Journal of Mechanics Engineering and Automation9 (2019) 111-119
doi: 10.17265/2159-5275/2019.04.001

Biomechanical Analysis of the STS Movement and Impact of an Assistive Device on the Elderly and People with Displacement Difficulties

Rodrigo Gonçalves de Souza Maciel, Daniel Kioshi Kawasaki Cavalcanti, Carlos Eduardo Guedes Catunda and Fernando Ribeiro da Silva

Department of Mechanical Engineering, Federal Center of Technological Education Celso Suckow da Fonseca (CEFET/RJ), Rio de Janeiro, RJ20271-110, Brazil

Abstract: This study presents a biomechanical analysis of the performance of the STS (sit-to-stand) movement applied to elderly and people with difficulty of displacement comparing it with and without an assistive technology device. A biomechanical kinemetry method was used for the movement evaluation. A two-dimensional virtual human model was developed through segmented ergonomic video analysis and the data obtained were numerically simulated to measure the inertial forces and torques of the complete execution of the STS movement. The device allows a vertical elevation of 0 to 400 mm and an anterior slope of up to 25°. A prototype was used to compare the movement with and without the assistive device. As a result, the torques in the lower limbs' joints and the vertical ground reaction forces were reduced by up to 60% and 23%, respectively. There was a reduction of up to 37° in the maximum trunk flexion angle during the cycle. The horizontal displacement of the center of mass was reduced by up to 70%.

Key words: STS, biomechanics, assistive technologies, kinemetry method, ergonomics, video analysis, human model, posture.

1. Introduction

With the accelerated development of new technologies aiming at the quality of life over the last few decades, an increase in the average life expectancy around the globe is observed [1] contributing to a considerable increase in the number of elderly living in society [2, 3].

A necessity, at first very simple, stands out in the daily lives of elders: sit and stand, technically known as the STS cycle (sit-to-stand and stand-to-sit). The functional capacity to carry out these movements, especially standing, is one of the most important in the life of an individual, since it is precursor to the realization of fundamental daily activities such as walking [4]. In addition, this capacity is directly linked to the condition of autonomy and independence of a person also affecting formal and/or informal caregivers, which often suffer injuries during transfer task practices [3].

The STS cycle can be hampered by diseases, injuries, advanced age and muscular weakening, mainly at lower limbs. Elderly and people with displacement difficulty become increasingly dependent on caregivers to perform this cycle undermining their autonomy and quality of life [5, 6].

Observing the importance of this cycle in an individual's life, the present study aims to assess the impact of an assistive device, which works coupled with a chair allowing vertical elevation and antero-posterior tilt, when executing the STS movement (sit-to-stand). Thus, it is intended to compare, through experiments and numerical simulations, the torques in the joints of the ankle, knee and hip, vertical forces and VGRF (vertical ground reaction forces), the displacement of the individual's center of mass (CoM) and the trunk flexion angle. At the same time the data obtained will be compared with other studies related to the movement.

Corresponding author: Rodrigo Gonçalves de Souza Maciel, BSc in mechanical engineering, research fields: biomechanics, assistive technology and STS cycle.

2. STS Movement

The actionable assistive device used in this study encompasses in its conception the three determining parameters in the STS movement [7]: (1) chair height, (2) use of armrests and (3) foot position. Fig. 1 illustrates the device.

The chair height adjustment is the most critical factor since for a very low seat standing without aid might be impossible. Whilst higher chair heights require less effort from the individual [4, 5, 7, 9]. This height can be determined as a percentage of the lower leg length (the distance between the feet and the knee). A chair height of about 120% of the lower leg length is considered the minimum for the elderly to successfully realize the cycle [7]. This height is directly related to the angular velocity of the hip, mainly to the angular displacement of the trunk, knees and ankles, to the maximum momentum in the hips and knees, and to the force required by the lower body [10].

A chair height adjustment while sitting allows that the thigh and leg are at approximately 90°, equaling the popliteal height of the individual to the chair height. That way, the same chair coupled to the proposed device, allows a comfortable and ergonomic position for users with different anthropometric characteristics [11].

