On possibility of unloading control at leg orthotics

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Abstract

In earlier papers the idea was formulated of programmed control of the limb unloading coefficient (CU) through change of the brace lateral compression when tightening. With regard to the shin the hypothesis was that due to circumferential compression the shin will be "pushed" out of the brace up, its contact with the sole be weakened, and the load be redistributed from the shin on the brace case.

With PAC DiaSled for measuring subplantar pressure fields conducted were systematic measurements in vivo of CU of the orthosed limb using under the brace both cotton and special slippery synthetic stockings. For cotton stockings no dependence of CU on compression was found, for slippery stockings in walking – as well, while in statics recorded was a sharp rise in CU with an increase of compression on the left side of the curve and the stabilization in the right part of it. For the simulation of unloading in orthotics measured were friction coefficients of pairs of leather-stocking and stocking-brace for cotton (0.48 and 0.57, respectively) and for synthetic stockings (0.42 and 0.16).

A model was considered of rough rigid wedge (the shin), covered by a conformal expandable rigid holder (brace sleeve) and loaded with vertical force (weight) and belt-like load (lateral compression). From the requirement of implementation of the regime of orthosis sliding with respect to the limb and based on their geometry the estimate of the value of the required friction coefficient between them k <= 0.25 was obtained. For cotton stockings, taking into account friction coefficients obtained the direction of the compression force acting on the shin is within the cone of friction. Therefore, conditions are realized of limb cohesion with the orthosis and lateral compression can not affect the unloading. For slippery synthetic stocking slippage regime is realized leading to change in CU by varying the circumferential compression.

Models of leg-brace system considering the deformability of soft tissues were studied numerically: conical models – based on the method of boundary integral equations, models of real forms – by finite element method using a specially made for the purpose computer tomograms of the shin in vivo. It was found that on the shin-brace contact surface slip areas arise. For a cotton stocking with a coefficient of friction between the shin and the orthosis k = 0.48 their size are strongly dependent on the geometry adopted, for synthetic stocking with k = 0.16 in all cases they are significant.

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1. Introduction

Nomenclature

\( F \)  force of shin lateral compression by orthosis  
\( G_m \)  shear modulus of elasticity of the lower leg muscles tissue  
\( k \)  frictional coefficient between the body (orthosis) and the base (shin)  
\( N \)  force of normal pressure on the part of the body (orthosis) on the base (shin)  
\( P \)  weight of trial subject  
\( Q \)  fictitious reaction of a smooth wall that appears when considering one of mirror-symmetrical halves of the model problem  
\( R \)  reaction of the support  
\( T \)  frictional force between the body (orthosis) and the base (shin)  
\( \alpha \)  shin «taper angle»  
\( \nu \)  Poisson's ratio of the lower leg muscles tissue  
\( \sigma \)  normal stresses at the boundary Shin-Orthosis  
\( \tau \)  tangential stresses at the boundary Shin-Orthosis

Unloading the damaged segment of lower limb is one of the major therapeutic functions of orthoses based on Nikitin (2016). The rate of healing and rehabilitation by Logvenkov and Stein (2006) depends on the level and the program of loading the damaged segment. Earlier Dashevskiy and Nikitin (2011, 2016) studied the mechanisms and degree of unloading at imposing orthosis to the lower limb. For the quantitative characterization of the degree of limb unloading in orthotics unloading coefficient (CU) was introduced, which was calculated as the ratio of the load on the unorthosed limb to the load on the orthosed one. In Dashevskiy and Nikitin (2011, 2016) the idea has been put forward of possibilities of efficiently controlling the unloading level of the affected area by varying the degree of orthosis tightening (lateral compression). With regard to the shin idea was that by circular compression occurring when tightening the brace, shin is "pushed" out of the brace up. This will weaken its contact with the bottom and the load will be redistributed from lower leg to the orthosis body.

However, the pilot experiments conducted by Dashevskiy and Nikitin (2011, 2016) did not reveal any explicit dependence of the CU on tightening. There was also formulated a hypothesis on the causes of such an effect. According to the biomechanical models of the "Foot-Orthosis" system, proposed by Dashevskiy and Nikitin (2011, 2016), in the absence of slippage between the leg and orthosis RC is independent of the tightness, in a slip the limb by increasing the lateral compression is pushed up and the CU will grow. In reality, conditions of contact between leg and orthosis correspond neither complete adhesion nor ideal slipping. It is rather dry friction conditions, which, depending on the values of the friction coefficient, mechanical and geometrical characteristics of the contacting bodies and the acting loads can lead to the formation of both areas of adhesion and sliding of different sizes. Thus, for theoretical analysis of the possibility of unloading control it is necessary to study the models of "Foot-Orthosis" system taking into account friction between the leg and the orthosis and find out for what values of parameters slippage can be realized.

