Energy dissipation caused by plantar-sole frictional movement during gait

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1. INTRODUCTION

Diabetic foot is a consequence of diabetes mellitus disorder. Lack of insulin production can lead to high blood glucose level and diabetic neuropathy which may affect the nerves of lower limbs and specially feet. Diabetic patients are in danger of suffering non sensitive lesions [1] which can occur because of slips and falls or a combination of adverse feet microclimate and high pressure. Slips and falls could depend on the slip-resistant properties of flooring and shoes, evaluated by measurements of the coefficient of friction (COF) of the shoe/floor interface [2]. Adverse microclimate and pressure are usually tackled by appropriate footwear and hosiery recommendations, not always efficient.

Appropriate footwear can help prevent lesions as it redistributes plantar pressure and reduces shear which are known to produce sores and ulcers [3]. Nevertheless, research on foot care showed that appropriate shoes and insoles are not enough, and foot/contact sole surface must also be studied [4]. The effects of soft tissue and other structural bony characteristics on the stress distribution of the plantar surface is complementary to achieve a good standard of foot care [5].

The evaluation of footwear used by diabetic foot patients is difficult due to the neuropathy or, in serious cases, due to foot deformity. Because their sensitivity for pressure and shear force has decreased, it is difficult for clinicians to estimate the effect of footwear, even relying only the clinical evidence throughout follow up [6]. However, intervention by footwear is one of the most widespread methods to improve gait parameters for both prevention and therapy. Gait has been widely and variously analyzed in clinical settings for diabetic foot [7].

Likewise, hosiery helps to regulate foot microclimate, relief pressure and protect the skin from abrasion. During gait a person takes about 8.000 to 10.000 steps a day, so foot presses and rubs against footgear and can produce blisters and calluses [2]. Repetitive pressure and shear on the foot are correlated with the development of blister and further ulcers [11]. Methods of pressure measurement can be either static or dynamic, but in all cases plantar pressure is important [8]. Despite the difficulties there are only very a few commercial pressure sensors and baropodometric instruments available [9] and it is clear that shear and friction coefficient measurements are yet to be extensively studied [10] [4].

The study the mechanical effect of sock with different frictional properties on foot during the stance phases of gait could help model the foot-insole interface to investigate the effect of insole cushioning on plantar pressure distribution. Even though direct measurement is difficult, a model of the foot-sock-insole interface using a 3D model with finite element models can be used to investigate the biomechanical effects of soft tissue stiffening in diabetic foot [5], [7].

The objective of this paper is to present a model to estimate the dissipated energy during gait based on acceleration and pressure data at selected contact points in plantar area.

2. MATERIAL AND METHODS

To establish the method needed to estimate the energy dissipated by friction during gait, we looked at the literature seeking for previous descriptions. In the absence of engineering data referred to skin energy dissipation, we develop a simple model, as a first approximation, with the following variables:

- Weight of the subject: M
- Number of foot sole contacts: $i = 1, \ldots, n$
- Fraction of the subject mass on sensor i: g_i
- Area of sensor i: a_i
- Pressure on sensor i: P_i
- Coefficient of skin to sole friction: $\mu_i = \mu i = 1, \ldots, n$
- Vertical force at point i: $Fv_i = P_i a_i$

- Horizontal force at point i: $Fh_i = Fv_i \mu_i$
- Foot displacement with respect to sole: d
- Energy at point i: $E_i = Fh_i d$
- Time of foot – sole displacement: T
- Power at point i: $W_i = E_i / T$
- Ratio of displacement time with respect to stance time: $r = T / stancetime$
- Mean power during gait: $Wm_i = W_i r$

The adopted model is derived from basic friction theory. Friction is verified when a horizontal force acts on the moving object over a fixed surface. The static coefficient of friction µ allows to estimate the horizontal force knowing the vertical force (Fig. 1). Because dynamic friction involves a lesser coefficient of friction and therefore a minor horizontal force, for the purpose of this research, we consider only static friction. The model explores friction in the forward direction but could also be applied sideways, thus describing a 3D phenomenon.