An increment in the chair height alone would entail the loss of contact between the individual's feet and the ground. Thus, it is necessary to add an anti-posterior tilt combined with the vertical elevation. In addition, this slope elevates the hip height further, reducing the total displacement of the CoM and, consequently, the efforts that should be made by the user [8].

The use of armrests assists the movement distributing the efforts that were concentrated on the lower limbs only, serving also as a support for the individual, reducing the required momentum by the hip and knees at about 50% [7, 9]. Similarly, the vertical ground reaction force is reduced by approximately 14% [5].

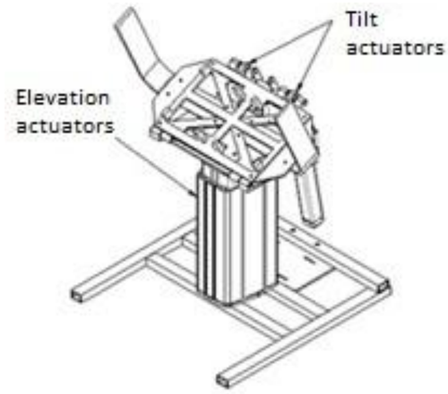


Fig. 1 Assistive device model in isometric view [8].

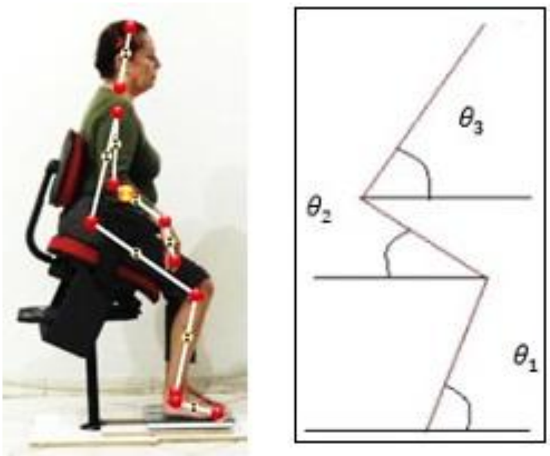
It is natural that the individual makes use of a "stabilization strategy" to stand from a chair [7]. This strategy consists of a posterior positioning of the feet (closer to the front edge of the seat) resulting in lower hip flexion, shorter length of the movement duration and greater antero-posterior stabilization in the instant in which the hip loses contact with the seat, moreover it approaches the individual's CoM to the base of support established by the feet [11, 12].

3. Materials and Experimental Procedures

3.1 Ergonomics Video Analysis—Kinemetry Method

To compare the movement of standing from a chair with the assistance supplied by the proposed device and without any assistance the kinemetry method was used. Images obtained in video were inserted into ergonomics video analysis software (SABIO) [13], developed and used by the Ergonomics Laboratory and by the Industrial Design Division of the INT (National Institute of Technology—Brazil/RJ). It allows the interaction of image and video files with a two-dimensional virtual human model. This model shows, through a coordinates system, the position of the individual's CoM, the limb's centroids, the joints (shoulders, knees, wrists, etc.) and the angles between the segments. Fig. 2 shows the overlap of this model with one video scene.

The data obtained were compiled into the MATLAB® software, where a computational model was created, which simulates the movements of an individual



(a) (b)

Fig. 2 Overlap of (a)the SABIO virtual model in a video scene and (b)angles used.

performing the STS movement, analyzed in the sagittal plane. The kynemetry was also used by other authors [14, 15].

3.2 Open Mechanism Virtual Human Body Model

The proposed model consists of a 3 bars open mechanism, and its measurements and masses are shown in Table 1. The percentages of the limbs' masses in relation to the whole-body mass, as well as the position of the segment's center of gravity, were obtained through Ref. [16]. The segments are represented by vectors, where the position of the ankle is determined as x = 0 and y = 0, and the movement is made through the increments of three angles: θ_1 (between the sole of foot and the leg), θ_2 (between an axis parallel to x line and the posterior part of the thigh) and θ_3 (between a parallel axis to x and the anterior part of the trunk), as shown in Fig. 2.