2. Rigid-body model

Consider a simplified flat model of "Leg-orthosis" system as a rough rigid wedge (shin), covered by a conformal expandable rigid holder (brace sleeve) and loaded with vertical force (body weight) and lateral forces (lateral compression). (Fig. 1). Coulomb’s friction law can be described as follows based on Kuz’michev (1989). If the resultant force acting on the body in the tangential direction (excluding base actions) does not exceed \( kN \) (where \( k \)
Fig. 1. The scheme of loading the wedge (left) and the holder (right). \( P \) stands for trial subject weight, \( \alpha \) is shin «taper angle», \( N \) – force of normal pressure on the part of the body (orthosis) on the base (shin), \( T \) – frictional force between the body and the base, \( Q \) – fictitious reaction of a smooth wall that appears when considering one of mirror-symmetrical halves of the model problem, \( F \) – the force of shin lateral compression by orthosis, \( R \) – reaction of the support.

\[ F = P \cdot \frac{(\cos \alpha + \sin \alpha)}{(\sin \alpha - k \cos \alpha)} \]

We see that when \( \sin \alpha - k \cos \alpha = 0 \) (or \( \tan \alpha = k \)) the denominator of the right side of the formula obtained vanishes; when \( k > \tan \alpha \) a negative value is obtained for \( F \), and the formula becomes meaningless. From this the condition for \( F \) follows of feasibility of sliding mode in the adopted model: \( k < \tan \alpha \).

For the shin "cone angle" \( \alpha \) direct measurements provide an estimate \( \alpha \sim 15^0 \). This makes it possible to obtain an approximate estimate of what should be the coefficient of friction between the shin and the orthosis for the implementation of sliding mode: \( k \sim \tan \alpha \sim \tan 15^0 \approx 0.25 \).

3. Experimental determination of friction coefficients between skin and stocking, stocking and orthosis

Since one of the key characteristics that determine the implementation of the slippage mode is interfacial friction coefficients of friction between the lower leg and orthosis were measured.

According to the existing procedure when applying orthosis for hygienic reasons at first regular cotton stocking is put on the leg and only then the orthosis is imposed. Thus, the patient's skin contacts the stocking which external side is in contact with the orthosis sleeve. Since in measuring CU by Dashevskiy and Nikitin (2011, 2016) with a cotton stocking there was not observed CU growth with an increase in lateral compression, as well as visually there was seen no slipping of lower leg relative to the orthosis, we assumed that the friction coefficients for cotton stocking are too large and do not satisfy slip conditions. For this reason, along with cotton special synthetic
would satisfy the condition that at the Shin-Orthosis seizing boundary zones would appeared where shear stresses reach respectively for the tangential and normal stresses at the boundary Shin-Orthosis, then the emergence of slip areas (79 kg), is placed into a rigid shin-conformal cone (orthosis), engaged with it and loaded with the body weight: lower limb by elastic isotropic double truncated cone (muscles) with a rigid cylindrical core (tubular bone), the foot Formulated and based on the boundary integral equation method considered was the problem of modeling of the bodies, as well as by the magnitudes of current loads.

To take into account deformability of soft tissues Leg-Orthosis system models were studied numerically. Formulated and based on the boundary integral equation method considered was the problem of modeling of the lower limb by elastic isotropic double truncated cone (muscles) with a rigid cylindrical core (tubular bone), the foot is placed into a rigid shin-conformal cone (orthosis), engaged with it and loaded with the body weight: \( P = 774 \text{ N} \) (79 kg), \( v = 0.5, G_m = 10.9\times10^3 \text{ Pa} \) where, respectively, \( P \) is trial subject weight, \( v \) – Poisson’s ratio and \( G_m \) – shear modulus of elasticity of the lower leg muscles according to Dashevskiy and Timanin (2014).

Since at dry friction by Coulomb’s law relation \( \tau \leq k\sigma \) must hold on the contact surface, where \( \tau \) and \( \sigma \) stands respectively for the tangential and normal stresses at the boundary Shin-Orthosis, then the emergence of slip areas need that at the Shin-Orthosis seizing boundary zones would appeared where shear stresses reach \( k\sigma \) values, i.e., would satisfy the condition \( \tau / \sigma \geq k \). Fig. 3 shows a graph of the ratio \( \tau / \sigma \) (\( \tau \) along the boundary changes its sign).
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Thus, the coefficients of friction, will also be affected by the mechanical and geometric characteristics of the contacting tissues and individual patient characteristics. However, the situation becomes complicated and the picture in addition respectively, for synthetics measured coefficients of friction with HDPE (orthosis) and skin were found to be: for cotton stocking 0.57 and 0.48 in a photo editor. In three such measurements calculated was the average value of the coefficient of friction.

Angle and corresponding coefficient of friction (equal to its tangent) were determined by treating the resulting photo gradually increased until the moment when the slip began. This moment was photographed (Fig. 2a), then the slip was placed of a 4-mm HDPE size ~ 100 x 100 mm, and then the construction angle of inclination to the horizon was made for a more realistic in vivo description of the geometry of the limb (as well as of the orthosis) multi-slice computer tomography (MSCT) of the shin according to the technology "CT scan -> Specialized processing software (Mimics) -> FEA package (ABACUS)". The shear modulus for muscle tissue was taken according to Timanin and Eremin (2012) equal to $G_m \approx 24.3 \times 10^3$ Pa, i.e. for shin muscles in the active (tight) state. Fig. 4 shows the distribution of areas of seizing (light), sliding (gray) and the lack of contact (dark) with $k = 0.4$ (left) and $k = 0.1$ (right). It can be seen that according to the calculations at $k = 0.1$ there are no areas of seizing at all.

