Tema A1a Diseño Mecánico: Modelado por elemento finito

“Determination of the impact force attenuation provided by trochanteric soft tissue in sideways falls using a simplified finite element model”

Francisco Javier Cadena Delgado, Gilberto Mejía Rodríguez*, Hugo Iván Medellín Castillo, Dirk Frederik de Lange


*Autor de contacto: gilberto.mejia@uaslp.mx

RESUMEN

Existe un creciente riesgo de caídas entre personas de la tercera edad, siendo la fractura de cadera en una caída lateral uno de los peores resultados. Esta investigación presenta un modelo dinámico de bajo costo computacional basado en el método del elemento finito para simular una caída sobre la cadera y determinar la fuerza que recibe el fémur, así como su atenuación bajo diversas condiciones del tejido blando. El modelo propuesto incluye el fémur, tejido blando y una masa efectiva que representa la porción de masa corporal que influye en el impacto. Los resultados muestran un incremento en la fuerza de impacto a medida que disminuye el grosor del tejido blando y conforme aumenta la masa efectiva. Con respecto a la atenuación de la fuerza, ésta incrementa a medida que aumenta el grosor del tejido blando y conforme disminuye la masa efectiva. Expresando la atenuación de la fuerza de impacto por mm de espesor de tejido blando se obtienen valores entre 0.6 y 13.3 N/mm.

Palabras Clave: Caída de lado, Simulación por elementos finitos, Tejido blando trocantérico, Masa efectiva, Atenución de la fuerza.

ABSTRACT

An increased risk of fall-related events exist among elderly people, where hip fracture in sideways falls is one of the most critical outcomes. This investigation presents a low-cost dynamic finite element model to simulate a sideways fall on the hip in order to determine the force the femur receives and the impact force attenuation provided by the trochanteric soft tissue. The proposed model includes the femur, soft tissue and an effective mass that represents the body mass portion (associated with the trunk, head and upper limbs) that influences the impact. The results show an increase in the impact force as the soft tissue thickness decreases and as the effective mass increases. Regarding the impact force attenuation, an increase is observed as the soft tissue thickness increases and as the effective mass decreases. Expressing the impact force attenuation per mm of soft tissue thickness, values between 0.6 and 13.3 N per mm are obtained.

Keywords: Sideways fall, Finite element simulation, Trochanteric soft tissue, Effective mass, Force attenuation.

Nomenclature

STT  Soft tissue thickness
m   Body mass
h   Body height
me  Effective mass
F_imp Peak impact force
F_imp,a Analytical impact force
F_imp1 Peak impact force with one contact pair
F_imp2 Peak impact force with two contact pairs
F_femur Force received by the femur
F_att Force attenuation provided by the soft tissue
%F_att Force attenuation as a percentage
F_att,STT Force attenuation per mm of STT

1. Introduction

One of the events most associated with elderly people are falls, which become more common as people get older. This is of great interest given that the distribution of population in industrialized countries has changed towards older age groups [1]. Hip fracture is one of the most severe fall-related injuries, with a greater occurrence in elderly women [2-5]. This type of fracture is related to sideways falls [6,7], and it is associated with a significant deterioration in the quality of life [8]. For these reasons, diverse investigations have focused on determining how hip fractures takes place, which involves knowing the bone strength, as well as determining the forces and stresses generated during the impact.

The finite element method (FEM) is a technique that has become increasingly used to analyze the effect of an impact due to a human fall. Several investigations can be found in the literature that use FEM to simulate this phenomenon,
since it allows flexibility in the implementation of the impact conditions, such as modifying the body configuration and the geometry of the body parts involved during the impact.

The investigations that use a FEM-based dynamic analysis can be differentiated by the body parts included in the model and by the way they are represented. For instance, Zhang et al. [9] presented a 3D model that only included the femur, which was used to investigate how different hip fracture types are affected by the impact directions. This study showed that falling posture was an important factor leading to different types of fracture. On the other hand, Dahlgren [10] used a full body model (THUMS) to quantify the contact force between a variety of different flooring systems and the soft tissue, as well as the force that the femur receives. In this work beam elements were used to represent the body parts, besides adding their mass through point masses. Majumder et al. [11] simulated hip fracture in sideways falls using a 3D finite element model of pelvis–femur–soft tissue complex with simplified representation of the whole body. To incorporate the inertia effect, the whole body was represented by a spring-mass-dashpot system.

