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The coefficient of friction alterations in stroke gait

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Abstract*— The aim of this study is to analyze the possible coefficient of friction (COF) pattern alterations during barefoot gait of patients after stroke compared to a control group. Twenty-four volunteers have attended to this study: 12 patients post Stroke and 12 healthy age-matched subjects as a control group. First, COF curve was calculated as the ratio of the shear to normal ground reaction force (GRF) during stance phase of gait cycle, then the specific peaks (P1, P2 and P3) and valleys (V1 and V2) of the COF curve were extracted and normalized by the walking velocity. Comparisons between the stroke group affected side (AS), stroke group non-affected side (NAS) and control group (CG) were evaluated using the two samples T-test comparing every 1% of support phase (SP) the groups. To compare the peaks and valleys differences among the groups, the one-way ANOVA and the Turkey post-hoc test were applied for the parametric data. Nonparametric data was analyzed by Kruskall-Wallis test and the Bonferroni post-hoc test. The comparisons between AS and CG reveal differences in initial contact (5% to 8% of the SP), loading response (13% to 29% of the SP), mid stance (46% to 67% of the SP), and terminal stance to pre swing phases (77% to 70% of the SP). For NAS and CG, differences were found in loading response (13% to 38% of the SP) and terminal stance to pre swing phases (79% to 100% of the SP). Differences between AS and NAS were seen in the mid stance phase (45% to 60% of the SP). The stroke group have shown a gait velocity reduction and, consequently, had reduced the COF to perform gait safe, which is possibly related to compensatory strategies due to the altered AS motion during swing. Moreover, when compared with the CG, the stroke groups AS and NAS presented higher P1, V2 and P3. Once these variables were normalized by the walking velocity, the results of this study show that in patients with stroke for the AS and NAS the initial contact, the mid stance and the terminal stance seem to be critical phases for the incidence of slips.

***Index Terms*—Coefficient of Friction, falls, gait, stroke.**

I. INTRODUCTION

Abnormal gait significantly limits the patients' autonomy and capacity of participation, and also contributes to decrease their life quality [1], [2]. Hemiplegia is one of the most common impairments observed after stroke and it contributes significantly to reduce gait performance. About 50% to 60% of patients that complete the standard rehabilitation after a stroke still experience some degree of motor impairment, and approximately 50% are at least partly dependent in activities-of-daily-living [3]. Thus, one of the earliest concerns of stroke patients and their families relates to walking issues [4], therefore one of the focuses on the intervention after strokes is to treat gait abnormalities [4].

The gait pattern of individuals post-stroke is often characterized by movement initiation delays, inefficient movement patterns on the hemi paretic side, decreased stance time on the paretic side, and premature toe off during terminal stance, when compared to healthy adults [5]-[7]. Studies have shown that cognitive deficits, functional impairment, and impaired balance are related to fall incidence in stroke patients [8], [9].

During walking, slips are the results of a loss of friction between the foot and the floor. A slip is likely to occur when the required coefficient of friction (RCOF) of an individual exceeds the available coefficient of friction at the foot floor interface. The RCOF is the minimum coefficient of friction (COF) that is necessary at the foot and floor interface to support the human locomotion and it is usually measured on dry surfaces.

In the research setting, the RCOF generated during walking is determined from the recordings of the ground reaction forces (GRF) by a force plate [10]–[12]. To determine the RCOF, the instantaneous COF curve is calculated as the ratio between the shear (resultant of lateral and anterior posterior GRF components) and the vertical GRF component generated by a person while walking across a given dry surface. Then, some peaks and valleys in instantaneous COF are extracted and the RCOF is typically considered to be the one of the local maximum of the instantaneous COF, generally it is observed during the loading response and terminal stance phases of the support phase [10]-[12]. The Figure 1 illustrates de COF peaks and valleys.

There is a variety of factors that need to be taken into consideration when the COF is analyzed. For example, previous studies have shown that the COF peaks vary with age [13], [14], gender [13], [14], walking speed [13] and the presence of a disability [15]-[18]. Those with a disability would appear to be at potentially greater risk owing to the larger changes in gait characteristics and GRFs [15], [18].

