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Finite Element Analysis of Donning Procedure of a Prosthetic Transfemoral Socket

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Abstract—Lower limb amputation is a severe psychological and physical event in a patient. A prosthetic solution can be provided but should respond to a patient-specific need to accommodate for the geometrical and biomechanical specificities. A new approach to calculate the stress–strain state at the interaction between the socket and the stump of five transfemoral amputees is presented. In this study the socket donning procedure is modeled using an explicit finite element method based on the patient-specific geometry obtained from CT and laser scan data. Over stumps the mean maximum pressure is 4 kPa (SD 1.7) and the mean maximum shear stresses are 1.4 kPa (SD 0.6) and 0.6 kPa (SD 0.3) in longitudinal and circumferential directions, respectively. Locations of the maximum values are according to pressure zones at the sockets. The stress–strain states obtained in this study can be considered more reliable than others, since there are normal and tangential stresses associated to the socket donning procedure.

Keywords—Lower limb amputee, Contact stress–strain state, Patient-specific model.

INTRODUCTION

Amputation of a limb is one of the most traumatic events in one's life. Apart from the obvious loss of functionality, the psychological consequences and economic loss on the amputee as well as the social consequences in the society are immense. A prosthetic solution can be envisaged when a person is amputated after an accident, a violent action or a vascular disease. Essentially, the prosthetic device aims to restore: (1) the self-esteem of the patient by using the prosthesis like a complement of his complete body shape and (2) a normal and independent ambulation as much as possible.

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Typically, lower limb prosthesis needs a socket to act as an interface or link between the human stump and the prosthetic device. This situation modifies completely the natural performance of the residual limb. The most important physiological change is suffered by the soft tissues that transfer the body loads generated during the gait. These new conditions can induce skin problems such as callosities, abrasions, and blisters, and can also affect the vascular system.[3,7,15,18,21,22,25,28,31]

The stress state in the soft tissues of a lower limb amputee has been established in experimentation procedures using force transducers.[1,3,6,13,18–21,23,24,30,31] However, the sensors used in the experimentation can produce stress concentrations over the soft tissues, can modify the gait, and the results are valid only at the point where the sensor is located.[22,23] All these experimental difficulties have favored the use of numerical methods like the finite element (FE) method to assess the stress–strain state in a lower limb stump.[29]

Numerical models for amputees above the knee,[23,28] and below the knee[5,7–9,13,16–19,26,27,30,31] have been developed to establish the stress–strain state in the interaction between the socket and the stump. To obtain the geometry of the model, most studies use computed tomography (CT) or magnetic resonance imaging (MRI). They use either static models,[9,10,16,17,19,26,27,30,31] quasi-static model,[5] or qeuasi-dynamic models.[7,8] The loads and boundary conditions (BCs) can be divided into two groups: (1) those which generate forces, moments, or displacements over the bone and applying a displacement or rotation restriction over the socket,[5,7,8,11,13,16–19,27,30,31] and (2) those which generate forces, moments, or displacements over the socket while restricting the bone.[10,26] In either group, data from gait analysis are used to define the magnitude and direction of the loads.

Generally, the FE analysis is separated in two steps. The first one corresponds to the stress–strain state generated to ensure that the stump is inside the socket; in a numerical simulation, it is equivalent to solve an initial overclosure. Maintaining the stress-state generated during the first stage, the second one starts, when the load is applied over the stump or over the socket. In general, the simulation of the real procedure of socket donning is a challenging task which involves large motions. For this reason, the first stress–strain state has been addressed by applying manually a radial nodal displacement over specific areas of the stump,[27,28] or using an automated contact method provided by the software.[7,8,10,11] In both cases the magnitude of the displacement represents the difference between the actual residual limb shape and the socket shape. However, those methodologies do not match with the actual socket donning procedure where a very large relative movement between socket and stump is applied.

Most studies use a linear elastic isotropic model to represent the soft tissue mechanical behavior.[5,7–11,13,18,26–28,30,31] However, recently Portnoy and colleagues have used viscoelastic[19] or hyperelastic[16,17] formulations. The mechanical properties of the bone and the socket are common in all models, a linear elastic isotropic behavior is assumed,[5,8,10,11,18,28,30,31] or rigid body is defined for the bone,[16,17,19,26] the socket,[7,13,27] or both.[9]All of those dissimilar characteristics that had been used and the fact that each model was developed for a different person have produced different results, as can be seen in Table 1. These differences suggest that it is not possible to have the same stress–strain state at the residual limb for different amputees.

Chronologically, these numerical models have improved over time. Zhang *et al.*[27] proposed the first model where the influence of the interface friction and slip between stump and socket over the stress–strain state was considered in detail. They found that the pressures, shear stresses, slip, and bone movement are very sensitive to the coefficient of friction. Zhang and Mak[28] developed a 2D model for above knee (AK) amputee where one of the BCs was air cavity between socket and stump. They found that the sealed air cavity plays a role in the suspension of the prosthesis during the swing phase, and in supporting the body weight during the stance phase. Jia *et al.*[7] presented a 3D model where the effects of material inertia over the stress–strain state were studied. They found that during the stance phase there is no effect while during the swing phase of the gait, interface pressures and shear stresses are considerably affected by inertia. Portnoy *et al.*[17] developed a 3D FE model for transtibial amputee where the contribution of bone length, tibial bevelment, stiffness of the muscle flap and scarring, over stress–strain state were tested. They established mainly that the sharp edges of the truncated bones cause higher stresses and can potentially injure the muscle flap over time.

TABLE 1. Stresses at socket–stump interface in a lower limb amputees.

