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Analysis of Behavior of Forces in a Pelvis-Soft Tissue Mechanical Model

Análisis del Comportamiento de Fuerzas en un Modelo Mecánico de Pelvis-Tejido Blando

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SUMMARY: Pressure ulcers are tissue damage resulting from the constant pressure on the underlying soft tissue to bony prominences for long periods. Some of the most common ulcers are developed at the ischial tuberosities area (ITs). It has been found that stresses produced in the underlying tissue to the ITs may exceed 5 to 11 times the surface stresses, making it necessary to estimate the forces generated between the soft tissue and the ITs. However, it is not possible to determine these stresses *in vivo* in a patient, due to ethical reasons. This paper presents a mechanical model of the pelvis-soft tissue in order to study the behavior of contact forces. The model simulates the load on the ITs of a male subject of 70 kg weight and 1.70 m height, which were recorded for 8 min. The registered forces in the model were compared with the surface forces estimated from pressure records measured by the Force Sensing Array system in a patient with spinal cord injury. After 2 min, both forces measured in the model, and the ones estimated in the patient followed the trend described by Crawford during clinical measurements of pressures during sitting. It was also found in the model that measured forces below the ITs are higher than those measured below soft tissue, which suggests that the model may be valid for the study of the forces generated inside the tissue.

KEY WORDS: Pressure ulcer; Pressure sensor; Mechanical model.

INTRODUCTION

Pressure ulcers are lesions on the skin and/or underlying soft tissue to bony prominences, caused by the constant pressure, or combined pressure of shear forces exerted by bony prominences for an extended period of time, causing capillary obstruction and, thus, damaging the surrounding tissue (Lyder, 2003; National Pressure Ulcer Advisory Panel and European Pressure Ulcer Advisory Panel, 2009). The formation of pressure ulcers is influenced by intrinsic factors such as paralysis, because the patient will suffer loss of muscle mass and muscle tone, causing a reduction in muscle tissue surrounding the bone prominences, mainly in the pelvis. Besides, loss of muscle tone impairs motor response to external stimuli (Bogie & Bader, 2005), and additionally, the motor paralysis affects the ability of the nervous system to respond to noxious stimuli and the ability to change position consciously. Many other intrinsic factors are recognized: nutrition, genetic predisposition, phycological status, chronic diseases, infections and physiological and biochemical response of the individual to ischemia-reperfusion, among other factors (Bogie & Bader; Bouten *et al*., 2005; Henzel *et al*., 2011). Together with the aforementioned, there are extrinsic factors, like sustained pressure over time, in combination with shear forces, which causes occlusion in blood vessels, and soft tissue damage (National Pressure Ulcer Advisory Panel and European Pressure Ulcer Panel, 2009); as well, friction and microclimate that are generated between the individual and the support surface must be considered (Ferguson-Pell, 1990; Wounds International, 2010).

Patients more prone to develop pressure ulcers, derived from their neurological condition, are the elderly

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and people with spinal injuries. In the case of spinal injuries, some studies have found that between 31 % and 79 % develop pressure sores throughout their lives (Henzel *et al*.). According to Bogie *et al*. (1995), 47 % of pressure ulcers are located in the area of the tuberosities and sacrum in patients with a spinal cord injury, and these injuries may be associated with conditions related to the way in which the patient is sitting.