5. Account for deformability. Model of real form

Analogous turned out the results for problems with the shin (and orthoses) of real form, obtained on the basis of made for a more realistic in vivo description of the geometry of the limb (as well as of the orthosis) multi-slice computer tomography (MSCT) of the shin according to the technology "CT scan -> Specialized processing software (Mimics) -> FEA package (ABACUS)". The shear modulus for muscle tissue was taken according to Timanin and Eremin (2012) equal to $G_m \approx 24.3 \times 10^3$ Pa, i.e. for shin muscles in the active (tight) state. Fig. 4 shows the distribution of areas of seizing (light), sliding (gray) and the lack of contact (dark) with $k = 0.4$ (left) and $k = 0.1$ (right). It can be seen that according to the calculations at $k = 0.1$ there are no areas of seizing at all.

6. Experimental study of the dependence of the coefficient of unloading on the circumferential tightness for cotton and synthetic stockings

Measurement of feet pressure on the support was carried out with the help of software and hardware complex (SHC) "DiaSled" using special tensometric insoles inserted in parallel directly beneath both feet: for the right (orthosed) foot – into the orthosis, for the left one – right into the shoes. Thereat a number of additional measures to reduce the possibility of errors and increase accuracy was undertaken. Tests were conducted on healthy trial subject with intact limbs. In the experiment, a specially made individual orthosis-halfboot from the toes to the mid-shin with three Velcro was used (Fig. 5a).

Changing circumferential compression was performed by tightening the top fastening. Tightening was measured using millimeter tape pasted on the orthosis sleeve below the fastening (Fig. 5b). In Fig. 5c a photo of the trial subject is given in the orthosis with sensors installed of the "DiaSled" system. Fig. 6 presents a typical view of SHC DiaSled windows: Fig. 6a – upon standing on each leg alternately, Fig. 6b – when walking; in the middle of each
window the graphs of integral load (GIL) for both feet are given. The coefficient of unloading (CU) was calculated from GIL (while standing or all over the single measurement when walking) as the ratio of the maximum load on the unorthosed limb to the maximum load on the limb in an orthosis. Tightness varied from 0 to 4 or 5 in increments of 1. For a fixed tightness three measurements were conducted, for each of them the CU was calculated, and then the average. The plots of CU on the tightness built on these data, have the form of Fig. 7.

As seen from Fig. 7a, when using cotton stockings experimental results do not show valid growth of CU with an increase in lateral tightening (compression) neither while standing nor walking.

For synthetic stockings (Fig. 7b) CU values when walking found themselves 20-25% higher compared to cotton stockings, with reliable change of CU on the tightness not observed. At statics recorded was a sharp increase in CU with increasing the tightness in the left half of the curve and stabilization in the right part of it. For both types of stockings CU values were higher while standing than when walking, but for synthetics the difference appeared to be more significant and amounted to 1.5-2 times.

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Fig. 4. Arrangement of areas of adhesion, sliding and the lack of contact on the Shin-Orthosis interface for: (a) $k = 0.4$; (b) $k = 0.1$.

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Fig. 5a  Fig. 5b  Fig. 5c

Fig. 5. (a) the orthosis used; (b) measuring tape; (c) trial subject in the orthosis with sensors installed of the "DiaSled" system.
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Fig. 6. Typical view of SHC DiaSled windows: (a) upon standing on each leg alternately; (b) when walking.

If the observed increase in static CU is really determined by mechanism of pushing the limb up at circumferential tightening, then cessation of this growth at high tightness may be due to purely geometrical constraint of pushing by the orthosis upper cover, against which the back side of the foot begins to bear. To verify this, it is necessary to come up with, make and test an orthosis, free from such restrictions. Another possible reason – occurring at high tightness sweating and adhesion, leading to sticking and seizing.

We can not yet explain why CU does not depend on the tightness when walking. Perhaps it requires bringing in dynamic considerations.

There might be other approaches to the problem of load programming at limb orthotics. As already indicated by Dashevskiy and Nikitin (2016), this approach may be associated with such positioning and fixing limbs, which would provide a low initial load on the segment. As the recovery proceeds the load can be gradually increased, easing the tightness. The limiting case here is the creation of the technical gap between the foot and the orthosis sole when applying orthosis (e.g., using an appropriate gasket inserted between the foot and the orthosis when taking the cast of the limb). More complex schemes may include adjustable gaps created with special devices (runners and screws, etc.), spring or other paddywhack tabs which would allow both to regulate and, coming together with the strain gauges in the sensor unit, to control the load on limb like Aidarov (2010).

Fig. 7. Plots of unloading coefficient on the tightness: (a) for cotton stockings; (b) for "slippery" stockings.
However, the main thing is controlling the level of loading a limb. To do this, apparently some simple sensor may be used placed between the heel and the orthosis and serving for orthosis individual customization to provide adequate load on the affected segment of the limb.

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References