		Stage 1: solving the initial overclosure		Stage 2: initial overclosure solved and load application			
Reference	Amputation level	σ (kPa)	τ (kPa)	Load	μ	σ (kPa)	τ (kPa)
Zhang *et al.*[27]	BK	NR	NR	800 N	0.5	226	53[a] 21[b]
Zhang and Mak[28]	AK	NR	NR	4 N/m	1	20	16[a]
					0.5	26	11[a]
					0.1	63	5[a]
Zhang and Roberts[30]	BK	90	NR	800 N	0.5	226	50[a]
Zachariah and Sanders[26]	BK	NR	NR	800 N	0.675	250.3	108.3[c]
Jia *et al.*[7]	BK	NR	NR	800 N	0.5	297	80[c]
Lee *et al.*[10]	BK	NR	NR	800 N	0.5	370	120[c]
Lee *et al.*[11]	BK	147	NR	800 N	0.5	300	110[c]
Lin *et al.*[13]	BK	NR	NR	600 N	0.5	783	314[c]
Jia *et al.*[8]	BK	NR	NR	800 N	NR	323	NR
Faustini *et al.*[5]	BK	NR	NR	800 N	NR	250	NR
Portnoy *et al.*[18]	BK	NR	NR	400 N	NR	56.6	NR
Lee and Zhang[9]	BK	NR	NR	800 N	NR	260	NR
Portnoy *et al.*[19]	BK	NR	NR	500 N	0.7	65	51.9[a]
Portnoy *et al.*[17]	BK	NR	NR	NR	NR	24	NR

AK above knee, *BK* below knee, *NR* not reported, μ coefficient of friction, σ normal stress (pressure), τ shear stress.
[a]Longitudinal shear stress, [b]circumferential shear stress, [c]resultant shear stress.

One of the main limitations of those studies is that they have not studied the stress–strain distribution due to the actual socket donning procedure. In this study, we hypothesized that it is essential to account for the stress generated during the donning procedure to generate the stress–strain state in an amputee under normal loading. Thus, our aim was to model the actual donning procedure of the socket in five transfemoral amputees avoiding the initial overclosure in the model.

MATERIALS AND METHODS

Five male patients with one side transfemoral amputation were selected. Table 2 shows their general information. All of them were relatively active in their daily life, and had not any additional physical, vascular, neurological, or psychological condition that could alter or modify the results of the simulation. The patients used a non-distal end support socket, a Solid Ankle Cushioned Heel (SACH) foot, and mechanical monocentric knee prosthesis. They did not use a liner or a sock. According to the Ethics committee of *National University of Colombia* a proper informed consent was provided by the patients before the procedure started.

Digital Geometric Reconstruction

Compared to the natural shape of the stump, the actual geometry of the socket had noticeable differences. The socket was smaller than the stump which may decrease or increase the contact stresses locally in some zones of the interface as was found by Faustini *et al.*[5] and Zhang and Roberts.[29] Because of this, the actual geometries of the socket and the stump were obtained separately. While the patient was standing up, a prosthetist made a cast that was used in the fabrication of a plaster positive of the socket and the stump. After these solid elements were done, a laser scanner was used to obtain a digital representation of them. Although, as showed by Portnoy *et al.*,[17] the surgical scars are important in numerical models, they were not included in this study because the main objective was to simulate the socket donning procedure.

To complete the geometric digital information of the model the femur was included into the solid representation of the stump by using CT scan information of the bone and its relative position to the soft tissues. The parameters used for this scan were: SIEMENS/Emotion6 Scanner, 112 mAs, 130 kV, 512×512 pixel matrix, pixel size 0.758 mm, gantry tilt 0.0°, slice increment 1 mm, and were the same for all patients. During the CT scan, the patients were not wearing their prosthetic socket or any additional element like sock or liner around his residual limb. Taking into account that during the CT scan the patients were in a laying supine position, which generates a natural deformation of the soft tissues, only those images related to the bone tissues were extracted from the CT scan. Using Mimics (Materialise, Belgium) the three-dimensional digital representation of the residual bone was obtained.

Some marks were made over the plaster positive of the socket and the stump at specific location as close as possible to the greater trochanter and ischial tuberosity. Then, matching that marks with the CT image information, the three different solids (socket, residual bone, and a bulk representation of the remaining soft tissues) were aligned in an anatomical and well-defined relative position using Solid Works (Dassault Systèmes, France). Figure 1 shows the final geometric configuration of the solids for patient P5, where it is possible to differentiate the bone, the socket, and the soft tissues. Additionally, it is possible to identify the initial overclosure between the soft tissues and the socket, which will be solved during the numerical simulation of the socket donning procedure. For the other patients, the configuration is similar.

Mechanical Properties

The socket–stump interaction for transtibial amputees has been studied more extensively than for transfemoral amputees. In this study a linear elastic homogeneous isotropic condition for the socket and the bone was considered, while for the soft tissues hyperelastic condition was used. All properties are in the same order of magnitude than those used by other authors.[7,10,11,13,26,27,30–32]

TABLE 2. General information of the transfemoral amputees selected.

Patient	Amputation side	Age (years)	Tall (cm)	Weight (kg)	Time since amputation (years)	Body mass index (kg/m^2)	Proximal–distal girth of the stump (m)
P1	Left	39	177	92	4	29	0.63–0.28
P2	Left	43	171	70	11	23	0.55–0.23
P3	Left	58	167	74	23	26	0.56–0.23
P4	Right	65	165	84	1	31	0.61–0.26
P5	Right	50	163	59	2	22	0.57–0.24

P1 patient 1, *P2* patient 2, *P3* patient 3, *P4* patient 4, *P5* patient 5.

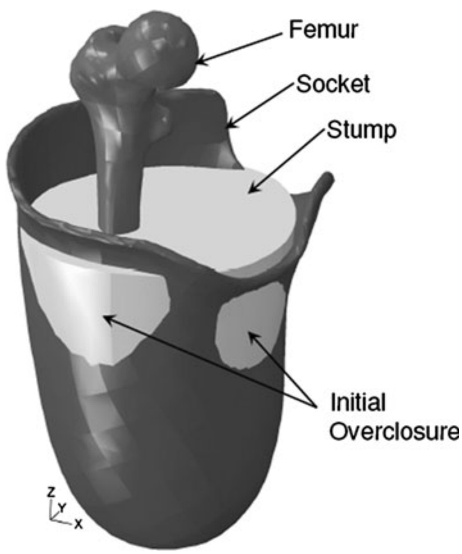

FIGURE 1. Socket, stump, bone, and initial overclosure between the stump and the socket.