Recently, the National Pressure Ulcer Advisory Panel (NPUAP) (National Pressure Ulcer Advisory Panel and European Pressure Ulcer Advisory Panel) has recognized a relative damage in the subcutaneous tissue found beneath the intact skin, known as suspected deep tissue injury (DTI), which is difficult to detect, and represents a clinical challenge. There are other factors to consider, such as the recommendation of support surfaces to help reduce internal stresses within the tissue. In clinical conditions for the evaluation of the support surfaces (seats for wheelchairs and mattresses), the surface pressures generated between the patient and supporting surface are measured, Crawford (2004), suggesting that clinical measurements of seats for wheelchairs, should last at least eight minutes; however, this measurements do not quantify the intensity of forces generated within the tissue, since according to the data found in a previous experimental model that simulates the loading of an ischial tuberosity on the soft tissue, proposed by Gefen & Levine (2007), the stresses just below the ischial tuberosity may exceed between 5 and 11 times those recorded at closer distances to the surface. In an animal model, Le *et al*. (1984) found that higher pressures (270 mmHg) were found directly below the bone prominences, at a depth of 12.5 mm with respect to skin, while they measured a contact pressure of 47 mmHg in the surface of the skin. That is the reason why it is necessary to determine the stresses that are generated directly below the ischial tuberosity; however, if we planned to measure these stresses *in vivo* in a patient, it would imply direct instrumentation inside a patient tissue, which is not permitted by ethical implications. As it was mentioned above, a mechanical model that will simulate the force produced by the ischial tuberosity on the soft tissue was proposed, for which a pelvis-tissue model was used, coupled to a Universal Testing Machine, Instron Model 4502 (Instron Inc., Norwood MA, USA) which simulates the load of the human trunk.

MATERIAL AND METHOD

A physical model was developed to measure the forces generated between the pelvic bone and soft tissue, simulating the load generated by the trunk of the human body in the area of the ischial tuberosity. The load was held for eight minutes, according to a clinical protocol for measuring pressures in a wheelchair seat proposed by Crawford. The tendency of these forces obtained experimentally was compared with the estimated clinical measurements of pressure of a patient with spinal cord injury.

A. **Description of the Model**. A synthetic model of a human pelvis for biomechanical testing (Alemán-Perez, 2015) was used. The model of the pelvis corresponds to a 70 kg and 1.70 m height individual; under the ischial tuberosities of pelvic bone, two 20 mm latex plates were placed, simulating the soft tissue found naturally below bony prominences (Matsuo *et al*., 2011). The model was coupled to a universal testing machine, Instron 4502, locating the pelvis and latex plates on a flat rigid metal plate. The load was applied to L4 vertebra through a metal cylinder with a spherical end. In all experiments, the pelvis was placed in the same position; in order to ensure this, two marks were made on each ischial protuberance and placed at 90 mm from the edge of the metal plate (Fig. 1).

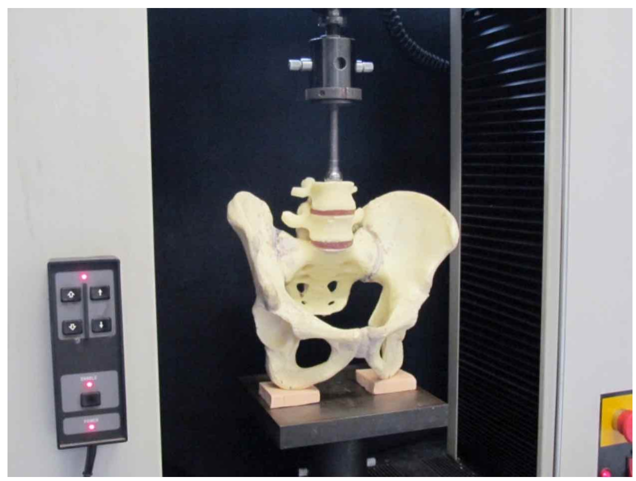

Fig. 1. Model of a pelvis coupled to an Instron 4502 Universal Testing Machine, with a 20 mm latex plate simulating the ischial tuberosities underlying tissue.

A load of 450 N was applied, corresponding to the load exerted by the trunk of a male subject weighing 70 kg and 1.70 m height. In order to measure the forces generated in the area of the ischial tuberosities, we employed Flexi-Force® (Tekscan Inc. Boston, MA, USA), a piezoresistive pressure sensor. These sensors have already been evaluated for this type of application, and we found it is the best choice over other piezoresistive sensors, as they have a wider range and low hysteresis, which implies that this sensor can be used for ascending or descending loads (Enriquez-Rivera *et al*., 2013). The model was placed on a polypropylene plate supported on the rigid metal plate (see Fig. 1). This configuration represents the worst case scenario for the patient, that is, the pelvis is in direct contact with a rigid material.