With the code to generate the human model, the movement was divided into three steps, according to the methodology of Schenkman [17]: (1) Initiation to the anterior rotation of the trunk, by reducing the θ_3 angle, with no movement in the legs and thighs (the other two bars). This step of Schenkman represents the trunk flexion; (2) The θ_3 angle continues to decrease until it reaches a minimum value, at the same time the θ_1 angle decreases up to a minimum value and the θ_2 angle starts to increase. At this stage the buttocks are lifted from the seat of the chair (lift-off). Schenkman's step 2 represents the momentum transfer; (3) The three angles grow to reach 90°, where the movement is completed, and the individual is standing (it was considered that all centroids align at the end of the movement). Schenkman's step 3 is the extension phase. To calculate the torque in the joints of the ankle, knee and hip, the Method of Sections was used and static balance during the movement was considered. Therefore, the torque in the hips is due from the weight of the bar 3, the torque in the knee results from the weights of the bars 2 and 3 and finally the torque in the ankle results from the weights of the three bars.

The weight of each limb is applied in the centroid of itself, previously located. P_1, P_2 and P_3 are the weights of the bars 1, 2 and 3, respectively. The length of the lever arm is the distance between the x coordinate of the articulation and the x coordinate of the centroid(s) of the limb(s) located above the joint. It was considered that during the whole duration of step 1, there is only torque applied in the hip joint, since the thighs (bar 2), the legs and the feet (bar 1) are supported on the seat and on the ground, thus the knee and ankle torques are null. From the beginning of step 2 (when the buttocks lose contact with the seat) it is considered that instantaneously all the body weight is on the feet, consequently, there are torques in the three joints. As a result, at any moment in step 2,

Table 1 Data from the virtual model developed in MATLAB® .

Segments	Length	Equivalent mass
Bar 1	Ankle to knee	Mass of both legs
Bar 2	Knee to hip	Mass of both thighs
Bar 3*	Hip to the top of the head	Mass of the trunk, arms, forearms, hands and head.

* The position of the centroid of segment 3 was determined as 60% of the distance between the hip and the top of the head (starting from the hip).

the torques are obtained according to the Eqs. (1)-(3), where M_q, M_j and M_t are respectively the torques in the hip, in the knee and in the ankle. The diagram of the forces and levers is shown in Fig. .

$$M_q = P_3 * d_3 \quad (1)$$

$$M_j = (P_2 * d_{21}) + (P_3 * d_{22}) \quad (2)$$

$$M_t = (P_1 * d_{11}) + (P_2 * d_{12}) + (P_3 * d_{13}) \quad (3)$$

An experiment was carried out with three individuals performing the STS movement using a prototype that simulates the proposed conditions of chair height and antero-posterior tilt of the seat. Each individual carried out the STS movement in three ways: (1) standing with the arms crossed around the trunk, from a position without any slope, with chair height corresponding to the popliteal height (UAC—unassisted arms-crossed); (2) standing with the aid of the assistive device (chair height and tilt) with arms crossed (AAC—assisted arms-crossed); (3) standing with the aid of the assistive device (chair height and tilt) using the armrests (AAR—assisted with hands on armrests).

The tilt angle used was 25° for all individuals and the chair heights were individualized. An "ideal height" was calculated for each individual with 25° anterior tilt of the seat. This height is the sum of the popliteal height of the individual with the vertical displacement of the knee due to the tilt of the seat, based on the length of the buttock-knee and the seat depth (41.5 cm). The individuals' anthropometric measures are found in Table 2.

A voice command warned the individual to stand at the desired speed. The initial position was with the trunk resting in the backrest of the chair (comfortable position). For the three movements (UAC, AAC and AAR) were compared: (a) variations in the CoM displacement, (b) variations in the maximum trunk flexion (minimum angle between the thighs and the trunk) and (c) the torques in the joints of the knees, ankles and hips.