One of the most important parts to represent in a human fall model is the trochanteric soft tissue, since it influences the force the femur receives. The trochanteric soft tissue thickness was found to be the most dominant parameter for hip fracture under sideways falls, surpassing body height and body weight [12]. In addition, Majumder et al. [13] concluded that as the trochanteric soft tissue thickness increased, the possibilities for hip fractures would reduce non-linearly. Other important consideration is to incorporate the correct body orientation during the impact, given that it directly affects the impact zone, as well as the portion of the body mass that influences the impact (associated with the trunk, head and upper limbs).

Two main limitations are identified as the model gets more complex. One is the difficulty to replicate and control the body configuration in the impact, which involves the correct implementation of boundary conditions. The other is the high computational cost, which increases as the model gets more complex.

In the literature it is more common to find works focusing on determining the impact force \( F_{imp} \) (contact force between the floor and soft tissue), compared to the force the femur receives \( F_{femur} \). A work that stands out in the literature is the experimental investigation of Robinovitch et al. [14], where the force attenuation is related to the soft tissue thickness \( (STT) \), and a value of 71 N per mm of \( STT \) was proposed. Nevertheless, to determine this value, all tests involved a pendulum effective mass of 44 kg and an impact energy of 140 J. In order to determine the force the femur receives, first the force attenuation \( (F_{att}) \) provided by the soft tissue is obtained by multiplying the value of 71 N/mm times a proposed soft tissue thickness value \( (F_{att} = 71 * STT) \), then this value of force attenuation is subtracted from the impact force to get the force the femur receives \( (F_{femur} = F_{imp} - F_{att}) \).

In the present investigation a simplified Finite Element (FE) model that includes the femur, soft tissue and the portion of body mass that influences the impact (effective mass \( m_e \)) is proposed. This model allows to simulate a sideways fall with low computational cost, and it is used to quantify the impact force attenuation \( (F_{att}) \), and as a percentage \( (%F_{att}) \), as well as to evaluate the force received by the femur \( (F_{femur}) \) due to variations in the \( STT \) and some anthropometric parameters (body mass \( m \) and body height \( h \)). Based on anthropometric data that correspond to a wide range of female population (60-69 years old), it was possible to express the force attenuation per mm of \( STT \) \( (F_{att,mm}) \). In addition a straightforward alternative to relate the anthropometric parameters to the effective mass was proposed.

The relevance of the proposed model lies in that it can facilitate the design of attenuation surfaces, it can be used to understand the effect of the soft tissue thickness on the impact force, and it may allow analyzing the resulting forces and stresses in order to understand how fracture develops during the impact.

2. Methods

This section consists of two parts, the first part describes the FE model used to simulate a sideways fall and the second part describes the methodology followed to obtain the force attenuation.

2.1. FE model

This subsection presents the simplified dynamic FE model used to simulate the hip impact during sideways fall, which was implemented in the ANSYS/LS-DYNA® software. In the model two forces will be identified, the impact force (contact force between the floor and soft tissue) and the force received by the femur, where the last one will result from the attenuation of the impact force due to the presence of soft tissue.

![Finite element mesh of the model to simulate the sideways fall.](image)

2.1.1. CAD Model

For the representation of the femur a standard model was used, which was obtained from computed tomography scans [15,16]. The femur was positioned in the same configuration that it takes just before the impact during a sideways fall, and it corresponds to the orientation used in experimental tests to determine the bone strength [17-19], where the femoral shaft is positioned at 10° with respect to the horizontal, and the femoral neck axis at 15° of internal
rotation. The proposed model includes the femur, soft tissue and an effective mass (Fig. 1). In order to save computational resources only portion of the soft tissue was mainly included in the impact zone and along the femur in the lower region. Additionally, a floor was included to represent the impact surface, which was defined as rigid (non-translational and non-rotational).

