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Analysis of compression/release stabilized transfemoral prosthetic socket by finite element modelling method

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Abstract

This project aimed to investigate the residual limb stress of a transfemoral amputee's Compression/Release Stabilized (CRS) socket by finite elemental modelling. The model was constructed from magnetic resonance images of the left residual limb of a 48-year-old male transfemoral amputee. Two conditions were simulated. In the donning condition, the prosthetic socket under the residual limb moved proximally until it reached the required donned position. The weight-bearing condition was subsequently simulated by applying body weight (800N) at the femoral head while keeping the distal end of the socket fixed. The maximum contact pressure was concentrated at the proximal anterior-medial regions of the residual limb surfaces in both conditions. In the donning condition, the maximum von Mises stress and the maximum contact pressure were 277.7 kPa and 254 kPa respectively. The respective values were 191.9 kPa and 218.5 kPa when body weight was applied. The CRS socket demonstrated higher stress and contact pressure as compared to that of other designs reported in the literature. Nevertheless, the higher stress and contact pressure may be acceptable since the values are tolerable and below the pain threshold of the previous study. Our findings provide important biomechanical information on the CRS socket that may help future design optimization.

Keywords: compression/release; transfemoral amputee; socket design; finite element

Introduction

Lower limb prostheses are designed to substitute the functions of the amputated limb and play an important role in facilitating daily activities such as standing and locomotion. The prosthetic socket, a crucial component of the prosthesis, makes direct contact with the residual limb and influences prosthetic fit and comfort. Inappropriate design or fitting may lead to various complaints, especially in those who use a prosthesis with a more conventional socket design (e.g., Quadrilateral or Ischial-ramal containment socket) [1-3]. These complaints include pain, abrasion, sweating, poor prosthetic control, and inconvenient donning and doffing. Improvement of socket design could hopefully alleviate some of these problems, which commonly arise during rehabilitation and activities of daily living.

The quadrilateral prosthetic socket (Quad) has a quadrilateral brim that is narrower in the anterior-posterior direction compared to the medial-lateral direction. The ischial tuberosity rests on the posterior brim for weight-bearing and force transfer [4]. Since the ischium is not secured, the ischial bone may medially slide on the posterior surface of the prosthetic socket during weight bearing causing femur abduction. As a result, the distal end of the residual limb would be pressed against the socket wall, leading to pain, discomfort and lateral lurch gait. The ischial-ramal containment (IRC) socket was developed in the 1980s aiming to remedy this problem [5-8]. The IRC socket extends the posteromedial brim of the socket to cover the medial border of the ischial tuberosity and ramus by an oblique and sloping contours design. Compared with the Quad socket, it is narrower in the medial-lateral dimension. The Quad and IRC Sockets share similar design features and can both compromise hip range of motion [9-11]. The IRC socket design was further modified to become the Marlo Anatomical Socket (MAS). Instead of containing the ischium, the MAS surrounds the ischio-pubic ramus and allows a release on the posterior gluteal portion. Improved comfort, stability and reduced energy cost of walking were reported [12,13].

Sufficient compression on the residual limb can reduce the relative motion between soft tissue, bone and the socket, which is essential for prosthetic control [14]. Traditional socket designs (Quad, IRC, MAS) may produce insufficient tissue compression and thus prosthetic control. The compression/release stabilization (CRS) socket design was recently introduced that features four longitudinal pre-compression bars and four adjacent release regions. The release regions allow some room to accommodate the compressed tissue for better comfort and reduced chance of ischemia while maintaining adequate compression on the residual limb through the pre-compression bars. Subsequently, a more compressed residual limb enables a higher efficiency of momentum transfer from the femur bone to the socket [15]. There may be no need for a proximal brim design to stabilize the socket around the ischial region and therefore no loss of hip range of motion. The control stability, sitting, donning and doffing comfort may be improved by the CRS design [16,17]. However, the condition of the residual limb under further compression is not clear. This study aimed to evaluate the pressure and stress distribution on the transfemoral residual limb using the CRS socket by the finite element (FE) modelling method.

Methods

Subject

A 48-year-old male transfemoral amputee (left side) was recruited for this study. His limb was amputated 28 years ago due to a bone tumor and has used a prosthesis ever since. He was active in daily activities and walked independently every day. Before the experiment started, the subject signed an informed consent and the study was approved by the institutional ethics committee.