In addition to the factors previously cited, the inertial forces inherent to the movement represent another cause for the difficulty of performing the STS movement [5]. To compare the behavior of the VGRF during the STS movement with and without the assistive device's aid, an experiment was done with a force platform (Dual-Top AccuSway, AMTI®, MA, USA), provided by the Physical Education Institute of

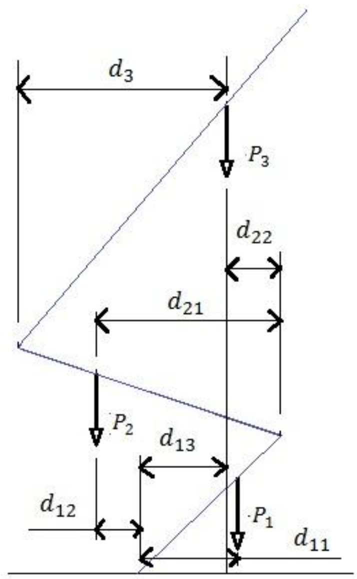


Fig. 3 Diagram of the forces and levers (parameters for calculations of torques).

Table 2 Individuals' anthropometric measures.

	Individual 1	Individual 2	Individual 3
Height (m)	1.64	1.72	1.56
Age	28	27	78
Mass (kg)	64.7	65.3	61.2
Popliteal height (mm)	410	440	380
Ankle-knee length (mm)	390	430	390
Knee-hip length (mm)	370	390	350
Hip-head length (mm)	810	810	720
Buttocks-knee length (mm)	560	570	550
Ideal height (mm)	470	500	437

UFF (Fluminense Federal University—Brazil). The tests were performed with individuals 2 and 3 (Table 2). Each individual performed the STS movement in three ways under the same conditions cited in the previous experiment (UAC, AAC and AAR). The software BALANCE CLINIC® was used to collect the data from the force platform. No standardized speeds were pre-established; the individual was free to stand in the desired speed. Graphics of the VGRF over time were generated and compared with the resulting curves obtained in other studies [5].

4. Results and Discussion

4.1 Validation and Evaluation of the STS Movement in the Virtual Human Model

After the image capture of the movement and obtaining the angles θ_1, θ_2 and θ_3 with the SABIO software, a simulation of the movement was made in MATLAB® through increments of the cited angles. Fig. shows the comparison of the videos' frames, with the overlay of the SABIO model in the images, with the two-dimensional virtual human model developed in the algorithm for the following positions: start of step 1, start of step 2, start of step 3 and end of the movement. Both the similarity between the video's image and the simulation's image and the correspondence of the steps' positions obtained to those obtained by Schenkman [17] reinforce the validity of the method.

The graphics with the torques of an individual (mass = 61 kg and height = 1.56 m), obtained from the method used, are shown in Fig. . The maximum values obtained for the ankle, knee and hip were 73.9 N·m, 97.1 N·m and 108.2 N·m, respectively.

After the start of the movement from a position where the angle between the thighs and trunk was greater than 90°, the inertial forces of the trunk generate a torque and, in response, the joint reacts with a torque in the opposite direction. At a specific time of step 1 the trunk [11] is perpendicular to a line parallel to the ground, in this point the torque in the hip is null. The torque then begins to increase to a peak that corresponds to the point of maximum flexion of the trunk (greater distance between the centroid of the trunk and the respective joint) when the extension phase begins, where the trunk's centroid starts to approach the hip joint again, reducing the torque until the end of the movement.

The torque in the knees is always negative due to the inertial forces of the thighs always generating a torque counter-clockwise, and the same occurs with the trunk in most of the movement. Even though at some point the trunk's centroid generates a clockwise torque (centroid of the trunk surpassing the knees position), the lever arm would be very small.

The torque on the ankle begins with a negative value at the beginning of step 2, reaches a null value during this same step, reaches a positive peak in the beginning of step 3 (extension phase) and begins to decay until it reaches zero.