2.1.2. Material properties

The material properties of the trabecular bone, cortical bone and soft tissue are shown in Table 1. The properties are taken from the work of Majumder et al. [11,13], where a range of values is presented for trabecular bone due to its anisotropic nature, however it is decided not to assign heterogeneous properties, thus facilitating implementation. The highest values were used, overestimating bone stiffness, possibly resulting in a lower limit of force attenuation. In Table 1 $E$ is the Young's modulus, $\rho$ is the density, $v$ is the Poisson's ratio, $E_t$ is the tangent modulus and $\sigma_y$ is the yield stress. Although the floor is modelled as rigid, Table 1 also shows the properties assigned to it, which are required for calculating the contact stiffness, and that correspond to rough values of low strength concrete.

<table>
<thead>
<tr>
<th>Parts of the model</th>
<th>Type of model</th>
<th>Properties</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trabecular bone</td>
<td>Bilinear isotropic</td>
<td>$\rho=0.042-0.541$ g/cm$^3$, $E=32\text{-}3340$ MPa, $\sigma_y=0.354-40$ MPa, $v=0.2$, $E_t=0.032\text{-}3.34$ MPa</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>Bilinear isotropic</td>
<td>$\rho=1.8$ g/cm$^3$, $E=22700$ MPa, $\sigma_y=158$ MPa, $E_t=1135$ MPa, $v=0.3$</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>Mooney-Rivlin</td>
<td>$\rho=0.749$ g/cm$^3$, $C_1=85.5$ kPa, $C_0=21.38$ kPa, $v=0.495$</td>
</tr>
<tr>
<td>Floor</td>
<td>Rigid</td>
<td>For contact stiffness: $\rho=2.38$ g/cm$^3$, $E=22.1$ GPa, $v=0.15$</td>
</tr>
</tbody>
</table>

2.1.3. Mesh properties

For the creation of the FE mesh two types of elements were used, the SOLID164 (eight-node element) and the MASS166 (point element). The SOLID164 element was used for meshing the whole model, while the MASS166 element was used to apply $m_e$.

Based on a mesh sensitivity analysis elements of uniform size of 0.003 m were assigned to the cortical bone, trabecular bone and soft tissue, while the mesh size was set to 0.005 m in the floor.

2.1.4. Boundary conditions

The model considers the actual hip impact velocity of 3.17 m/s [20], as well as the effect of gravitational acceleration. In addition, node-to-surface contact elements were included between the soft tissue and floor, with a friction coefficient of 0.5 [11,21], and initially in the first analysis the femur and soft tissue are considered rigidly connected. Other important considerations are the simulation time (40 ms) and the implementation of the mass scaling option to reduce the CPU time. The simulation time was enough to capture the peak impact force ($F_{imp}$) between the floor and soft tissue, and the step size (time increase) was set at $4\times10^{-4}$ ms, resulting in a mass increase of 0.35%. As a result, it was possible to reduce the simulation time from an initial time of 9.6 h to approximately 1.65 h.

In order to represent the movement of the femur during the impact, special attention was given to the rotation of the knee joint [18,19]. For this purpose, some nodes were vertically constrained in the joint (Fig. 2(a)) and the displacement was allowed horizontally [18]. It is important to point out that care was taken to avoid over-constraining or under-constraining the joint movement.

As discussed above, it is necessary to include in the model the body mass portion ($m_e$) that influences the impact, which is transmitted through the contact surface between the femur and the acetabulum. This mass was placed in some nodes of the outer surface of the femoral head with the use of lumped masses (MASS166 element) of equal magnitude (Fig. 2(b)).

2.2. Methodology to obtain the force attenuation percentage $\%F_{att}$ and the force the femur receives $F_{femur}$

In the model two forces will be identified, the first one corresponds to the impact force (contact force between the floor and soft tissue), while the second one corresponds to the force received by the femur (Fig. 3). This second force will result from the attenuation of the impact force due to the presence of soft tissue.

![Figure 2](image2.png)

![Figure 3](image3.png)
This subsection will first describe the procedure to find the relationship between $F_{\text{imp}}$, $m_e$, and $STT$, and an alternative is proposed to find the relationship between the $m_e$ and the anthropometric parameters $m$ and $h$, since no general expression exists to relate them. Subsequently, equations that depend on $m$, $h$ and $STT$ will be proposed to determine the $\%F_{\text{Att}}$ and $F_{\text{femur}}$.

2.2.1. Relationship between effective mass, impact force and anthropometric parameters

Using the FE model previously described, first the impact force $F_{\text{imp}}$ is obtained for the proposed values of $m_e$ (7 to 18 kg) and $STT$ (30 to 65 mm), as shown in Table 2, where each of these simulations took about 4 hours of CPU time. The proposed thickness values for the soft tissue correspond to 7 to 18 mm, as shown in Table 2.