Geometric reconstruction of the residual limb and design of CRS socket

MR images of the subject's residual limb were taken (GE Signa HDxt 1.5T, Canada). Image parameters were: slice thickness, 2mm; acquisition time, 0.42s/35mins; matrix, 320×160, Field of view, 40×32cm, 1-mm pixel size. The patient was supine lying on the MRI patient table with the hip in a neutral position. An unmodified plaster cast was worn to minimize gravity induced residual limb tissue distortion [18]. Soft tissue and bone segmentation were carried out using image segmentation software Mimics v10 (Materialise, Leuven, Belgium).

The CRS socket was constructed by offset the surface of the 3D geometry of residual limb in SolidWorks (Corporation, MA, USA). The compression was started 40mm below the upper brim of the socket. Four compression areas with an equal compression width (the upper circumference of residual limb divided by 8) were centered between median sagittal and coronal plane with the adjacent openings for the soft tissue displacement. The compression depth was set to 15mm (Figure 1).

Material properties and mesh creation

The material properties of the bone encapsulated soft tissue and socket were assumed to be linear elastic, homogeneous, and isotropic. As the CRS socket is a new socket design approach, the simplification of material properties can quickly assess the CRS socket biomechanics and the results can be a starting point for further study. The Young's modulus of the bone and socket were 15,000 MPa and 1,500 MPa respectively [19]. The Poisson's ratio for the bone and socket was 0.3 [20,21]. As reported by Malinauskas, the average modulus from nine transfemoral amputees ranged from 53.2 to 141.4 kPa [22]. However, the underlying muscles would be in tension during the socket donning, which increased the soft tissue modulus. The Young's modulus for the encapsulated soft tissue was set to 0.2 MPa, and 0.49 for Poisson' ratio [20, 23, 24]. For easy modelling of complex geometries of the CRS socket, bone and encapsulated soft tissue, three-dimensional 4-node tetrahedra (C3D4) mesh elements were assigned for all parts and created by the finite element software package Abaqus (Dassault Systèmes, RI, USA). The muscles, ligaments, tendons, subcutaneous fat and skin etc. were not segmented, but were integrated as the encapsulated soft tissue. The element size was 3 mm for both the bone and socket, and 5 mm for the encapsulated soft tissue which adopted from the study which performed meshed convergence testing [25]. There was a total of 349, 828 elements in the model. The set values of material properties and mesh types are summarized in Table 1.

Table 1. Material properties

Material	**Young's modulus (MPa)**	**Poisson's ratio**	**Elements**	**Element type**
Bone	15000	0.3	58,982	C3D4[19]
Encapsulated soft tissue	0.2	0.49	166,663	C3D4[19]
Socket	1500	0.3	124,183	C3D4[19]

Boundary and loading conditions

The interfaces between the femur and soft tissue were tied. The coefficient of friction between the socket and soft tissue was assumed to be 0.4 [25,26].

Two conditions were simulated sequentially: (1) simulation of donning the prosthetic socket. Before donning, the socket was placed underneath the residual limb with the upper inner compression area just touching the residual limb (Figure 2a). The donning was simulated by a 106 mm upward displacement of prosthetic socket along the z-axis with x and y-axis fixed.

The femoral head was fixed while the socket moved vertically upward until it reached the fully donned position. The upward displacement was kind of the donning force. The actual donning force was generated by the Abaqus software. After the socket fully donned, it maintained at that position.

(2) weight-bearing on the prosthetic socket and the residual limb. After the socket fully donned, a simulated body weight load of 800N was applied on the femoral head while the distal end of the socket was fixed (Figure 2b).

Results

Contact pressure and von Mises stress in the donning condition

The model predicted contact pressure distribution and von Mises stress are shown in Figure 3 and Figure 4 respectively. The maximum contact pressure and the maximum von Mises stress of the encapsulated soft tissue were 254.1 kPa and 227.7 kPa respectively. The residual limb was loaded primarily at the compression areas as indicated by the higher pressure distribution. In addition, there was a little pressure concentration on the residual limb corresponding to the upper brim of the CRS socket.

The maximum von Mises stresses of encapsulated soft tissue were concentrated in the proximal region of the compression areas except for the anterior-lateral side where the von Mises stress spread out from the compression center. The posterior-medial compression area appeared to have the smallest maximum von Mises stress (186.8 kPa) among the four compression areas.