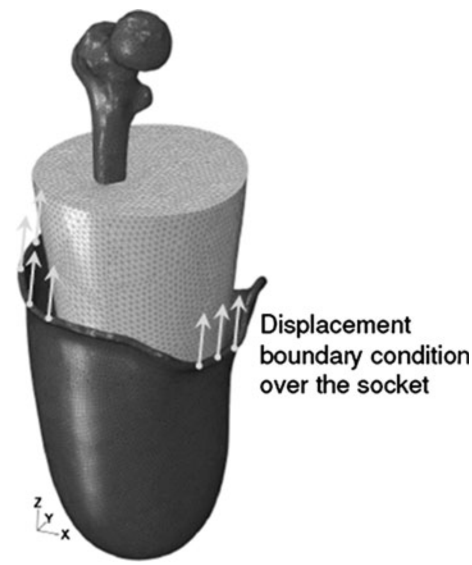

FIGURE 2. Relative position at the beginning of the simulation.

For the bone, the Young's modulus was 15 GPa and the Poisson's ratio was 0.3.[4] For the socket made of polypropylene, the Young's modulus was 1.5 GPa and the Poisson's ratio was 0.3.[10,11,28]

For soft tissues a hyperelastic, linear, homogeneous, and isotropic behavior was defined using the Generalized Mooney–Rivlin Solid strain energy function:

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

where the invariants of the principal stretch ratios are $I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2$ and $I_2 = \lambda_1^{-2} + \lambda_2^{-2} + \lambda_3^{-2}$, the relative volume change is $J = \lambda_1\lambda_2\lambda_3$, and C_{10}, C_{11}, D_1 are the constitutive parameters. For this study, $C_{10} = 4.25$ kPa, $C_{11} = 0$ kPa, and $D_1 = 2.36$ MPa^{-1} were established following the average flaccid muscle property used by Portnoy *et al.*[17]

Loads and Boundary Conditions

The BCs were applied according to the model of the donning procedure of the socket. The interaction between the bone and the residual bulk representation of the soft tissues was modeled using a tie condition that simulates perfect bonding between the two materials.[5,8,16–19] The interaction between the socket and the stump was modeled using surface to surface ABAQUS V6.10-2 contact condition, which impede the stump nodes (slave nodes) to trespass or penetrate into the socket (master surface) during the relative displacement between socket and stump. Taking into account that during the donning procedure the amputee uses some kind of sock to insert stump into the socket, a friction coefficient of 0.415 was assigned to the contact.[2] Additionally, the hip joint action was represented with a restriction on all degrees of freedom relative to displacement in the femoral head, specifically over the zone where the acetabulum acts.

Before the simulation, the socket and the stump were not in contact (Fig. 2). During the simulation, displacement vector on the proximal part of the socket was applied. The value of the vector is equivalent to the displacement that is needed to put the stump in the actual final position within the socket. It is different for each patient and was calculated using the CT information and the marks located over the positive plaster of the socket and the stump.

The displacement applied to the socket was made in a quasi-static step, where the velocity needs to be as low as possible to minimize the dynamic effects but using a reasonable time of calculation. Thus, according to the actual donning procedure where the amputee uses a short time to put his stump inside the socket, in this study 15 s were assigned as model duration and depending on the socket displacement for each patient it generates an approximate velocity from 6 to 9 mm/s.

The quasi-static model was developed using ABAQUS V6.10-2/Explicit. Due to the complexity of the geometry, tetrahedral elements were used for all models; its approximate global size (inter-nodal spacing) is 5 mm for the stump and 3 mm for the bone and the socket, and was defined after a mesh sensitivity analysis. A mesh distribution for patient P5 can be seen in Fig. 2, and it is similar for the other patients since the mean size of the elements is the same for all patients. The total number of elements ranges from around 300,000 to 480,000 depending on the patient.

The runtime for each model ranges from around 6 to 8 h, using a Quad Core Processor Core i7-880, 3.06 GHz and 16 GB RAM, Windows 7 Pro 64 bits OS. The pressure and shear stresses at the interaction face between the socket and stump were calculated; also maximum and minimum principal logarithmic strain, displacements, and von Mises stresses over the stump were calculated too.

RESULTS

The stress distributions can be shown in any part of the model (socket, stump, and bone), but focus on the surface of the stump with the socket–stump interaction will be made. The application of the displacement at the proximal part of the socket produces not only normal, but also shear stresses. Beside the peak values of the stresses at the socket–stump interface shown in Table 3, normal stress (pressure) distribution is shown in Fig. 3 for all patients. The peaks of pressure were shown in Fig. 3 and therefore the orientation of the bone was not always the same. It is found that peaks of pressure occur in the anterior side for patients P1, P2, P4, and P5, while for patient P3 it is located posteriorly. Pressure peaks are about 1.3–3.7 times higher for P2 and P4 than P1, P3, and P5. The differences of stump geometry among all patients can also be appreciated in Fig. 3. Zones of high pressure can easily be identified for all patients indicating that the stress distribution is not homogeneous. The circumferential peak shear stress is about 2–4 times higher for P1, P2, and P4 than for the other patients (Table 3). The positive and negative circumferential peak shear stresses are located in the anterior side of the stump for patients P1, P4, and P5, while for patient P2 it is located in the medial side and for patient P3 it is located in the posterior side as shown in Fig. 4. The longitudinal peak shear stress is similar for P2 and P4 and about 1.3–3.5 times higher than for the rest of the patients. Positive longitudinal peak shear stresses are all located in the anterior side of the femur while negative longitudinal peak shear stresses are located in the posterior side (Fig. 5). As in the pressure distribution, zones of high shear stresses can be identified while the rest of the stump is under low stress.

For all patients the displacement peak values (Table 3) are located at the posterior–proximal part of the stump, except for patient P1 where it is located at medial side as shown in Fig. 6. According to the displacement applied over the socket, and the initial over closure between the socket and the stump, the magnitude of the displacement for each patient is different.