The results show that as the $STT$ increases $F_{\text{imp}}$ decreases, while $F_{\text{imp}}$ increases proportionally with the value of $m_e$. This information will be used later to find the $m_e$ that corresponds to a specific $F_{\text{imp}}$ and $STT$, for which it will be necessary to interpolate between the data. Also from Table 2 it can be obtained that the $F_{\text{imp}}$ attenuates by about 18% by increasing $STT$ from 30 to 65 mm.

### Table 2 – Impact force (N) for different values of effective mass and soft tissue thickness.

<table>
<thead>
<tr>
<th>$STT$ (mm)</th>
<th>7</th>
<th>9</th>
<th>11</th>
<th>13</th>
<th>15</th>
<th>17</th>
<th>18</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>4425.8</td>
<td>4938.3</td>
<td>5613.0</td>
<td>6264.8</td>
<td>6818.2</td>
<td>7295.4</td>
<td>7454.4</td>
</tr>
<tr>
<td>35</td>
<td>4297.9</td>
<td>4850.3</td>
<td>5439.5</td>
<td>6043.7</td>
<td>6636.3</td>
<td>7112.5</td>
<td>7286.2</td>
</tr>
<tr>
<td>40</td>
<td>4155.9</td>
<td>4723.7</td>
<td>5342.0</td>
<td>5913.7</td>
<td>6447.5</td>
<td>6917.0</td>
<td>7064.7</td>
</tr>
<tr>
<td>45</td>
<td>3994.3</td>
<td>4578.9</td>
<td>5146.4</td>
<td>5737.3</td>
<td>6272.0</td>
<td>6701.0</td>
<td>6941.8</td>
</tr>
<tr>
<td>50</td>
<td>3886.7</td>
<td>4459.7</td>
<td>5009.7</td>
<td>5589.0</td>
<td>6109.9</td>
<td>6609.2</td>
<td>6833.8</td>
</tr>
<tr>
<td>55</td>
<td>3780.7</td>
<td>4326.2</td>
<td>4843.5</td>
<td>5358.5</td>
<td>5904.7</td>
<td>6439.4</td>
<td>6693.3</td>
</tr>
<tr>
<td>60</td>
<td>3736.7</td>
<td>4238.9</td>
<td>4711.4</td>
<td>5213.1</td>
<td>5714.6</td>
<td>6189.0</td>
<td>6428.8</td>
</tr>
<tr>
<td>65</td>
<td>3688.8</td>
<td>4132.6</td>
<td>4583.9</td>
<td>5048.3</td>
<td>5549.8</td>
<td>5978.8</td>
<td>6223.4</td>
</tr>
</tbody>
</table>

The following expression that depends on some anthropometric data ($m$ and $h$) will be used as the reference impact force and it will be linked to the proposed model through the effective mass and the resulting simulated impact force [20,23]:

$$F_{\text{imp}_{\text{a}}} = \sqrt{\frac{g}{k}} \left(0.51 h \left(0.37 m\right)\right)$$

where $k$ is the effective stiffness of the body measured in pelvis release experiments, with values of 71060 N/m and 90440 N/m for women and men respectively [24]. Note that this equation already includes any effect or influence of any part of the body on the impact force, and these are included in the proposed model through the effective mass.

In order to find the $m_e$ that corresponds to the proposed data of $m$, $h$ and $STT$ of a person under study, it is considered that the value of $m_e$ assigned to the model is the required one so that $F_{\text{imp}_{\text{a}}}$ is obtained in the FE model ($F_{\text{imp}_{\text{a}}} = F_{\text{imp}}$). The procedure would be first to use the proposed $m$ and $h$ to evaluate eq. (1) to determine $F_{\text{imp}_{\text{a}}}$. Then, assuming that $F_{\text{imp}_{\text{a}}}=F_{\text{imp}}$, $F_{\text{imp}}$ and $STT$ are used in Table 2 to find the corresponding value of $m_e$.