In our previous papers [14], [18], we explore the in influence of the flooring type in the RCOF in elderly and stroke gait. We saw that more than differences in the flooring

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type this variable is able to distingue the stroke affected to the stroke less affected sides in the loading response and toe off phases. However, to the best of our knowledge, the COF curves during the gait of stroke patients have not yet been fully studied. So, our aim is to analyze the COF instantaneous curves of these patients during the barefoot gait and consequently kinetics aspects of hemiplegic gait.

II. METHODS

The Research Ethics Committee has approved this study (protocol No. 319/2011) and the volunteers have given written inform consents to participate at it.

Participants

The hemi paretic group (HG) was formed by 12 individuals affected by stroke (5 females and 7 males). The HG average characteristics were: age = 62.83 ± 6.86 years; body mass = 69.50 ± 13.96 kg; height = 1.68 ± 0.06 m; Fugl-Meyer = 88.25 ± 6.95; Berg Balance Scale = 47.16 ± 8.13; DGI = 16.25±4.13; Mini-mental = 21.33±4.61; months after stroke = 6.1 ± 2.8 months. The control group (CG) consisted of 12 healthy adult (5 females and 7 males) and the average characteristics were: age = 63.58 ± 6.94 years; body mass = 73.08 ± 14.31 kg; height = 1.69 ± 0.05 m.

Experimental Procedures for motion analysis

Each participant was oriented to walk barefoot along the pathway covered by the experimental flooring and over two force platforms (Kistler 9286BA), embedded in the data collection room floor, at himself/herself chosen speed. The participants were aware of the position of the force plates. Three trials were performed for each subject in order to guarantee the consistency of the data. The force plates' vertical, anterior-posterior and lateral ground reaction force components were normalized by the subject body weight (%BW) and expressed in function of the percentage of support phase. Data acquisition was performed using BioWare software (Version 4.0.x). Kinetic raw data were filtered using a 2nd order low-pass digital Butterworth filter, with a cut-off frequency of 10 Hz. An algorithm developed in Matlab was used to filter the raw data and calculate dependent variables.

First, the COF curve was calculated as the ratio of the shear to normal ground reaction force (GRF) during stance [10], [11], as described in Equation 1.

$$COF = \frac{\sqrt{(FY)^2+(FX)^2}}{FZ} \qquad (1)$$

In which FY is the anterior-posterior GRF, FX is the lateral GRF and FZ is the vertical GRF.

Then, as illustrated in Figure 1, the following parameters of the COF curves were calculated:

RCOF1 (P1): was calculated as the maximum value between the 9-15% of the COF curve;

Valley (V1): was calculated as the minimum value between the 15-80% of the COF curve;

RCOF2 (P2): was calculated as the maximum value between the 81-100% of the COF curve.

Since the COF can be affected by walking velocity, these variables were also normalized by the walking velocity (stride length/stride duration).

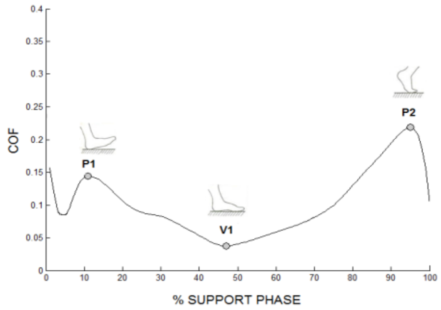

Fig 1. Illustration of COF curve variables. Legend: COF: coefficient of friction; %SUPPORT PHASE: percent of support phase; P1 = RCOF1; V1 = Valley1; P2=RCOF2.

In order to calculate the statistical analysis, the data normality was tested by Kolmoronov-Smirnov test. Then, to compare the differences between stroke hemi body (affected side - AS and non-affected side - NAS) and control group, the parametric data was analyzed by one-way ANOVA and the Tukey post-hoc test (α<0.05); the nonparametric data was analyzed by Kruskall-Wallis test and the Bonferroni post-hoc test (α<0.05). Also, comparisons between the AS, NAS and CG COF instantaneous curves were made by the two sample T-test (α<0.05) comparing every 1% of gait cycle. The software SPSS (version 19) was used to perform all statistical analysis.

III. RESULTS

The ANOVA one-way test have revealed no significant differences between stroke and control groups when considering age (F1,23=0.071; p=0.793), body mass (F1,23=0.385; p=0.541) and height (F1,23=0.352; p=0.559).