The von Mises stress distribution within the stump is shown in Fig. 7, where it is possible to identify that stresses are higher at the stump–bone interface than at the stump–socket interface. The stresses are mostly concentrated in the proximal part of the stump for all patients, except for patient P5, where the peak value is in the distal part of the bone–stump interaction. The maximum von Mises stresses are similar for P1 and P2 and about 1.5–4.7 times higher than for the rest of the patients.

The maximum (tensile) and minimum (compressive) principal logarithmic strain distribution is shown in Fig. 8, where it is possible to identify that the peak values are located at stump–bone interaction or at the proximal and distal side of the stump. For all patients the compressive strains are higher than tensile strains. For patient P1 tensile strains are about 1.9–3.6 times higher than for the rest of patients, while for the compressive strains those are 1.3–2.3 times higher than for the rest of the patients.

DISCUSSION

According to the information shown in Table 1, it is possible to identify that the numerical simulations of stress–strain state for the interaction between socket and stump for AK amputees are less than for below knee (BK). In that table, there is only one reference to AK model which can be partially used to make a comparison against the results obtained in this study.

Zhang and Mak[28] obtained a maximum pressure of 26 and 63 kPa using a coefficient of friction of 0.5 and 0.1, respectively. These results are considerably higher than those presented in this study where the maximum

TABLE 3. Peak values of stresses and displacements at the socket–stump interface.

Patient	Maximum initial overclosure (m)	Normal stress (kPa)	Circumferential shear stress (kPa)	Longitudinal shear stress (kPa)	Displacement (m)
P1	3.04E−2	4.37	−0.93	1.48	4.77E−2
P2	1.73E−2	5.38	−0.79	1.99	4.25E−2
P3	1.03E−2	3.15	−0.42	−1.16	2.03E−2
P4	2.85E−2	5.61	−0.89	−2.00	3.62E−2
P5	2.64E−2	1.54	−0.23	−0.57	1.90E−2

P1 patient 1, *P2* patient 2, *P3* patient 3, *P4* patient 4, *P5* patient 5.

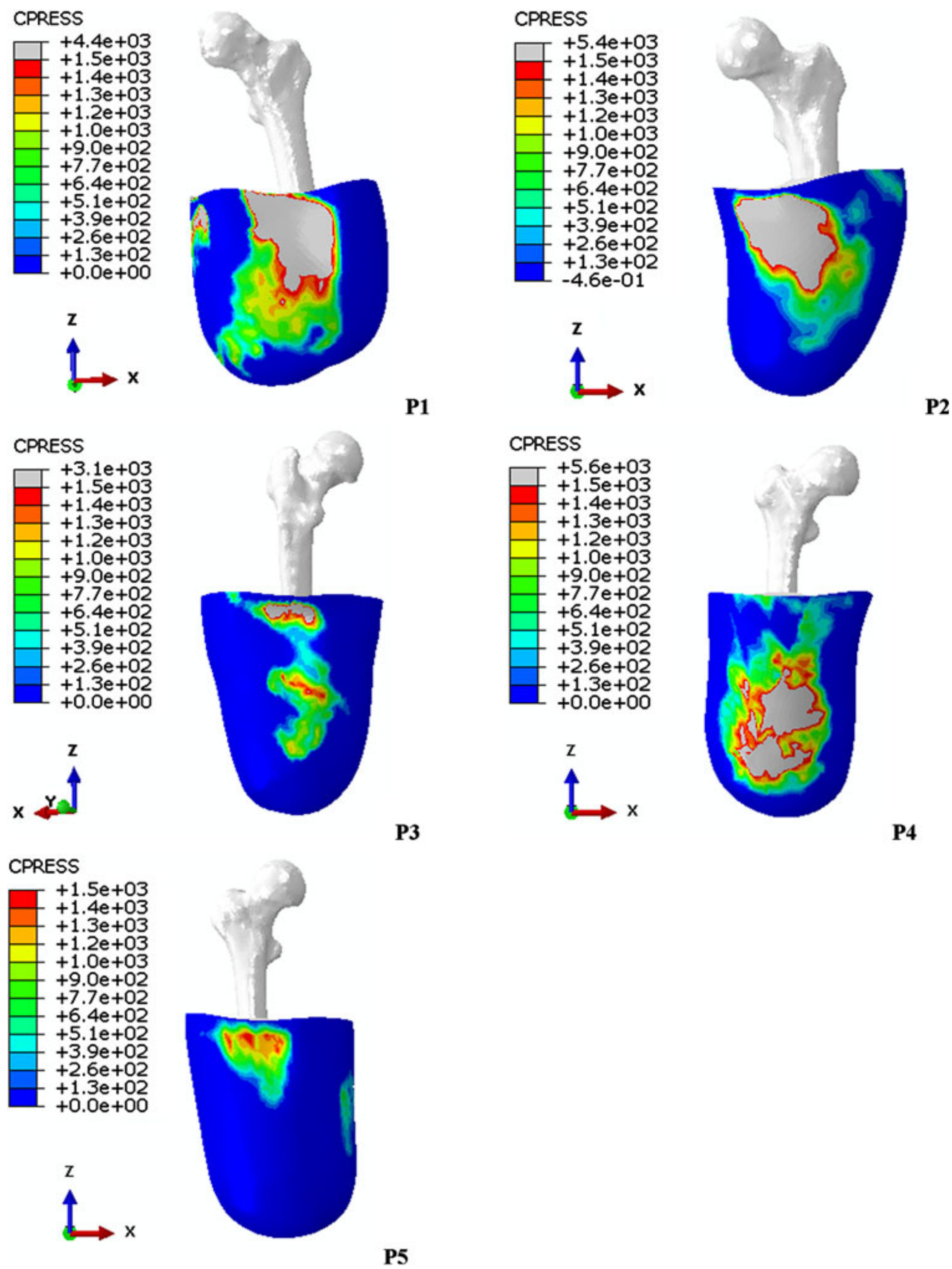

FIGURE 3. Normal stress (pressure) in Pascal at the stump for the five patients.