2.2.2. Determination of the impact force attenuation and the force received by the femur

This part of the paper proposes expressions to quantify the percentage of the impact force attenuation ($\%F_{\text{Att}}$) provided by the soft tissue for different values of $m_e$ and $STT$. These equations will be used in a case study to determine the range of $\%F_{\text{Att}}$ that corresponds to some values of $m_e$ and $STT$.

In order to obtain the force received by the femur ($F_{\text{femur}}$), located between the femur and soft tissue, and subsequently $\%F_{\text{Att}}$, additional node-to-surface contact elements are incorporated to the model, also with a friction coefficient of 0.5. It is important to note that the $F_{\text{femur}}$ values that result from incorporating this second contact pair differ from those found with a single contact pair in Table 2. Due to this variation in the results, in order to differentiate $F_{\text{imp}}$ either with one or two contact pairs, from now on they will be referred to as $F_{\text{imp}1}$ and $F_{\text{imp}2}$ respectively. Once this has been clarified, $\%F_{\text{Att}}$ is obtained as follows

$$\%F_{\text{Att}} = \frac{\left(F_{\text{imp}2} - F_{\text{femur}}\right)}{F_{\text{imp}2}} \times 100$$

where the difference between $F_{\text{imp}2}$ and $F_{\text{femur}}$ turns out to be the impact force attenuation ($F_{\text{Att}}$). Note that with the above clarification, the impact force values in Table 2 correspond to $F_{\text{imp}1}$.

### Table 3 – Percentage of impact force attenuation ($\%F_{\text{Att}}$).

<table>
<thead>
<tr>
<th>$STT$ (mm)</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>10</th>
<th>13</th>
<th>15</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>6.9</td>
<td>7.5</td>
<td>5.2</td>
<td>5.6</td>
<td>4.7</td>
<td>6.8</td>
</tr>
<tr>
<td>35</td>
<td>10.2</td>
<td>8.1</td>
<td>7.1</td>
<td>5.6</td>
<td>4.4</td>
<td>6.1</td>
</tr>
<tr>
<td>40</td>
<td>11.9</td>
<td>10.9</td>
<td>8.8</td>
<td>6.7</td>
<td>7.2</td>
<td>7.3</td>
</tr>
<tr>
<td>45</td>
<td>15.6</td>
<td>12.4</td>
<td>12.1</td>
<td>10.0</td>
<td>8.4</td>
<td>6.7</td>
</tr>
<tr>
<td>50</td>
<td>17.8</td>
<td>15.3</td>
<td>14.6</td>
<td>11.0</td>
<td>7.1</td>
<td>7.6</td>
</tr>
<tr>
<td>55</td>
<td>19.1</td>
<td>16.3</td>
<td>15.4</td>
<td>12.7</td>
<td>8.3</td>
<td>6.6</td>
</tr>
<tr>
<td>60</td>
<td>21.7</td>
<td>20.0</td>
<td>18.8</td>
<td>13.8</td>
<td>9.4</td>
<td>7.7</td>
</tr>
<tr>
<td>65</td>
<td>23.3</td>
<td>23.6</td>
<td>21.7</td>
<td>15.2</td>
<td>10.3</td>
<td>8.4</td>
</tr>
</tbody>
</table>

Table 3 shows values of $\%F_{\text{Att}}$ for different values of $STT$ and $m_e$. It can be observed that $\%F_{\text{Att}}$ increases proportionally with the $STT$, while $\%F_{\text{Att}}$ decreases as $m_e$ increases. In addition, it can also be observed that as $STT$ increases, a wider range of values is obtained for $\%F_{\text{Att}}$. For instance, with $STT=30$ mm the range of values for $\%F_{\text{Att}}$ is 2.8% (7.5% – 4.7%), while with $STT=65$ mm the range is 15.2% (23.6% – 8.4%). On the other hand as $m_e$ increases, a...
smaller range of values for $\%F_{At}$ is obtained. For instance, with $m_e = 5$ kg the range of values for $\%F_{At}$ is 16.4% (23.3% – 6.9%), while with $m_e = 15$ kg the range is 2.3% (8.4% – 6.1%). Because of this, it could be expected that by further increasing $m_e$, the values of $\%F_{At}$ would tend to stabilize towards a constant value.