B. Description of the Acquisition System. Eight A201 Flexi-Force sensors were used for this test. The acquired signal was conditioned according to the specification sheet, and the connection between the circuit and the computer was performed using a data acquisition card (DAC) model NI-DAQmx (National Instruments, Austin, Tx, USA). The LabView 2011 interface was used (National Instruments, Austin, Tx, USA), which receives andstores the data of force measured by the sensors.

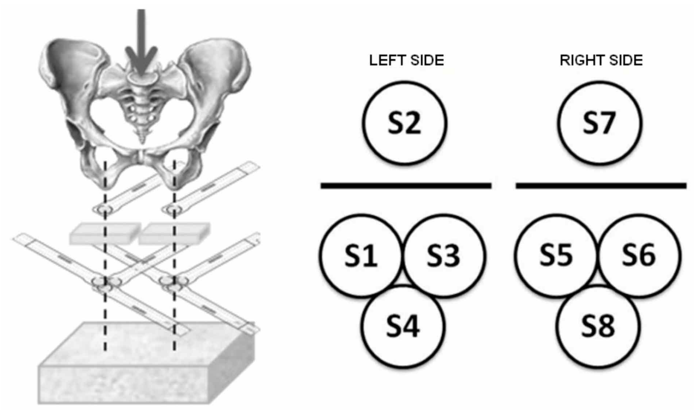

Fig. 2. Sensor array. Sensors 2 and 7 were placed under the tuberosities while the rest remained positioned below the test tissue.

C. Testing. For testing, an ascending load was applied up to 450N, simulating the weight supported by the trunk of a 70 kg person; once reached this load, it was maintained for eight minutes. According to Crawford, it is in the first eight minutes of measurement when the most significant changes are recorded in the pressure between the patient and the seat. Data collected by the sensors during the test were recorded. The sensors were placed in the configuration shown in Fig. 2.

A clinical trial of pressure measurement was performed following the protocol proposed by