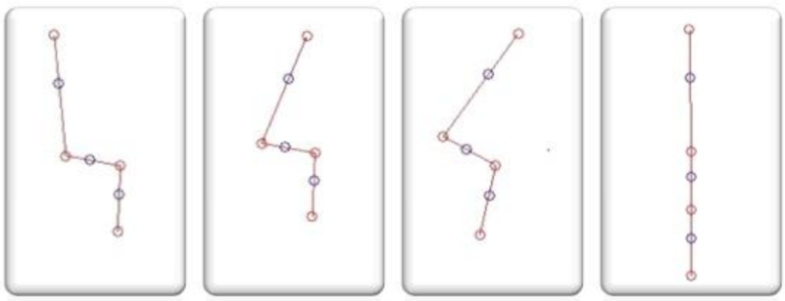
All torques end with a null value because it was considered that at the end of the movement the centroids are perfectly aligned. Table 3 shows the results obtained for each individual in the STS movement in the three researched conditions.

The use of "ideal heights" has shown to be satisfactory. As cited earlier, the slope allowed a grater chair height increment, while the whole feet kept in contact with the ground. Figs.5 and 6 show the comparison of the torques and the whole-body CoM displacement, respectively, in the three conditions for individual 3.

4.2 Optimized Sit-to-Stand Movement

Based on the experimental results, it was observed that the maximum torques in the AAC and AAR conditions in the hip and ankle are considerably reduced when compared to regular movement, 57% to 24% reduction on the hip and 66% to 48% on the ankle. It was verified that the maximum torque in the

(a)



(b)

Fig. 4 Comparison between the steps in the simulation of the STS movement: (a) image capture of the movement with the SABIO software in a Dual-Top AccuSway force platform, and (b) two-dimensional virtual human model provided by segmented ergonomic video analysis in the MATLAB® algorithm; in the simulated images the red circles indicate the beginning and end of the bars and the blue circles represent the respective centroids.

Table 3 Reaction forces, torques and displacements obtained for each individual in the algorithm of STS movement.

	Individual 1			Individual 2			Individual 3		
	UAC	AAC	AAR	UAC	AAC	AAR	UAC	AAC	AAR
M_t (N·m)	122.6	45.2	44.4	57.4	30.1	35.2	73.9	34.7	25.7
M_j (N·m)	148.9	89.5	87.5	102.2	101.5	111.7	88.0	97.1	91.2
M_q (N·m)	103.1	78.4	78.7	125.9	81.1	74.9	108.2	47.0	50.5
Max. trunk flexion (°)	81.7	100.1	104.9	72.0	104.3	101.4	76.9	114.3	111.5
CoM-H. Displ. (mm)*	319	167	155	379	129	130	317	114	91
VGRF (%)**				123.0	100.7	100.2	115.9	112.9	108.2

* Horizontal displacement of the CoM.

**Vertical ground reaction force—the percentages refer to the whole-body weight.

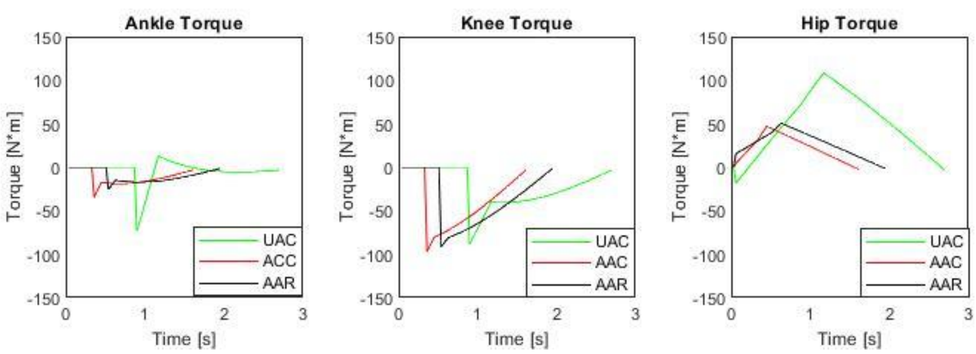


Fig. 5 Comparison between the torques in the joints of the individual 3 in three conditions (UAC, AAC and AAR).