![Graph](image)

**Figure 4** - (a) slope $c_1$; (b) Y-intercept $c_2$ for different values of $m_e$ in order to approximate $\%F_{At}$.

In order to smooth the data in Table 3 and to facilitate obtaining $\%F_{At}$, these were linearly approximated for each value of $m_e$ by the following equation

$$\%F_{At} = c_1 \cdot STT + c_2 \tag{3}$$

where $STT$ is in mm, $c_1$ is the slope and $c_2$ is the Y-intercept. The values of $c_1$ and $c_2$ for different values of $m_e$ are shown in Fig. 4 to facilitate its interpretation. As seen in Figs. 4(a)-(b), two zones can be identified where the data can be approximated, being the first zone $5 \leq m_e \leq 7$ kg and the second one $7 \leq m_e \leq 15$ kg. Linear approximations to these data are shown in Table 4, and they will be used to facilitate obtaining the parameters $c_1$ and $c_2$ required in eq. (3), in order to determine the $\%F_{At}$ that corresponds to a specific $STT$. Note that in Fig. 4 at $m_e = 7$ kg there is a drastic change in the behavior of the data ($c_1$ and $c_2$). Higher force attenuation values will result for values of $m_e$ below 7 kg, and the drastic change may be due to fact that for these very low values of $m_e$ the $STT$ turns out to be excessive.

**Table 4** – Force attenuation percentage ($\%F_{At}$).

<table>
<thead>
<tr>
<th>Interval</th>
<th>Slope, $c_1$ (Fig. 4a)</th>
<th>Y-intercept, $c_2$ (Fig. 4b)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$5 \leq m_e \leq 7$</td>
<td>$c_1 = 0.4633$</td>
<td>$c_2 = -1.4670 \cdot m_e + 1.1497$</td>
</tr>
<tr>
<td>$7 &lt; m_e \leq 15$</td>
<td>$c_1 = -0.0524 \cdot m_e + 0.8329$</td>
<td>$c_2 = 1.7312 \cdot m_e - 21.6057$</td>
</tr>
</tbody>
</table>

It will be verified in the next section with a case study that the proposed values for $m_e$ cover a wide range of anthropometric parameters, and it is very unlikely that values of $m_e$, even lower than 7 kg could be required. On the other hand, care must be taken when extrapolating beyond $m_e = 15.895$ kg, since negative values of $c_1$ would be obtained, which is not physically possible. Further discussion on the latter point will be given in the conclusions.

Finally, based on $\%F_{At}$ and $F_{imp}$, the force received by the femur can be obtained as follows

$$F_{femur} = F_{imp} \left(1 - \frac{\%F_{At}}{100}\right) \tag{4}$$

Next, a case study is presented where the aforementioned equations are applied.

### 3. Case study

The present section proposes a case study to show the procedure to follow on how to use the tables and expressions presented above, in order to determine the range of values of $\%F_{At}$ and $F_{femur}$ for anthropometric data that correspond to an interval of 70 % (centered around the median) of the women’s population with ages between 60 and 69 years [25]. The resulting data ranges from 154.7 cm to 166.7 cm for $h$, and from 55.5 kg to 86.8 kg for $m$. Regarding the $STT$, values of 35, 50 and 65 mm were proposed, as mentioned above, based on the work of Bouxsein et al. [22]. These values of $STT$ and the limit values of $h$ and $m$ are used in Table 5. First $F_{imp}$ (column 4) is obtained by eq. (1), then this force and $STT$ are used to interpolate $m_e$ using Table 2, which is used to determine the ratio of $m_e/m$ as percentage ($\%F_{At}$). Subsequently, $\%F_{At}$ was obtained by means of eq. (3) using $m_e$ and $STT$. This required determining the $c_1$ and $c_2$ coefficients by making use of the linear approximations shown in Table 4. Finally, the last column shows the force received by the femur ($F_{femur}$), which was obtained using eq. (4).

For the proposed anthropometric parameters the results in Table 5 show a range of values for $m_e$ between 8 and 18 kg. In addition the results show that $\%F_{At}$ varies approximately between 3 and 13%, which means that most of the impact force is transmitted to the femur, with values between 4000 and 6000 N.