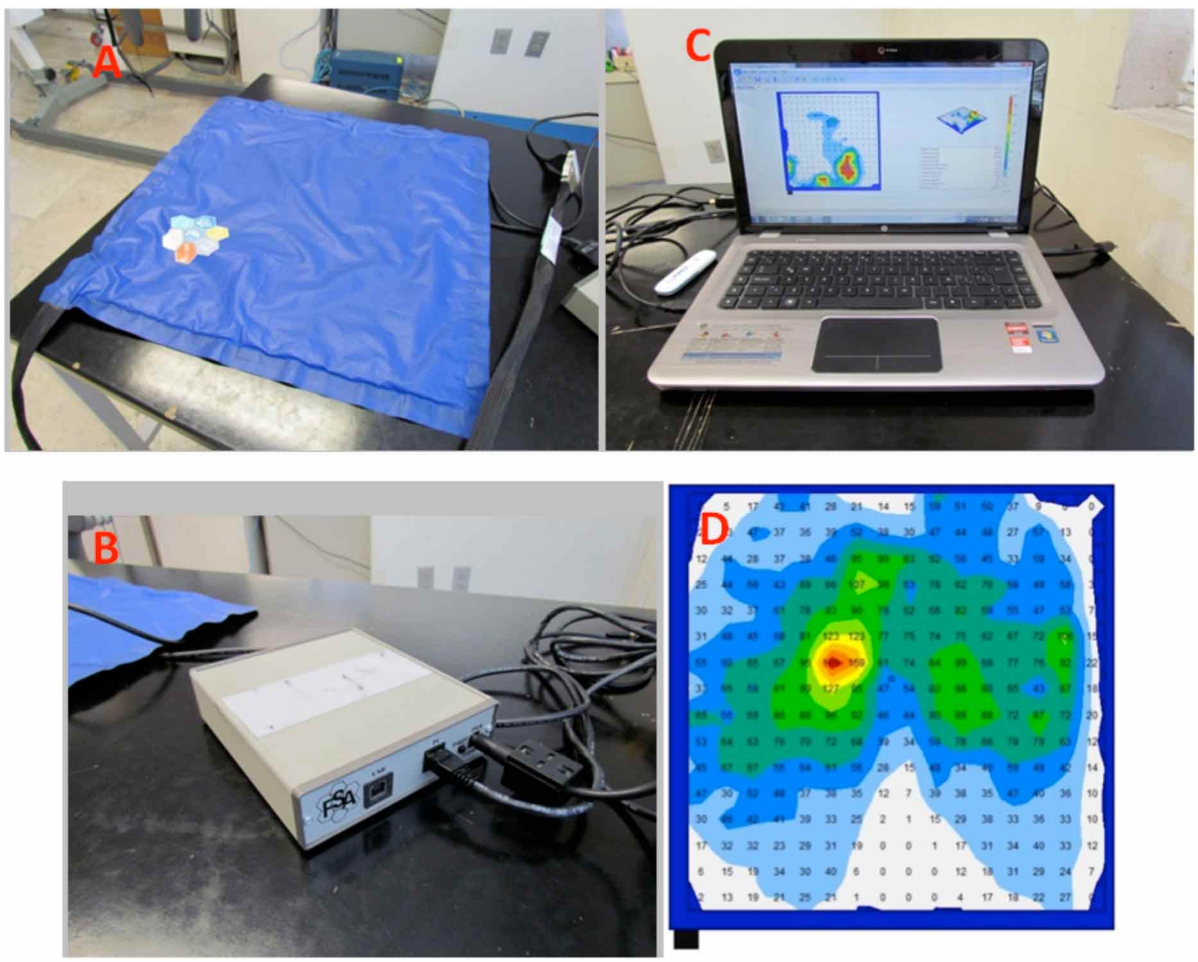

Fig. 3. Pressure Measurement System, Force Sensing Array (FSA®). A. FSA® Mat. B. Device that processes data. C. laptop, for pressure map deployment and data analysis. D. FSA® pressure map. Light colors show lower pressure areas, whereas dark colors show higher pressure areas.

Crawford in a patient with complete spinal cord lesion, located on the fourth thoracic vertebra (T4): a male, 65 kg and 1.68 m height, with a history of pressure ulcers in sacrum, who uses a wheelchair and a seat both made-to-measure; the seat was manufactured using polypropylene 6.35 mm (1/4 in.) thick and foam rubber padding. The informed consent form with the patient's handwritten signature was obtained to perform these measurements. In order to register contact pressures clinically, a Force Sensing Array (FSA), (Vista Medical Ltd., Winnipeg, MB, Canada) was employed. This system has an array of 16 by 16 piezoresistive pressure sensors. Each sensor has a useful sensing area of 7.22 cm^2, which are contained in a thin, flexible material, which form a mat sensor placed between the subject and his wheelchair. This equipment has an interface on a personal computer, and displays a map of topographical and numerical pressure profile. (Fig. 3).

D. Data Analysis. The data collected by sensors were stored by the interface Labview 2011 (National Instruments Corp., Austin, TX, USA) and exported to Excel 2010 (Microsoft Corp., Redmond, WA, USA) where graphs corresponding to the response of each sensor were made.

Figure 4 shows the change over time of the sensor data in the relaxation step subsequent to a force equal to 450 N. From the pressure maps obtained from FSA measured on the patient, the forces (force = Area x Pressure) which are generated in the area of the ischial tuberosities were estimated, so that the active area was measured corresponding to ischial tuberosities (see Fig. 3), located on maps of pressure, processed with Image J software (National Institutes of Health, Bethesda, MD, USA), and recorded as pressure. The average value of area corresponding to the area of the ischial tuberosity was above 80 mmHg in active sensing (Bennett *et al*., 1979).

RESULTS

The forces measured by the FSA between the pelvis model and latex plate are slightly higher than those measured between the metal and the latex plates, although both follow the typical behavior of a relaxation test (Fig. 4). This is because the latex plate partially absorbed the loads exerted by ischial tuberosities.