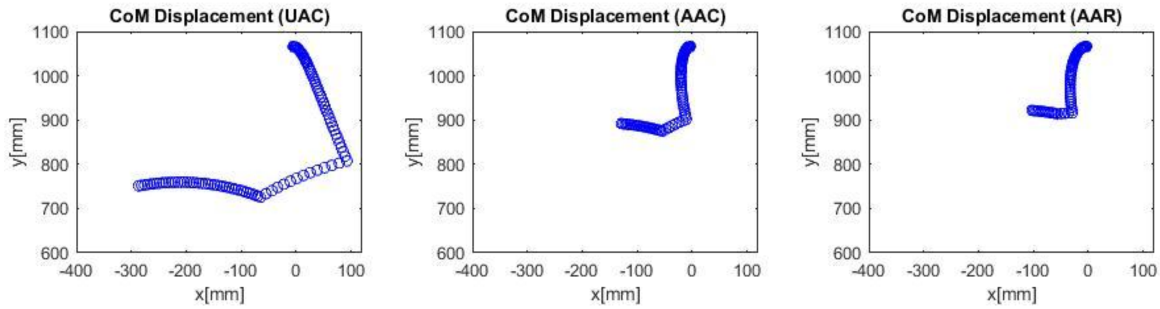


Fig. 6 Whole-body CoM displacement of individual 3 in the three conditions UAC, AAC and AAR.

knee was slightly minor in regular movement conditions (UAC). By the used method, the higher the trunk flexion, i.e., the more the individual tilts his trunk forward, the more the centroid of the trunk approaches the knee joint. As a result of the shortening of the lever arm there is a lower torque in the knee. As in the AAC and AAR conditions occurs less trunk flexion, in the lift-off instant (when the loss of contact with the seat occurs), the centroid of the trunk is a little further from the knee, leading to a larger lever arm and consequently a greater torque.

As to the maximum trunk flexion, a big difference is noticed when comparing condition UAC with the conditions AAC and AAR. A smaller angle of maximum flexion corresponds to a larger anterior tilt of the trunk and a larger torque in the hip. In the regular condition of movement (UAC), these angles reach up to 72° indicating a very sharp slope. The conditions with the use of the device produced a decrease of this angle in a range of 22.12% to 32.72%. This reduction implies a lower risk of falls and a lower initial impulse by the individual for the execution of the movement, showing the benefits of the proposed assistive device.

The comparison of horizontal CoM displacement also shows the reduction of the horizontal displacement during the STS movement. When compared to condition UAC, this displacement in the AAC and AAR conditions had a reduction ranging from 51% to 71%. Thus, the significant reduction of the horizontal movement of the CoM results from a lower impulse at the beginning of the movement, which implies in a lower risk of accident for the user.

Experiments with the force platform proved to be satisfactory, due to the fact that, as well as demonstrating a reduction in peak VGRF, they presented experimental data with the same behavior seen in Ref. [5]. In the UAC condition for instance, a reduction of the VGRF in the initial phase of the movement was observed, then increasing gradually until it reaches its peak (with a value greater than the individual's body weight) followed by a recoil (called "rebound") and stabilizing with a value equal to the body weight, as shown in Fig. 6. For individual 2 (weight: 640.59 N), the maximum value for VGRF was 793.52 N, corresponding to 123% of the body weight. In the AAC and AAR conditions this maximum value decreased to 100.66% (644.84 N) and 100.18% (641.80 N), respectively. For individual 3 (weight: 600.37 N) the values under conditions UAC, AAC and AAR were respectively 115.90%, 112.87% and 108.21%. Comparing the results of the two individuals, a drop in the VGRF's peak value was noticed, especially in condition AAR (initial position with incremented height, anterior tilt of the chair and use of armrests). These results indicate that the proposed conditions reduce the movement's extension, consequently reducing the inertial forces inherent in it, thus facilitating its execution.

Fig.7 shows a comparison of the VGRF (as percentages of the individual's body weight) during the STS movement between the results obtained in this study for the conditions UAC, AAC and AAR and Ref. [5]. The similarity between the curves' behaviors obtained in the proposed method and the one used by Refs. [5, 15] can be observed, such as the reduction of the VGRF's peak under conditions AAC and AAR.

