface. *Med Eng Phys* 1996;18:207–214.
4. Meulenbelt HE, Geertzen JH, Jonkman MF, Dijkstra PU. Determinants of skin problems of the stump in lower-limb amputees. *Arch Phys Med Rehabil* 2009;90:74–81.
5. Goh JCH, Lee PVS, Chong SY. Stump/socket pressure profiles of the pressure cast prosthetic socket. *Clin Biomech (Bristol, Avon)* 2003;18:237–243.
6. Silver-Thorn MB, Steege JW, Childress DS. A review of prosthetic interface stress investigations. *J Rehabil Res Dev* 1996;33:253–266.
7. Portnoy S, Yarnitzky G, Yizhar Z, et al. Real-time patient-specific finite element analysis of internal stresses in the soft tissues of a residual limb: a new tool for prosthetic fitting. *Ann Biomed Eng* 2007;35:120–135.
8. Zhang M, Roberts C. Comparison of computational analysis with clinical measurement of stresses on below-knee residual limb in a prosthetic socket. *Med Eng Phys* 2000;22:607–612.
9. Zhang M, Mak AF, Roberts VC. Finite element modelling of a residual lower-limb in a prosthetic socket: a survey of the development in the first decade. *Med Eng Phys* 1998;20:360–373.
10. Geerligs M. *Skin Layer Mechanics*. Eindhoven, The Netherlands: Technische Universiteit Eindhoven; 2010.
11. Hendriks FM. *Mechanical Behaviour of Human Epidermal and Dermal Layers in Vivo*. Eindhoven, The Netherlands: Technische Universiteit Eindhoven; 2005:107.
12. Zhang M, Mak AF. A finite element analysis of the load transfer between an above-knee residual limb and its prosthetic socket-roles of interface friction and distal-end boundary conditions. *IEEE Trans Rehabil Eng* 1996;4:337–346.
13. Colombo G, Filippi S, Rizzi C, Rotini F. A new design paradigm for the development of custom-fit soft sockets for lower limb prostheses. *Comput Industry* 2010;61:513–523.
14. Portnoy S, Siev-Ner I, Yizhar Z, et al. Surgical and morphological factors that affect internal mechanical loads in soft tissues of the transtibial residuum. *Ann Biomed Eng* 2009;37:2583–2605.
15. Lee WCC, Zhang M, Jia X, Cheung JTM. Finite element modeling of the contact interface between trans-tibial residual limb and prosthetic socket. *Med Eng Phys* 2004;26:655–662.
16. Duchemin L, Bousson V, Raossanaly C, et al. Prediction of mechanical properties of cortical bone by quantitative computed tomography. *Med Eng Phys* 2008;30:321–328.

17. Hendriks FM, Brokken D, Oomens CWJ, Baaijens FPT. Influence of hydration and experimental length scale on the mechanical response of human skin in vivo, using optical coherence tomography. *Skin Res Technol* 2004;10:231–241.

18. Sellgren U, Björklund S, Andersson S. A finite element-based model of normal contact between rough surfaces. *Wear* 2003; 254:1180–1188.

19. Zhang M, Lord M, Turner-Smith AR, Roberts VC. Development of a non-linear finite element modelling of the below-knee prosthetic socket interface. *Med Eng Phys* 1995;17:559–566.

20. Goldstein B, Sanders J. Skin response to repetitive mechanical stress: a new experimental model in pig. *Arch Phys Med Rehabil* 1998;79:265–272.

21. Lapeer RJ, Gasson PD, Karri V. A hyperelastic finite-element model of human skin for interactive real-time surgical simulation. *IEEE Trans Biomed Eng* 2011;58:1013–1022.

22. Limbert G. A mesostructurally-based anisotropic continuum model for biological soft tissues—decoupled invariant formulation. *J Mech Behav Biomed Mater* 2011;4:1637–1657.

23. Zhong H, Peters T. A real time hyperelastic tissue model. *Comput Methods Biomech Biomed Eng* 2007;10:185–193.

24. Ramírez JF. *Nivel de Confort y Distribución de Esfuerzos en la Interfaz Socket—Muñón en Amputados Transfemorales* [dissertation]. Antioquia, Colombia: Universidad Nacional de Colombia; 2011.