peak pressure was 5.6 kPa using a coefficient of friction of 0.415 for patient P4. The models developed here only consider the donning procedure while Zhang and Mak[28] not only include the donning procedure but also a vertical force; it is therefore difficult to compare both studies. The normal (pressure) and the tangential stresses due to the interaction between socket and stump obtained in this study compared against those by Zhang and Mak[28] are lower. The differences in the magnitude of the stresses can be associated mainly with: (1) in this study for the bulk soft tissues the hyperelastic behavior was defined, while Zhang and Mak[28] used a Young's modulus of 150 kPa and the Poisson's ratio of 0.45, (2) the tangential stresses obtained by Zhang and Mak[28] are associated with the second stage of the model (load stage), because for the first stage (solving initial overclosure) the radial displacement applied over the nodes does not produces a tangential stresses. These differences may indicate the need for a patient-specific analysis since variations from one patient to another can have tremendous effect in the stress–strain distributions within the soft tissues. This also suggests the need for a patient-specific analysis for the design of sockets.

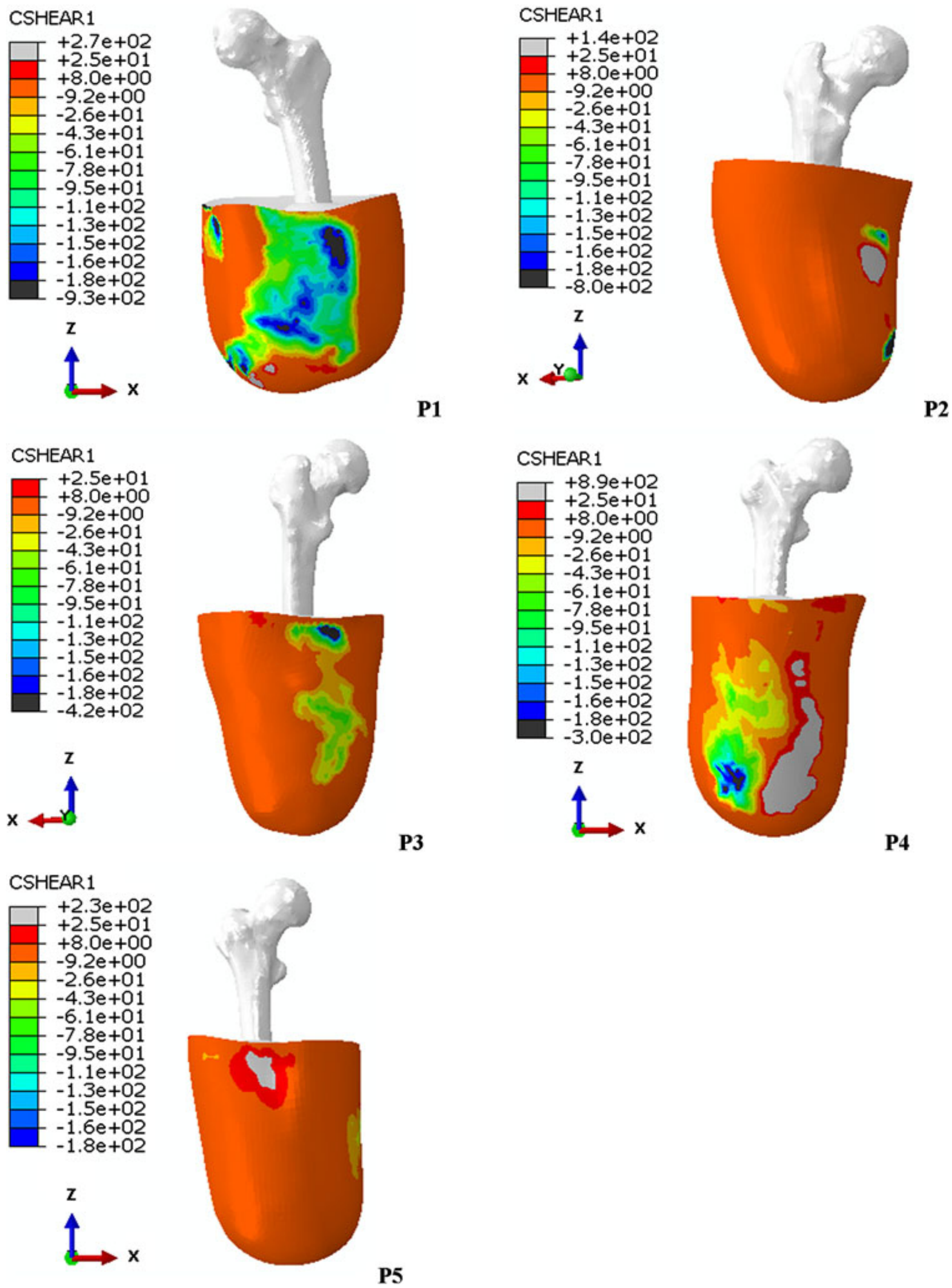

FIGURE 4. Circumferential shear stress in Pascal at the stump for the five patients.

As shown in Table 1, there are only two results available for the normal stresses at the first stage, those are 147 and 90 kPa which are related to Lee *et al.*[11] and Zhang and Roberts[30] studies, respectively. Despite the fact that both of them are for transtibial amputees, the normal stresses obtained in this study are lower than those. This difference is attributed to the use of models that are geometrically and mechanically different. Moreover, comparing the stresses obtained for all patients (Table 3; Figs. 3, 4, and 5) it is possible to identify that no matter that all of them are AK amputees, the models are different and as a consequence stresses are different too.

Portnoy *et al.*[17–19] studies for transtibial amputees were developed using an hypereleastic constitutive equations for the soft tissue, as those presented in this study. The normal and tangential stresses for Portnoy *et al.*[17–19] studies are larger than those obtained in this study, the differences can be attributed mainly to the amputee condition and the load condition used into the models.