Contact pressure and von Mises stress in the donned condition with weight-bearing

The maximum contact pressure and maximum von Mises stress of encapsulated soft tissue were found 218.5 kPa and 191.9 kPa respectively in the anterior-medial compression area (Figure 5 and Figure 6). The values were decreased by 14% for the maximum contact pressure and 15.5% for the von Mises stress compared with the donning procedure.

Maximum Contact pressure and maximum von Mises Stress on different compression areas

The maximum contact pressure and the maximum von Mises stress of encapsulated soft tissue on individual compression area are shown in table 2. From the donning to the weight-bearing condition, the maximum contact pressure was decreased by 14.2%, 15.5%, 5.7%, 17.2% in the anterior-medial, anterior-lateral, posterior-medial, and posterior-lateral sides respectively. Corresponding maximum von Mises stress was also decreased by 16.2%, 19.8%, 7.0%, and 24.9%. Both contact pressure and von Mises stress were reduced the least in the posterior-lateral compression area.

Table 2. Maximum contact pressure and maximum von Mises stress on individual compress area

	Contact pressure		**von Mises stress**	
Location	Donning (kPa)	Weight-bearing (kPa)	Donning (kPa)	Weight-bearing (kPa)
Anterior-medial	254	218	228	191
Anterior-lateral	219	185	212	170
Posterior-medial	212	200	187	174
Posterior-lateral	204	169	201	151

Discussion

The CRS prosthetic socket design was claimed to reduce the lost motion induced by soft tissue-socket interaction in the conventional prosthetic socket. However, the condition of the residual limb under further compressions was not well understood. By simulating the donning and weight-bearing responses, the stress and contact pressure of a transfemoral amputee's residual limb utilizing the CRS socket were predicted. The maximum von Mises stress and the contact pressure were both located on the anterior-medial side of the residual limb. The posterior-medial compression area was the region which the value of the maximum contact pressure and maximum von Mises stress decreased the least from donning to weight bearing. This may indicate that the posterior-lateral compression area bore higher body weight during loading. A previous study suggested that thicker soft tissue can absorb more impact force and energy [27]. From the cross-sectional view of the subject's residual limb, the soft tissues (fat and muscles) around the posterior-medial shaft of the femur was thicker than the other three locations (Figure 7). It is likely that this structural difference contributed to the location-dependent differences in the pressure and stress distribution.

Socket interfacial pressure is an important factor which affects the comfortability of prosthesis user. The interfacial pressure has been studied by many researchers with different modalities to evaluate socket structure and function. Zhang's group [28] found the maximum normal stress in the donning procedure was less than 55.51 kPa, and the pre-stress was more evenly distributed for the conventional transfemoral prosthetic socket. In their study, the simulation of donning was created by applying 50N on the upper surface of the soft tissue. It was likely that this adding force was too small to cause force concentration and may not accurately reflect the donning action. As donning is an action of putting the residual limb into the prosthetic socket, providing a vertical displacement of the prosthetic socket seems to be a valid way to simulate the donning action. Kahle and Highsmith [29] studied nine transfemoral amputees wearing IRC and brimless socket designs. The maximum pressure was 254 kPa with a mean of 112 ± 80 kPa in the IRC socket and 222 kPa with a mean of 109 ± 61 kPa in the brimless socket. Consistent with the present results, they found the largest pressure and stress occurred in the medial-proximal aspect of the socket, suggesting that high pressures are more likely to occur in this region. The largest pressures of both sockets were even larger than the value (218 kPa) recorded in the present study.

Colombo et al. [30] reported a 200 kPa maximum contact pressure on the transfemoral residual limb surface which is slightly smaller than present findings. However, they evaluated the walking condition with several times of the body weight applied at the hip joint. Velez et al. [31] found the maximum pressure during donning and loading with Quad and IRC socket designs was 151 kPa in one transfemoral amputee, about 30% smaller than the present study. The maximum compression force in the CRS socket was much higher than the maximum pressure in conventional socket designs. Nevertheless, it was also far less than the previous study of transtibial amputee pressure tolerance and pain threshold [32]. It seems that further compression of the CRS socket may be possible. However, possibility of blood vessel damage and skin break should be evaluated further for patient using the CRS socket as well as under walking gait condition.