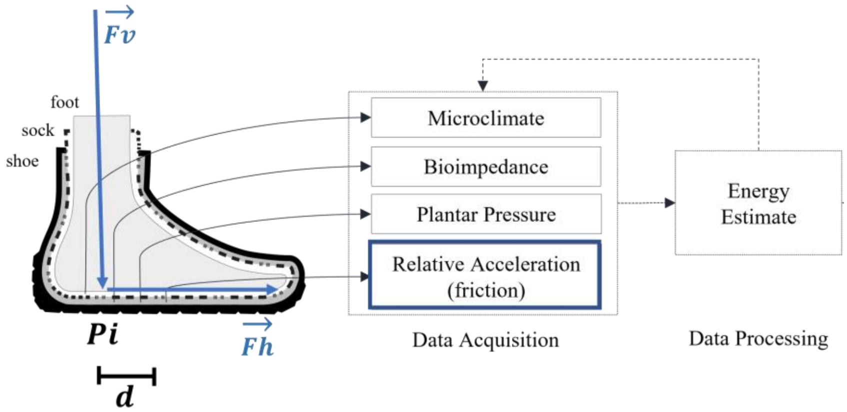

Figure 1. Main variables of the model to study the friction of gait between skin and immediate shoeing layer.

Using the model, we have calculated the dissipated energy as a function of displacement, for all combinations of possible variables as shown in Table 1. Having fixed one of the variables, sensor area $a_i = 4cm^2$, we have calculated the instant power according to the remaining combinations of variables. Finally, having fixed the 'duty cycle' of dissipated energy within the gait cycle at $r = 0.1$, we have calculated the mean power over a gait cycle.

3. Results

To obtain practical figures of the phenomena involved, and specifically the orders of magnitude of energy and mean power at the skin level, we have assigned values as shown in Table 1.

Table 1. Variables values for the model of skin sole friction

Pressure	Pressure on sensor	Area of sensor	Foot sole displacement time	COF	Foot sole displacement	Ratio displacement time – stance time
P Kg/cm2	P_i Kpa	a_i cm2	T s	μ_i	d mm	r
2	196.133	0.75	0.1	1	0.10	0.05
4	392.266	2	0.2	1	0.20	0.10
8	784.532	4	0.4	1	0.40	0.20
10	980.665	12.25	0.8	1	0.80	0.40

[a.] COF: coefficient of friction

Fig. 2 shows the energy $Ei = f(d)$ for 4 values of the parameter sensor area from 0.75 cm^2 to 12.25 cm^2. Considering that a constant pressure with increasing contact areas is equivalent to having heavier subjects, the exact pressure on each sensor depends upon the weight of the subject as well as the weight distribution among the *i* contact points. The simulated energy, with the values adopted, spans from 0.15 Nmm to 19.6 Nmm.

Fig. 3 shows the power $Wi = f(d)$ for 4 values of the parameter foot sole displacement time from 0.1 s to 0.8 s. This figure is based on a constant pressure (2 Kg/cm^2) and a constant contact area (4 cm^2) which is equivalent to considering a subject exercising a vertical force of 8 Kg on one of the *i* contact points. The simulated power spans from 1 mW to 64 mW.

To evaluate the long-term effects of friction we have considered the "duty cycle" of friction over the whole gait cycle. In addition to a fixed pressure, a fixed sensor area of 4 cm^2 and a displacement time of 0.01 s, we explore in Fig. 4 the effect of four friction durations expressed as fractions of the gait cycle. If the friction time is a small fraction of gait, the mean power will be lower than when the friction time is a significant part of stance phase, such as during quick walking or running. Fig. 4 thus explores this effect of speed on mean power due to foot sole friction. Considering a constant pressure of 2 Kg/cm^2, a constant contact area of 4 cm^2 and a displacement time of 0.01 s, we simulated 4 stance times of 1 s to 4 s. The simulated mean power spans from 0.03 mW to 12.80 mW.

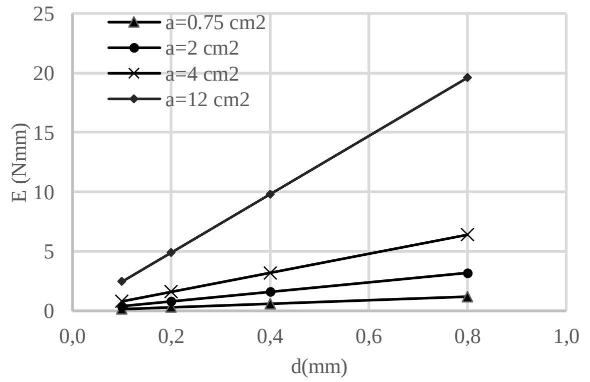

Figure 2. Energy as a function of displacement with 4 values of sensor area.