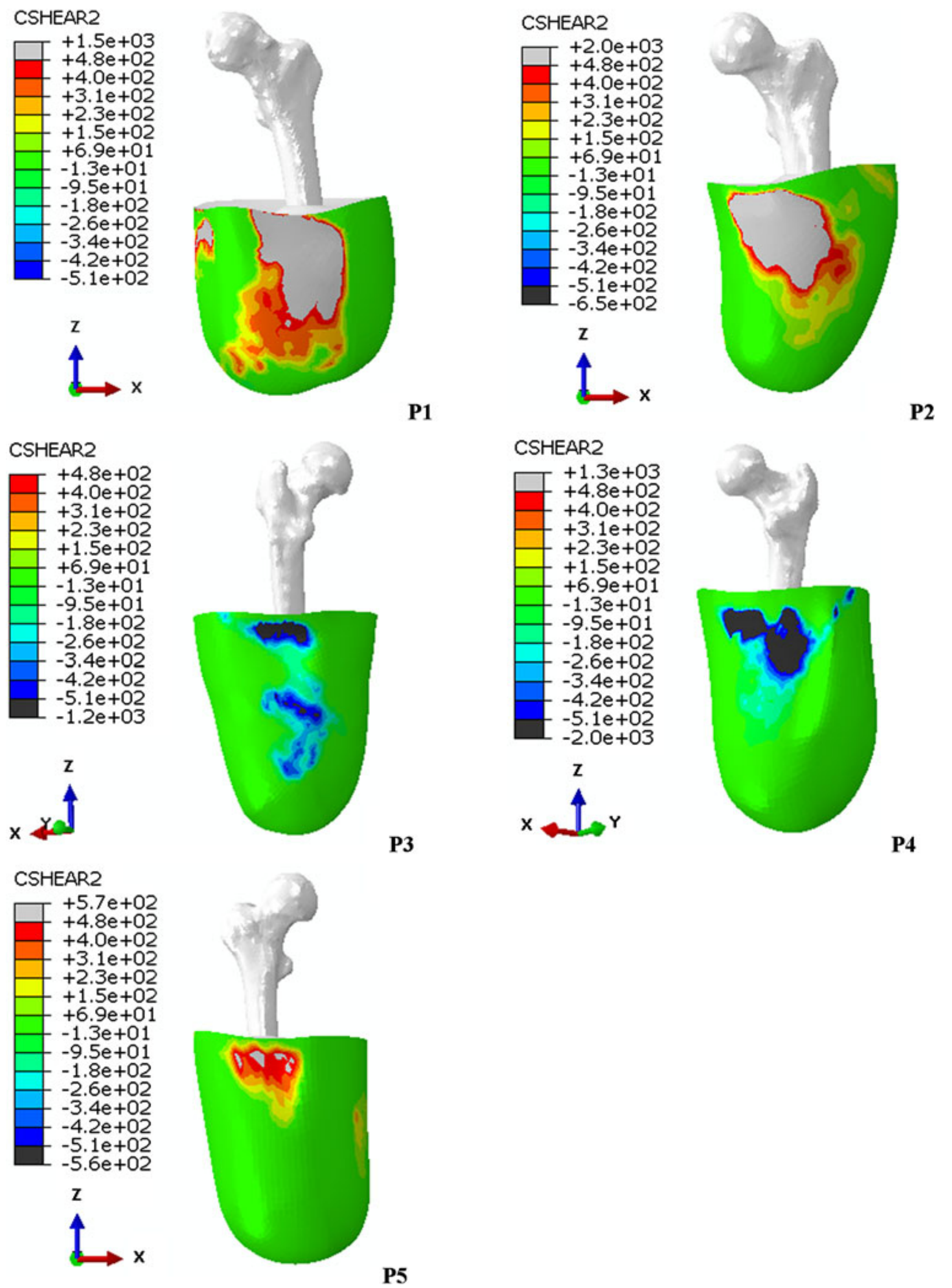

FIGURE 5. Longitudinal shear stress in Pascal at the stump for the five patients.

Another reason to explain the differences could be the way overclosure is solved. The models developed in this study represents more accurately the actual donning procedure, where the relative displacement between the socket and stump generates not only normal but also shear stresses, while the other models presented at the literature (Table 1) usually solve the initial overclosure applying a manual or automatic radial displacement over the stump's nodes which were trespassing the socket, and generates mainly normal stresses.

There is no information related with the stress–strain state generated during the socket donning procedure, specifically for transfemoral amputees. Because of this, it is necessary to validate partially the results, verifying that the pressure at the interface is lower than tolerable normal stresses (pressure threshold). Unless the pain threshold is specific for each person and for each zone of the body,[12] the lower pain threshold assessed by Lee and Zhang[9] in a transtibial amputee was 690 kPa which is higher than the maximum contact pressure obtained in this study (5.6 kPa). This