Table 1 presents the discrete variable means and standard deviation for each group as well as the statistical results. When the COF curves' peaks and valley were compared, differences were found in V1nor, to which during the mid-stance phase both stroke AS and NAS presented higher values than the matched control group. Differences were also observed in P2nor, during the terminal stance the control group presented lower values than the stroke AS and NAS.

Table 1. The discrete COF variables mean and standard deviation for each group and variable and the statistical results.

Var	Groups			KW	P
	AS	NAS	Control		
P_1nor	0,25±0,08	0,28±0,15	0,17±0,03	1,083	0,582
V_1nor	**0,09±0,06•**	**0,11±0,13°**	**0,03±0,01•°**	**7,407**	**0,025**
P_2nor	**0,46±0,34•**	**0,40±0,18°**	**0,29±0,03•°**	**8,722**	**0,013**

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Legend: Var = Variables; V1nor = Valley1 normalized by the gait velocity; P1nor = RCOF1 normalized by the gait velocity; P2nor = RCOF2 normalized by the gait velocity; AS = stroke group affected side; NAS = stroke group non-affected side; Control = Control group; • = differences between AS and Control; ° = differences between NAS and Control.

The COF instantaneous curves' analysis highlights the phases during the support phase where the Stroke patients AS and NAS have presented alterations compared to the control group, and it have shown differences in between the stroke symmetry (AS versus NAS).

When comparing AS and the control group (Figure 2a), differences were seen on initial contact (5% to 8% of the support phase), loading response (13% to 29% of the support phase), mid stance (46% to 67% of the support phase) and terminal stance to pre swing phases (77% to 70% of the support phase). When the NAS and the control group (Figure 2b) were compared, differences were found on loading response (13% to 38% of the support phase) and terminal stance to pre swing phases (79% to 100% of the support phase). Differences between AS and NAS (Figure 2c) were found in the mid stance phase (45% to 60% of the support phase).

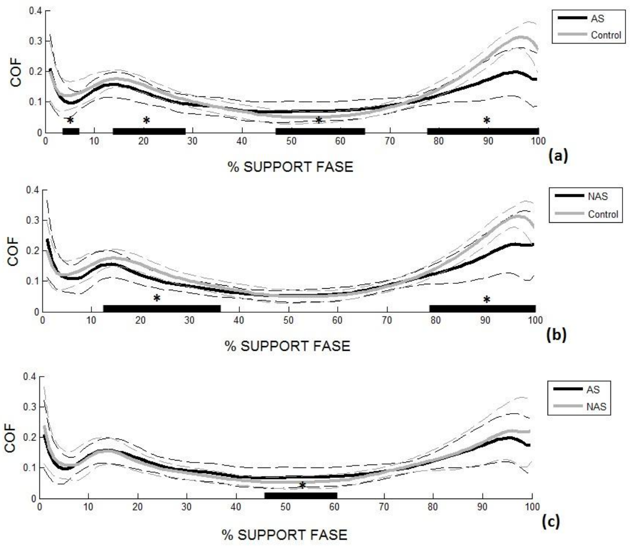

Fig 2. COF curve's mean and standard deviation in the comparisons between: (a) Stroke affected side (black solid line – mean, and black dashed line - standard deviation) and Control Group (grey line – mean, and grey dashed line - standard deviation); (b) Stroke non-affected side (black solid line – mean, and black dashed line - standard deviation) and Control Group (grey line – mean, and grey dashed line - standard deviation); and, (c) Stroke affected side (grey solid line – mean, and grey dashed line - standard deviation) and Stroke non-affected side (black solid line – mean, and black dashed line - standard deviation).

The bars and asterisks on the x-axes indicate the moments of the support phase that presented significant differences ($P \leq 0.05$) between the groups. Legend: AS = Stroke Group Affected Side; NAS = Stroke Group Non-affected Side; Control = Control Group; %SUPPORT PHASE = normalized by the percentage of the support phase.

IV. DISCUSSION

This study compared the gait of stroke patients and healthy age matched peers in an effort to quantify differences that may be predisposing the stroke population to falls.