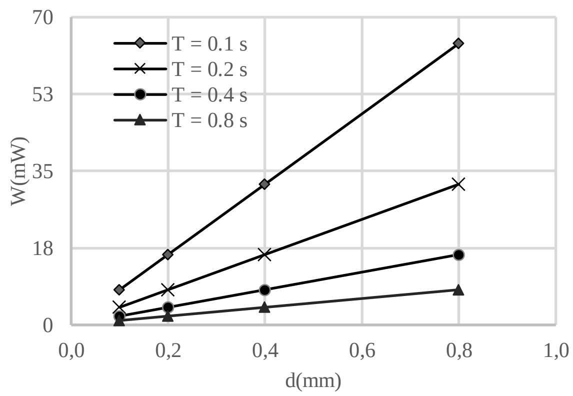

Figure 3. Power as a function of displacement with 4 values of friction time

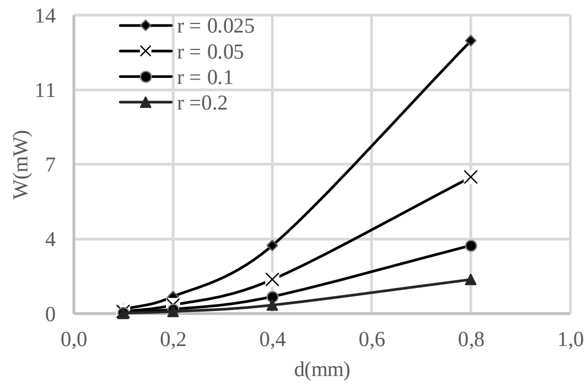

Figure 4. Mean power as a function of displacement with 4 values of gait speed.

4. Discussion

The foot contacts the support surface by adapting to it, releasing itself, then stiffens, becoming a lever to react to the surface itself [5]. The reaction of the foot to the surface implies a compression of tissues and some friction, which we have address in this paper. Several studies have investigated the effects of the friction coefficient on the risk of slip and slip-induced falls, resulting in proposals for appropriate friction coefficients at the shoe-floor interface to prevent slipping [10]. Our problem is concerned nevertheless with skin to hosiery friction and not barefoot to floor, as the flooring industry is interested. The design of DIAPODAL[8] would benefit to have a threshold of energy above which the risk of skin lesion increases. The energy and power involved in gait is an indication of the effect of friction on the skin.

As was shown in Fig. 2, the energy dissipated at a skin sole contact point as a function of foot displacement with an area of 0.75 cm^2, supporting a fraction of the mass of a subject estimated at 2 Kg/cm^2, is below 1 Nmm, whereas for a much heavier subject and a contact area of 2 cm^2 the energy is tenfold, of the order of 20 Nmm. The highest frictional dissipated energy in Fig. 2 is associated with the largest frictional displacement of 0.8 mm between skin and hosiery. All other things being equal, more displacement is equivalent to more dissipated energy. Supposing we could estimate the horizontal forward displacement of each contact point, DIAPODAL could calculate the energy dissipated at each step. A previously determined threshold could help deduce an excess energy, eventually leading to skin lesions.

Fig. 3 shows the instant power dissipated at the skin sole contact point with instant power values from 1 mW to 65 mW. The skin reacts to friction by physiological energy dissipation distributed over time. An intense power must be followed by neutral times. Fig. 4 allows for energy dissipation from one friction episode to the next one during the following step. A slower gait allows more time for the skin to cool down. The mean friction power according to our model applied to plausible values spans from 0.03 mW to 12.80 mW.

The present preliminary values of dissipated energy, power, and mean power at each contact point of a diabetic foot with underlying hosiery is a further step towards the design of DIAPODAL, as a real time wearable footwear device to protect skin from lesions during gait. A more detailed analysis will be performed with finite elements. The numerical figures given here should be taken as the initial iteration of an approach of increasing precision to the problem of real time energy estimation during gait.

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