Although 800 N vertical load was added at the femoral head, the maximum von Mises and the maximum contact pressure were both smaller compared to the donning condition. While body weight was applied, the vertical load would force the residual limb further into the socket. It might increase the residual limb contact area with the socket and distal end bearing which in turn helped the maximum von Mises stress and maximum contact pressure decreased from donning to weight bearing. This study utilized the upward displacement to simulate the donning procedure. Although it is more approach to the clinical scenario, the limitation is that the displacement driven upward force was generated by the computer which is uncontrollable and may cause a larger donning pressure.

The von Mises stress distribution pattern illustrated that stress was distributed along the middle portion of the anterior-lateral compression area of the residual limb. In contrast, stresses were concentrated at the upper portion for the other three compression areas. These concentrated stresses can be found internally or externally of the residual limb. Knowing the approximate location of stress concentration is enough regardless it is superficial or deep.

To improve the CRS socket design, the von Mises and the normal stresses should be more evenly distributed to enhance comfort. From the base of the inner CRS socket, the structure was depressed equally 15mm to achieve compression. With stress predicted, the socket structure can be modified to reduce the compression depth of upper parts of anterior-medial, posterior-medial, and posterior-lateral compression bars. After modification, stresses on the upper part of the residual limb would be reduced for better comfort.

This study did not exam the shear stress. However, we expected the shear stress elevation after removing of the upper brim of the socket. While the upper brim is removed, the pressure and shear stress previously taken by the upper brim of the conventional socket would shift to the lower part of prosthetic socket which causes shear stress elevation. It remains crucial to evaluate shear stress of the residual limb wearing the CRS socket as excessive shear force may cause pressure ulcer.

A single case study incorporating simplified material properties is a limitation. Another limitation is that the 800 N vertical load simulating weight bearing may not accurately reflect the real loading of the hip joint. In vivo, the joint reaction force is not vertical, but occurs in three-dimensional [33]. The loading magnitude and direction of the hip joint are determined by the angle of the femur head and muscle force. However, these parameters are difficult to measure and vary between different persons. The FE results may not be accurately validated if we are unable to measure the hip joint loading force and its direction. Our study simplified the load as vertical, which was quite often used by previous FE prosthetic socket studies. The CRS socket is a new transfemoral socket design. Our study is an essential starting point to understand basic biomechanics of CRS socket design through simplified FE model.

Conclusion

In this study, the design of the CRS socket was based on 3D reconstructed geometry of the residual limb. This approach of CRS socket design allowed assembly of the FE model to achieve a more reliable FE prediction. The simulation results provide an initial look into the design and optimization of the CRS socket structure. Further work is necessary to validate this process in a larger sample.

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Reference

[1] Dudek NL, Marks MB, Marshall SC. Skin problems in an amputee clinic. Am j Phys Med Rehabil 2006; 85:424-429.

[2] Dudek NL, Marks MB, Marshall SC, et al. Dermatologic conditions associated with use of a lower-extremity prosthesis. Arch Phys Med Rehabil 2005; 86:659-262.

[3] Lyon CC, Kulkarni J, Zimerson E, et al. Skin disorders in amputees. J Am Acad Dermatol 2000; 42: 501-507.

[4] Radcliffe CW. Functional considerations in the fitting of above-knee prostheses. Artif Limbs 1955; 2:35-60.

[5] Long IA. Allowing normal adduction of femur in above-knee amputations. Orthotics and Prosthetics 1975; 29: 53-54

[6] Schuch CM. Report from: international workshop on above-knee fitting and alignment techniques. Prosthet Orthot Int; 12: 81-98.

[7] Sabolich J. Contoured adducted trochanteric-controlled alignment method (CAT-CAM): introduction and basic principles. Clinic prosthetics and orthotics 1985; 9.

[8] Pritham CH. Biomechanics and shape of the above-knee socket considered in light of the ischial containment concept. Prosthet Orthot Int 1999; 14: 9-21.

[9] Schuch CM and Pritham CH. Current transfemoral sockets. Clin Orthop Relat Res 1999; 363: 48-54.

[10] Klotz R, Colobert B, Botino M, et al. Influence of different types of socket on the range of motion of hip joint by the transfemoral amputee. Ann Phys Rehabil Med 2011; 54: 399-410.

[11] Hagberg K, Haggstrom E, Uden M, et al. Socket versus bone-anchored trans-femoral prostheses: hip range of motion and sitting comfort. Prosthet Orthot Int 2005; 29:153-163.