**Table 5** – Results of the case study.

<table>
<thead>
<tr>
<th>$h$ (m)</th>
<th>$m$ (kg)</th>
<th>$STT$ (mm)</th>
<th>$F_{imp}$ (N)</th>
<th>$m_e$ (kg)</th>
<th>$%F_{At}$</th>
<th>$m_e/m$</th>
<th>$F_{femur}$ (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.667</td>
<td>86.8</td>
<td>35</td>
<td>6169.9</td>
<td>13.5</td>
<td>6.2</td>
<td>4.9</td>
<td>5789.7</td>
</tr>
<tr>
<td>1.667</td>
<td>55.5</td>
<td>35</td>
<td>4933.6</td>
<td>9.3</td>
<td>16.7</td>
<td>6.6</td>
<td>4608.4</td>
</tr>
<tr>
<td>1.547</td>
<td>86.8</td>
<td>35</td>
<td>5943.6</td>
<td>12.7</td>
<td>14.6</td>
<td>6.2</td>
<td>5572.5</td>
</tr>
<tr>
<td>1.547</td>
<td>55.5</td>
<td>35</td>
<td>4752.7</td>
<td>8.6</td>
<td>15.6</td>
<td>6.7</td>
<td>4436.3</td>
</tr>
<tr>
<td>1.667</td>
<td>86.8</td>
<td>50</td>
<td>6169.9</td>
<td>15.2</td>
<td>17.6</td>
<td>6.5</td>
<td>5769.0</td>
</tr>
<tr>
<td>1.547</td>
<td>55.5</td>
<td>50</td>
<td>4933.6</td>
<td>10.7</td>
<td>19.4</td>
<td>10.5</td>
<td>4416.0</td>
</tr>
<tr>
<td>1.547</td>
<td>86.8</td>
<td>50</td>
<td>5943.6</td>
<td>14.3</td>
<td>16.5</td>
<td>7.3</td>
<td>5510.5</td>
</tr>
<tr>
<td>1.547</td>
<td>55.5</td>
<td>50</td>
<td>4752.7</td>
<td>10.0</td>
<td>18.1</td>
<td>11.1</td>
<td>4224.6</td>
</tr>
<tr>
<td>1.667</td>
<td>86.8</td>
<td>65</td>
<td>6169.9</td>
<td>17.8</td>
<td>20.5</td>
<td>2.8</td>
<td>6000.0</td>
</tr>
<tr>
<td>1.667</td>
<td>55.5</td>
<td>65</td>
<td>4933.6</td>
<td>12.5</td>
<td>22.6</td>
<td>11.6</td>
<td>4362.8</td>
</tr>
<tr>
<td>1.547</td>
<td>86.8</td>
<td>65</td>
<td>5943.6</td>
<td>16.8</td>
<td>19.4</td>
<td>4.3</td>
<td>5086.6</td>
</tr>
<tr>
<td>1.547</td>
<td>55.5</td>
<td>65</td>
<td>4752.7</td>
<td>11.7</td>
<td>21.2</td>
<td>12.9</td>
<td>4141.3</td>
</tr>
</tbody>
</table>

In a similar way to what was done in Table 5, more intermediate values of $h$, $m$, and $STT$ were evaluated, and the results were grouped in Fig. 5 to understand the relationship between $\%F_{At}$ and $STT$. It can be observed that smaller values of $\%F_{At}$ are obtained as $m_e$ increases, which can be attributed to the strong dependence on the energy involved in the fall. It can also be observed that for the maximum value of $m_e$ (15 kg) the slope of the line is almost zero, meaning that the $\%F_{At}$ is almost independent of the $STT$ with an approximate value of 6%.

Processing the results of Fig. 5 and expressing the impact force attenuation per mm of $STT$ ($F_{At/mm}$), the values of 13.3,
9.9, 5.6 and 0.6 N/mm were obtained for $m_e$ values of 9, 11, 13 and 15 kg respectively. The corresponding energy values of the simulated falls range from 45 to 75 J.

Based on these results, another alternative is proposed to determine the force the femur receives

$$F_{\text{femur}} = F_{\text{imp1}} - STT F_{\text{Att} = 50}$$ (5)

Additionally, in eq. 1 it was presented the expression to determine the impact force analytically, where it can be noted that the anthropometric parameter that dominates the most is $m$, for this reason this parameter was picked to relate it to $m_e$. Such relationship is shown in Table 5 as the ratio $\%m/m$, which was used to propose another alternative to establish a straightforward procedure to relate the anthropometric parameters to $m_e$.