In Figure 5, the behavior of the contact forces generated between the ischial tuberosities and the simulated soft tissue is observed. Figure 6 shows the behavior of the forces generated below the soft tissue simulated, which are generally measured superficially by devices like FSA®; it is noted that the forces generated below the simulated soft tissue are smaller than those measured directly beneath the ischial tuberosities. In Figure 7, the estimated forces measured from pressures obtained with FSA® on the ischial tuberosities of the patient are shown. In the case of the right ischial tuberosity, it was found that the estimated force is approximately five times greater than those obtained in the model, while the force in the left ischial tuberosity estimated in the patient, compared with the model, showed no noticeable differences. Both force values obtained in the model, and those estimated for the patient follow the trends of pressure values described by Crawford, who found differences in the behavior of the values of pressure in a patient group with multiple sclerosis and who are wheelchair users, versus a group of patients with multiple sclerosis and who are not wheelchair users.

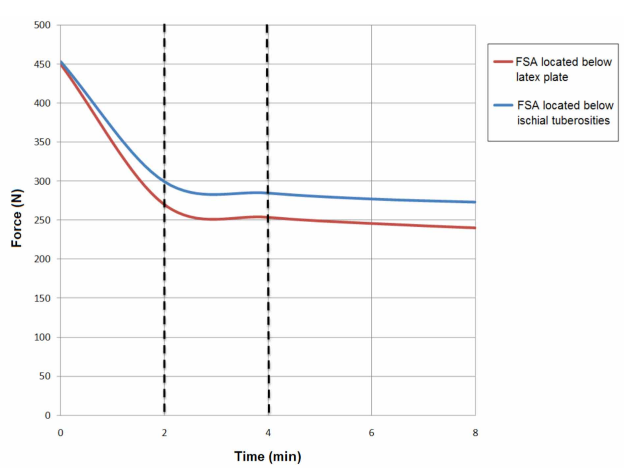

Fig. 4. Change in the forces applied to the experimental model measured by the FSA system below the latex plate, and between the latex plate and the ischial tuberosities.

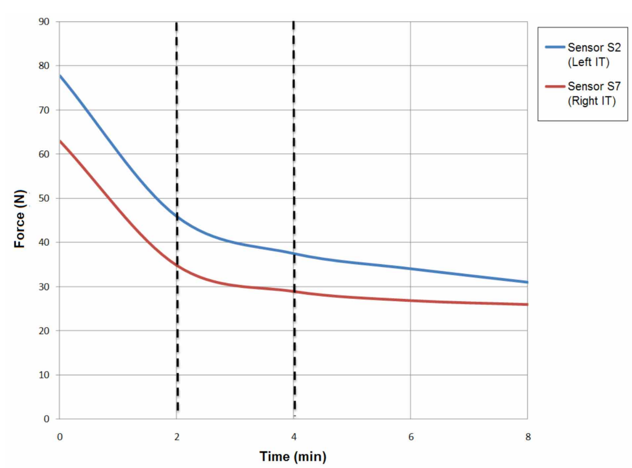

Fig. 5. Variation of forces in the S2 and S7 sensors located below the ischial tuberosity, measured over time, according to Figure 2 scheme.

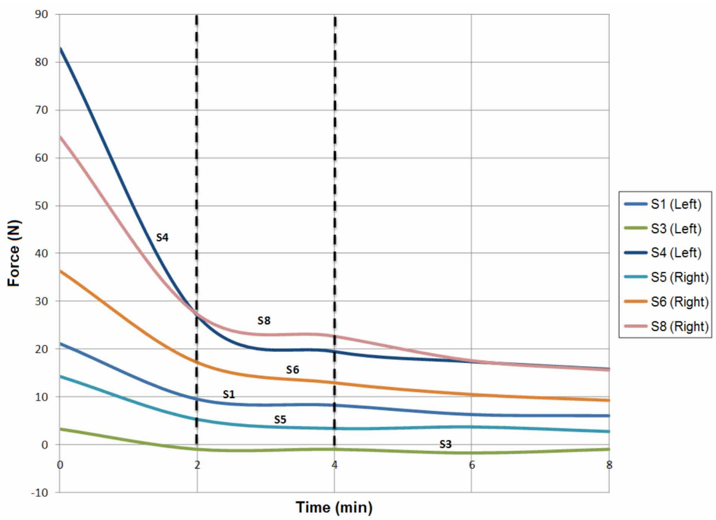

Fig. 6. Change in forces in the S1, S3, S4 and S5, S6, S8 sensors, located under the simulated tissue, with respect to time, according to the Figure 2 scheme.