When analyzing the instantaneous COF curves, it was noted that in normal gait patients the COF were higher than the stroke group near to the loading response and terminal stance phases. During these phases, the COF was actually higher when compared to the other stance phases, to firstly permit the deceleration phase for the loading acceptance and secondly the acceleration phase for guaranteeing the gait progression. It permits the right grip and consequently the transmission of the developed forces to the kinematic chain, reducing the slipping and the risk of falls. The loading response and the terminal stance are the critical phases in which slips often occur: the lower the friction is in these phases, the higher is the slipping risk [10], [11].

The analysis of the COF curves of the pathological group – for both AS and NAS sides – have evidenced lower values of COF on the same phases, pointing out a diminished grip on deceleration and acceleration phases. It seems that the stroke group reduced the gait velocity and, consequently, reduced the necessary (required) COF to perform the gait safely. In normal gait, once the deceleration phase starts, the COF decreases and the inertia forces sustain the gait progression: in that phase, the mid-stance, the contribution of the shear forces decreases in order to invert the decelerated movement and prepare the following acceleration phase. Considering the COF curve of AS of stroke participants, the COF showed higher values on this phase, pointing out a greater grip and a constraint in the motion inversion: this behavior breaks the contralateral side swing (NAS) decreasing the smoothness and increasing the level of balance uncertainty. The comparison between AS and NAS evidenced a statistical difference on the COF values in the mid-stance: on this phase the calf muscle is at its maximum stretching (the ankle reaches the maximum dorsiflexion and the knee the maximum extension – 19) and spasticity [6], [20] probably playing an important role in the progression constraint in AS.

Moreover, when compared to the control group, the stroke group AS and NAS presented higher V1nor and P2nor. Once these variables were normalized by the walking velocity, the results of this study show that in patients with stroke for the AS and NAS the mid stance and the terminal stance seems to be critical phases for slips incidence. This behavior can be explained by the stroke patients dropped foot. It is commonly described by kinematic deviations at the ankle – foot including forefoot or flat foot initial contact leading to reduced stability during stance [5]. Stroke related to ankle impairments causes inadequate dorsiflexion control during gait, including weakness of dorsiflexors, spasticity of plantar flexors, passive stiffness of the plantar flexors, and abnormal muscle co activation [21]. Moreover, limited ankle dorsiflexion and knee flexion during swing on AS often result in the use of compensatory strategies (i.e. pelvic hiking and circunduction) to achieve foot clearance [22]-[24].

The shear forces are higher near the initial contact, loading response and terminal stance-to-pre swing phases in the CG COF curve; this pattern is not the same for the stroke group. The main differences between the stroke patients and the control group is that both AS and NAS performed lower shear forces during the loading response and terminal-to-pre swing phases. During these phases, the lower the friction was the higher was the risk of falling. Moreover, the AS group performed higher COF values in the mid stance than the NAS group and CG. In this case, the higher the friction, the higher was the risk of tripping.

V. CONCLUSION

The stroke group have reduced the gait velocity and, consequently, have reduced the COF to perform the gait safely, what is probably related to compensatory strategies due to the altered AS motion during swing. The COF normalized by the walking velocity can be useful in predicting the real fall propensity of a stroke patient and to develop more effective therapy for the gait improvement. Moreover, the normalized COF shows that the mid stance and the terminal stance are phases of critical importance in determining if the frictional capabilities of the foot/floor interface to prevent slips in Stroke patients.

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REFERENCES

[1] T.B. Wyller, and M. Kirkevold, "How does a cerebral stroke affect quality of life? Towards an adequate theoretical account", Disabil. Rehabil, vol. 21, pp. 152–161, 1999.

[2] J. Perry, M. Garrett, J.K. Gronley, and S.J. Mulroy, "Classification of walking handicap in the stroke population", Stroke, vol. 26, pp. 982–989, 1995.

[3] J.D. Schaechter, "Motor rehabilitation and brain plasticity after hemi paretic stroke", Progress in Neurobiology, vol. 73, pp. 61-72, 2004.

[4] N.E. Mayo, N.A. Korner-Bitensky, and R. Becker, "Recovery time of independent function post-stroke", Am J Phys Med Rehabil, vol. 70, pp. 5–12, 1991.

[5] S.J. Olney, and C. Richards, "Hemi paretic gait following stroke. Part I: Characteristics", Gait Posture, vol. 4, pp.136-148, 1996.