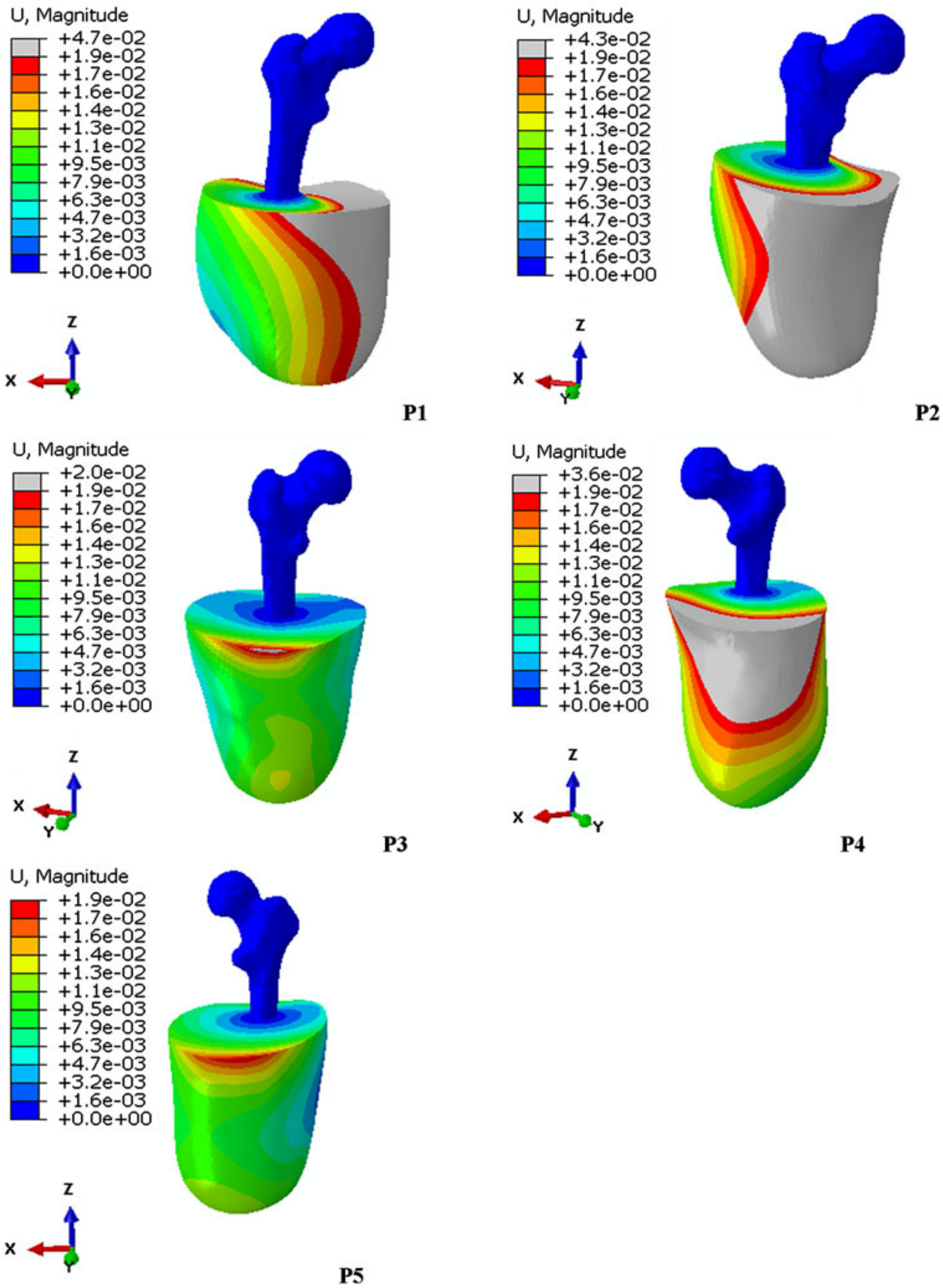

FIGURE 6. Displacement in metre at the stump for the five patients.

comparison allows establishing that the order of magnitude of the pressure contact obtained in this work is according to tolerable stresses for one human.

Considering all patients, the mean peak compressive and tensile principal strain are 53.2% (SD 13.7%) and 32.4% (SD 16.7%), respectively. For a BK amputee, Portnoy *et al.*[19] found 85 and 129% peak compressive and tensile strain, respectively, while Linder-Ganz *et al.*[14] for six patients during sitting found 75 and 72% mean peak compressive and tensile strain, respectively. Since the mean tensile and compressive strain values obtained in this study are lower than those showed by Portnoy *et al.*[19] and Linder-Ganz *et al.*[14] its order of magnitude can be considered appropriated.

Interface contact stress–strain state due to the interaction between socket and stump of an amputee is influenced by different factors such as alignment, gait, knee prosthesis, and foot type. In this study these aspects were not considered, because the stage of donning procedure is not affected by them. This novel simulation can be improved modifying the contact BC between bone