[12] Trower TA. Changes in lower extremity prosthetic practice. Phys Med Rehabil Clin N Am 2006; 17: 23-30.

[13] Traballesi M, Delussu AS, Averna T, et al. Energy cost of walking in transfemoral amputees: comparison between Marlo Anatomical Socket and ischial containment socket. Gait Posture 2011; 34: 270-274.

[14] Alley RD, Walley TW 3rd, Albuquerque MJ, et al. Prosthetic sockets stabilized by alternating areas of tissue compression and release. J Rehabil Res Dev 2011; 48: 679-696.
[15] Alley RD, Williams TW 3rd. Method for use of a compression stabilized prosthetic socket interface. In: Google Patents 2012.
[16] Paterno L, Ibrahimi M, Gruppioni E, et al. Socket for limb prostheses: a review of existing technologies and open challenges. IEEE Trans Biomed Eng 2018; 65: 1996-2010.
[17] Cutler, TJ. Rehabilitation of an individual with transfemoral amputation combining Aquatic ambulation with prosthetic socket incorporating high-fidelity skeletal capture. J Prosthet Orthot. 2017; 29: 206-212.
[18] Lee WC, Zhang M, Jia Z, et al. Finite element modeling of the contact interface between trans-tibial residual limb and prosthetic socket. Med Eng Phys 2004; 26: 655-662.
[19] Restrepo V, Villarraga J, Palacio JP. Stress reduction in the residual limb of a transfemoral amputee varying the coefficient of friction. J Prosthet Orthot. 2014; 26: 205-211.
[20] Zhang LL, Zhu M, Shen L, et al. Finite element analysis of the contact interface between transfemoral-femoral stump and prosthetic socket. Conf Proc IEEE Eng Med Biol Soc 2013; 2013: 1270-1273.
[21] Lee WC, Zhang M, Jia X, et al. Finite element modeling of the contact interface between trans-tibial residual limb and prosthetic socket. Med Eng Phys 2004; 26: 655-662.
[22] Malinauskas M, Krouskop TA, Barry PA. Noninvasive measurement of the stiffness of tissue in the above-knee amputation limb. J. Rehab. Res. Dev. 1989; 26: 45-52.
[23] Jia X, Zhang M, Li X. A quasi-dynamic nonlinear finite element model to investigate prosthetic interface stresses during walking for trans-tibial amputees. Clin Biomech 2005; 20: 630-635.
[24] Dickinson AS, Steer JW, Worsley PR. Finite element analysis of the amputated lower limb: a systematic review and recommendations. Med Eng Phys 2017; 43:1-18.
[25] Lacroix D, Patino JF. Finite element analysis of donning procedure of a prosthetic transfemoral socket. Ann Biomed Eng 2011; 39: 2972-2983.

[26] Restrepo V, Villarraga J, Palacio JP. Stress reduction in the residual limb of a transfemoral amputee varying the coefficient of friction. J Prosthet Orthot. 2014; 26: 205-211.

[27] Robinovitch SN, McMahon TA, Hayes WC. Force attenuation in trochanteric soft tissues during impact from a fall. J Orthop Res 1995; 13: 956-62.

[28] Zhang M, Turner-Smith AR, Roberts VC, et al. Frictional action at lower limb/prosthetic socket interface. Med Eng Phys 1996; 18:207-214.

[29] Kahle JT, Highsmith MJ. Transfemoral sockets with vacuum-assisted suspension comparison of hip kinematics, socket position, contact pressure, and preference: Ischial containment versus brimless. J Rehabil Res Dev 2013; 50: 1241-1252.

[30] Colombo G, Morotti R, Rizzi C. FE analysis of contact between residual limb and socket during simulation of amputee motion. Computer-Aided Design & Applications 2014; 11: 381-388

[31] Velez ZJ, Bustamante GL, Villaraga OJ. Relation between residual limb length and stress distribution over stump for transfemoral amputees. Rev EIA 2015; 12: 107-115.

[32] Lee WC, Zhang M, Mak AF. Regional differences in pain threshold and tolerance of the transtibial residual limb: including the effects of age and interface material. Arch Phys Med Rehabil 2005; 86: 641-649.

[33] Monk AP, Simpson DJ, Riley ND, et al. Biomechanics in orthopaedics: considerations of the lower limb. Surgery 2013; 31: 455-451.