[6] D.A. Winter, "Biomechanical motor patterns in normal walking", J Mot Behav, vol. 15, pp. 302–30, 1983.

[7] D.A.Winter, "Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences", Clin Orthop Relat Res, vol. 1, pp. 147–154, 1983.

[8] R.W. Teasell, and L. Kalra, "What's new in stroke rehabilitation. Stroke", vol. 35, pp. 383–385, 2004.

[9] J.E. Harris, J.J. Eng, D.S. Marigold, C.D. Tokuno, and C.L. Louis, "Relationship of Balance and Mobility to Fall Incidence in People with Chronic Stroke", Phys Ther, vol. 85, pp.150-158, 2005.

[10] W.R. Chang, C.C. Chang, and S. Matz, "Comparison of different methods to extract the required coefficient of friction for level walking", Ergonomics, vol. 55, no. 3, pp. 308-315, 2012.

[11] M.S. Redfern, R. Cham, and K. Gielo-Perczak, "Biomechanics of Slips", Ergonomics, vol. 44, no. 13, pp. 1138-1166, 2001.

[12] J.P. Hanson, M.S. Redfern, and M. Mazumdar, "Predicting slips and falls considering required and available friction", Ergonomics, vol. 42, no. 12, pp. 1619-1633, 1999.

[13] J.M. Burn field, and C.M. Powers, "Influence of age and gender of utilized coefficient of friction during walking at different speeds", In: Marpet MI, Sapienza MA, editors, 2003.

[14] A.F.R. Kleiner, M. Galli, A.A. Carmo, and R.M.L. Barros, "Effects of flooring on required coefficient of friction: elderly adult vs. middle-aged adult barefoot gait", Applied Ergonomics, vol. 50, pp. 147-152, 2015.

[15] J.V. Durá, E. Alcántara, T. Zamora, E. Balaguer, and D. Rosa, "Identification of floor friction safety level for public buildings considering mobility disabled people needs", Safety Sci, vol. 43, pp. 407-423, 2005.

[16] T. Lockhart, S. Kim, R. Kapur, and S. Jarrott, "Evaluation of Gait Characteristics and Ground Reaction Forces in Cognitively Declined Older Adults With an Emphasis on Slip-Induced Falls", Assist Technol, vol. 21, no. 4, pp. 188–195, 2009.

[17] C.A. Haynes, and T.E. Lockhart, "Evaluation of Gait and Slip Parameters for Adults with Intellectual Disability", J Biomech, vol. 45, no. 14, pp. 2337–2341, 2012.

[18] A.F.R. Kleiner, M. Galli, C. Rigoldi, AA. Carmo, and R.M.L. Barros, "Effects of Flooring and Hemi Body on Ground Reaction Forces and Coefficient of Friction in Stroke Gait", International Journal of Neurorehabilitation, vol. 1, pp. 1-6, 2014.

[19] L. Bensoussan, S. Mesure, J.M. Viton, and A. Delarque, "Kinematic and kinetic asymmetries in hemiplegic patients' gait initiation patterns", J Rehabil Med, vol. 38, p. 287–94, 2006.

[20] S.J. Olney, M.P. Griffin, T.N. Monga, and I.D. McBride, "Work and power in gait of stroke patients", Arch Phys Med Rehabil, vol. 72, pp. 309–314, 1991.

[21] A. Lamontagne, F. Malouin, C.L. Richards, and F. Dumas, "Mechanisms of disturbed motor control in ankle weakness during gait after stroke", Gait Posture, vol. 15, pp. 244-255, 2002.

[22] C. Chen, C. Patten, and D.H. Kothari, "Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds", Gait Posture, vol. 22, pp. 51-6, 2005.

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International Journal of Engineering and Innovative Technology (IJEIT)
Volume 4, Issue 9, March 2015

[23] D.C. Kerrigan, M.E. Karvosky, and P.O. Riley, "Spastic paretic stiff-legged gait: joint kinetics", Am J Phys Med Rehabil, vol. 80, pp. 244–249, 2001.

[24] A.R. Lindquist, C.L. Prado, and R.M.L. Barros, "Gait training combining partial body-weight support, a treadmill, and functional electrical stimulation: effects on post-stroke gait", Phys Ther, vol. 87, pp. 1144-1154, 2007.

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