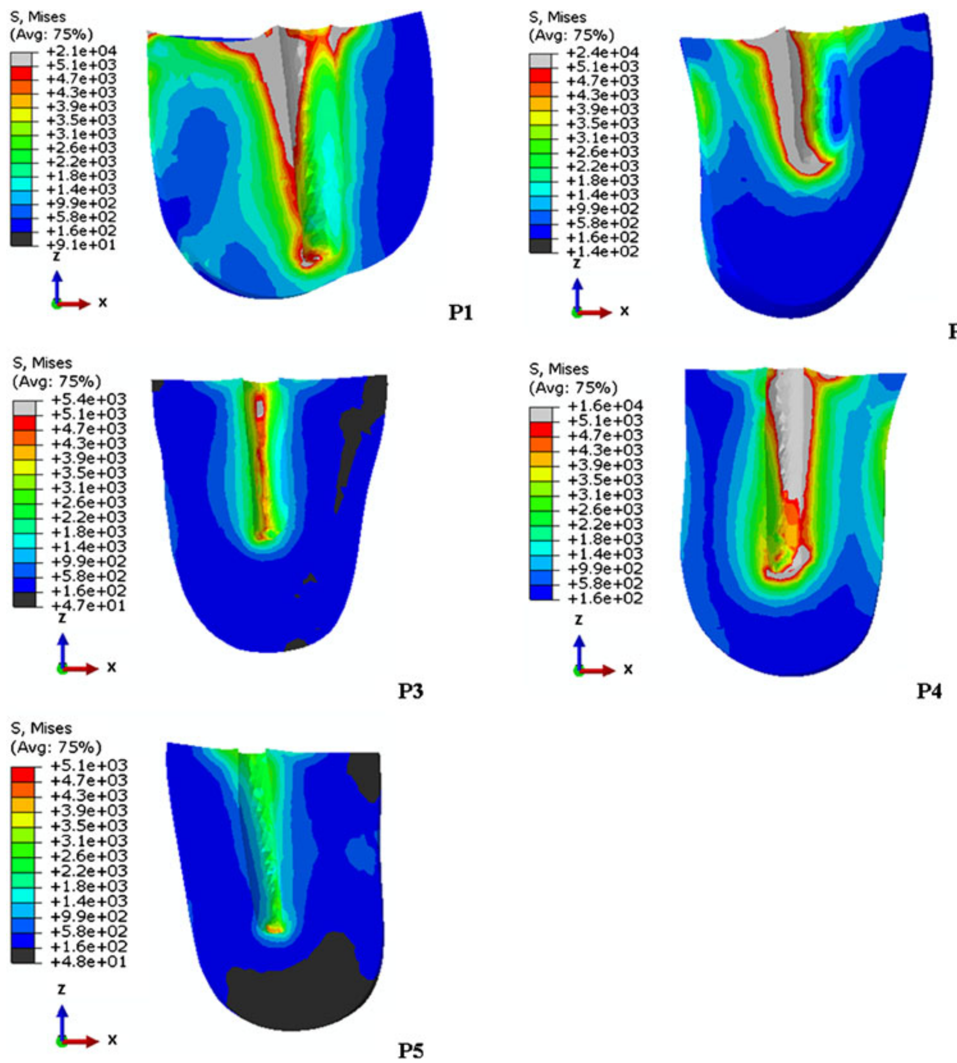

FIGURE 7. Von Mises distribution in Pascal in a cross-section of the bone, the stump and the socket for the five patients.

and soft tissues where a friction contact can be more realistic than the tie constrain used and representing the stump as a multilayer body, where skin, fat, and muscle should be differentiated. However, as have been demonstrated by other authors, the use of a tie BC between bone and soft tissues and the representation of the soft tissues as a bulk volume generate accurate results.

Finally, it is possible to say that the stress distribution is according to what to expect, because in those areas where the socket has been modified to increase the contact pressure, the normal stress distribution obtained from the numerical model agree with them. Also, accordingly with the relative displacements between the socket and the stump, tangential stresses appear at stress state. In spite of these, it is necessary to develop validation process that allows to verify completely the results shown in this study.

CONCLUSIONS

The actual socket donning procedure simulation was done avoiding the use of radial displacement over the stump nodes to solve the initial overclosure. Instead, a progressive deformation was generated over the stump while it was introduced into the socket during simulation. As a consequence of this, the stress–strain state obtained in this study is more reliable, since there are normal and tangential stresses, while applying radial displacements over the socket the stresses and strains are mainly normal.

Although, stress–strain patterns and magnitudes have shown similar behavior for all patients, this study has shown that a patient-specific solution is needed, not only to FE analysis, but also to socket design and manufacture.

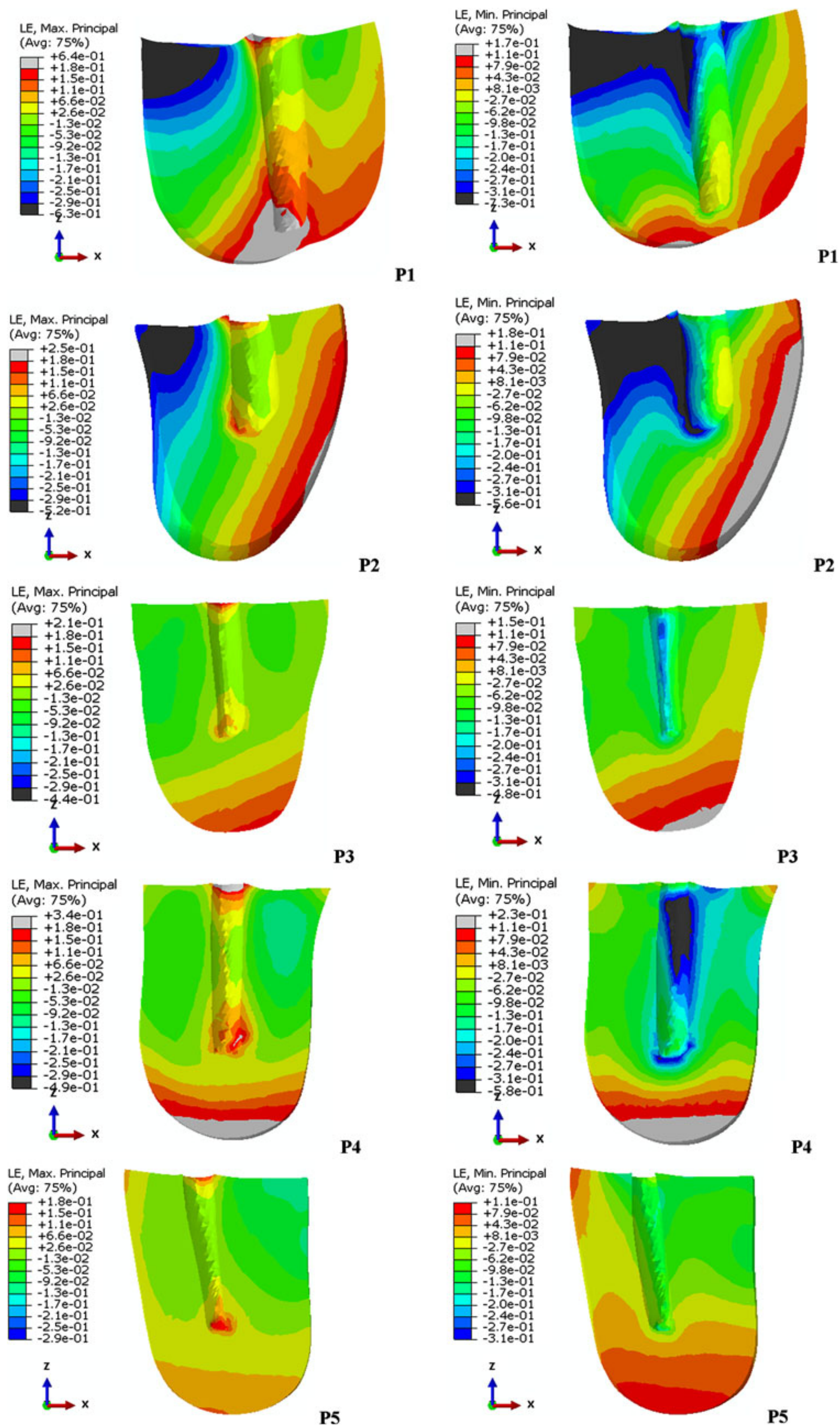

FIGURE 8. Maximum and minimum logarithmic principal strain for the five patients.

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