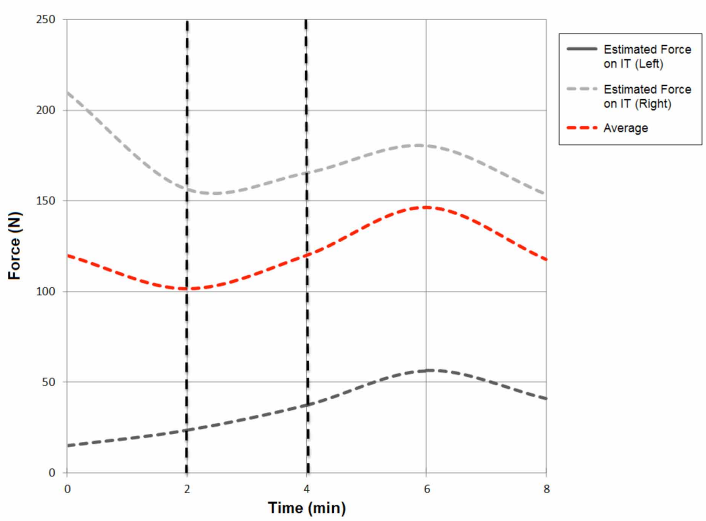

Fig. 7. Estimated forces in a patient using FSA® system, with respect to time.

DISCUSSION

It was found that the FSA® system is not suitable for measuring the contact forces, since it measures pressure and the area was estimated from the graphic display. In addition, the pressure sensors are saturated at 200 mmHg. However, it is clinically useful and widely used to evaluate bearing surfaces due to its characteristics, because placing sensors individually in a patient is technically complex, as in order to positioning the sensors, it is necessary to position the patient in decubitus prone, and it is required to locate the bony prominences (sacrum, ischial tuberosities and trochanter). Additionally, sensors must be attached to the skin, with some sort of adhesive tape (3M Micropore®, Transpore® 3M, among others), which can cause injury to the skin of a patient with a history of pressure ulcers; besides, in these cases, placing individual sensors is contraindicated.

A reduction is observed in forces estimated for the patient, based on data obtained from FSA®, after the first two minutes, which is consistent with those reported by Crawford. After two minutes of testing, in the case of estimated ischial tuberosity forces for the patient, the behavior is consistent with trends reported by Crawford in patients using wheelchairs. This behavior may be due to the fact that the ischiogluteal area is subject to gradually increasing contact with the seat, suggesting an effect of envelopment. This phenomenon involves that the body sinks into the support surface to be surrounded by it, (Crawford; Wounds International) as well as the creep phenomenon, which is the inherent tendency of the pressure to increase in time when the load remains constant. In the case of the model, the behavior observed was that the forces were stabilized, which may be due to the intrinsic characteristics of the material used to simulate soft tissue, in addition to the conditions in which the direct contact of soft tissue was simulated, without cushioning on a rigid surface, so envelopment and creep phenomena were not evidenced.

Contact forces registered for the patient increased after four minutes, which coincides with that reported by Crawford for patients using wheelchairs, while forces in the model follow the trend of contact force values reported for patients who are not wheelchair users.

After six minutes, the estimated forces in the patient descended, which can be caused by saturation of the FSA® sensors. In this period of time, the behavior of the forces in the model is similar to that found in patients who are wheelchair users during the clinical measurement (Crawford), which may suggest that the model may be valid to simulate the load exerted by the trunk in the area of the ischial tuberosities, despite the forces estimated in the patient, which are much greater than those measured in the model. This may be caused by the fact that, during the measurements, the patient wore his underwear and sports clothing, which could interfere with measurements, like the surface where the patient is supported; however, this point should be taken into account for measurements made in the model, since most clinical measurements are performed with clothes and on a wheelchair. Other materials used for mechanical simulation of tissues must be also considered.