Regarding the last point, it could be anticipated that by further increasing $m_e$, the $\%F_{\text{Att}}$ would stabilize towards a single constant value regardless of the $STT$, because the soft tissue would reach its maximum energy absorption capacity. However, the proposed model does not capture this behavior, so later this is addressed when talking about the limitations of the model.

In addition, with the use of the case study, a small range of $\%F_{\text{Att}}$ (3 - 13 %) was obtained for anthropometric data that cover a large percentage of the female population (60-69 years old). Consequently, it can be anticipated that in a falling event it will be difficult to get a force attenuation percentage greater than 13 %, since the energy of the fall exceeds the energy absorption capacity of the soft tissue. Expressing the impact force attenuation per mm of STT, values of $F_{\text{Att},m}$ between 13.3 and 0.6 N/mm were obtained for values of $m_e$ between 9 and 15 kg respectively.

Finally, the authors would like to point out some possible causes of the difference between the force attenuation value of 71 N/mm [14] provided by the soft tissue and the values reported in this work (maximum of 13.3 N/mm):

- This investigation considers separately the influence of $STT$ on $F_{\text{imp}}$ and on $F_{\text{Att}}$. The force attenuation percentage $\%F_{\text{Att}}$ does not include the effect of $STT$ on $F_{\text{imp}}$, since the FE model implicitly includes its effect on $F_{\text{imp}}$. On the other hand, the value of 71 N per mm considers jointly the influence of $STT$ on both $F_{\text{imp}}$ and $\%F_{\text{Att}}$.
- A different configuration of the femur is used. The experiment presented by Robinovitch et al. [14] does not use the configuration of a sideways fall.
• The impact zone in the experiment [14] is larger than that existing in a sideways fall, resulting in a greater impact force attenuation of the soft tissue.

• In the work of Robinovitch et al. [14] a single energy value (140 J) was considered to quantify the force attenuation value of 71 N/mm. This represents a limitation in its use given that, as already stated, the $\% F_{Att}$ provided by the soft tissue depends on the specific energy involved in a fall. In our investigation, the maximum value of impact energy considered is 75 J.

The limitations of the present investigation are the following:

• Trabecular bone was assumed to be bilinear, elastoplastic, isotropic and homogeneous although it is highly anisotropic. This could be one of the causes of the low force attenuation values, since the highest values of the property range were assigned, which increases the stiffness of the femur.

• The model can be improved considering the bone as heterogeneous and dependent on the apparent bone density.

• In Figure 5 it is shown the force attenuation percentage as a function of soft tissue thickness for different values of $m_c$, and we can note that as $m_c$ increases the slope of the lines decreases, being almost zero for a value of $m_c$ $=15$ kg. This means the soft tissue is reaching its maximum energy absorption capacity.

• Using values of $m_c$ higher than 15.895 will result in a negative slope, meaning that force attenuation percentage will decrease by increasing STT, which is not physically possible. Therefore, the use of values higher than this should be avoided, since the model does not capture the correct physics probably due to a contacts issue.

• The response of the model depends on the effective mass included in the model, which has a direct relationship with $m$ and $h$, but a one-to-one relationship cannot be explicitly stated.

• The model is tuned to the impact force analytical expression proposed by van den Kroonenberg et al. [20], which does not consider the influence of the STT.

5. Conclusions

This investigation presented a low-cost finite element model to simulate a sideways fall on the hip in order to determine the force the femur receives and the impact force attenuation provided by the trochanteric soft tissue. The results show an increase in the impact force as the soft tissue thickness decreases and as the effective mass increases. With respect to the impact force attenuation, an increase is observed as the soft tissue thickness increases and as the effective mass decreases. In addition, a small range of $\% F_{Att}$ (3 - 13 %) was obtained for anthropometric data that cover a large percentage of the female population. Expressing the impact force attenuation per mm of soft tissue thickness, values between 0.6 and 13.3 N per mm were obtained, which are low compared to the most used value reported in the literature. Nevertheless, our study best agrees with a real sideways fall.

The relevance of the proposed model lies in that it can facilitate the design of attenuation surfaces, it can be used to understand the effect of the soft tissue thickness on the impact force, and it may allow analyzing the resulting forces and stresses in order to understand how fracture develops during the impact.

Acknowledgements

This work has been conducted under the financial support of CONACYT during the Master’s Degree studies of the first author.

REFERENCES