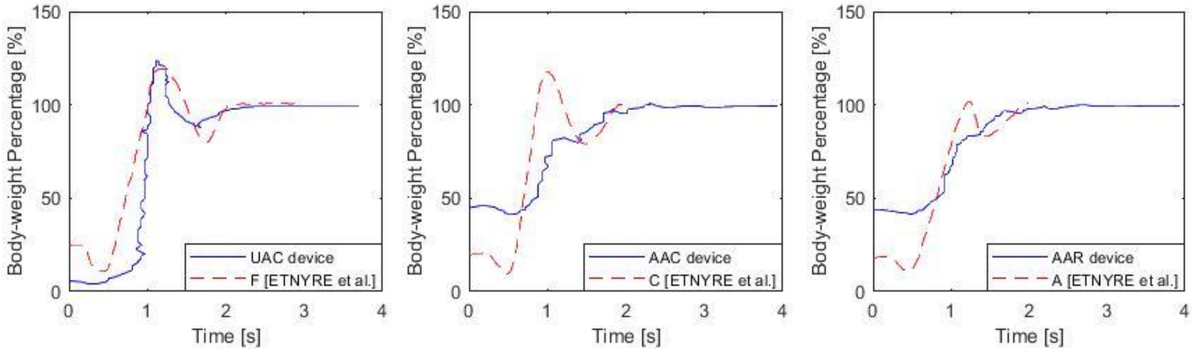


Fig. 7 Comparison of VGRF as percentages of the individual's body weight during the STS movement for the conditions UAC, AAC and AAR with the ones obtained by Etnyre et al. (without aid of any devices) [5]. F = arms-free condition, C = arms-crossed condition and A = hands-on-armrests condition.

5. Conclusions

The comparison between the behavior of the curves and peaks of the torques in the joints found in Refs. [5, 15], during the STS movement in order to assess the impact of an assistive device, demonstrates the validity of the method used in this study. Algorithms were developed in MATLAB® for analysis and simulation of the STS movement. These algorithms have ample adherence to the current studies on the biomechanics of the movement and can be used for future researches.

The experiments made with the prototype revealed a positive outcome regarding the ergonomic position and the execution of the STS movement. As for ergonomics, the assistive device allows a chair height adjustment, making a common chair ideal for users with different anthropometric measures. In terms of the execution of the STS movement, significant reductions in the torques required by both the hip and ankle's joints were observed, respectively, 57% and 66%. The peak of the VGRF (just after the loss of contact with the seat) reached 100.66% of the individual's body-weight, whilst typically this peak ranges from 111.0% to 125.5% of the weight [5]. Furthermore, the use of the assistive device discards the need of impulse by the individual at the beginning of the movement, specifically in the horizontal direction. Consequently, the maximum trunk flexion angle was reduced by up to 32.72% and the horizontal displacement of the individual's CoM was reduced by up to 71%. Regarding the "seat tilt", Rasmussen et al. [18] suggest the possibility of an anterior slope of 0 to 30° in the experiments conducted during this study, it was determined from experimental observations a 25° ± 1 angle since this angulation provided considerable help in the execution of the STS at the same time that allowed a comfortable posture to the user.

It is expected that the present work serves as a basis and incentive to future studies regarding the development of assistive technologies to aid the STS movement. In this concern, this study suggests an assessment on the impact of the anterior tilt of the chair and possible sliding between the seat and the individual. Moreover, it is suggested an evaluation of the ideal velocities for the device to move in a speed not too high, which could cause discomfort or even accidents, or too low, making the movement too time-consuming.

Acknowledgments

The support from CEFET-RJ, UFF and INT is greatly acknowledged.

References

[1] Pereira, K. C. R., Alves, A. M., and Traebert, J. L. 2011. "Contribuição das condições sociodemográficas para a percepção da qualidade de vida em idosos." *Rev. Bras. Geriatr. Gerontol.* 14 (1): 85-95.