Regarding the behavior of the sensors, it was found that the greatest force was concentrated in the S2 and S7 sensors (Fig. 5), which is consistent to what was observed by Gefen & Levine about the forces compared between the inner tissue and the surface. We calculate the average of the sensors that are located under the latex, S1, S3 and S4 (left side) and S5, S6 and S8 (right side), finding that the force recorded in S2 is 2.1 to 4.9 times higher than the average of S1, S3 and S4, while the force in S7 is 1.6 to 2.8 times higher than the average of S5, S6 and S8, see Fig. 6. In both the left and right sides, the ratio increases as the signal is stabilized and over time. In addition, these results are consistent with those forces reported in a Finite Element Method (FEM) model developed by Araujo-Monsalvo *et al*. (2016). In this model, greater stresses are appreciated at the points of contact between the ischial tuberosity and the soft tissue, finding the greatest stresses when the simulated injuries wereoriginated inside the soft tissue.

Furthermore, a mirror effect is noted among the sensors, which can be observed in Fig. 6 where it is appreciated that the values obtained from the sensors show that the forces in S1 and S5 are similar, as they are for S3 and S6 and also for S4 and S8, suggesting that the pelvic anatomy has an influence in the phenomenon, as the forces on the left side are larger, there is a symmetrical relationship in the distribution thereof. This may be because the model was obtained from a subject without motor disabilities. It could be expected that the symmetry of the forces was lost by traumatic events which cause malformations in the pelvis, despite in the patient the measurement on the right tuberosity is five times larger than the left. This same trend is observed after the first two minutes.

The analysis of the contact forces in this model provides results on the interaction of forces on the ischial tuberosity and underlying tissues, which will help design seats, not just for people with spinal cord injury, but for people who, by the type of disability they have, suffer deformities in pelvis, and also for those who need special designs not only to prevent pressure ulcers, but to prevent greater deformity in their pelvis, causing alterations in the trunk.

CONCLUSION

It was observed that, in the model, the contact forces over time follow the trends of the results of clinical tests conducted by Crawford in patients with multiple sclerosis, non-users of wheelchairs. It was also found that measured forces in the model below the ischial tuberosity are much higher than under the latex, which is consistent with the literature, and a finite element computer model, made in a previous work (Araujo-Monsalvo *et al*.), suggesting that this model is valid with some considerations to simulate the interactions between the ischial tuberosities and soft tissue.

It is believed that the mirror effect found between the sensors may be due to the symmetrical configuration of the pelvis, because the model was constructed from a healthy subject. It is expected that the model can be applied in the future for the design of support surfaces to help reduce harmful forces to the soft tissue in people with motor disabilities.

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RESUMEN: Las úlceras por presión son daños en el tejido, derivados de la presión constante por periodos prolongados sobre el tejido blando subyacente a una prominencia ósea, algunas de las úlceras más comunes se desarrollan en la zona de las tuberosidades isquiáticas (TI´s). Se ha detectado que esfuerzos generados en el tejido subyacente a las TI´s pueden exceder entre 5 a 11 veces a los esfuerzos superficiales, lo que hace necesario conocer las fuerzas que se generan entre el tejido blando y las TI´s, sin embargo medir estos esfuerzos *in vivo* en un sujeto, no es posible por razones éticas. Este trabajo presenta un modelo mecánico del sistema pelvis-tejido blando con la finalidad de estudiar el comportamiento de las fuerzas, el modelo simula la carga en las TI´s de un sujeto masculino de 70 kg y 1,70 m, en el cual se registraron por 8 min. Las fuerzas registradas en modelo fueron comparadas con las fuerzas superficiales estimadas a partir de los registros de presión medidas por el sistema Force Sensing Array, en un paciente con lesión medular. A partir de 2 min, tanto fuerzas medidas en el modelo, como estimadas en el paciente, siguen la tendencia descrita por Crawford para mediciones de presiones clínicas durante la sedestación, también se encontró en el modelo que las fuerzas medidas por debajo de las TI's son mayores a las medidas debajo del tejido blando; lo que sugiere que el modelo puede ser válido, para el estudio de las fuerzas que se generan al interior del tejido.

PALABRAS CLAVE: Úlcera por presión, Sensor de presión; Modelo mecánico.

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