[2] Alexandre, T., Cordeiro, R., and Ramos, L. R. 2008. "Fatores associados à qualidade de vida em idosos

com osteoartrite de joelho." *Fisioter Search.*15 (4): 326-32.

[3] Farias, C. E. S. 2007. "Avaliação Funcional do Idoso." MSc thesis, Veiga de Almeida University.

[4] Faria, C. D. M., Saliba, V. A., and Teixeira-Salmela, L. F. 2010. "Musculoskeletal Biomechanics in Sit-to-Stand and Stand-to-Sit Activities with Stroke Subjects: A Systematic Review." *Fisioter. Mov.* 23 (1): 35-52.

[5] Etnyre, B., and Thomas, D. Q. 2007. "Event Stadardization of Sit-to-Stand Movements." *Phys Ther.* 87 (12): 1651-66.

[6] Santos, R. G. T., Meneguci, J., Bastos, L. L. A., Damião, R., and Júnior, J. S. V. 2013. "Força de membros inferiores como indicador de incapacidade funcional em idosos." *Driving. Rev. Ed. Fis.*19 (3): 535-42.

[7] Janssen, W. G. M., Bussmann, J. B., and Stam, H. J. 2002. "Determinants of the Sit-to-Stand Movement: A Review." Phys Ther. 82 (9): 866-79.

[8] Maciel, R. G. S., and Cavalcanti, D. K. K. 2016. *Análise da Biomecânica do Movimento STS e Projeto de Dispositivo Assistivo para Idosos e Pessoas com Dificuldade de Deslocamento*. TCC—CEFET/RJ, Brazil.

[9] Prata, M. G., and Scheicher, M. E. 2012. "Correlation between Balance and the Level of Functional Independence among Elder People." *Sao Paulo Med J.* 130 (2): 97-101.

[10] Moraes, R., and Mauerberg-DeCastro, E. 2010. "Relação entre Percepção e Ação durante os Movimentos de Sentar e Levantar em Indivíduos Idosos." *Psicologia: Terapia e Pesquisa*26 (2): 253-64.

[11] Alexander, N. B., Koester, D., and Grunawalt, J. 1996. "Chair Design Affects How Older Adults Rise from a Chair." *J Am Geriatr Soc.* 44 (4): 356-62.

[12] Goulart, F., Chaves, C. M., Vallone, M. L. D. C., Carvalho, J. A., and Saiki, K. R. 2003."O movimento de passar de sentado para de pé em idosos: implicações para o treinamento funcional." *ACTA FISIÁTR.* 10 (3): 138-43.

[13] Guimarães, C. P. 2012. *Guia de Uso do SABIOv 1.1*. Industrial Design Division of the National Institute of Technology, Brazil.

[14] Salah, O., Asker, A., El-Bab, A. M. R. F., Assal, S. M. F., Ramadan, A. A., and Sessa, S. 2013. "Development of Parallel Manipulator Sitto Stand Assisi Device for Elderly People." In *Proceedings of IEEE Workshop on Advanced Robotics and Its Social Impacts (ARSO)*, Vol. 7, 27-32.

[15] Cao, E., Inoue, Y., Liu, T., and Shibata, K. 2011. "A Sit-to-Stand Trainer System in Lower Limb Rehabilitation." Presented at IEEE/ASME International Conference on Advanced Intelligent Mechatronics (AIM 2011), Budapest, Hungary.

[16] Tözeren, A. 2000. *Human Body Dynamics: Classical Mechanics and Human Movement*. New York: Springer-Verlag. 300-2.

[17] Schenkman, M., Berger, R., O'Riley, P., Mann, R. W., and Hodge, W. A. 1990. "Whole-Body Movements during Rising to Standing from Sitting." *Phys Ther.* 70 (10): 638-48.

[18] Rasmussen, J., Torholm, S., and De Zee, M. 2009. "Computational Analysis of the Influence of Seat Pan Inclination and Friction on Muscle Activity and Spinal Joint Forces." *International Journal of Industrial Ergonomics* 39 (1): 52-7.